The Effects of Loaded Squat Exercise Variations on Tri-Planar Hip Net Joint Moments and Deep Gluteal Requirements in Resistance Trained Females

by

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Abstract

Gluteus medius and minimus (hereafter, the *deep gluteals*) contribute to everyday movement by providing both supportive and propulsive actions, such as during gait. Deep gluteal weakness has been associated with chronic low back pains, knee pains, and abnormal skeletal development. Due to these associations, physical training may be required to maintain or enhance their function. The objectives of this cross-sectional investigation were to: (i) identify the loaded squat exercise(s) that generated the greatest hip net joint moments (NJMs) in planes associated with primary deep gluteal actions in flexed hip postures to facilitate extrapolations to potential muscle utilizations, and (ii) identify a potential relationship between ground reaction forces (GRFs) and hip NJMs to be used as approximation method for estimating frontal and transverse plane hip NJMs. Tri-dimensional lower limb NJMs, GRFs, and kinematics were collected during four resisted squat variations in female participants to satisfy these objectives. Exercises were selected based on their potential to require substantial frontal and/or transverse plane hip NJMs, as the deep gluteal muscles are postulated to be primary contributors to both planes. The bilateral squat (BS) was expected to have substantial transverse plane NJMs; the forward split squat (FSS), lateral split squat (LSS), and single limb squat (SLS) were hypothesized to have large frontal plane NJMs. Nineteen female participants who could BS ≥80% bodyweight attended three sessions which consisted of exercise familiarization and/or five repetition-maximum (RM) testing and/or data collection. Findings revealed significant between-exercise differences in all three planes for both hip NJMs and GRFs. Specifically, the SLS elicited the greatest frontal (abductor) and transverse (internal rotator) plane NJMs of any variation. In addition, tri-planar hip NJM strategies varied the least during the FSS, LSS, and SLS; these exercises exhibited hip NJM polarities associated with the deep gluteal muscles.

Medial GRF magnitudes did not parallel frontal and transverse plane NJM magnitudes as hypothesized, indicating that GRF alone are not sufficient to predict hip NJMs in hip flexed postures. Further investigations should seek to improve upon methodological inconsistencies, investigate causes of presented NJM strategy patterning, and study the longitudinal effects of squat variations on deep gluteal strength.

Preface

This thesis is an original work by Zachary Fielding. The research project, of which this thesis is a part, received research ethics approval from the University of Alberta Research Ethics Board, titled "HOW DO VARIATIONS OF THE FREE WEIGHT SQUAT EXERCISE ALTER THE DEMANDS OF THE HIP AND DEEP GLUTEAL MUSCLES?", study identification Pro00107448, on March 4th, 2021.

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1.0 Introduction

The deep gluteal muscles (gluteus medius and gluteus minimus) are strong hip abductors and internal rotators that play critical roles in activities of daily living. These roles include supporting and moving the trunk during walking, running, and stair climbing[1,2]. Their primary function is traditionally ascribed to gait, where deep gluteal contraction generates frontal and transverse plane net joint moments (NJM) at the hip to prevent excessive pelvic motion and facilitate forward progression[3]. Deep gluteal weakness has been correlated with chronic and pregnancy low back pain[4,5], knee pain[6], iliotibial band syndrome syndrome[7], and abnormal femoral development[1]. Therefore, determining optimal exercises for strengthening the deep gluteal muscles is critical for developing rehabilitative, preventive and performance training programs.

Current deep gluteal exercise recommendations for rehabilitative[8] and performance[9] goals are based off electromyography (EMG) investigations. Though a potentially valid measure of muscular electrical activity, EMG is an indirect and commonly invalid measure of muscular effort on its own[10,11]. Current evidence also fails to address the inversion of muscular actions of the deep gluteal muscles with progressive hip flexion[12–15], limiting the scope of *'feasible'* exercises.

The current thesis seeks to address these limitations and build upon our understanding of the role of the complex three-dimensional hip musculature in resisted squatting. Tri-planar (analogous to three-dimensional) hip NJM strategies (computed using inverse dynamic solutions) opposed to EMG, will be used to extrapolate to deep gluteal requirements during exercises. Both inverse dynamic and EMG methodologies have their limitations, but the limitations of EMG (particularly with the deep gluteal muscles during dynamic exercises [see *2.4.2 Limitations of EMG Techniques*]), may outweigh those of the muscle indeterminacy[16] and co-contraction[17] limitations of inverse dynamics. The current thesis extrapolates NJM findings to *potential* deep gluteal utilization by assuming (i) the deep gluteals are the primary abductors and internal rotators of the hip (posture dependent; see *2.2.0 Deep Gluteal Actions* and *2.2.1 Deep Gluteal Functions*), and (ii) muscles are the primary contributors to internal NJMs[16].

In addition, squatting posture allows for the safe addition of external resistance, mimicking real-world training environment intensities. High sagittal (flexion-extension) hip range of motion (ROM) can also be achieved while squatting[18], facilitating exploration of the inversion of action of the deep gluteal muscles. Finally, as inverse dynamic solutions consider externally applied forces (i.e., ground reaction forces [GRFs])[16,17], which may vary considerably due to squat-specific demands such as foot stance, exploring this relationship can yield insight into hip NJM deterministic factors.

1.1 Purpose

The purpose of this thesis is to develop a better understanding of how resisted squat variations effect tri-planar hip NJMs; particularly, that of the frontal and transverse planes, as these correspond to the deep gluteal muscles *'dominant'* moment-generating planes. Exploration of the hip NJM – GRF relationship during resisted squat exercises seeks to better understand the underlying mechanical attributes that modulate hip NJMs, and thus, potentially deep gluteal requirements.

1.2 Hypotheses

Two primary hypotheses were tested: (i) the bilateral squat (BS) exercise will generate the greatest internal hip rotator NJM (transverse plane) across resisted squat variations, and (ii) the single limb squat (SLS) will generate the greatest hip abductor NJM (frontal plane) across resisted squat variations. As the deep gluteal muscles are some of the dominant internal rotators in deep hip flexion[13,15], and dominant abductors in neutral hip postures[14,15], support for these hypotheses *may suggest* the BS and SLS require deep gluteal contributions, supporting future muscle modeling and/or training investigations.

The secondary hypothesis is non-directional, and states that medial GRFs are associated with hip abductor and internal rotator NJMs – irrespective of the squat variation.

2.0.0 Literature Review

2.1.0 Pelvic and Deep Gluteal Anatomy

One of the greatest methodological gaps the current thesis seeks to address, is harmonizing measured outcomes with a thorough understanding of the underlying anatomical structures in non-anatomical postures. Researchers may often become myopic with their findings and fail to consider the scalability to complex, three-dimensional, co-related structures. The following section paints a more detailed picture of gluteal muscle structural properties, which act as a foundation for understanding commonly ascribed muscle actions. These muscular actions are the basis for selection of the included squat variations and corresponding NJM anatomical planes of interest.

2.1.1 Osteology of the Hip

The hip is a unique structure as it's truly the abstraction of six bones (sacrum, coccyx, ischium, pubis, and femur)[2,19] and four joints (sacrococcygeal, sacroiliac [SI], pubic symphysis, and acetabular-femoral)[20]. The pelvis (hip excluding the femur), provides attachment sites for muscles and ligaments, facilitates locomotion, supports and protects abdominal tissues, enables childbirth[21], and transmits force from the torso and upper extremity, to the ischial tuberosities during sitting and femur during standing[2]. Majority of hip bones are tightly fused facilitating minimal movements (e.g., right and left pubes forming the pubic symphysis)[2,19]. The acetabular-femoral joint is the exception, as the three bones that comprise the *os coxae* (*pelvis*; includes: ilium, ischium, and pubis)[2], articulate with the large femoral head.

The structure of the acetabular-femoral joint facilitates large femoral sagittal (flexion-extension) ROM and tri-planar NJMs. The cup-shaped acetabulum houses the large, nearly perfectly spherical femoral head, covered in articular cartilage and reinforced with a dense labrum[22]. The acetabulum is reinforced on the posterior-superior aspect, allowing the joint to handle high joint contact forces associated with walking and during hip flexed positions (e.g., squatting)[22]. It is the discrepancy between the diameter of the femoral head and neck that give rise to the large flexion-extension ROM, as the neck is only 3/4^{ths} the diameter of the head; allowing a greater ROM prior to labral impingement[22]. Compounding size geometrical discrepancies, only 2/5^{ths} of the bony acetabulum is in contact with the femoral head at any position – furthering the femurs sagittal ROM[22]. In a state with complete muscle relaxation and least labral tension, the femur sits in 10° of extension, abduction, and external rotation[22].

The proximal aspect of the femur is comprised of the femoral head, neck, and shaft[2,19]. At the most inferior-lateral aspect of the neck, two large bony prominences (*trochanters*), are present: the lesser trochanter (directed *medially*), and the greater trochanter (directed *laterally*)[19]. The latter serves as a common distal bony anchor site for the deep gluteal and quadriceps coxa (piriformis, gemelli, and obturator internus) muscles[19]. Abnormalities in femoral neck structure such as coxa brevis can directly reduce the mechanical efficacy of the deep gluteal muscles[23].

2.1.2 Deep Gluteal Anatomy

The gluteal complex is comprised of three muscles (from superficial to deep): gluteus maximus (gmax), gluteus medius (gmed), and gluteus minimus (gmin)[2,19]. The term *'deep gluteal'* (an aggregate synonym representing gmed and gmin as one) is used in the current thesis due to their actions being quite similar[14,15] and the traditional *'primary hip abductor'* terminology failing to consider their *inversion of actions* in postures outside of anatomical position[15].

Gmed originates in a relatively more superior position, running in-close proximity to the iliac crest as it spans from the posterior superior iliac spine[24] and sacroiliac ligaments[25] anteriorly, to the anterior superior iliac spine (ASIS)[24]. Fascial origins along the gluteal aponeurosis have also been detailed[25]. Gmin arises much more inferiorly along the dorsum (*back*) of the iliac wing, spanning fore between anterior and inferior gluteal lines at heights between anterior iliac spines[24–27].

All descriptions of gmed's insertion detail a termination at the femoral greater trochanter (GT)[22,24,25,28–31]. Due to the large size of the GT, exact locations of gmed termination varies between reports. Lateral[25], anterior-superior[24], anterior and posterior[31], posterior-superior and anterior[29], and the apex of the GT[30] have been previously reported. Flack et al.[25] further detailed termination of gmed fibres with both the piriformis and gmin muscles. It's unlikely these variations in reports have clinically meaningful effects as demonstrated by marginal changes in deep gluteal frontal plane leverages when the *entirety* of the GT was displaced in computer models[32].

Flack et al.[25] described gmin's distal flat aponeurotic tendon which few (0-19) fascicles inserted to the superior-anterior aspect of the hip joint capsule. The remaining gmin fascicles arch around the apex of the GT to an anterior termination on the GT. Majority of primary and anatomical texts also detail a hip capsule and GT distal insertion [2,27,29,33,34]. Gmin's hip capsule insertion draws similarities to the glenohumeral joint's rotator cuff muscles; noting their capacity to retract the capsule during movement to prevent impingement[2,34]. Additionally, the smaller capsular termination may allow tightening of the capsule to increase robustness of the femoral head in the acetabulum[29].

Description of gmed fibre geometry is highly heterogenous, with reports of two[30], three[24,26,31] or four[25] numbers of distinct compartments. Flack et al.[25] reported the most detailed fibre descriptions, noting the most anterior fascicles have the greatest volume and 20° fibre orientation running posterior-inferiorly. Traveling from the most anterior to the most posterior fibres, the angle of insertion decreased to a maximum of -39° running in the anterior-inferior direction[25]. Similarly, Gottschalk et al.[24] and Al-Hayani et al.[26] noted that the middle and anterior fibres ran vertically from insertion to origin, with the posterior fibres running parallel to the femoral neck. This suggests the most anterior gmed fibres may play larger roles in hip abduction (neutral hip posture) and internal rotation (flexed hip posture), as vertical fibres would lead to greater leverages in these planes. The more horizontal, posterior gmed fibres, may then be responsible for the large joint compression forces ascertained to gmed[1], as this fibre orientation would minimize muscle leverages and maximize linear resultant forces (i.e., compression)[16].

Al-Hayani et al.[26] described anterior and posterior gmin compartments, in which the former contained fibres running vertically between the ilium and GT, and the latter, horizontal fibres. Beck et al.[27] reported similar gmin fibre orientations though no distinct compartments, noting that in 90° of hip flexion, all fibres ran straight from ilium to the femur (suggesting a large transverse plane moment arm). Flack et al.[25] described four compartments based on fascicle geometry and nerve patterning, though none differed significantly in volume or physiological cross sectional area (PCSA).

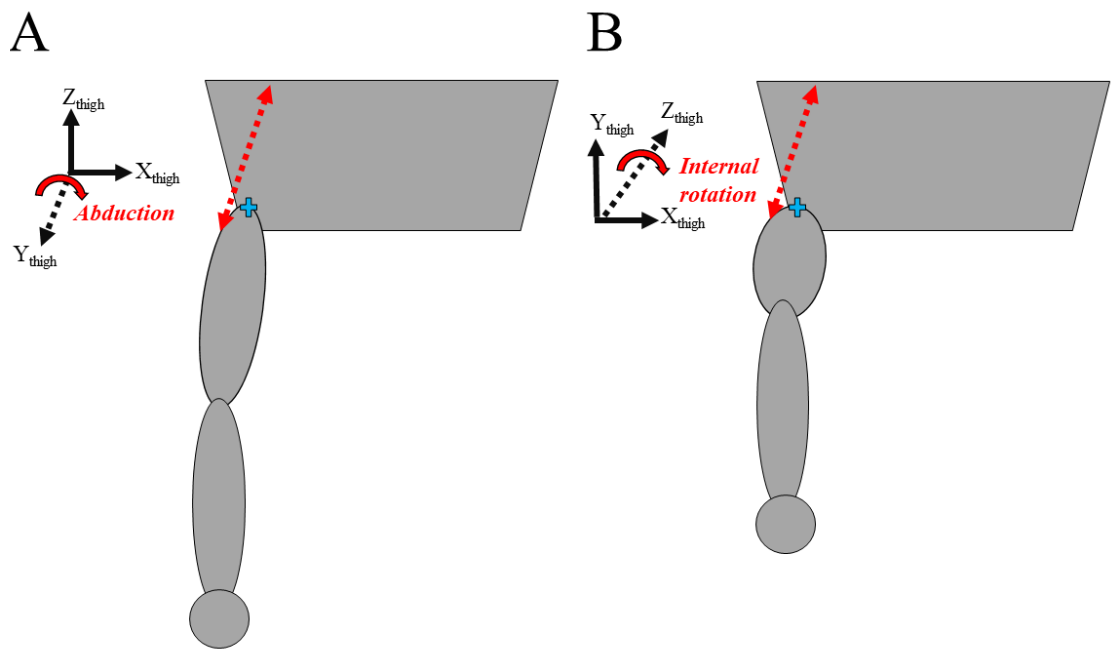
2.2.0 Deep Gluteal Actions

A muscles actions are a direct result of its anatomical construct, as its leverages are the product of its tri-planar orientation relative to the joint(s) it crosses. When a muscle develops tension (either due to stretch or contraction), it can produce two types of forces across all joints which it crosses: compression forces and rotational forces (*joint moments*), the latter of which is responsible for moving limbs[2]. What determines the relative magnitude of each is dependent on a muscle's tri-planar moment arm. Smaller moment arms reduce a muscles contribution to limb movement[2], acting more to push/pull/resist dislocation via it's linear action[2,19] (these muscles are sometime referred to as *shunt* muscles[19]). As the linear compressive forces are not relevant to the current thesis, they will not be discussed further.

A moment is the product of the applied force (realized at the origin, insertion, or other structural interactions), and the shortest perpendicular distance between the muscle's line of action and joint centre(s)[2]. The latter describes a *moment arm*, and is the primary mechanical property used to ascribe muscle actions. Muscles can posses' moment arms in each plane (sagittal, frontal, and transverse) – though whether their respective magnitudes are sufficient to noticeably contribute to joint moments, varies with the muscle.

The magnitude of a muscle's tri-planar moment arm can have substantial impact on its contribution to movement. If a muscles force is held constant, increasing moment arm magnitude will equivalently increase the resultant joint moments. Hip muscle moment arms are particularly dynamic due to the acetabular-femoral joints large tri-planar ROM[2]. Dynamic moment arms result in dynamic, posture-dependent muscular actions. Nineteen muscles that cross the hip joint change actions in at least one of three planes during tri-planar thigh movement due to changing moment arms[15]. Even when thigh motion is restricted to the sagittal plane, at 40° of thigh flexion *four* muscles have flipped frontal plane moment arm polarities (rel. to a 20° hip extended position)[14]. The transitioning of muscular actions with changing moment arms is referred to as the *inversion of muscular actions*[12]. An example of this using a simplified gmed segment is presented in Figure 1. Note, the example provided in Figure 1 is just one way a muscle can undergo inversion of muscular actions (desynchronised adjacent local coordinate system rotation), but this was selected as the deep gluteal action inversion is of direct relevance to the current thesis.

Figure 1. Inversion of muscular action using a single simplified two-dimensional gluteus medius segment



A = a neutral hip posture results in a simplified gluteus medius segment (hashed double-arrowed red line) abducting the thigh via clockwise rotation about the thigh's embedded local coordinate system, y-axis; B = when the hip is flexed, the simplified gluteus medius segment's line of action does not change relative to the hip joint centre, but as the thigh's local coordinate system has rotated (as it's embedded in the thigh segment), contraction results in an internal rotation rather than abduction due to the rotated local coordinate system; blue cross represents location of emedded thigh local coordinate system

Only two studies have quantified tri-planar hip muscle moment arms: Dostal et al.[14] reported moment arm magnitudes in a neutral hip position and provided insight on the effects of hip flexion up to 90°, and Pressel and Lengsfeld[15], who computed *muscle relative moments* up to 130° of hip flexion. A relative muscle moment refers to the percentage of the total moment a muscle can produce about a joint in the corresponding plane[15]. For example, an 80% relative flexor moment means 80% of the moment the muscle generates will contribute to a flexor moment. Pressel and Lengsfeld did not provide moment arm lengths but did provide a full list of muscle action changes and the corresponding joint angle. Both Dostal et al. and Pressel and

Lengsfeld used a straight-line model which comes with its share of limitations, explored in subsequent sections.

In a neutral hip position, all gmed partitions posses a hip abductor moment arm that ranges from 6.7 to 4.3 cm from the most anterior to the most posterior segments[14]. Gmin partitions follow suit with smaller magnitudes, ranging from 5.8 to 3.9 cm[14]. Pressel and Lengsfeld's findings support strong deep gluteal abductor moment arms in neutral postures, as authors reported relative abductor moments of 89% and 94% for gmed and gmin, respectively[15].

In the transverse plane, both gmed and gmin follow a similar pattern: the anterior segment has a modest internal rotator moment arm, a trivial moment arm magnitude in the middle, and equal-and-opposite moment arm in the posterior (relative to anterior)[14]. Pressel and Lengsfeld reported marginal external rotator relative muscle moments, though authors failed to consider gmed and gmin as polysegmental all while both underwent an inversion to internal rotator moment arms at *only* 10° of hip flexion[15] (this may signify potential differences in how authors defined a *'neutral'* hip position). Delp et al. segmented gmed and gmin into four and three partitions, respectively[13]. Authors reported the most anterior segments as internal rotators, with successive posterior segments displaying greater external rotator capacities. Intriguingly, none of the hip muscles were identified as primary or secondary internal rotators by Dostal et al. (secondary defined as a moment arm ≥ 50% of the primary)[14]. The threshold would have had to been lowered to *35%* for a muscle to be classified as a secondary internal rotator in the neutral hip posture[14]. This is supported by the maximum ~19% relative transverse plane muscle moments across internal rotators, with most residing < 10%[15]. According to Dostal et al., the entirety of gmed as well as the posterior gmin are marginal to trivial hip extensors[14].

Deep gluteal muscle moment arms undergo substantial changes as the hip is flexed. The large frontal plane moment arm of the anterior gmed segment is nulled by 90° of hip flexion, while internal rotator capacity increases in parallel[14]. The same trend was found by Pressel and Lengsfeld, which reported both gmed and gmin flipping to adductors at 100° of hip flexion[15], and Delp et al., which reported mean moment arm changes of 6.0 and 4.2 cm for gmed and gmin, respectively[13]. Frontal to transverse plane deep gluteal patterning with hip flexion has also been demonstrated during gait[35]. Anterior gmed segment flipped from a primary hip abductor to primary hip internal rotator when traveling from an extended/neutral hip posture, to points of maximal flexion (maximal hip flexion angle of ~25°)[35]. These findings suggest the inversion from primary abductor (frontal plane) to primary internal rotator (transverse plane) may occur in much shallower hip flexion angles than those demonstrated by Dostal et al.[14] and Pressel and Lengsfeld[15].

At 90° of hip flexion, muscles with abductor moment arms are limited to the quadriceps coxa, tensor fasciae latae, sartorius, and rectus femoris[14,15,36]. The quadriceps coxa (*'four-headed muscle of the hip'*) includes the piriformis, gemelli, and obturator internus. Vaarbakken et al. demonstrated the quadriceps coxa muscles moment arm patterning parallel one-another with progressive hip flexion, resulting in substantial abductor moment arms and minor extensor and internal rotator capacities[36]. TFL, sartorius, and rectus femoris abductor capacity is currently unknown, yet their modest flexor moment arms[14] may render them suboptimal for use during extensor-intensive tasks (e.g., squatting).

Though not a primary focus of the current thesis, the sheer size and proximity of gmax warrants discussion. Gmax has a substantial extensor and external rotator moment arms in neutral - both of which drastically decrease with hip flexion[13,37]. At 90° of hip flexion, the majority of gmax segments posses an internal rotator moment arm[13]. Frontal plane actions in neutral are dependent on the distal fibre terminations; gmax fibres inserting into Gerdy's tubercle via the iliotibial tract, are abductors, in direct contrast to the adductor fibres of those terminating into the gluteal tuberosity[15]. Both segments demonstrate abductor moment arms at 70° of hip flexion[15].

In summary, neutral hip posture abduction is the primary result of the deep gluteal muscles. In deeper magnitudes of hip flexion, the quadriceps coxa and gmax may become the primary abductors. Hip internal rotator capacity in neutral is likely the weak combination of anterior gluteal segments, iliacus, and psoas major[15]. Gmed, gmin, and majority of gmax (Gerdy's tubercle terminating fibres) are presumably the primary internal rotators in deep hip flexion (≥90°).

Dostal et al.[14] and Pressel and Lengsfeld[15] determined muscle moment arms using *straight-line* models, which though computationally simpler, has its limitations. The straight-line model does not consider the three-dimensional reality of the bag-of-bones and other tissues our muscles reside and interact within. For example, muscle bulging due to contraction could alter an adjacent muscles orientation, altering its moment arm. Gmax likely occurs the most *wrapping* due to adjacent tissues as it's the most superficial posterior coxa muscle and the inferior fibres must arch about the prominent ischial tuberosity and associated tissues. Additionally, both Dostal et al.[14] and Pressel and Lengsfeld[15] did not consider muscle stretch. Despite a muscle possessing a large moment arm, if the corresponding joint angle stretches the muscle to a disadvantageous force-generating length (e.g., over-stretched sarcomeres placing it on the descending limb of the force-length relationship[38]), it's unlikely the muscle can make practically meaningful contributions to the NJM.

2.2.1 Deep Gluteal Functions

The deep gluteal muscles are substantial in size, comprising of 80% of the total hip abductor cross sectional area[3], with gmed ranking third in largest physiological cross sectional area of the lower extremity[39]. Their unique low *fibre-to-muscle* length ratio suggests they produce greater force over a narrower fibre excursion window[40]. In regards to having the greatest potential to produce the frontal plane moments in slight hip extension/neutral[41], it's no surprise the primary role of the deep gluteal muscles is often ascribed to maintaining pelvic frontal plane equilibrium during gait[3].

Frontal plane demands at the hip during gait are mechanically best described as a 1st class lever system (*see-saw*)[2]. A first-class lever system is comprised of two forces acting in the same direction yet yielding *opposing* moments, as they reside on opposite sides of fulcrum (axis of rotation)[2]. Residing on the medial side of the see-saw is the strong downward pull of gravity realized at the systems centre of mass (COM); the point-mass in which all mass of the system is distributed equally[16], located at approximately the umbilicus in anatomical position[42]. Countering this strong gravitational rotation is the downward pull of the deep gluteal muscles, residing on the lateral side of the fulcrum (i.e., the hip joint centre [HJC]). During gait, the weight of the body and the pull of the deep gluteal muscle are in constant opposition – a battle best visualized with Trendelenburg sign. Trendelenburg sign is characterized by excessive pelvic sagging towards the unloaded limb during single limb support[2], demonstrating the inability of the deep gluteal muscles to counter the strong downward gravitational forces.

At the hip, the centrally located gravitational forces generate an *external* hip adductor moment. The strong laterally located downward pull of the deep gluteal muscles on the pelvis serve to counter this adductor moment by generating an *internal* hip abductor moment. External and internal in this case refer to the location of the moment-generating force (gravity being external and muscles being internal)[2]. Yet this is not a fair battle, as the COM moment arm can be >2.5-fold larger than the deep gluteal moment arms, requiring countering muscular forces greater than 2x body mass[3,12]. For example, a 70 kg person with a frontal plane COM moment arm of 0.168 m (based on gmed values reported by Dostal et al.[14]), would require the gmed to generate 1722.4 N of force to counter 686.7 N of gravitational pull.

The battle between gravitational and deep gluteal forces is an oversimplification as it fails to consider dynamic attributes such as inertial and accelerative segment properties. This has led to concerns regarding the validity of this 'static' model[43]. Though it's not a goal of this thesis to investigate these concerns, acknowledging the potential for inaccuracies in the underlying theory is critical for comprehensive and just interpretations of the findings.

Prior research has demonstrated the deep gluteal muscles play major supportive and propulsive roles in gait. The *support moment* was first introduced by Winters[44], who stated the sum of extensor moments at the ankle, knee, and hip, must be a net extensor NJM to prevent collapse during gait[44]. As it considers the *net* lower-limb NJM, extensor moments at one joint can compensate for insufficiencies at another[44]. Vertical ground reaction forces can be reduced to muscular contributions to support during gait as muscles have been demonstrated to be the dominant contributors[45]. During early single limb stance, the deep gluteals generate a supportive force alongside gmax and vastus lateralis; by midstance, the deep gluteals are the sole muscular support[45]. The critical supportive role of the deep gluteal muscles is further corroborated by gmed ranking third largest contributor to vertical COM accelerations (greater than that of gmax)[46]. The deep gluteals also act as a medial-lateral propulsor, trivially contributing to anterior-posterior ground reaction forces[46]. At all phases in gait, John et al.[47] identified the abductors (gmed, gmin, and tensor fasciae latae), as the primary contributors to medial-lateral GRFs.

At a normal walking speed, peak medial-lateral COM excursion is approximately 1.3 percent body height[48]. As 40% of the gait cycle is spent in single limb stance[2], cyclical changes in base of support area require constant medial-lateral COM control to prevent falling. Medial-lateral GRFs then directly control frontal plane COM stability in gait. As the *'hip abductors'* are the primary generator of medial-lateral GRFs in all phases of gait[47], the deep gluteal muscles serve as frontal plane propulsors during gait.

Intriguingly, increases in gait speed results in substantial decrements in medial-lateral COM excursions despite trivial changes in medial-lateral COM velocities[49]. Lulic and Muftic found an 81% decrement in peak medial-lateral COM excursion from normal to fast walking speeds[48]. The dichotomous changes of excursions and velocities may explain why Liu et al.[46] found gmed to be the only muscle to decrease its contributions to vertical COM acceleration with increases in gait speed. Stagnated velocities at constant mass require no further increases in forces to counter momentum (medial-lateral COM excursion is a sinusoidal curve which requires ***redirecting*** the COM towards the swing leg during midstance[48,49]). Increasing gmed forces could result in undesired medial-lateral COM perturbations (i.e., overcompensation) due to its major contribution to medial GRFs; thus, requiring greater contributions from other *vertical support* musculature with reduced medial GRF generating capacity (e.g., vastii, soleus)[46,47].

The deep gluteal muscles play substantial roles in controlling our vertical and horizontal movements during gait. GRFs have proven to offer great insight into muscle functions during gait. The potential of GRF analysis during squatting tasks has been mostly left unexplored. This has left a large gap in our understanding of the deep gluteal muscle function – one in which the current thesis begins to address.

2.3.0 Exploring GRFs During Squatting

Forces generated primarily by muscles throughout the body can travel down the lower extremity as intersegmental forces[16]. These forces are then applied to the ground through the foot and can be segmented into three orthogonal components. When a force is applied to an object, an equal and opposite force is applied back to the original force propagating object in accordance with Newton's third law. In biomechanics, this opposing force is referred to as the ground reaction force (GRF) and is commonly assessed using a tri-axial force platform[16].

A medial GRF (the force vector pointing towards the midline of the body), may in theory scale positively with hip abductor moments, as the abducting moment would *push* the lateral aspect of the foot away from the body's midline. As the deep gluteals are situated optimally for generating abductor moments in a neutral hip posture, their major contribution to medial GRFs at all points of the gait cycle[47] support this theory. Hip internal rotator moments may also produce medial GRFs, as internal rotation could also laterally push the foot in this hip posture.

Gmed's substantial contribution to vertical GRF is particularly intriguing. Anderson and Pandy[45] did specify if it was the posterior gmed segment which had the greatest contributions to vertical GRFs. Posterior gmed segments do have the largest extensor moment arms, though compared to frontal plane leverages, are quite small[14]. As the majority of gmed's moment capacity is abductor[15] despite contributing the greatest to vertical GRFs in midstance[49], the lower extremity must be spatially-oriented in such a way to facilitate vertical GRF production from hip abductor moments. A thigh ***adducted*** posture during gait could satisfy both abductor moments generating vertical GRFs and facilitation of optimal medial-lateral COM excursion.

In both neutral and flexed hip postures, the deep gluteals have the potential to be the largest contributors to medial GRFs during squatting. As the GRF is a three-dimensional vector, both anterior-posterior and vertical GRFs must also be simultaneously considered for accurate interpretations. Additionally, thigh segment adduction angles need to be considered in parallel to better elucidate the deep gluteal-GRF relationship.

2.4.0 Review of Current Deep Gluteal Strengthening and Tri-Planar Hip NJM Research

Previous research purposed at quantifying deep gluteal effort during exercise can be categorized into two groups based on research methodologies: electromyography (EMG) and muscle modeling (e.g., static optimization models). Majority of the literature is of the former type. Though EMG was not implemented in the current thesis, it's majority stake in the literature warrants discussion. The following sections are non-systematic reviews of studies quantifying acute deep gluteal efforts during weight-bearing exercise using EMG and/or muscle modeling techniques. In addition, studies quantifying tri-planar hip NJMs during weight-bearing movements are included as they serve as pivotal comparisons to the current thesis' findings by bearing the greatest similarity in methodologies. Only weight-bearing movements have been included to remain relevant as possible to the current thesis. Tables 1 to 4 present the findings from all included investigations.

2.4.1 EMG Investigations

EMG studies commonly report outcomes as a percentage of maximum voluntary isometric contraction (MVIC). This process allows for the normalization of findings, facilitating comparison between subjects and (to a lesser degree), across studies[11]. Thus, a 100% MVIC means the EMG activity during the exercise was the same as during a maximal isometric contraction, typically performed in a posture that's thought to best elicit the action of that muscle.

Table 1 presents nine weight-bearing resistance exercise studies which quantified gmed surface EMG[50–58]. Except for Felício et al.[51], all samples had some resistance training background. Three studies utilized female-only samples[51,54,57], three male and female[53,55,56], and three male-only[50,52,58]. Seven studies used a barbell[50,52–56], one a loaded backpack[51], one unilaterally held dumbbells[58], and one did not report their means of resistance[57]. Intensities ranged from 10-repetition maximums (RM)[52] to 4 RMs[50]. A single investigation utilized bodyweight-relative intensity (25%)[51].

Table 1. Review of previous deep gluteal EMG findings during weight bearing external resistance exercises

Study	Intensity	Exercise(s)	EMG Magnitudes
Eliassen et al.[50]	100% 4 RM	BS SLS (trail limb anterior) SLS (trail limb posterior)	153 ± 17 µV (ascent)[a] 171 ± 18 µV (ascent)[a] 166 ± 20 µV (ascent)[a]
Felício et al.[51]	25% of bodyweight	BS BS (band-resisted abduction) BS (solid object-resisted adduction)	33 ± 27 %MVIC[a] 47 ± 20 %MVIC[a] 59 ± 22 %MVIC[a]
Marchetti et al.[52]	100% 10 RM	FSS (traditional) In-line FSS (feet 50% hip width apart)	~ 55-60 %MVIC[b] ~ 55-60 %MVIC[b]
Mausehund et al.[53]	100% 6-8 RM	Rear-foot elevated split squat SLS (step down from box) FSS	~ 50-60 %MVIC[c] ~ 80 %MVIC[c] ~ 40-50 %MVIC[c]
McCurdy et al.[54]	85% 3 RM	BS Rear-foot elevated split squat	57.85 ± 79.94 mV[c] 72.17 ± 81.37 mV[c]
Muyor et al.[55]	60% 5 RM	SLS Forward lunge Lateral step up	~ 60 %MVIC[a] ~ 30 %MVIC[a] ~ 30 %MVIC[a]
Reece et al.[56]	80% 1 RM	BS (no band) BS (light band) BS (extra heavy band)	~ 80-90 %MVC (ascent)[c] ~ 80-90 %MVC (ascent)[c] ~ 80-90 %MVC (ascent)[c]
Simenz et al.[57]	100% 6 RM	Step up Lateral step up Diagonal step up Crossover step up	69.57 ± 16.75 %MVIC (ascent)[a] 54.66 ± 13.77 %MVIC (ascent)[a] 65.87 ± 15.73 %MVIC (ascent)[a] 76.47 ± 23.40 %MVIC (ascent)[a]
Stastny et al.[58]	100% 5 RM	FSS (ipsilateral load) FSS (contralateral load) Forward lunge (ipsilateral load) Forward lunge (contralateral load)	49 ± 27 %MVIC (descent)[a] 46 ± 23 %MVIC (descent)[a] 51 ± 17 %MVIC (descent)[a] 90 ± 22 %MVIC (descent)[a]

RM = repetition maximum; BS = bilateral squat; SLS = single limb squat; MVIC = maximum voluntary isometric contraction; FSS = forward split squat; MVC = maximal voluntary contraction (peak EMG during 80% 3 RM BS); ~ = approximated from graphical representation; [a]average; [b]integrated EMG; [c]peak EMG

The SLSs tended to elicit the greatest activities within investigations[50,53,55]. The exceptionally high BS values reported by Reece et al.[56] is an exception, but these may be inflated as investigators did not quantify a maximal isometric contraction, but the peak EMG during an

80% 3 RM BS set. Forward split squat (FSS) gmed EMG magnitudes trended on the lower side with reported values of ~30%[55], ~55%[52], and 49% MVIC[58]. Investigations which assessed the BS in conjunction with another weight-bearing lower-limb exercise, reported the BS eliciting relatively lower gmed EMG magnitudes[50,54]. Exercises with more pronounced lateral COM excursions elicited gmed EMG intensities of ~30% MVIC[55] and ~76% MVIC[57] during the lateral step up and crossover step up, respectively. Holding a single dumbbell during a forward lunge at a 5 RM intensity elicited the greatest relative gmed EMG activity at ~90% MVIC[58].

Studies investigating the effects of externally applied medial-lateral knee loads reported conflicting results. Felicio et al. demonstrated that both external and internal rotation against a fixed object between the knees while performing a resisted isometric wall squat, elicited greater gmed EMG intensities relative to the no rotation condition, while no differences between rotations were found[51]. Reece et al. found no significant differences between no, light, and heavy resistance band conditions, wrapped about the knees during 80% 1 RM BSs[56].

Inferring global relationships from current deep gluteal exercise EMG research is difficult due to the range in intensities, exercises, and reporting methods. With caution, it does appear the SLS elicits the greatest gmed activity, while the BS the least. Inferring beyond superficial observations is difficult as EMG intensities were typically reported as a mean throughout squat ascent/descent, or a peak at an unknown time-point. For these limitations and more (described in the subsequent section), utilizing inverse dynamics to quantify hip muscle requirements during exercise can offer greater insight and begin to fill the current knowledge gap.

2.4.2 Limitations of EMG Techniques

EMG is a frequently used tool as it requires nominal set up, is minimally invasive[11], and in-expensive relative to three-dimensional motion capture technology. Though appealing for these reasons, EMG comes with its own set of limitations that result in *unknowns* – some of which can be addressed with other tools.

EMG measures the electrical activity generated (in-part) by the motor unit action potentials (MUAP) that traverse the sarcolemma of muscle fibres[11]. These MAUP's are the signal for muscular contraction, traveling down t-tubules, stimulating calcium release, resulting in cross-bridge shortening and force production[38]. Surface EMG is the most frequently used in exercise investigations as it's the least invasive and gives a more *'whole muscle'* perspective[11].

There are limitations to this whole muscle view. In-vivo investigations have demonstrated an optimal surface EMG *depth-of-reach* of 20 mm, with a roughly linear decline in amplitude with increased skin thickness[59]. Computational investigations have demonstrated a maximal 34 mm range for large motor units (8 mm for small) before amplitudes become equivalent to that of noise[60]. Gmed MUAPs must travel through thick muscular (gmax) and fat (subcutaneous) mediums before being arriving at surface electrodes. Though the superior-lateral part of gmed is not always covered by gmax, it is covered in thick subcutaneous fat and fascia – potentially pushing the distance between gmed and surface electrodes beyond optimal. Previous work in cats has demonstrated specimens with increase local fat deposits suffer from much greater crosstalk generated by nearby muscles[61]. The large gmax and subcutaneous fat act as a substantial MUAP generator and conductor, increasing the noise of gmed surface EMG measurements.

EMG signals are also affected by the force-length and force-velocity relationships in skeletal muscle[10]. Longer muscle lengths result in both a reduction in MUAP amplitudes as well as a skew towards lower frequency components[11]. The type of contraction also greatly effects the EMG signal, with eccentric contractions resulting in significantly less amplitudes relative to concentric when displacing the same external resistance[62], as it fails to consider the passive force properties of muscle[38].

Though both neural and mechanical factors during contraction are important for strength adaptations[63], the latter has yet to be explored to the same degree. Adequately addressing this methodological gap is critical as both rehabilitative[8] and performance-oriented[9] deep gluteal training recommendations are based solely on EMG studies. An inverse dynamic methodology is great starting point for reasons that will be explored in the subsequent section.

2.4.3 Inverse Dynamics

Inverse dynamics describes the process of translating external forces (e.g., ground reaction forces [GRF]), segment properties (e.g., moment of inertia), and segment kinematics (e.g., angular velocity), into linear (joint reaction) and rotational (moment) segmental forces[16,64].

As internal NJMs are directly attributed to the forces generated by muscles/ligaments/other tensile-producing structures that cross a corresponding joint[16], muscular requirements can be (in-part) extrapolated from tri-planar NJMs.

Deducing muscular effort from inverse dynamic solutions does suffer from a major limitation; *indeterminacy*[16]. If more than one muscle crosses a joint, attributing proportions of the total NJM to each muscle is impossible without further computations[16,64] (e.g., static optimization[65]). A common solution for muscle indeterminacy is to assume all muscular forces act through one common muscle. This technique is referred to as the *single muscle equivalent*[16], and idealizes the net effect of all forces and resulting moments about a joint to a single actuator. Three idealized single muscles can be computed at each joint that correspond to each of the three NJM axes. Yet, not just muscles generate internal NJMs, as ligaments, capsules, and even adjacent segment collision can contribute[16].

Though precise muscle force values will not be computed due to the added complexity of current modeling approaches[65], inferences regarding 'required muscles' can be made by localizing key muscles in plane-specific moment production. Due to the deep gluteal muscle's frontal and transverse plane majority moment stake, limitations regarding the single muscle equivalent model are assumed to be less impactful.

The inverse dynamic approach is also limited by the inability to distinguish co-contraction[17]. The computed net moment can not have its constituents broken down to determine the magnitude of each polarity that factored into the net result. For example, using the current thesis' approach, it's not possible to determine if a net *adductor* moment is masking a substantial *abductor* moment – one much greater in magnitude than another exercise the elicits a *net* abductor moment. For this reason, the current thesis avoids using NJM magnitudes as an indicator of greater muscular loads across NJMs of the same polarity. In addition, the current thesis avoids the use of absolutes when extrapolating to muscles – instead opting for terminology such as *"greatest potential to require"*, affectively treating NJMs as a *potential* yes/no indicator of muscular contributions.

2.4.4 Hip Kinetics During Weight-Bearing Unloaded Exercises

Table 2 presents six investigations that reported multi-planar hip NJMs during weight-bearing exercises with no additional external resistance (i.e., bodyweight exercises)[66–71]. Five investigations quantified NJMs during the BS[66,68–71], and two during the SLS[67,69].

Table 2. Hip net joint moments and corresponding sagittal joint angle during body weight squatting

Study	Exercise (variant)	Sagittal Hip Angle (°)	Sagittal Hip NJM (Nm/kg)	Frontal Hip NJM (Nm/kg)	Transverse Hip NJM (Nm/kg)
Bagwell et al.[66]	BS	113.0 ± 6.7	-0.56 ± 0.12[a]	0.09 ± 0.17[a]	-0.05 ± 0.10[a]
Khuu and Lewis[67]	SLS (trail limb middle)	59.9 ± 15.1[b]	-1.1 ± 0.52[b]	-0.89 ± 0.16[b]	NR
	SLS (trail limb back)	65.8 ± 12.1[b]	-0.84 ± 0.39[b]	-0.87 ± 0.15[b]	NR
Li et al.[68]	BS (full depth)	102.5 ± 7.37[c]	~ -0.35[c]	~ -0.15[c]	~ 0.15[c] ER
Malloy et al.[69]	BS	106.1 ± 11.8	-1.2 ± 0.2	-0.1 ± 0.1 0.6 ± 0.2	-0.1 ± 0.1
	SLS	94.7 ± 13.1	-2.0 ± 0.5	-1.1 ± 0.3 Null[d]	-0.5 ± 0.1
Slater and Hart[70]	BS	NR	~ -1.1[c]	~ 0.25[c]	NR
Van Houcke et al.[71]	BS	107 (104.6 to 109.4)[c,e]	-0.56 (-0.506 to -0.617)[c,e]	0.22 (0.184 to 0.248)[c,e]	0.12 (0.081 to 0.151)[c,e]

BS = bilateral squat; SLS = single limb squat; all net joint moments (NJM) were explicitly reported as internal (except Li et al. which is assumed); all values are reported as mean ± SD unless otherwise stated; all values are reported from control/healthy experimental groups; negative polarity denotes extensor (sagittal), abductor (frontal), and external rotator (transverse) NJMs; NR = not reported; [a]mean across entire repetition; [b]at peak knee flexion (); ~ = approximated from graphical representation; [c]at bottom of squat; [d]SLS never demonstrated an adductor NJM; [e]mean (95% confidence interval)

Bagwell et al. and Malloy et al. both investigated lower-limb kinematic and kinetic differences between those with cam femoracetabular impingement (FAI) and healthy, age and sex matched controls[66,69]. Remaining studies investigated general hip kinetics during deep BSs[71], the effects of sagittal and frontal plane knee alignment on BS hip kinetics[70], differences in full- and half-BS lower limb kinetics and kinematics[68], and technique and sex differences during the SLS[67]. Though majority of investigations included female participants[66–70], only two reported sex-isolated findings[67,68]. Technique only varied partly between investigations. Of the five BS studies, two used goal depths of 1/3 tibial tuberosity height[66,71], one was self-selected[69], one parallel[70], and one as low as possible[68]. Of the two studies which assessed the SLS movement, one used a self-selected technique[69] while the other assessed three variations which specified the non-weight bearing limb position relative to the weight bearing limb: anterior, middle (non-bearing foot held parallel to load-bearing foot), and back (non-bearing limb knee flexed at 90° and thigh held vertical)[67]. One study assessed limbs as one[67], two assessed the *'preferred limb to kick a ball'*[68,70], one only right limbs[71], and both FAI investigations measured the effected limb[66,69] (though Bagwell et al.[66] never report if limbs were matched in the control analyses). Two studies[67,69] implemented a CODA pelvis[72] which uses a modified Bell method[73,74] for approximating the hip joint centre (HJC). One study[70] used the Bell method[73] while the remainder[66,68,71] did not specify HJC approximation method. Three investigations filtered kinematic and kinetic forces at different frequencies[66–68] (at-least in high-impact movements, this has been demonstrated to result in substantially different hip NJMs)[75], while one did not specify filter characteristics[71]. Slater and Hart[70] were the only to sample kinetic data at frequencies not evenly divisible by the kinematic sampling frequency, which desynchronizes kinematic and kinetic data during kinematic data reduction, prior to computing NJMs.

When excluding graphically estimated values, inter-study sagittal plane NJMs for both the BS and SLS varied greatly, ranging between 0.56 Nm/kg[66,71] and 1.2 Nm/kg[69] in the BS, and 0.84 Nm/kg[67] and 2.0 Nm/kg[69] in the SLS. Peak adductor NJMs ranged between 0.22 Nm/kg[71] and 0.6 Nm/kg[69] in the BS and were substantially larger than peak abductor NJMs (0.1 Nm/kg[69]). Two studies reported external rotator NJMs in the transverse plane during the BS, demonstrating magnitudes ≤0.1 Nm/kg[66,69], which approximated the sole peak internal rotator NJM[71]. Malloy et al. were the only to report transverse plane NJMs during the SLS, demonstrating magnitudes much greater than that elicited during the BS (0.5 Nm/kg [external rotator][69]).

Four studies presented temporal NJM figures[66,68,70,71]. Sagittal plane hip NJMs demonstrated a U-shaped trend, with a peak extensor NJM at the lowest point of the squat[66,68,70,71]. Majority of investigations demonstrated abductor-adductor-abductor trend, starting and ending the repetition with an abductor NJM, with a peak adductor NJM at the bottom. Intriguingly, Li et al.[68] was the only outlier which reported an adductor-abductor-adductor trend in the full-depth BS, though adductor magnitudes at start of ascent were extremely trivial. Intriguingly, the adductor-abductor-adductor NJM trend was inverted during the half-depth BS variant[68], which was more similar to the other included investigations. Li et al. demonstrated a U-shaped transverse plane trend that always demonstrated an external rotator NJM, that peaked at the bottom of the exercise. Bagwell et al. and Van Houcke et al. both demonstrated peak external rotator NJMs half-way through descent and ascent phases[66,71]. Bagwell et al. demonstrated near-null values at the start, bottom, and end of the repetition[66], whereas Van Houcke et al. demonstrated a clear internal rotator peak at the bottom[71]. Transverse plane NJM magnitudes are relatively small in comparison to sagittal and frontal planes, which may result in greater relative influence from differences in factors such as HJC approximation methods[76].

Slater and Hart[70] findings are of particular relevancy to the current thesis. Participants performed three unloaded BS variations: medial knee malalignment (purposeful internal hip rotation), anterior knee malalignment (purposeful anterior tibial rotation, i.e., *knees past toes*), and control (*knees in-line with toes*). Likely due to the cue given during the control condition (*'push your knees out at the bottom')*, the medial- and lateral-maligned squats did not differ much in magnitude of frontal plane excursion (~80 mm v. ~60 mm, respectively)[70]. In the medial maligned squat, hip abductor NJMs were significantly greater during the middle ~1/3 of the squat cycle than the control[70]. This is realized as a much-reduced adductor NJM, as all three variations had similar trends but varying peaks[70]. Hip extensor NJMs were reduced in both maligned squats relative to the control, though scaling of the figure makes discerning these differences difficult (all demonstrating a peak of ~1.0 Nm/kg)[70]. Anterior knee malignment resulted in greater hip adductor NJMs for the majority of descent and tail-end (20%) of ascent[70].

2.4.5 Hip Kinetics During Weight-Bearing Loaded Exercises

Four studies assessed hip kinetics and/ or kinetics during loaded, weight-beating exercises[76–79] (Table 3).

Table 3. Hip net joint moments and corresponding sagittal joint angle during resisted bilateral squats

Study	Intensity (condition)	Sagittal Hip Angle (°)	Sagittal Hip NJM (Nm/kg)	Frontal Hip NJM (Nm/kg)	Transverse Hip NJM(Nm/kg)
Lahti et al.[77]	85% 1 RM (narrow stance)	105.7 ± 6.1	-4.08 ± 0.65	-1.59 ± 0.76	0.85 ± 0.40
	85% 1 RM (wide stance)	107.7 ± 6.2	-4.36 ± 0.74	-1.75 ± 1.04	1.01 ± 0.34
Southwell et al.[78]	80% 1 RM (running shoe)	64.5 ± 20.7	-3.1 ± 0.5	-0.67 ± 0.21 0.46 ± 0.31	-0.30 ± 0.16 0.31 ± 0.12
Sinclair et al.[76]	70% 1 RM (1/4 GT)	86.1 ± 13.1	-1.8 ± 0.4[a] -139.59 ± 29.65[b]	1.0 ± 0.3[a] 77.51 ± 22.43[b]	-0.2 ± 0.2[a] -26.10 ± 13.86[b]
Swinton et al.[79]	70% 1 RM (traditional)	104.3 ± 4.9	-2.6 ± 0.3[a] 256 ± 35[b]	-0.7 ± 0.3[a] -70 ± 30[b]	0.4 ± 0.2[a] 43 ± 24[b]
	70% 1 RM (powerlifting)	112.6 ± 5.8	-2.8 ± 0.3[a] -281 ± 32[b]	-0.9 ± 0.3[a] -94 ± 26[b]	0.5 ± 0.2[a] 55 ± 22[b]

RM = repetition maximum; all net joint moments (NJM) are presumed to be an internal based on the sagittal plane polarity as many authors were not explicit; all values are reported as mean ± SD; negative polarity denotes extensor (sagittal), abductor (frontal), and external rotator (transverse) NJMs; [a]normalized using reported group mean body mass (Sincliar et al. = 79.47 kg[76], Swinton et al. = 100.2 kg[79]); [b]non-normalized data as presented in paper

None of the studies assessed between-sex differences, though half did use heterogenous samples[77,78]. Two studies investigated the effects of squat techniques on hip outcomes. Lahti et al. assessed lower-limb kinetics and kinematics during wide (1.52-fold inter-GT distance) and narrow (0.99-fold inter-GT distance) foot stances during 70% and 85% 1 RM BSs[77]. Swinton et al. investigated biomechanical differences between a traditional (knee-over-toes), powerlifting, and box BSs at 30%, 50%, and 70% 1 RM intensities[79]. One investigation was interested in the effects of shoe style (barefoot, running shoes, and weightlifting shoes) on kinetic and kinematics during an 80% 1 RM BS[78]. Sinclair et al. was interested the effects of divergent HJC approximation methods on lower-limb kinematics and kinetics during 70% 1 RM BSs[76]. In the two studies which did not identify technique as an independent variable[76,78], participants used self-selected/preferred squat techniques and did not report a minimum squat depth. Majority of studies provided incomplete reports of their data processing/modeling procedures. Only Sinclair et al. reported HJC approximation methods[76]. Three investigations reported only filtering kinematic[78,79] or kinetic data[77], while one did not report kinetic sampling frequencies[76]. Only Lahti et al. reported the reference frame in which NJMs were expressed (distal segment)[77]. Peak values were reported for all studies which were either the average between limbs[77–79], or only the right[76]. Only Southwell et al. reported assessing between-limb differences prior to aggregating, which yielded non-significant findings[78]. All studies which reported sampling frequencies[77–79] reported kinetic rates equally divisible by kinematic.

Peak sagittal hip NJMs during the BS show great interstudy variation. The three investigations which assessed the BS at 70 % 1RM intensity, reported peak values ranging from 1.8 Nm/kg[76] to 4.06 Nm/kg[77] (this may be in-part be the result of normalizing NJMs using sample mean masses for two of the investigations[76,79]). Peak abductor NJMs were quite similar across majority of investigations (~0.7 to 1.0 Nm/kg) including both 70%[76,79] and 80%[78] 1 RM intensities. The sole study to report both abductor and adductor values recorded larger abductor magnitudes[78]. Transverse plane peak internal rotator NJMs were similar across majority of reporting studies (0.3 Nm/kg[78] to 0.5 Nm/kg[79]), except for the reported 1.0 Nm/kg peak by Lahti et al.[77]. Studies which reported both internal and external rotator NJMs reported similar magnitudes for opposing polarities[77,78]

Of the studies which reported temporal NJM trends[76–78], all reported similar sagittal plane U-shaped trends, with peak extensor NJMs at the bottom of the BS. In the frontal plane, Sinclair et al. and Southwell et al. reported mostly U-shaped trends, with the start and end of the repetitions marked by abductor NJMs, and a peak adductor NJM at the bottom[76,78]. The mean frontal plane narrow stance trend reported by Lahti et al. never appears to demonstrate an adductor NJM (just an extremely trivial abductor NJM), at the bottom of the BS[77]. The wide stance variation does demonstrate an adductor NJM, though due to scaling of the figure axes, impossible to accurately compare with the other investigations[77]. Transverse plane NJM trends varied the most between loaded squat investigations. Sinclair et al. reported a constant external rotator NJM for all HJC approximation methods[76]. In contrast, Southwell et al. demonstrated a clear sinusoidal trend for all footwear types, starting at a near-null value, followed by a peak internal rotator at approximately half-way through descent, which rapidly switched to an external rotator that peaked at the bottom of the BS[78]. Ascent paralleled that of descent, with an internal rotator peak at approximately half-way, that transitioned to a near-null value at completion of the repetition[78]. Lahti et al. was the only investigation to report differences in temporal hip NJM trends between experimental groups[77]. While no observable differences in sagittal and frontal plane NJMs occurred between wide and narrow BS stances, the transverse plane was marked by a consistent internal rotator NJM, while the wide stance, an external rotator NJM[77].

2.4.6 Gluteal Forces During Weight-Bearing Loaded Exercises

Only one study to date has modeled deep gluteal forces during weight-bearing exercises[80]. Kipp et al. recruited nine male NCAA division I athletes who had at least one year of structured resistance training experience[80]. The BS, step-up, and FSS were assessed at 0%, 25%, 50%, and 70% bodyweight intensities using an Olympic barbell. The authors approximated forces of 11 lower-limb muscles (including gmax and gmed), using a static optimization approach. Compartmentalized forces generated by different gmax and gmed segments, were summed to produce a single resultant force for the gmax and a single resultant force for gmed. Subjects performed the BS to the preferred depth and were asked to best match that depth during the step-up and FSS. Minimal information regarding the inverse dynamics model was reported – thus, it's unclear what HJC approximation method was used and what reference frame/system in NJMs were reported. Though only data from one limb was reported, selection of the assessed limb was never clarified. Peak gluteal force findings demonstrated the BS elicited the lowest gmax and gmed forces. Both FSS and step-up generated significantly greater gmed forces than the BS.

Gmax forces at 75% bodyweight load were significantly greater than 0%, 25%, and 50% intensities for *all* exercises[80]. Gmed forces at the 75% intensity demonstrated identical findings, except during the step-up. Differences in gmed forces during the step-up were significant between 0% & 75%, and 25% & 75%, but not 50% & 75% (though there was a non-significant increase). These findings suggest a *'threshold'* for external resistance during completely unilateral exercises. A potential reasoning for this threshold could be related to the external adductor NJM exceeding the upper limits of the internal abductor generators (i.e., gmed). Once at this threshold, the body may need to compensate in other ways to ensure successful movement completion. For example, one may hip hike or laterally spinal flex to reduce COM-HJC moment arm to reduce the external adductor moment, facilitating greater COM magnitudes with mostly stagnant frontal/transverse plane muscle contributions.

Though temporal trends were not reported, authors did provide normalized force relative to muscle length plots which allow for interpolation. Due to gmax's relatively more posterior origin[19], it's assumed that increases in hip flexion result in linear changes in gmax stretch. For all exercises, gmax produces the greatest forces at the greatest stretches (e.g., the assumed bottom)[80]. The only exception being the step up, which had some minor reductions just prior to maximal gmax stretch. Gmed in contrast demonstrated the opposite relationship, producing its greatest forces at its lowest lengths. This suggests that postures that elongate gmed during squatting (e.g., flexion-adduction-external rotation), may result in suboptimal gmed force production. In contrast, this trend could also mean that postures that correlate with the greatest gmed lengths require minimal abductor and/or internal rotator moment contributions. Total changes in normalized muscle length were similar between gluteal muscles within squatting exercises.

2.4.7 Hip Kinetics During Weight-Bearing Activities of Weekly Living

Understanding tri-planar hip demands during activities of weekly living (e.g., walking, running, and jumping), paints the broader picture of everyday gluteal demands. Four studies[81–84] which reported multi-planar hip kinetics during activities of week living have been included (Table 3).

Table 4. Hip net joint moments and corresponding sagittal joint angle during activities of weekly living

Study	Movement	Sagittal Hip Angle (°)	Sagittal Hip NJM (Nm/kg)	Frontal Hip NJM (Nm/kg)	Transverse Hip NJM(Nm/kg)
Bennett et al.[81]	Walking	24.0 ± 8.9	-0.87 ± 0.21 0.99 ± 27	-0.80 ± 0.22 Null[a]	0.09 ± 0.06 -0.16 ± 0.04
	Running	30.2 ± 10.8	-1.30 ± 0.40 0.87 ± 0.30	-1.40 ± 0.42 0.28 ± 0.12	0.16 ± 0.12 -0.01 ± 0.04
	Single limb landing	41.9 ± 15.9	-3.01 ± 1.64 2.76 ± 1.55	-2.12 ± 0.91 0.81 ± 0.54	0.44 ± 0.21 -0.16 ± 0.14
Costigan et al.[82]	Stair ascent	NR	-0.80 ± 0.24[b]	-0.80 ± 0.12[b]	-0.31 ± 0.07[b]
	Walking	NR	-1.13 ± 0.30[b]	-0.95 ± 0.14[b]	-0.15 ± 0.06[b]
Moisio et al.[83]	Walking (female)	NR	-0.91 ± 0.25[b] 0.65 ± 0.21	-0.87 ± 0.14[b] 0.21 ± 0.11[b]	0.10 ± 0.04[b] -0.15 ± 0.04[b]
Novak and Brouwer[84]	Stair ascent (YDL)	NR	-0.56 ± 0.19 0.11 ± 0.08	-0.61 ± 0.16	NR
	Stair descent (YDL)	NR	-0.23 ± 0.19 0.39 ± 0.14	-0.76 ± 0.14	NR

All net joint moments (NJM) are expressed as internal NJMs; all values are reported as mean ± SD; hip angles are reported as flexion; negative polarity denotes extensor (sagittal), abductor (frontal), and external rotator (transverse) NJMs; Walking is either explicitly stated to be level ground or assumed in the case of Moisio et al.[83]; [a]frontal plane hip NJM never demonstrate an adductor polarity; [b]NJM polarity has been flipped as authors reported external NJMs; NR = not reported; YDL = young, dominant limb condition

All investigations used a sex-heterogenous sample of males and females[81–84], though only one reported sex-isolated outcomes[83]. All investigations used healthy subjects, free of obvious pathology, that would limit independent completion of tasks. Methodological descriptions of three-dimensional models were overall, poorly reported. As Bennett et al. used model parameters as their primary independent variable (HJC approximation method)[81], their descriptions were the most thorough. Kinematic and kinetic data was sampled at different frequencies (8 Hz and 15 Hz, respectively), and five non-functional HJC approximation methods were compared (not including the ¼ GT method used in the current thesis)[81]. Internal NJMs were reported in the distal segment and an X-Y-Z Cardan rotation sequence was used for segment angles. Costigan et al. had the most intriguing approach: both kinematic and kinetic data was sampled at 80 Hz, a much lower value than all aforementioned studies, and motion data was not collected for the foot segment (mass of foot was added to the shank and corresponding moment of inertia properties modified)[82]. HJC was approximated by X-RAY for each participant (centre of the femoral head)[82]. Novak and Brouwer used the same HJC approximation (1/4 GT) and cut-off frequency (6 Hz) as in the current thesis but sampled kinetic data at 1/10th the rate (100 Hz)[84]. Limb(s) included in the analysis varied, with one only measuring the dominant limb (never stated how this was determined)[82], one using both[84], one limiting the reported analysis to the right limb after observing *'comparable'* results[83], and one unclear in their limb analysis method[81].

The largest sagittal plane extensor NJMs were experienced during single limb landing[81], while stair descent elicited the lowest[84]. In walking and stair ambulation, peak abductor values ranged from 0.60 Nm/kg (stair descent)[84] to 0.95 Nm/kg (walking)[82]. Level-ground running and single limb landing elicited much greater hip abductor NJM magnitudes than walking and stair ambulation[81] (Table 3). When reported, hip adductor NJM magnitudes were less than that of the abductor polarity[81,83]. Transverse plane NJMs were much more variable yet did not exceed 0.50 Nm/kg for any of the included activities of weekly living[81–83].

The single limb landing movement described by Bennett et al., is presumably the most closely related to resisted squatting due to the larger vertical axis loading and hip flexion angles[81], relative to the other activities of daily living (Table 3). Participants used a two-step approach while taking-off and landing on one limb[81]. Participants were required to jump to 90% of a previously determined maximum. The single limb landing generated much greater tri-planar hip NJMs relative to both level-ground walking and running[81]. A clear, double-peak abductor NJM is demonstrated during stance phase in both level ground walking, stair ascent, and descent[81]. In single limb landing, a double abductor peak is present, though more subtle. In the traverse plane, a consistent and steady increase in internal rotator NJM is demonstrated – a stark contrast to running's U-shaped internal rotator trend and walking's double-peak (of which both are in opposing directions)[81]. Both frontal and transverse plane trends demonstrate a *peak-and-hold* pattern for abductor and internal rotator NJMs, where peaks were realized at ~50% of cycle and remain constant until completion[81].

2.5.0 Literature Review Summary

The deep gluteal muscles are optimally situated to control frontal plane pelvic motion while also assisting with gait support and propulsion. The deep gluteal muscles demonstrate posture-specific actions due to the *inversion of muscular actions* with hip flexion, which needs to be considered in-order to fully elucidate muscular loading during exercise. Current deep gluteal strengthening recommendations use EMG methodologies, which is limited when assessing gmed and gmin due to their depth and proximity to other tissues. Inverse dynamics offers greater mechanical insights into muscular efforts while facilitating exploration of the properties that elicited said demands. Current weight-bearing, multi-planar hip kinetic exercise data, is limited to the BS and SLS. Sagittal plane temporal NJMs demonstrate similar trends across reports for the BS and SLS, though more varied in frontal and transverse planes. Tri-planar hip NJMs during activities of weekly living can aid in estimating an exercise's deep gluteal requirements, as gait and related activities are often ascribed as their primary function.

3.0 Methods

3.1 Investigation at a Glance

The primary purpose of this investigation was to examine the tri-planar hip NJMs during loaded squat variations with distinct mechanical loading patterns. Hip NJMs are used to extrapolate and theorize potential deep gluteal utilizations due to the presumed dominance of these muscles in the selected anatomical planes. The findings from this investigation will support planning of future studies that utilize muscle force modeling and/or longitudinal training methodologies, and may begin to aid practitioners in exercise selection for utilizing the deep gluteal muscles. The findings will have mostly scientific significance as this thesis begins to narrow the deep gluteal efficacious exercise pool by identifying mechanical measures that may associate with hip NJMs. Findings from the current thesis may also have practical significance due to the various musculoskeletal pathologies correlated with deep gluteal weakness[5–7], though this can not be certain as muscle forces and/or strength adaptations are not being quantified, thus, are hypothetical.

The primary goal of the current thesis is to quantify tri-planar hip NJMs during resisted squat variations, which will be used to approximate deep gluteal utilizations based on known hip muscle leverages. The secondary goal of the current thesis is to investigate the hip NJM – GRF relationship to further understand the stimuli that elicit tri-dimensional hip NJMs and improve exercise selection for both researchers (future investigations) and practitioners (clients).

3.2 Hypotheses

The primary hypotheses are: (i) the BS will generate the greatest hip internal rotator NJM (transverse plane), as the wide base of support (BOS) allows for substantial medial-lateral GRFs, and (ii) the SLS will generate the greatest hip abductor NJM (frontal plane), as the pure single limb stance demands the greatest frontal plane pelvic support. The secondary hypothesis states that medial-directed GRFs will be associated with hip abductor and internal rotator NJMs.

3.3 Overview

The current investigation is a cross-sectional investigation in resistance trained females, to determine the effect of resisted squat variations on hip NJMs. Each participant completed a minimum of three study sessions. Informed consent, exercise familiarization, and 5 RM BS testing occurred during the first session. Further familiarization and 5 RM testing of the remaining exercises occurred during the second session. Data collection, where participants underwent three-dimensional motion and force analysis for all assessed resisted squat variations, occurred on the third and final session.

3.4 Participant Screening

Females with resistance training experience were included and recruited via convivence sampling (social media posts, undergraduate class presentations, word-of-mouth, etc.). A female only sample was used as female's have a higher prevalence of physical activity related back and lower extremity pain[6]; a measure that insufficiencies in deep gluteal strength has been previously correlated with[5,7]. A female-only sample controls for sex-specific differences in hip kinetics, previously demonstrated in bodyweight SLSs[67].

Included participants met the following inclusion criteria: (i) female, (ii) 18 to 50 years of age, and (iii) a current BS 5 RM $\geq$ 80% of body mass. These criterions did not contain all of the original seven-point list approved by the University of Alberta Review Ethics Board. Non-confirmed inclusion/exclusion criterion included: no current lower limb and back injuries, no history of low back and/or hip surgeries, 1 year of resistance training experience, and currently performing resistance training. Due to inexperience of the study's primary investigator (the current thesis' author) and absence of a complete list of inclusion/exclusion criteria on the informed consent form, non-confirmed criterion points were not validated during the initial study session. Despite this, many participants were recruited through pitches to undergrad classes and laboratories (Appendix A1) and social media posts (Appendix A2, A3, A4).

As many of these recruitment mediums did contain most of the inclusion criterion in addition to the *tested* 5 RM BS $\geq$ 80% body mass in the initial study session, it's likely that many participants did meet the full list of criteria despite not being adequately screened. Previous (one year) and current resistance training and no current lower limb and back injuries were selected to reduce risk of injury and familiarization time due to the intensity of the included exercises. No history of low back and/or hip surgeries was selected to mitigate potential effects on hip NJMs as demonstrated in previous, unloaded squat studies[66,69]. Effects of including those who did not meet the origin seven-point inclusion/exclusion criterion list on the primary outcomes, is discussed in the *5.3 Strengths and Limitations*.

Ad hoc distribution analyses (e.g., Lilliefor's Test) revealed non-normality in some of the outcomes. While considering the relatively small final sample size (N = 19) in conjunction with the non-normality findings, an ad hoc decision was made to use non-parametric opposed to parametric statistical tests which a priori sample sizes were based on (N = 22). The final sample size of 19 approximates that of similar investigations[76–79]. Twenty-one participants were recruited over a seven-month period, with two lost to attrition, resulting in 19 participants completing the required three sessions. One participant withdrew from the study due to aggravation of a previous injury believed to have arisen during the following day(s) of session one, which was ascribed to single limb dominant exercise familiarizations, and a second participant due to cessation of communication following session one.

3.5 Selected Resisted Squat Variations

Four loaded squat variations were selected based on their potential to elicit substantial hip abductor and/or internal rotator NJMs. The BS was selected as the wide base of support and the presumably near-equal bilateral vertical limb loading, could facilitate large internal rotator NJMs without greatly perturbing balance. The single limb squat (SLS) was selected as the unilateral limb support could facilitate large hip abductor NJMs in accordance with the static hip model. As even moderate medial GRFs could perturb the system in the SLS, the SLS will offer great insight into the GRF-tri-planar hip NJM relationship. The forward split squat (FSS; synonymous with a non-forward progressing forward lunge), was selected as the split nature could result in similar frontal plane pelvic demands as the SLS, yet with added balance robustness due to trail limb support; potentially facilitating greater intensities and joint ranges of motion. The FSS is predicted to generate the largest anterior-posterior GRFs given the lower-limbs elongated sagittal plane posture. The lateral split squat (LSS; synonymous with a lateral lunge) was selected as large median-lateral plane COM excursions may require equally large medial GRFs (potentially a combination of hip abductor and internal rotator NJMs), to accelerate the COM back to starting position. Similar with the BS, it's not believed the LSS will produce substantial anterior-posterior GRFs given their minimal sagittal plane BOS length. Thus, the BSS, FSS, LSS, and SLS are believed to have unique tri-axial GRF profiles and thus, hypothesized, unique tri-planar hip NJMs. All included exercises were performed with a 20 kg or 15 kg Olympic barbell based on participant preference. Participants began with a 7 kg barbell for the SLS and either progressed to their preferred Olympic barbell, remained at the 7 kg barbell (n=1), or regressed to a 0.5 kg metal bar due to balance difficulties (n=1). The sole participant who regressed to the 0.5 kg bar during the SLS, could not successfully complete three repetitions in succession with the 15 kg barbell. As no minimum intensity for the SLS used as inclusion criterion, this participant was not identified as an outlier. Data was collected for all squat variations at 100% 5 RM intensities.

3.6 Familiarization and RM Testing

Successful study completion required participants to attend three separate study sessions at the biomechanics laboratory.

The first session began with participants providing informed consent, followed by demonstration and familiarization with the less-traditional squat variations (FSS, LSS, and SLS). The BS was not practiced as participants would already be familiar due to the required qualifying intensity minimum. Minimal coaching was provided for all exercises to capture broader techniques and facilitate greater finding generalizability. Successful BS repetitions during 5 RM testing required participants to reach a minimum *thigh-parallel* position, which was visually approximated by a single spotter. Thigh-parallel is a commonly used metric for quantifying a full BS repetition[77,79,85]. Successful FSS repetitions required the trail limb knee (i.e., the minority weight bearing limb) to never contact the ground for safety concerns, as failure during contact had the potential to result in a high load, half kneeling posture, which would be uncomfortable and potentially unsafe. Successful LSS repetitions required majority of the plantar surface of the shoe to maintain ground contact (i.e., purposeful inversion), approximated visually by the spotter. Continued plantar surface contact during the LSS maximized base of support to reduce risk of falls. Successful SLS repetitions required the non-weight bearing limb to be held in a hybrid middle-back position to maximize hip flexion angles[67]. Ground contact of the non-weight bearing limb would deem the repetition unsuccessful, due to loss of single limb mechanics. Outside of these restraints, participants could use their preferred technique (e.g., low- or high-bar, wide or narrow stance, etc.).

A general warm-up prior to familiarization and testing for all sessions consisted of four minutes of static and dynamic movements. Familiarization for the FSS, LSS, and SLS involved three sets of five repetitions at an easy to moderate intensity during session one. Session two familiarization involved two sets of five repetitions at an easy to moderate intensity. Familiarization was always performed on both limbs, acting as a general warm-up prior to 5 RM testing. Only one limb was the lead or measured limb for 5 RM testing and analysis. Lower limb dominance has been traditionally determined by the limb one would use to kick a ball[86]. Though simple, lower limb preference and dominance do not appear synonymous in all contexts[86]. People tend to produce greater forces in their non-preferred limb during single joint tasks, and preferred-dominant limb overlap can be quite variable during multi-joint tasks such as jumping and changing directions[86]. The lead limb used to step onto a raised platform was used to determine the lead/measured limb during all squat variations. The goal was to maintain test simplicity while being task specific. The lead limb used while stepping onto a slightly raised platform (e.g., scale), was assigned the lead/measured limb for *all* exercises.

The first session concluded with a 5 RM BS test. The 5 RM testing protocol was similar for all variations. Participants began with a *very easy* intensity performed for five to eight repetitions for the initial set. Hereafter, five repetitions were performed for each successive set no matter the relative intensity. A set was deemed unsuccessful if a failure to lift the load to starting position or observable technique breakdown occurred. The spotter would take notice of squatting technique as the participant progressed in intensity during 5 RM testing. If thigh angle was noticeably shallower compared to previous sets, the repetition and set was deemed unsuccessful.

Due to the unfamiliarity of performing high load split- and single limb-squats, combined with the equal difficulty of single observer spotting, participants were informed *"if the previous repetition felt difficult, the next repetition feels out of reach, and you've completed only the first few repetitions, you can stop"*. Thus, by definition, some participants did not complete a *true* RM test, which is believed to be a justified trade-off for increased participant safety and confidence.

All sessions were separated by a minimum of 72 hours. There was also a 72-hour minimum separation from any moderate to high intensity workout sessions taken place outside of the study. The 72-hour minimum was believed to be a sufficient window to mitigate fatigue[87].

The second session began with further familiarization and concluded with FSS, LSS, and SLS 5 RM testing. Session two's 5 RM testing was performed in a pre-determined randomized order to mitigate the effects of fatigue at the sample level. Five opposed to one or 10 RMs were implemented, as it maintains relatively high resistance but at intensities that an exercise with a high balance component is more likely to be performed at in a real-world setting.

3.7 Data Collection

The third and final session comprised of data collection. Participants were first outfitted with 26 individual reflective markers (1.0-1.2 cm) and six rigid tracking clusters. Two additional reflective markers were placed on either end of the barbell to track barbell kinematics for a total of 54 reflective markers. All anatomical landmark identification and reflective marker placements were performed by the same researcher. Individual markers were placed bilaterally on the first- and fifth-metatarsal heads, medial- and lateral-malleoli, medial- and lateral-tibial condyles, medial- and lateral-femoral epicondyles, femoral GTs, iliac crests (highest point which tended to be the most lateral aspect), left and right acromioclavicular joints, and single markers on L5/S1, lumbar spine, thoracic spine, and sternal notch. Rigid clusters were used bilaterally on foot (11.0 x 7.2 cm; 3 markers), leg (8.5 x 11.5 cm; 5 markers), and thigh segments (8.5 x 11.5 cm; 5 markers). Neoprene wraps were placed bilaterally around the thigh and leg at the visually approximated segments region of greatest circumference. Neoprene wraps acted as an intermediate layer between the rigid clusters and skin to increase participant comfort and minimize skin- and clothing-related movement artifact.

Following marker placement, participants performed the same general warm-up as the previous two sessions. Next, participants underwent a brief, static-standing calibration trial with all reflective markers. This static trial is used to define three-dimensional segment locations[16]. Participants stood erect with the arms bilaterally raised in a position that simulated holding a barbell to prevent occlusion of pelvic region markers which can occur when the upper limb is in anatomical position. Static trials were held constant at ten seconds in duration and multiple were recorded if: (i) occlusion/ flickering of markers during static trial or (ii) tracking markers fell off or were noticeably displaced during data collection trials. Following the calibration trial, anatomical markers inferior to the femoral GT were removed, leaving only the rigid cluster markers to track lower extremity limb motion.

Exercise order was pre-determined and randomized to mitigate the effects of fatigue at the sample level. Four sets of each exercise were performed which consisted of two warm-up sets of easy to moderate intensity loads, followed by two data collection sets performed at the pre-determined 5 RM load. Due to the 5 RM loads of the SLS often being 0-20 kg heavier than the barbell, commonly only one SLS warm up set was used. Only three repetitions were performed for each of the four sets, and in the case of the FSS, LSS, and SLS, only on the pre-determined lead/measured limb. Participants rested for a minimum of two minutes between sets but were allowed more as desired with no maximal limit. In cases where the researcher noticed marker occlusion or aberrant marker movement during data collection, the set would be repeated. Earlier participants would perform the additional set following the occluded set, where later participants performed the additional sets at after completion of all other exercises. Participants 01 and 02 also performed five to six sets of each exercise opposed to four. Five to six was the original plan, but after discussions following one participant failing during data collection due to fatigue, reduction to four sets was implemented. It's unlikely the differences in fatigue related to total number of sets and additional set order would have made a meaningful difference on outcomes. This is due to the already high levels of fatigue associated with the data collection. The participant who failed during the BS data collection also squatted the highest absolute load while expressing concerns post-test regarding fatigue related to recent strength competition events.

Kinematic data were collected using eight infrared Miqus 3 cameras (Qualisys, Kvarnbergsgatan, Göteborg, Sweden) at 100 Hz. Force data were collected with two, tri-axial strain gauge force plates (AMTI, Waterdown, Massachusetts, United States of America) at 1000 Hz. Both kinematic and kinetic sampling frequencies are suited for the slow, low-impact nature of heavy resisted squatting. Only the foot of the measured limb was required to be in contact with the respective force plate for the entirety of the set for the FSS, LSS, and SLS. For the BS, participants squatted with the feet from both limbs on the force plates.

3.8 Data Processing

Data processing was compartmentalized into three stages: (i) marker labeling and gap filling, (ii) filtering and inverse dynamics, and (iii) aggregation and statistics.

All saved trials were imported into Qualisys Track Manager (Qualisys, Kvarnbergsgatan, Göteborg, Sweden) for the first stage of data processing. Trials were first trimmed to align the start of the file to be just prior to descent of the first repetition, while the end of the file was aligned with the frames immediately proceeding completion of the final repetition's ascent (i.e., end of repetition three). Markers were then manually labeled by the same researcher. Gap filling was classified as minor and major. Minor gap filling was automatically performed by the software; performed when maker occlusion occurred for <10 consecutive frames. The implemented gap filling algorithm was manually selected for major gap filling when markers were occluded for ≥10 consecutive frames. The algorithm that resulted in the most realistic interpolated path was selected. If all algorithms resulted in noticeably deviant trajectories or markers were missing for entirety of repetitions, the marker was left un-tracked. In cases where this marker was critical to tracking a segment (e.g., L5/S1), the entirety of the set was excluded from further analysis.

Following marker labelling, files were imported to Visual 3D (C-Motion, Boyds, Maryland, United States of America) for filtering and biomechanical analysis. Raw kinematic and kinetic data were run through a low-pass, fourth-order Butterworth filter, set to a 6 Hz cut-off frequency. A 6 Hz cut-off will ensure majority of human motion is captured while optimizing noise reduction[88]. Phase-lag was eliminated by running the filter once forward (2nd order) and reverse (4th order)[17,88]. A bottoms-up inverse dynamics approach was implemented to compute foot, leg, and thigh kinetics[16]. NJMs for all joints were expressed in the distal segments local coordinate system (LCS; e.g., hip NJMs were expressed in the thigh's LCS). Joint angles were computed using a Cardan X-Y-Z rotation sequence and expressed in the proximal segments LCS (e.g., hip flexion is the angle of the thigh relative to the pelvis segment). Segment LCSs positive Z-axis was directed vertically, the Y-axis anteriorly, and the X-axis laterally (right segments) or medially (left segments) when segments were aligned with the global coordinate system (GCS; i.e., relaxed, erect standing). HJC was computed using the ¼ GT prediction method which projects the HJC at one quartile the inter-GT distance, medially[89]. The ¼ GT method yields reliable results and positions the HJC medial-laterally similar to that of the commonly used Bell and Functional methods[89]. Medial-lateral HJC discrepancies have the largest effect on frontal plane hip NJMs[90].

Start and completion of repetitions were defined by the point of zero pelvic vertical velocity (manually identified by the same researcher). A combination of the vertical velocity (points in which velocity changes directions; ~0 m/s) and vertical pelvic displacement (local minimum and maximums) were used in cases where pelvic velocity was erratic. Likely due to the unfamiliarity with higher intensities during the FSS, LSS, and SLS, lots of participant 'adjusting' took place prior to descent (e.g., shifting of the feet). In these cases, identifying precise starting and end points was difficult. For consistency, the first zero velocity that did not rapidly cross zero again prior to the bottom of the squat (i.e., a prolonged downward velocity), was identified as the start. The bottom event was identified as the second crossing point (~0 m/s), while the end was identified as the third crossing point. Three-dimensional segment angles (segment angle relative to the GCS), joint angles (angle between adjacent segment LCSs), tri-planar NJMs, external forces (GRF, COP), as well as event labels, were exported to MATLAB (Mathworks, Natick, Massachusetts, United States of America) for further analysis.

Data aggregation and statistical analyses was performed in MATLAB using custom scripts. Start and end event labels were used to bin kinematic and kinetic outcomes into 9.9° epochs ranging from 10° to 99.9° of hip flexion for both ascent and descent phases, yielding 18 bins. Data was averaged within repetitions, across repetitions, and then across sets, resulting in a single within-exercise value for each bin.

Tri-planar hip NJMs and GRFs that corresponded to the greatest hip flexion angle during ascent was used for zero-dimensional statistical analysis in accordance with primary and secondary hypotheses. To assist in development of theories made post-hoc, tri-planar knee NJMs and thigh segment angles underwent identical zero-dimensional analyses. Ideally, the greatest hip flexion angle corresponded to the 90-99° ascent bin, but as maximal hip flexion angles varied between participants and exercises, so did the analyzed data's associative bins. Selecting for the greatest hip flexion angle focuses the analysis on a characteristic squat trait: deep hip flexion. Histograms, Q-Q plots, and the Lilliefors Test were generated for each primary outcome to determine normality. As some plots and tests revealed non-normality (in conjunction with the small sample size), non-parametric tests were used. Friedman's Tests (non-parametric repeated measures ANOVA analog) and Sign tests (non-parametric t-test analog) were selected. Friedman's Test alpha was held at p = 0 .05. A Bonferroni correction[91] to mitigate inherent Type I error risks[92] was applied to adjust for the Sign tests six pairwise comparisons (p = 0.0083).

4.0 Results

4.1 Participant Characteristics

Of the nineteen participants who completed all study sessions, one participant was not per-protocol due to their test-selected lead limb becoming aggravated with some of the exercises during testing. This resulted in the contralateral limb being used for testing and analyses. All presented analyses exclude this protocol-deviant participant.

Table 5. Participant characteristics

Participant age	25.3 ± 6.4 years
Participant mass	70.8 ± 10.8 kg
Participant height	169.0 ± 6.3 cm
Bilateral squat load, absolute load	75.6 ± 15.2 kg
Bilateral squat load, relative to body mass	106.7 ± 13.8 %
Forward split squat load, absolute load	65.8 ± 10.3 kg
Forward split squat load, relative to body mass	93.8 ± 13.7 %
Lateral split squat load, absolute load	51.4 ± 10.0 kg
Lateral split squat load, relative to body mass	73.0 ± 11.8 %
Single limb squat load, absolute load	24.8 ± 9.8 kg
Single limb squat load, relative to body mass	35.4 ± 13.1 %

All values are expressed as mean ± SD

Mean (±SD) age of participants was 25.3 years (±6.4) with a body mass of 70.8 kg (±10.8) and height of 169.0 cm (±6.3) (Table 4). Participants 5 RM were 75.6 kg (±15.2) for the back squat, 65.8 kg (±10.3) for the forward split squat, 51.4 kg (±10.0) for the lateral split squat, and 24.8 kg (±9.8) for the single limb squat. 5-RMs relative to body mass ranged on average from 0.4-fold (SLS) to 1.1-fold (BS).

4.2 Tri-planar NJMs

Group differences in hip NJMs were found in the sagittal ($X^2(3) = 34.7$, $p < 0.001$), frontal ($X^2(3) = 25.2$, $p < 0.001$), and transverse ($X^2(3) = 25.0$, $p < 0.001$) planes. Median and interquartile ranges for hip and knee NJMs are found in Table 5 and plotted in Figure 2. Significant pairwise differences were found for sagittal hip NJM (x) between the BS→FSS ($p < 0.001$), BS→LSS ($p < 0.001$), FSS→LSS ($p = 0.008$), and FSS→SLS ($p = 0.001$). Hip frontal and transverse plane NJM pairwise differences were identical, demonstrating significant differences between the BS→SLS (frontal: $p = 0.001$, transverse: $p = 0.001$), FSS→SLS (frontal: $p = 0.001$, transverse: $p < 0.001$), and LSS→SLS (frontal: $p = 0.001$, transverse: $p = 0.008$).

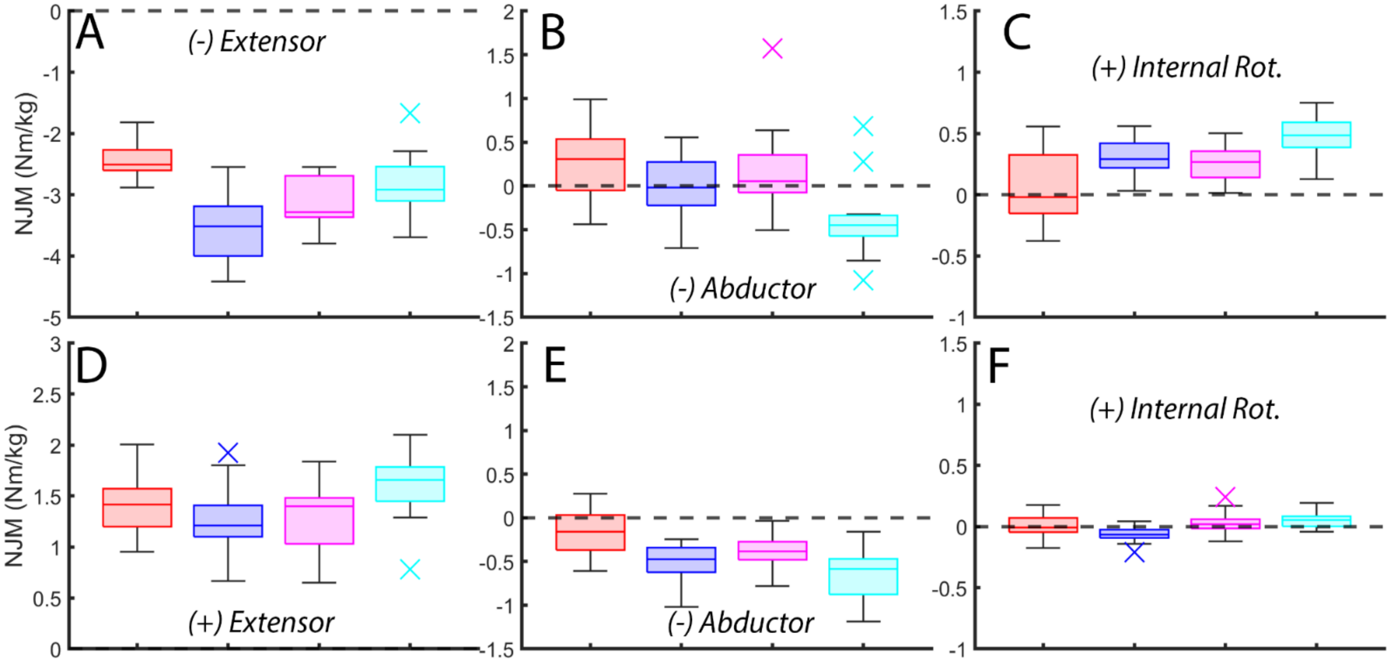
Group differences in knee NJMs were found in the sagittal ($X^2(3) = 19.7$, $p < 0.001$), frontal ($X^2(3) = 27.9$, $p < 0.001$), and transverse ($X^2(3) = 22.7$, $p < 0.001$) planes. Significant pairwise differences were found between the FSS→SLS ($p = 0.001$) and LSS→SLS ($p = 0.001$) in the sagittal plane. In the frontal plane, significant pairwise differences were found between the BS→FSS ($p = 0.008$), BS→SLS ($p = 0.001$), FSS→LSS ($p = 0.008$), FSS→SLS ($p = 0.001$), and LSS→SLS ($p = 0.001$). Only the FSS→SLS ($p < 0.001$) pairwise NJM difference was found to be significant in the knee's transverse plane.

Table 6. Median lower limb net joint moments during resisted squat variations

Joint, plane	Bilateral Squat (Nm/kg)	Forward Split Squat (Nm/kg)	Lateral Split Squat (Nm/kg)	Single Limb Squat (Nm/kg)
Hip, *sagittal*	-2.50 [-2.60 to -2.26]	-3.51 [-4.00 to -3.19]	-3.28 [-3.36 to -2.69]	-2.92 [-3.1 to -2.54]
Hip, *frontal*	0.31 [-0.05 to 0.53]	-0.02 [-0.22 to 0.28]	0.05 [-0.07 to 0.36]	-0.45 [-0.57 to -0.34]
Hip, *transverse*	-0.02 [-0.15 to 0.32]	0.29 [0.22 to 0.42]	0.27 [0.14 to 0.36]	0.48 [0.39 to 0.59]
Knee, *sagittal*	1.42 [1.20 to 1.57]	1.21 [1.1 to 1.41]	1.4 [1.03 to 1.48]	1.66 [1.45 to 1.79]
Knee, *frontal*	-0.16 [-0.37 to 0.03]	-0.47 [-0.62 to -0.34]	-0.38 [-0.48 to -0.27]	-0.59 [-0.88 to -0.47]
Knee, *transverse*	-0.01 [-0.04 to 0.07]	-0.07 [-0.09 to -0.02]	0.02 [-0.01 to 0.06]	0.05 [0.00 to 0.08]
Ankle, *sagittal*	-0.86 [-1.0 to -0.70]	-1.09 [-1.12 to -0.98]	-1.1 [-1.33 to -0.96]	-1.22 [-1.43 to -1.08]
Ankle, *frontal*	-0.18 [-0.3 to -0.11]	-0.11 [-0.17 to -0.01]	-0.21 [-0.35 to -0.05]	-0.07 [-0.14 to 0.01]
Ankle, *transverse*	-0.19 [-0.25 to -0.11]	-0.08 [-0.15 to -0.06]	-0.23 [-0.28 to -0.17]	0.01 [0.00 to 0.06]

All values reported as median [25^{th} quartile to 75^{th} quartile]. Values correspond to greatest binned hip flexion angle during ascent. Hip flexor, knee extensor, and ankle dorsiflexor (sagittal), adductor (frontal), and internal rotator (transverse) are denoted as positive values

Figure 2. Net joint moments median and interquartile ranges during resisted squat variations



Red represents the bilateral squat, blue the forward split squat, pink the lateral split squat, and green the single limb squat. X-markers represent outliers. A = sagittal hip plane, B = frontal hip plane, C = transverse hip plane, D = sagittal knee plane, E = frontal knee plane, and F = transverse knee plane

4.3 Tri-planar GRFs

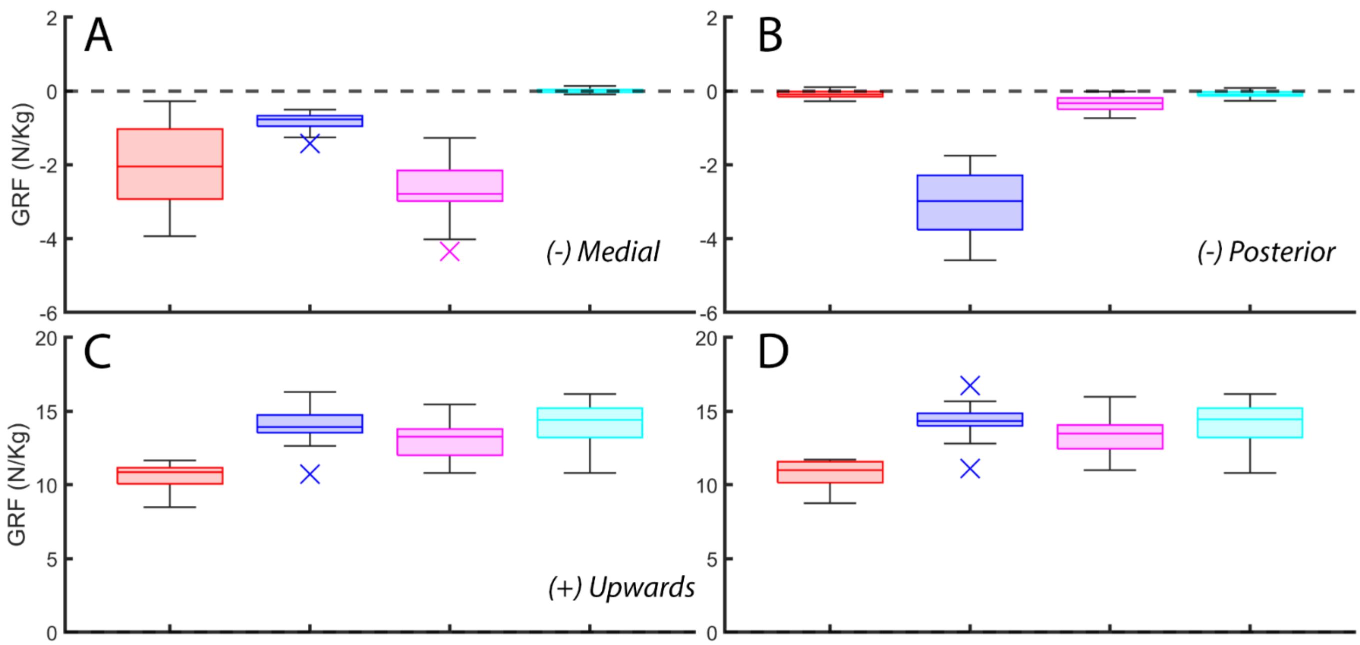
Group differences in GRFs were found in the medial-lateral ($X^2(3)$ =47.6, $p < 0.001$), anterior-posterior ($X^2(3) = 45.3$, $p < 0.001$), vertical ($X^2(3) = 36.3$, $p < 0.001$), and resultant ($X^2(3) = 37.7$, $p < 0.001$) axes. Median and interquartile ranges for GRFs are found in Table 6 and plotted in Figure 3. In the medial-lateral plane, significant pairwise GRF differences ($p \leq 0.001$) were found for all but between the BS→LSS ($p = 0.03$). In the anterior-posterior plane, significant GRF differences ($p < 0.001$) were found for all but between the BS→SLS ($p = 0.8$). In the vertical plane, significant differences were found between the BS→FSS ($p < 0.001$), BS→LSS ($p < 0.001$), BS→SLS ($p < 0.001$), and LSS→SLS ($p = 0.008$). The resultant GRF pairwise comparisons found significant differences between the BS→FSS ($p < 0.001$), BS→LSS ($p < 0.001$), BS→SLS ($p < 0.001$), and FSS→LSS ($p = 0.008$).

Table 7. Median ground reaction forces during resisted squat variations

Axis	Bilateral Squat (N/kg)	Forward Split Squat (N/kg)	Lateral Split Squat (N/kg)	Single Limb Squat (N/kg)
Medial-lateral (x)	-2.04 [-2.92 to -1.02]	-0.77 [-0.95 to -0.66]	-2.79 [-2.99 to -2.16]	0.02 [-0.04 to 0.04]
Fore-aft (y)	-0.09 [-0.16 to -0.02]	-2.98 [-3.75 to -2.29]	-0.33 [-0.49 to -0.19]	-0.09 [-0.14 to -0.03]
Vertical (z)	10.87 [10.06 to 11.17]	13.92 [13.55 to 14.73]	13.25 [12.00 to 13.78]	14.43 [13.2 to 15.21]
Resultant (r)	10.99 [10.14 to 11.58]	14.34 [13.99 to 14.86]	13.49 [12.43 to 14.07]	14.43 [13.21 to 15.21]

All values reported as median [25^{th} quartile, 75% quartile]; values correspond to greatest binned hip flexion angle during ascent; lateral, anterior, and upward forces are denoted as positive values

Figure 3. Ground reaction force median and interquartile ranges during resisted squat variations



Red represents the bilateral squat, blue the forward split squat, pink the lateral split squat, and green the single limb squat; X-markers represent outliers; A = medial-lateral axis, B = anterior-posterior axis, C = upward-downwards axis, D = resultant

4.4 Thigh Kinematics

Group differences in thigh segment angles were found in the sagittal ($X^2(3) = 45.0$, $p < 0.001$), frontal ($X^2(3) = 46.9$, $p < 0.001$), and transverse ($X^2(3) = 49.9$, $p < 0.001$) planes. Median and interquartile ranges for joint and segment kinematics are found in Table 7 & 8, respectively. Significant pairwise differences in the sagittal thigh segment angle were found between the BS→LSS ($p < 0.001$), BS→SLS ($p < 0.001$), FSS→LSS ($p < 0.001$), FSS→SLS ($p < 0.001$), and LSS→SLS ($p = 0.001$). In the frontal plane, pairwise comparisons between the BS→FSS ($p < 0.001$), BS→SLS ($p < 0.001$), FSS→LSS ($p = 0.001$), FSS→SLS ($p < 0.001$), and LSS→SLS ($p < 0.001$) were found to be significant. Finally, all pairwise comparisons in the thigh segments transverse plane were found to be significant ($p \leq 0.008$).

Table 8. Median lower limb joint angles during loaded squat variations

Joint, plane	**Bilateral Squat (°)**	**Forward Split Squat (°)**	**Lateral Split Squat (°)**	**Single Limb Squat (°)**
Hip, *sagittal*	95.11 [91.25 to 95.35]	92.24 [85.16 to 95.25]	91.64 [84.47 to 95.17]	91.7 [81.27 to 94.75]
Hip, *frontal*	-19.05 [-23.3 to -16.99]	6.70 [4.62 to 8.35]	-10.77 [-13.62 to -2.10]	12.38 [8.60 to 14.49]
Hip, *transverse*	13.29 [10.39 to 15.95]	5.94 [4.09 to 8.21]	8.61 [7.18 to 10.39]	6.35 [3.09 to 9.09]
Knee, *sagittal*	-93.48 [-103.11 to -83.56]	-77.05 [-83.99 to -66.27]	-75.31 [-83.21 to -67.73]	-74.26 [-79.25 to -61.59]
Knee, *frontal*	12.49 [10.51 to 18.03]	12.65 [9.55 to 16.17]	12.43 [9.28 to 15.00]	13.29 [9.95 to 16.94]
Knee, *transverse*	5.22 [2.40 to 8.84]	3.82 [-0.73 to 4.60]	0.32 [-2.46 to 6.33]	-0.44 [-4.94 to 1.49]
Ankle, *sagittal*	25.36 [21.57 to 27.08]	7.27 [5.00 to 11.75]	20.65 [17.97 to 23.06]	25.25 [23.99 to 28.3]
Ankle, *frontal*	6.31 [-0.94 to 10.93]	-3.86 [-6.80 to -1.55]	6.91 [3.09 to 8.93]	-11.17 [-13.08 to -8.66]
Ankle, *transverse*	-14.77 [-18.75 to -10.23]	-16.14 [-20.74 to -12.75]	-17.3 [-20.94 to -13.47]	-20.97 [-26.32 to -17.92]

All values reported as median [25^{th} quartile, 75% quartile]; values correspond to greatest binned hip flexion angle during ascent; joint hip flexion, knee extension, and ankle dorsiflexion (sagittal), adduction (frontal), and internal rotation (transverse) are denoted as positive values

Table 9. Median lower limb segment angles during loaded squat variations

Joint, plane	**Bilateral Squat (°)**	**Forward Split Squat (°)**	**Lateral Split Squat (°)**	**Single Limb Squat (°)**
Thigh, *sagittal*	65.76 [57.91 to 70.62]	67.01 [58.91 to 69.33]	51.85 [48.37 to 57.08]	45.62 [39.4 to 49.99]
Thigh, *frontal*	-20.35 [-21.84 to -16.60]	-8.45 [-11.06 to -6.98]	-15.44 [-19.25 to -12.81]	0.24 [-1.71 to 1.85]
Thigh, *transverse*	-13.85 [-16.73 to -8.51]	6.34 [4.66 to 10.79]	-2.06 [-5.36 to 1.19]	15.9 [10.17 to 19.86]
Pelvis, *sagittal*	-25.23 [-34.77 to -16.21]	-21.00 [-28.74 to -19.08]	-37.05 [-40.94 to -30.39]	-41.84 [-46.65 to -38.76]
Pelvis, *frontal*	0.26 [-1.77 to 1.47]	-4.04 [-8.3 to 3.03]	-4.93 [-6.74 to -1.67]	-2.46 [-7.79 to 2.12]
Pelvis, *transverse*	1.48 [-0.64 to 3.00]	-2.19 [-4.05 to 1.15]	3.35 [-1.22 to 6.02]	1.47 [-1.29 to 5.32]
Leg, *sagittal*	-28.86 [-30.52 to -23.35]	-10.06 [-12.39 to -5.67]	-23.26 [-25.12 to -19.27]	-29.34 [-31.24 to -24.67]
Leg, *frontal*	-6.4 [-10.22 to -2.26]	3.54 [1.54 to 4.81]	-4.30 [-7.27 to -2.82]	11.37 [9.14 to 12.72]
Leg, *transverse*	-15.07 [-19.01 to -10.79]	6.93 [4.23 to 9.90]	-7.40 [-10.18 to -2.51]	7.57 [3.16 to 12.69]
Foot, *sagittal*	-1.78 [-2.30 to -0.97]	-1.47 [-2.34 to -0.74]	-2.71 [-3.76 to -1.17]	-1.13 [-2.81 to -0.19]
Foot, *frontal*	-0.61 [-3.93 to 1.00]	-1.48 [-3.58 to 1.05]	1.38 [0.08 to 3.03]	0.3 [-2.23 to 1.27]
Foot, *transverse*	-29.9 [-33.55 to -25.84]	-10.1 [-15.31 to -1.23]	-24.17 [-27.62 to -19.38]	-15.51 [-19.24 to -8.10]

All values reported as median [25th quartile, 75% quartile]; values correspond to greatest binned hip flexion angle during ascent; segment angles are expressed relative to a global coordinate system with upwards, medial, and anterior as positive polarities in correspondence with the right hand rule

5.0 Discussion

Resisted squat exercise variations demonstrate divergent tri-planar hip NJM strategies in resistance trained females. The BS does not elicit greater internal rotator hip NJMs than the FSS, LSS, or SLS. In contrast, the SLS elicited the greatest internal rotator NJM. As hypothesized, the SLS does demonstrate the greatest hip abductor NJMs – larger than that of the BS, FSS, and LSS variations. As the SLS simultaneously demonstrated the lowest medial GRFs of any exercise while eliciting the largest frontal and transverse plane NJMs, medial GRF magnitudes alone are not valid surrogates for approximating tri-planar hip NJMs during resisted squats.

5.1 NJM & GRF Analyses and Muscular Implications

Tri-planar hip NJMs during resisted squats follow a pattern according to lead-limb loading. The BS, a variation which presumably vertically loads both lower limbs *more-or-less* symmetrically, tends to require a hip adductor NJM (0.31 [-0.05 to 0.53] Nm/kg), with a slight bias towards an external rotator NJM in the transverse plane (-0.02 [-0.15 to 0.32] Nm/kg). Failing to support the first hypothesis, the BS was the only variation of the four exercises to demonstrate participants utilizing an external rotator NJM, resulting in the BS having the lowest and most variable potential to utilize the deep gluteal muscles. Predominantly single limb variations - the FSS and LSS - demonstrate practically null median frontal plane NJMs, with interquartile ranges and maximum/minimums extending into both abductor and adductor directions. The FSS, LSS, and SLS variations demonstrated internal rotator NJM strategies, extending from 0.27 [0.14 to 0.36] Nm/kg (LSS) to 0.48 [0.39 to 0.59] Nm/kg (SLS). The consistent across-participant internal rotator suggests the FSS, LSS, and SLS have the greatest potential to utilize the deep gluteal muscles. The SLS produced the narrowest-ranged and largest front plane NJMs in participants, with a clear abductor bias (-0.45 [-0.57 to -0.34] Nm/kg), supporting hypothesis two. As the SLS had the greatest abductor NJM in conjunction with the lowest sagittal hip flexion angle, the SLS is presumably the only variation to demonstrate consistent potential to utilize the deep gluteal muscles in both frontal and transverse planes at the start of ascent.

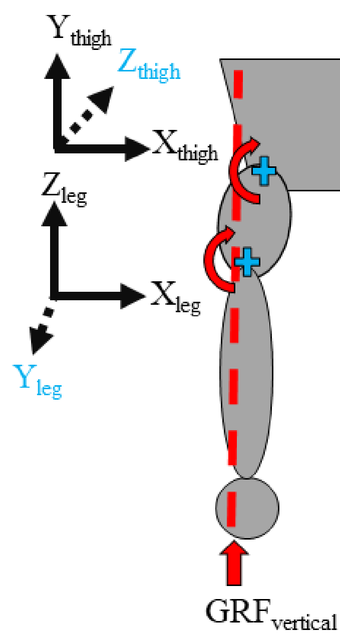
The GRF profile findings are as expected. The SLS generates practically null medial-lateral (0.02 [-0.04 to 0.04] N/kg) and anterior-posterior (-0.09 [-0.14 to -0.03] N/kg) forces, as these would easily perturb the system due to its small base of support. As the approximate area of the stance foot (i.e., the BOS) is greatly reduced compared to the relatively larger BOSs of the BS, FSS, and LSS, non-vertical forces must be minimized due to the reduced room for the COM to displacement prior to leaving the BOS and resulting in potential loss of balance. In other terms, the stability of the system is (in-part) proportional to its BOS area[93]. The LSS produced the greatest medial GRFs, presumably due to the COM of the system's large frontal plane excursions, requiring substantial medial GRFs to accelerate the system back to the starting position. As predicted, the BS produced substantial medial GRFs (-2.04 [-2.92 to -1.02] N/kg), yet unexpectedly, these medial GRFs were not statistically different than the those during LSS. These large medial GRFs are likely facilitated by the large base of support (allowing large perturbations prior to balance loss) and approximately symmetrical lower limb loading (allowing bilateral medial GRF components to nullify one another). For the similar reasoning as why the LSS generates the largest medial GRFs, the FSS generated the largest posterior GRFs (-2.98 [-3.75 to -2.29] N/kg).

As the SLS generated practically zero medial-lateral GRFs despite demonstrating the largest frontal and transverse plane hip NJMs, GRFs alone are insufficient in approximating/predicting tri-planar hip NJMs during resisted squats. Yet, these findings seemingly directly oppose previous findings which suggest the deep gluteal muscles (presumed dominant abductor/ internal rotator contributors), are the primary contributors to medial GRFs at all points of the gait cycle[47]. The thighs relatively more adducted posture during the SLS (0.24°) relative to the abducted postures of the other variations (-8° to -20°) may explain the dichotomy.

In conjunction with being the largest medial GRF generators, the deep gluteal muscles are also substantial vertical GRF contributors in gait[45,46]. Gmed was on average the third largest contributor to vertical COM accelerations - trailing only the soleus and vasti, while contributing more than gmax[46]. Given gmed's small extensor yet substantial hip abductor moment arms[14,15], it's likely the deep gluteal abductor (opposed to extensor) generated moments that are the primary contributors to the vertical GRFs. The abductors contribution to vertical GRFs would only be possible with the thigh in an adducted posture, as thigh abduction would need to push the leg/foot segment downwards rather than laterally. Thus, the more adducted thigh position during the SLS explains the absence of medial-lateral GRFs despite eliciting the largest hip abductor NJMs.

The preference towards the hip adductor NJM may be explained by the BS's large thigh abduction angle (-20.35 [-21.84 to -16.60]°) at the start of ascent. Applying the floor reaction force vector (FRFV) model, which provides a more simplistic base for theorizing NJM casual factors at the expense of accuracy[94], a more abducted femoral mechanical axis would increase the likelihood of the vertical GRFs generating an external hip abductor NJM, and thus, an internally (muscularly) generated hip adductor NJM. The external rotator strategy unique to the BS may be explained by the knee's frontal plane NJM. The BS was the only exercise where participants demonstrated a knee adductor NJM. Of the nine participants who utilized a hip external rotator NJM during the BS, five also utilized a knee adductor NJM, with the remaining four producing the lowest knee abductor NJMs (i.e., the binning method is likely masking adductor NJMs). As the leg segment had undergone much less rotation in the sagittal plane than the thigh (-28.86° v 65.76°), the leg's y-axis is more aligned with the thigh's z-axis than its y-axis. Applying a relatively more laterally positioned (relative to thighs mechanical axis) vertical GRF could then elicit both an external knee abductor and hip internal rotator NJMs (Figure 4). Internally, knee adductor and hip external rotator NJMs are produced by surrounding musculature to counter the externally GRF-generated NJMs.

Figure 4. Floor reaction force vector explanation of the bilateral squat's unique external rotator hip net joint moment



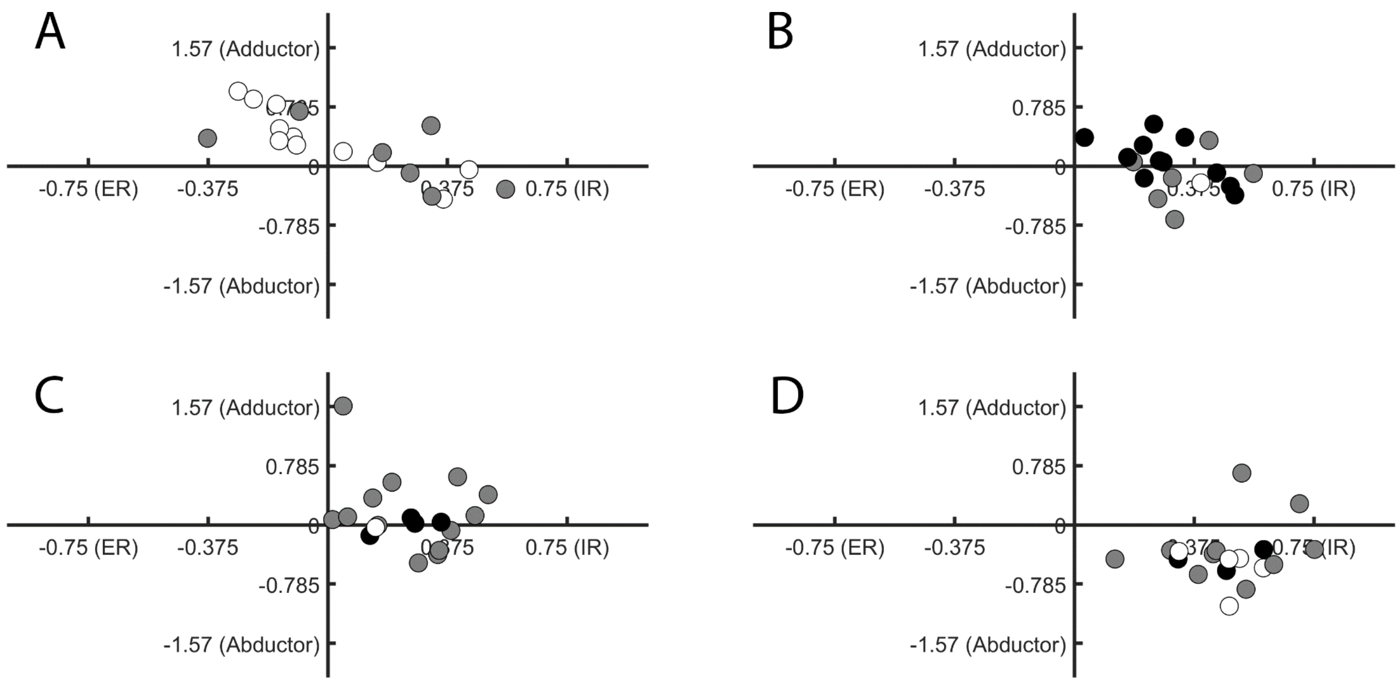
As the thigh has rotated relatively more in the sagittal plane (hip flexion) than the leg has (dorsiflexion), the thigh's z-axis and the leg's y-axis are now approximately aligned (though at a ~180° relative angle); applying a laterally placed vertical ground reaction force results in an external knee abductor and external hip internal rotator net joint moments (red solid curved arrows) when expressed in the knee's and thigh's local coordinate systems, which are then countered internally by a knee adductor and hip internal rotator moments (not depicted); GRF = ground reaction force; blue crosses represent the location of the local coordinate systems depicted to their left; hashed line = line of action of vertical GRF

The thigh segment's 45° sagittal plane angle in the SLS uniquely positions the thigh's z- and y-axis to be equally sensitive to the same vertical GRF. This sagittal plane rotation directs the positive z-axis between vertical and posterior, and the positive y-axis between anterior and vertical. Application of a relatively medial (relative to the thighs mechanical axis) vertical GRF would then require relatively equal in magnitude internal hip abductor (y-axis) and internal rotator (z-axis) NJMs. This relationship was demonstrated by the nearly identical frontal (-0.45 [-0.56 to -0.34] Nm/kg) and transverse (0.48 [0.39 to 0.59] Nm/kg) plane hip NJMs in the SLS. Additionally (as previously discussed), the SLS's relatively neutral frontal thigh segment angle (0.24 [-1.71 to 1.85] °) relative to the moderate abductor angles of the other variations (-8° to -

20°), results in a thigh mechanical axis optimally positioned for the requirement of an internal hip abductor NJM.

Relatively narrow interquartile ranges in both frontal and transverse plane hip NJMs in the SLS suggest constrained/minimal number of successful strategies. The large variability in frontal plane NJMs in the FSS and LSS suggests leniency in strategy; a potential product of the large base of support despite being predominantly single limb loaded. To further explore the relationship of exercise and successful NJM strategies, cartesian plots allowing cluster analysis were generated (Figure 5).

Figure 5. Hip net joint moment strategies during resisted squat variations



Both vertical and horizontal axis units are Nm/kg; marker shading denotes sagittal plane hip NJM magnitudes; white represents values less than the 25th quartile, grey equal to and between 25th and 75th quartiles, and black greater than the 75th quartile; A = bilateral squat, B = forward split squat, C = lateral split squat, D = single limb squat

Four subplots depicting each squat variation represent the frontal plane hip NJM on the vertical axis, the transverse plane NJM on the horizontal axis, and the sagittal plane NJM as marker shading (darker shading representing high quartile position). Both the plot axes as well as the marker shading are scaled to each of the respective planes maximal at the global exercise level. For example, the vertical axis maximal is 1.57 Nm/kg – the peak frontal plane NJM to occur across all squat variations. As all participants across all exercises demonstrated an extensor strategy in the sagittal plane, the following tri-planar strategies assume and do not *explicitly* state the sagittal polarity. Cluster patterning further emphasizes the relationship between limb loading and viable hip NJM strategies during resisted squatting. All but two participants utilized an abductor-internal rotator strategy in the SLS. In the predominantly single limb loaded variations (FSS and LSS), an approximately even number of participants utilized an abductor-internal rotator or adductor-internal rotator strategy. In the more symmetrically lower limb loaded BS, three viable strategies were demonstrated: abductor-internal rotator (n=5), adductor-internal rotator (n=4), and adductor-external rotator (n=9). Thus, it appears the number of viable hip NJM strategies during resisted squatting may be related to the robustness (i.e., balance or resistance to perturbance) and/or the relative bilateral lower limb loading of the squat stance. With reductions to the base of support and/or increased relative single limb loading, the number of viable strategies is reduced. The abductor-external rotator strategy was the only strategy not demonstrated in any of the squat variations.

The FSS, LSS, and SLS strictly utilize an internal rotator NJM. In addition, half of participants during the BS demonstrate an internal rotator strategy approximate of FSS and LSS magnitudes. It's plausible given the deep gluteal muscles internal rotator moment arm in deep hip flexion (~90°)[13] and gmed's substantial PCSA and mass[39], that deep gluteal loading is required for predominantly and completely single limb resisted squatting in resistance trained females. Reported median internal rotator NJMs in the FSS, LSS, and SLS are on average similar or greater than those reported during activities of daily living, including walking[81–83], stair ascent[82], and running[81]. Single limb landing generated internal rotator NJMs similar to the SLS but greater than FSS and LSS[81]. These findings demonstrate *potential* for predominantly and completely single limb, resisted squatting, to elicit strength adaptations of the deep gluteal muscles. Future training studies are still required to adequately determine if and the magnitude of these potential strength adaptations.

As sagittal plane extensor strategies were constant for all exercises and participants, gmax and/or adductor magnus posterior head contributions are likely required no matter the frontal and/or transverse plane demands. Due to their monoarticular course[19], large PCSA[39], and dominant extensor moment arms (especially that of adductor magnus posterior head in deep hip flexion[37]), gmax and adductor magnus are optimally situated for extending the hip in squatting.

Cartesian NJM plots facilitate a deeper level of analysis critical for theorizing muscle strategies. By considering all axes simultaneously, more accurate assessments can be made. For example, despite the FSS and LSS requiring an internal rotator and thus deep gluteal muscle strategy, the frontal plane strategies were nearly evenly split. An abductor-internal rotator strategy may require no further muscle contributions to satisfy non-sagittal requirements (due the deep gluteal muscles abductor capacities) or require further contributions from the quadriceps coxa (due to their strong abductor moment arm in deep hip flexion)[36]. An adductor-internal rotator strategy may require contributions from the adductor group (e.g., adductor magnus and/or adductor longus) in addition to the deep gluteal muscles.

The only NJM strategy not utilized in any of the squat variations was an abductor-external rotator. It's unclear why the abductor-external rotator strategy was never utilized in any of the exercises. It's plausible the hip NJM strategy is inefficient and/or results in poor kinematic technique. Further work is required to fully elucidate the link between hip NJM strategies and resisted squatting.

5.2 Comparing NJMs to Previous Squatting Investigations

Despite the current thesis utilizing the highest intensity of previous resisted squat studies[76–79] (to report multi-planar hip NJMs [see Table 3]; 100% 5 RM ≡ 86% 1 RM[85]), NJM magnitudes from the current investigation fall below most. In the sagittal plane, the median BS finding of 2.50 Nm/kg is the most equivalent to the 3.1 Nm/kg finding of Southwell et al. at 80% 1 RM in the running shoe condition[78].

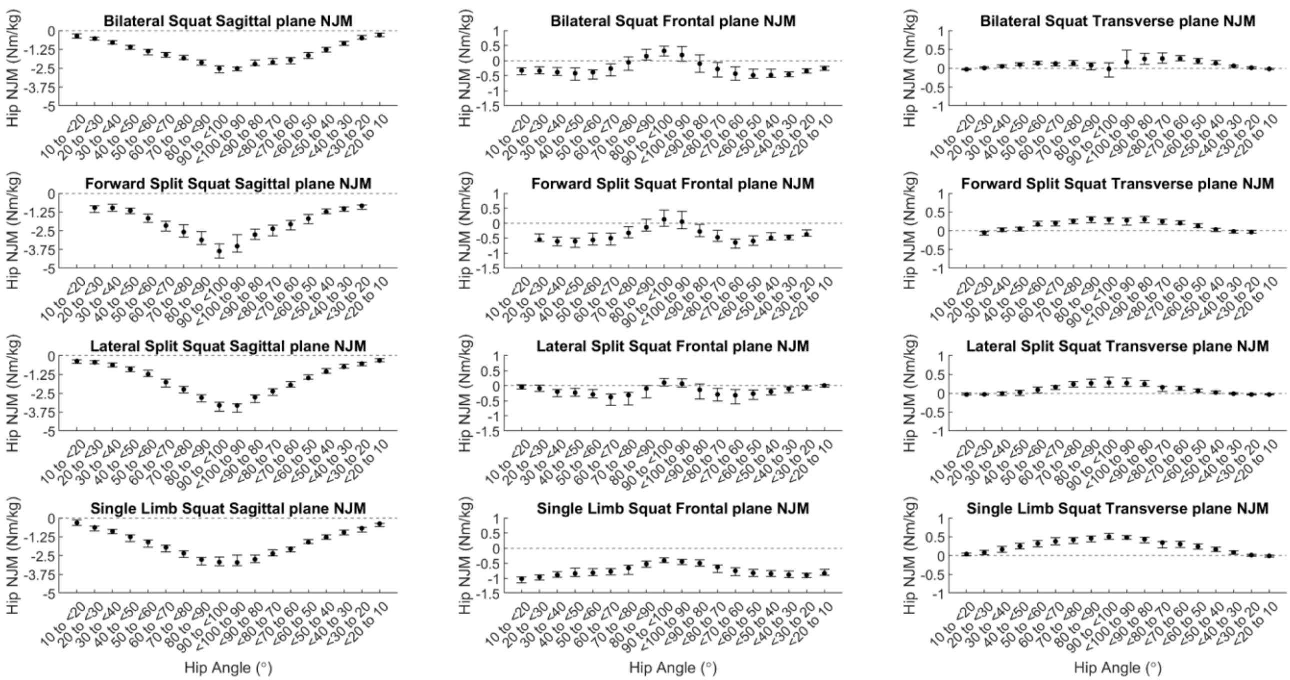
The gap between magnitudes further grows in the frontal and transverse planes. Lahti et al.[77] and Swinton et al.[79] reported peak abductor NJMs opposed to adductor NJMs of the current thesis. These peak abductor NJMs are even greater than the median abductor NJM of the SLS (0.45 Nm/kg) found in the current thesis; the largest frontal plane finding of the four variations. Lahti et al.'s values do appear as outliers, with magnitudes up to 1.75 Nm/kg[77] (~2-fold greater than Southwell et al.[78] and Swinton et al.[79]). Sinclair et al. reported a peak adductor NJM *3-fold* greater than that found in the current findings (note, this is not a completely accurate comparison as authors did not provide body mass normalized NJMs)[76]. As the epoch of the squat in which

NJMs were extracted varies between investigations, the addition of time can aid in inter-study analyses.

Temporal plots depicting median and interquartile ranges for each variation are presented in Figure 6. As sagittal hip angle progression was in good agreement with time progression, these plots are still referred to as *temporal* despite their lack of any direct time descriptor. As repetitions were not scaled to a relative reference such as *percent-cycle-completion*, not all bins have an equal number of participants (i.e., greater hip angle bins tend to have less than the 18 participants included in the statistical analyses, which may in-part explain the greater interquartile ranges). Despite Swinton et al. not providing any NJM figures, Lahti et al.[77], Sinclair et al.[76], and Southwell et al.[78] reported trends that paralleled the current thesis' findings. Frontal plane NJMs display a peak adductor NJM at the bottom of the BS and *trough* abductor NJMs during ascent and descent. Transverse plane NJMs follow a rough sinusoidal trend with approximate null NJMs at the start and end of the repetition, a peak external rotator NJM at the bottom, and peak internal rotator NJMs at ~25% and ~75% of the squat cycle. Frontal and transverse plane magnitudes appear to agree with Southwell et al.[78], are much less than those of Sinclair et al.[76], and are not possible to extract from Lahti et al.[77] due to figure scaling factors. Despite differences in relative NJM magnitudes across resistance investigations, a general agreement in temporal patterning is present.

Differences in frontal and transverse plane NJM magnitudes are greatly reduced when compared to bodyweight squat investigations. Sagittal plane extensor NJMs were greater in magnitude than any of the previously reported body weight squat variations (Table 2). This is logical as the greater system load would require greater vertical GRFs, likely primarily achieved via greater extensor contributions. Intriguingly, frontal and transverse plane NJMs in body weight squats are much more similar to those of the current *resisted* squat NJM findings (Table 2). All but one[68] previous investigation reported adductor NJMs at the lowest point of the BS[66,70,71]. Adductor NJM magnitudes were generally comparable to the current investigation's 0.31 Nm/kg, except Malloy et al.[69] which reported a peak adductor NJM 2-fold greater. Khuu and Lewis[67] reported a nearly 2-fold greater abductor NJM in the SLS. Khuu and Lewis' greater abductor NJM was reported at a much lower hip flexion angle[67] (65° v. 95°), which may in-part, explain the larger magnitude. As demonstrated in Figure 6, hip abductor magnitudes decrease with increasing hip flexion angle – thus, frontal plane hip moment comparisons may require sagittal plane hip angle standardization for accurate comparisons. Two studies reported external rotator NJMs[66,69] and one internal rotator[71] (an internal rotator peak can be estimated from Li et

Figure 6. Hip net joint moments during resisted squat variations expressed relative to hip sagittal angle



As the sagittal hip angle progressed in agreement with time, these are referred to as *temporal* plots; NJM – net joint moments; Extensor (sagittal), abductor (frontal), and external rotator (transverse) are denoted as negative polarities

al.'s figures[68]); magnitudes were in approximation of the current thesis' practically null median.

Despite similar NJM magnitudes, temporal trends of transverse plane NJMs in both Van Houcke et al.[71] and Bagwell et al.[66] were *inverse* to the current thesis'. Participants in Van Houcke et al. and Bagwell et al. began and completed squat repetitions with an external rotator NJM – transitioning to an internal rotator NJM at the bottom of the squat[66,71]. Majority of temporal frontal plane NJM trends paralleled the current findings[66,70,71], except Li et al., which demonstrated an *inverse* trend (repetitions began and completed with an adductor NJM and transitioned to an abductor NJM at the bottom of the *full-depth* squat variation)[68].

Differences in both methodologies and sample characteristics may explain the heterogeneity in interstudy hip NJMs. HJC approximation method likely has the greatest impact.

Unlike other lower limb joints, easily accessible and bilateral bony landmarks are limited in the proximal thigh region. As a result, many methods exist, which leads to inconsistencies between studies. Only 40% of previous squat investigations that reported multi-planar hip NJMs, reported their HJC approximation methods[67,69,70,76]. Including the current thesis, majority of previous investigations have strictly implemented *predictive* approximation methods[95], including the ¼ GT approach[89] (current thesis), Bell method[73] (Slater and Hart[70]), CODA pelvis[72] (Khuu and Lewis[67] and Malloy et al.[69]), and the projected 0.089m[76] (Sinclair et al.)[76]. These approaches approximate the HJC based on identifiable landmarks. For example, the Bell method places the HJC 30% distal, 14% medial, and 19-22% posterior, the inter-ASIS distance of the ipsilateral ASIS[73,74]. The CODA pelvis, a modified version of the Bell method, places the HJC, places the HJC 30% distal, 36% medial, and 19% posterior[72]. The >2-fold medial difference just between the Bell and modified Bell (CODA) approaches may have substantial ramifications, as frontal plane hip NJMs are the most sensitive to differences in differences in this axis[90]. The current thesis' HJC ¼ GT method has been demonstrated to approximate the HJC medial-lateral coordinates[89,96] and mechanical axis angle[96], to that of the Bell and Functional methods. Specific to resisted squatting, Sinclair et al. have demonstrated marginal differences in peak frontal plane NJMs between predictive approaches in males performing a 70% 1RM BS, though differences jump to nearly *2-fold* when contrasting functional and predictive methods[76]. The functional approximation also resulted in differences in temporal frontal plane trends, as it was the only method to ***never*** demonstrate an abductor NJM during the BS[76]. Despite some evidence suggesting marginal differences, the overall poor reporting (75% of resisted squat studies did not clearly state their approach) and wide range of possible methods, suggests differences in HJC approximation methods may be a large contributing factor to interstudy hip NJM heterogeneity.

Variability in lower limb kinematics may also contribute to interstudy differences in hip NJMs. Differences in tri-planar thigh segment tracking during data collection would result in different instantaneous thigh LCS rotations and thus, projected hip NJMs (if NJMs are expressed in the distal segment, in which the only two studies to report NJM expression, did[71,77]). Discrepancies in segment kinematics can also lead to errors that propagate proximally from the foot, but as this adds even more unknowns, this will not be discussed. Differences in kinematics can also directly affect HJC approximations when computed with functional methods, due to their dependence on tri-planar thigh motion[97]. Coyne et al.[98] demonstrated large differences in both frontal and transverse plane knee angles when using different thigh segment tracking markers during overhead squats in males. As computing knee angles requires thigh LCSs, reasonable extrapolations can be made from the following findings. Sagittal plane knee angles did not vary much between marker sets (~2°)[98]. In contrast, some marker sets demonstrated knee adduction angles while others abductor angles in the same squat position[98]. Differences in knee joint angles reached magnitudes of 7° across frontal and transverse planes[98]. Of the previous body weight and resisted squat studies, two were unclear in their marker sets[71,77]. One used an electromagnetic opposed to a passive, reflective motion capture system[70]. Outside of minor differences in distal thigh and pelvis markers, two studies stick out as outliers – Li et al.[68] and Swinton et al[79]. Both studies did not mention using a rigid tracking cluster on the thigh unlike the remaining studies with sufficient reporting[66,67,69,76,78]. Even when investigations do utilize identical marker sets, inter- and intra-assessor landmark identification variability may still affect NJMs. ASIS identification between assessors can vary between 15 mm and 30 mm[99,100] – a critical landmark for many predictive and functional HJC approximations. This error range aligns with magnitudes which have been observed to result in noticeable change in hip NJMs[90]. Even the same assessor can vary up to 20 mm when localizing ASIS landmarks[100]. The femoral GT is a particularly tricky landmark due to variability in localized soft-tissue presence and its broad, mushroom-cap-like structure, which makes determining a specific 'peak', difficult. Difficulty in localizing the femoral GT has been demonstrated by a resultant ~18 mm within and between assessor variability[100]. The largest discrepancies both between and within assessors in the formal GT location were in the medial-lateral axis (~12 mm)[100]. Similar magnitudes in error were reported for the right ASIS marker (~10 to ~12 mm)[100] – another commonly used marker for predictive HJC approximation methods[72–74]. Frontal plane hip NJMs have been demonstrated to be the most sensitive to changes in the medial-lateral HJC position, with a more lateral position decreasing the frontal plane hip NJM magnitude and medial increasing the frontal plane NJM magnitude[90]. Della Croce et al. did not ascribe a polarity to these differences in landmark locations[100], thus making it difficult to theorize the expected direction of these errors. In addition, changes in hip NJMs were reported during gait[90], likely utilizing greatly reduced hip flexion ranges compared the current thesis' squat variations. The smallest error in femoral GT position (~7 mm) yet the greatest in the right ASIS (~15 mm), was in the anterior-posterior axis[100], which has been demonstrated to result in the greatest discrepancies in the transverse plane hip NJMs in gait[90].

Variations in squat techniques and sex-differences may also play parts in inter-study hip NJM variability. The current thesis, two of the prior resistance squat studies[76,78], and two of the prior body weight squat studies[69,71], used minimal cueing. Reported values then contain a range of squatting techniques (kinematic and kinetic), minimizing technique influences on NJMs. The remaining investigations either purposefully modified squat technique[67,68,70,77,79] or provided substantial feedback[66]. Khuu and Lewis noted differences in hip NJMs both between sexes and tasks[67]. Li et al. demonstrated greater hip extensor (ascent phase), abductor (majority of descent and start of ascent), and external rotator (majority of descent and start of ascent) NJMs, during full-depth body weight squats compared to half-depth squats (90° of knee flexion)[68]. Slater and Hart demonstrated the high sensitivity of tri-planar hip NJMs to knee alignment during body weight squats – both increasing and decreasing depending on the condition and epoch[70]. Lahti et al. demonstrated no significant differences between sexes or stance-width conditions for hip NJMs[77]. At the 70% 1 RM (the highest resistance assessed), Swinton et al. found only the powerlifting BS technique to elicit a significantly greater hip abductor NJM than the traditional BS technique[79]. The current thesis in addition to Li et al.[68] used female-only samples, while three investigations included male-only samples[71,76,78], and the remaining assessed mixed samples[66,67,69,70,77,78].

The uncertainty in participant inclusion criteria of the current thesis may also factor into the inter-study hip NJM heterogeneity. Studies included that reported hip kinetics in Table 2 and 3 state excluding those with current lower extremity pain/injury[66–70,76,78] and/or history of surgeries[66,67,70]. Van houcke et al.[71] excluded those with hip issues that had the potential to affect squat kinematics. Lahti et al.[77] excluded those with non-specific issues that could both affect and/or be exacerbated by resisted BSs. Swinton et al.[79] did not explicitly state any inclusion/exclusion criteria. Bagwell et al.[66] and Malloy et al.[69] both included hip pathology samples – specifically, those with FAI syndrome. Malloy et al. reported ~30% lower peak hip abductor and 70% lower hip extensor NJMs during the SLS in the FAI sample[69]. The FAI sample also demonstrated 20% reduced peak extensor differences during the bodyweight bilateral squat[69]. Bagwell et al. also noted a decrease in mean hip extensor NJM during the bodyweight bilateral squat in the FAI sample, while also finding no between group differences in the remaining hip NJMs[66]. Some outlier values matching the reduced hip extensor and abductors NJMs during the SLS is seen in Figure 2. Though these differences could be the product of an injury/pathology, this is unclear, and these inferences are representative of only a single pathology. The wide IQRs found across outcomes in the current thesis suggests inclusion of a diverse sample. Whether this diversity is related to the inclusion of participants that should have been excluded is unknown. Thus, absence of adequate participant screening in the current thesis may explain some of the inter-study hip NJM heterogeneity.

Large inter-study hip NJM variability is not unique to squat investigations, as sagittal hip NJMs (which show the most *temporal* agreeability in resisted squats), were found to have low consistency ($R^2 = 0.66$) when the same participant was assessed across multiple laboratories[99]. Thus, a range of factors including computational methods, marker sets, human variability, exercise technique, and sex, likely result in the mosaic of tri-planar hip NJMs reported during squatting. The great variability in hip NJM findings and both the variability and poor reporting of methodologies, make it difficult to identify 'best' studies and whether the current thesis utilized 'best' practices. Until further research is designed to delineate reasoning for between-study variation and isolates for discussed parameters, all that can reasonably be done is acknowledge the presence of variation and *potential* reasoning. Future investigations should focus on better understanding between-study hip NJM variations prior to any recommendations made at the global study level for hip NJM related outcomes.

5.3 Strengths and Limitations

The strength of the current investigation is the generalizability to the greater female-resistance trained population. The application of this generalizability should be taken with caution due to the uncertainty surrounding whether included participants met the entirety of the inclusion/exclusion criterion. There is potential that both those with and without current back and lower extremity injuries and/or history of operations, would express NJM magnitudes different than that found in the current thesis. The large IQRs demonstrated for most outcomes is suggestive of a diverse sample that could be the result of including those who should have been excluded. Hip NJM polarity patterning showed clear clustering suggesting despite large variations in magnitudes, participants demonstrated distinct strategies. As the key finding from the current thesis is hip NJM strategies (i.e., polarities) due to the limitations of the assumptions surrounding muscle utilization extrapolations, findings are most applicable to females who can perform high intensity squat variations with minimal impedance (i.e., those who can perform multiple high intensity lower extremity strength tests in a single training session). The magnitude of NJMs can not be generalized in the same fashion, and caution should be taken when comparing intensities both within and between variations (i.e., comparing magnitudes within the current thesis and to that of other investigations).

Managing fatigue was one of the greatest challenges of the current investigation. To standardize intensity, each of the four exercises needed to be independently 5 RM tested. Yet, the time each participant spent in the lab needed to be minimized to increase study retention. This resulted in the selection of three study days to familiarize, test, and collect data for four high intensity squat exercises. FSS, LSS, and SLS 5 RM tests in addition to data collection order for *all* exercises, were randomized to minimize the effects of fatigue at the sample level. Still, this was not sufficient in completely minimizing fatigue – as demonstrated by the participant who failed a BS set *during* their data collection session. The effects of fatigue on tri-planar hip NJMs during loaded squat variations is currently unknown and would be a fascinating topic for future investigations.

A product of single investigator overload, BS depth was higher than predicted. Median thigh flexion angle was only 65° during the BS; much less than the ~90° goal. A sole assessor had to simultaneously spot and observe for technique deviations, rendering identification of *insufficient depth* repetitions impossible, as single-spotter squat technique places the spotter directly behind the squatter. Though unclear the magnitude of the effect, compression garments (neoprene wraps) wrapped about the thigh and leg segments used to aid marker cluster placements during data collection, likely warped the *'bottom of squat'* feeling; presumably due to unfamiliarity with segment compression during high intensity exercise. A combination of these and potentially other factors likely skewed 5 RM BS intensities higher than reality. Despite this, most (if-not-all) squats, visually appeared similar in depth and technique to that of an average squat performed in a high-performance training setting (based on the authors years as a strength and conditioning coach). Additionally, hip- and not thigh-flexion was the critical kinematic metric as muscle moment arms are reported relative to joint and not segment, angles.

The significant differences found between hip and knee sagittal plane NJMs may suggest intensity was not mechanically standardized across exercises. Yet, assessing intensity through an extensor-focused perspective does not consider differences in rate-limiters between squat variations. For example, the SLS is likely limited by the progressively increasing balance challenge with hip flexion, as displacing limbs make constraining the COM to the small area of a single foot, difficult. In contrast, the BS has a wide base of support, minimizing balance-related difficulties. Additionally, the larger non-sagittal NJMs displayed in single-limb dominant exercises suggests they may play larger roles in determining squat success than just sagittal NJMs. It's unlikely the differences in sagittal plane NJMs are limitations of the current investigation for two reasons: (i) 5 RMs were still tested for each exercise resulting in *global equivalency* (training intensities are commonly based on tested RMs), and (ii) sagittal plane hip NJM magnitudes don't appear to visually correlate with frontal/ transverse plane magnitudes (Figure 5). Thus, differences in sagittal plane NJMs should not be seen as a limitation of the current investigation, but an area of future exploration in quantifying rate-limiters in predominantly and completely single limb loaded exercises.

It's also unclear the accuracy of the stepping test in determining limb preference/dominance for the assessed squat variations. As both limbs were used as lead limbs during familiarization, many participants made remarks regarding the selected limb *"feeling like the weaker of the two"* during the FSS, LSS, and/or SLS 5 RM tests. As authors did not consciously decide whether to select the dominant or preferred limb, quantifying the level of inaccuracy is impossible. Future investigations incorporating single limb movements should clearly state limb selection criteria and provide thorough justification.

5.4 Future Investigations

Biomechanical researchers interested in hip NJMs during squatting exercises should focus on improved reporting methods to facilitate inter-study comparisons. As noted in previous sections, the current range of reported tri-planar hip NJMs in similar resisted squatting investigations, is substantial. Focus should be first emphasized on using a vetted and consistent methodology. This includes but is not limited to HJC approximation methods, marker sets, filtering thresholds, and other computation considerations. By minimizing variability in these factors, more confidence can be placed in ascribing differences to the independent variables.

Despite the near-null frontal plane hip NJM magnitudes in the BS, FSS, and LSS, transverse plane NJMs for the latter two (and the SLS) approximate those of activities of weekly living that there is a potential for strength adaptations from a training study. Additionally, participants were only exposed to two brief familiarization periods for the FSS, LSS, and SLS. Thus, participants were likely still unfamiliar with some of the variations where 12- to 24-weeks of consistent training could solidify NJM strategies and magnitudes. Additionally, even when cross-sectional studies estimate muscle load/stressed using muscle modeling[80], these alone are insufficient in determining adaptations. The substantial frontal plane NJMs demonstrated at the start and end of SLS repetitions also warrants exploration; potentially via resisted gait (e.g., farmer's walks) to mimic the combined near-neutral hip and single limb postures.

Elucidating the unexpected patterning of NJM strategies and squat variations would be an additional beneficial investigational follow-up. As GRFs alone have been demonstrated to be insufficient in approximating hip NJMs, reasoning for squat- and participant-specific strategies may require additional kinematic considerations. Delineating the reasons for divergent hip NJM strategies within and between resisted squat variations, could aid physical health practitioners in exercise prescription and cueing.

5.5 Concluding Remarks

Resisted squat variations require unique tri-planar hip NJMs, with the FSS, LSS, and SLS variations potentially utilizing the deep gluteal muscles. As hypothesized, the SLS did demonstrate the greatest frontal plane, hip abductor NJMs, despite generating practically null median-lateral GRFs. In contrast, the BS did not elicit the largest transverse plane, hip internal rotator NJMs, despite eliciting medial GRFs similar in magnitude to the LSS. The BS and SLS appear to elicit unique hip NJM profiles, with the former demonstrating a sample-level bias for adductor and external rotator NJMs. In contrast, the SLS elicited the largest abductor and internal rotator NJMs. The FSS and LSS, despite their differences in anterior-posterior and medial-lateral GRFs, have similar frontal and transverse plane hip NJM profiles. Due to participants noting the difficulty and poor comfort levels associated with resisted LSSs, in addition to the challenges of spotting the LSS, physical training professionals can likely utilize the FSS without any worries of missing any LSS-specific hip muscle demands.

Medial GRFs alone are not valid measures in approximating frontal and transverse plane NJMs during resisted squatting in resistance trained females. When selecting exercises designed to target/mimic real-world or sport settings, physical performance and rehabilitative professionals need to consider the stimulus of interest. For example, if a performance coach was interested in selecting exercises that simulate change of direction actions, they should consider whether it's the internal NJMs or the external force profile they're looking to recreate. The current thesis demonstrates that selecting resisted squat exercises based solely on external force profiles leads to incorrect internal hip NJM assumptions, and that coaches must be specific in both (a) the stimulus of interest and (b) knowing that the selected movement(s) will elicit the stimulus of interest.

Current findings support a longitudinal training study to determine if the NJM intensities are sufficient to result in deep gluteal muscle strength adaptations. Findings from future training studies are required prior to recommending performance and rehabilitation practitioners implement resisted squat variations into clientele physical training programs.

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Appendix

A1. Oral presentation template used to recruit participants from undergraduate courses and/or laboratories at the University of Alberta

"Hello everyone,

My name is Zack Fielding and I am a graduate student here the Faculty of Kinesiology, Sport, and Recreation as well as a strength and conditioning coach for the Panda's Volleyball team. I am a Master's student in the biomechanics laboratory here though you may recognize me from various undergraduate courses as I've TA'd a few of them now.

I am currently recruiting for a biomechanical research study investigating how different squat exercise variations load the hips and corresponding gluteal muscles (Study ID: Pro00107448). **Females with no current lower back and/or lower extremity injuries and no history of spine and/or hip surgery, with at least 1 year of resistance training experience including the barbell squat exercise may be considered for inclusion.** The main reason this study focuses on physically active females is due to the higher rate of physical activity related back and knee pain associated with weak hip muscles in females.

Voluntary participation in this study would require three separate visits to the Sports Biomechanics Laboratory at the University of Alberta (North Campus; Van Vilet Complex). Each visit would last between 45-105 minutes and be separated by 72 hours (minimum). Protocols to mitigate COVID-19 exposure risk are in place. You would be asked to perform four squat exercise variations at a moderate intensity determined by a five-repetition maximum test. Data collection on the final day would require you to perform each exercise at the predetermined intensity while standing on force platforms and outfitted with reflective markers on the legs and torso for 3D motion analysis.

The findings from this investigation will update current physical training guidelines for strengthening the hip muscles.

If you are interested in the study or have any questions regarding the study, please contact me at zfielding@ualberta.ca (will be written on board)"

<< copies of the flyer will be left behind >>

Bolded passages highlight inclusion criteria

A2. Social media recruitment written component

"A new weightlifting study in females is recruiting participants!

The Neuromusculoskeletal Research Program at the University of Alberta has recently begun a study investigating how free weight squat exercise variations load the hips and corresponding musculature (Study ID: Pro00107448). **Females 18-50 years of age with 1 year of resistance training experience including the barbell squat exercise, no current lower back and/or lower extremity injuries, and no history of spine and/or hip surgery may be eligible for inclusion**.

Voluntary participation in this study would require three separate visits to the Sports Biomechanics Laboratory at the University of Alberta (North Campus; Van Vilet Complex). Each visit would last between 45-105 minutes and be separated by 72 hours (minimum). Protocols to mitigate COVID-19 exposure risk are in place. You would be asked to perform four squat exercise variations at a moderate intensity determined by a five-repetition maximum test. Data collection on the final day would require you to perform each exercise at the predetermined intensity while standing on force platforms and outfitted with reflective markers on the legs and torso for 3D motion analysis.

The findings from this investigation will update current physical training guidelines for strengthening the hip muscles; particularly important in physically active females due to a higher prevalence of physical activity related back and knee pain.

If you are interested in the study or have any questions regarding the study, please contact me at zfielding@ualberta.ca.

<< attach flyer and/or social media recruitment image >>"

Bolded passages highlight inclusion criteria

A3. Study recruitment flyers posted across campus and on social media

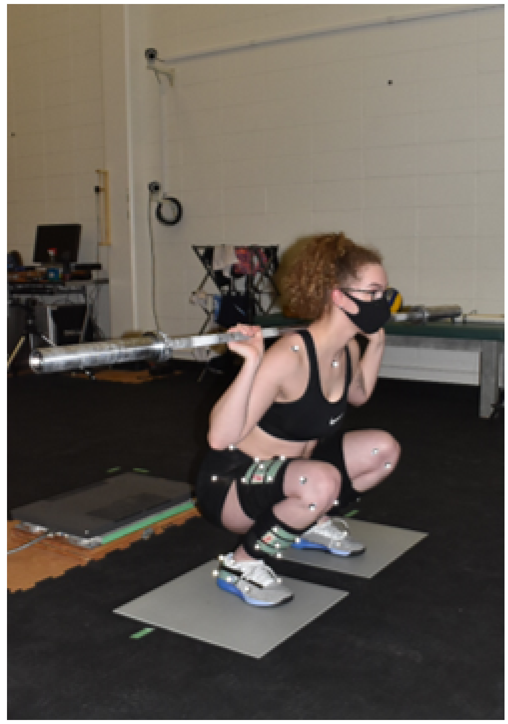
What are the Effects of Free Weight Squat Variations on Deep Gluteal Muscles in Females?

Eligibility criteria:

- Female, 18-50 years of age
- Previous and current resistance training experience
- Can perform 5 repetitions of 80% body mass during the back squat exercise
- No current lower back and/or lower extremity injuries and no history of spine and/or hip surgery

Commitment:

- Three, 45-105 minute in-laboratory sessions (Sports Biomechanics Lab, University of Alberta)



Interested or have questions? Contact the primary investigator Zachary Fielding (zfieldin@ualberta.ca)

Study ID: Pro00107448

Scan here for an email template

Flyer Version Date: May 3rd, 2021
Study ID: Pro00107448

A4. Social media study recruitment image used at times in conjunction with other social templates