SUPPORT MOMENT DISTRIBUTION AND INDUCED ACCELERATION ANALYSIS OF THE BARBELL BACK SQUAT

by

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ABSTRACT

The barbell squat exercise is performed in settings ranging from rehabilitation through to developing muscle size, strength and power. Unfortunately, the lower extremity coordination producing the squat is not clearly understood. This thesis involves two studies evaluating how lower limb joints and muscles coordinate varied squat performance. Study one included 19 females who performed squats at three randomized depths (above parallel, parallel, below parallel) and three loads (unloaded, 50%, 85% 1 repetition maximum). Inverse dynamics analysis revealed that peak hip and ankle extensor moments varied with load but not depth and were greatest when using 85% 1 repetition maximum. Within each depth, as load increased so did peak knee extensor moments. Peak knee extensor moments were greatest when squatting below parallel with load. Within each depth as load increased contribution of the hip increased whereas the knee decreased. Ankle contribution was only influenced by load. When squatting to deep depths with load, the contribution of the hip decreased while the knee increased. In study two, 13 females squatted to the same 3 depths using 85% of their 1 repetition maximum at each respective depth. Performance was evaluated by estimating the individual muscle force production and the individual muscle contribution to whole body acceleration using a musculoskeletal model. The gluteus maximus and adductors increased peak force to parallel while the hamstrings and rectus femoris increased to below parallel. At deep depths, the vasti decreased peak force while the hamstrings and rectus femoris increased peak force. The induced acceleration of the vasti at transition decreased with depth while the hamstrings and rectus femoris increased. Because muscles can instantaneously accelerate all joints in the body, it’s possible that at transition the hamstrings accelerated the hip and knee into extension while the rectus femoris also accelerated the knee and hip into extension while the soleus accelerated the ankle and knee into extension. In conclusion, a complex coordination of the lower extremity is used performing the squat. Varied coordination indicates that depth and load specificity is important and should be taken into consideration when programming based on the status and goals of the individual.

CHAPTER 1

INTRODUCTION

The Barbell Back Squat

The barbell squat is a multi-joint, lower body movement that is used in a variety of exercise situations ranging from development of strength and power in elite athletes, through to rehabilitation from musculoskeletal injury. Because of the wide array of applications for the squat exercise there are many ways in which a barbell squat can be performed. For example, the front squat is performed with a barbell across the clavicles on the anterior side of the body, while the back squat is performed with a barbell on the posterior side of the body. The back squat can also be performed with the barbell placed in different positions such as high on the back or low on the back. Within each general technique there are subtle differences for example the high bar back squat is performed with the barbell across the trapezius muscles above the C7 spinous process, while the low bar back squat is performed lower on the back across the spine of the scapula resting on top of the posterior deltoids (24). Additionally, barbell squats can be performed using various loading schemes and range of motion (depth of squat) depending on the desired outcome of training.

The range and scope of differences performing the squat have led to debate over which load, and depth is optimal; it is a nuanced debate because the answer likely depends on the desired training outcomes. For example, if an athlete is competing in a sport, such as powerlifting, where they have a specific criterion that is required the athlete must squat specifically to the demands of that sport. However, if an athlete is rehabilitating an injury, the depth and load may need to be modified to allow the athlete to work within a range of motion that avoids undesirable joint load and/or pain.

Anecdotally evidence suggest that some strength coaches believe that there are superior ways in which the squat can be performed to increase strength (43), hypertrophy or power. For example, Mark Rippetoe is a well-known strength coach who has written many books on strength training. Rippetoe believes that the low bar squat is superior to increasing strength compared to the high bar squat due to biomechanical and anthropometric factors (43). He believes that it is more advantageous to squat low bar compared to high bar because it requires greater forward torso lean which increases the moment arm of the bar about the hip and decreases the moment arm of the bar about the knee.

Anecdotally, other coaches in the strength community have pushed back against this notion and appear to believe that the high bar squat is superior to producing quadriceps hypertrophy suggesting that because the knee must travel further forward over the toes knee extensor moments are increased (leading to greater mechanical tension on the quadriceps) thus over time more hypertrophy. Other coaches have proposed that there are more “athletic” squats compared to other styles of squats. It has been suggested that the front squat and high bar back squat are more “athletic” than the low bar squat because it exhibits a more upright torso.

There also appear to be many opinions about squat technique and injury risk. Anecdotally, some coaches and physiotherapists seem to believe that high bar squats are “bad” for the knees because the knees must travel further over the toes and that low bar squats may be worse for the lower back because it requires greater forward lean and a more horizontal torso. This debate also continues relative to the consideration of squat depth. It appears some trainers and clinicians advocate against deep squats because they are “bad” for the knees and to avoid deep squats because the pelvis may go into posterior tilt colloquially called “butt wink” and should be avoided because it is “bad” for the low back. Although there is evidence that greater external loads an deeper depth increase knee joint moments (12,20) and patellofemoral joint loading (50), the opinions of practitioners seem to assume that greater forces and moments about the joints are worse in terms of injury potential. While tissue load does play a role, these opinions do not consider an individual’s prior training history, their ability to adapt or the dose in which these stressors are applied. A high-level lifter may impose large amounts of force and torque about a joint, but this alone does not necessitate an increased risk of injury.

Although these claims are popular in practice, there is little empirical scientific data to support these hypotheses and they remain conjecture. Because the squat is a common exercise implemented in a variety of settings, it is important to understand the differences in technique and their consequences, be they positive or negative, on both performance and wellbeing.

Back Squat Application in Athletic Performance and Clinical Settings

The barbell squat is used in a variety of applications ranging from improving athletic performance through to clinical rehabilitation (45). In athletic performance settings, the squat is used to promote hypertrophy of the lower body musculature, and to develop muscular strength and power. When implemented in an athletic performance setting there is a variety of loading, repetition and movement speeds used. For example, when hypertrophy is the goal the load is often light to heavy and loads are lifted for between 6-20 repetitions with consistent tempo because hypertrophy is similar between low and high loads when volume and relative intensity are similar (46); when strength development is the goal loads are heavy and often near maximal, lifted fewer times and the movement tends to be slower due heavy loads (46); power development tends to use moderate loads lifted a moderate number of times (3-6 repetitions) but at high movement velocity. The squat is also used to test force production of the lower body musculature at varying velocities (45).

In clinical settings, the squat is used for rehabilitation often to assist in the progression back to full lower body function following common injury. In particular the squat is often used to rehabilitate injury sustained at the knee joint that damages the Anterior Cruciate Ligament (ACL) because it has been found to generate less strain on the ACL when compared to other movements such as knee extensions (17). Quadriceps strengthening exercises are often implemented in populations suffering from knee osteoarthritis (OA) because quadriceps weakness has been associated with knee OA (4) and the squat can effectively strengthen the quadriceps. It is likely that pain associated with knee OA may contribute to quadriceps weakness; however, it is important that individuals' experiencing knee OA continue to train to tolerance to continue to build quadriceps strength. Importantly, within both clinical and athletic performance settings, modifying squat parameters such as the depth and load of the exercise can influence both training effect and the lower extremity joint loading (17).

Laboratory Based Research of the Squat

Investigations into the efficacy of squat performance have detailed, to some degree, the relative role of the lower limb joints. These investigations have approached the problem by either using kinematic analysis which generally describes the movement of the body segments and joints, or kinetic analysis which integrates kinematics with measures of external forces through the process of inverse dynamics. Inverse dynamics analysis calculates and allows the evaluation of kinetic variables such as the joint forces and moments that occur throughout the squat movement.

These variables are of interest because they provide insight to the quantitative aspects of movements and allow the researcher to infer or theorize what neuromuscular parameters are important to improve performance or reduce the risk of injury. In general, research examining how the ankle, knee, and hip contribute to squat performance when squatting with different depth and load combinations has can be categorized as using the following analytical methods: net joint moment (NJM), relative muscular effort (RME), electromyography (EMG), and support moment (M_s) (6,7,11,45,47,52,53).

Squat Studies Using Net Joint Moment Analysis. Net joint moment is a measure of the net flexor and extensor moments of a joint (19,47) and provides the sum of all the moments acting at the joint (7). Research investigating the individual NJM's of the ankle, knee, and hip using four depths, (45°, 90°, parallel, and deep) found that NJM of the hip increased from 90° to parallel, but not parallel to deep (53). Net joint moment of the knee was not found to differ between 45° to 90° and parallel, but did increase at deep depths (53).

A study investigating parallel and deep squat depth using 65% 1RM in a group of trained powerlifters, squatting using a low bar position, and Olympic lifters, using a high bar position, found that deep squats had greater peak hip and knee NJM than parallel squats (52). Furthermore previous research investigating how three squat depths (above parallel, parallel, and below parallel) and three loads (unloaded, 50% 1RM, and 85% 1RM) effect knee extensor moments found that as depth and load increased so too did knee extensor moments (12,20).

A separate study investigating the effects of external load and squat depth on knee joint loading found that as external squat load increased, from unloaded to 35% of bodyweight, and squat depth increased, from 0° to 90° of knee flexion, knee extensor moments increased (50). Contrary to these findings a study investigating the difference in knee joint loading when squatting with 85% 1RM to 70°, 90° and 110° of knee flexion found that the knee extensor moments were not different between depths (21); however, this study had an extremely small sample size of only five subjects. These findings suggest that at a fixed or increasing squat load, depth influences the moments generated about the hip and knee.

Squat Studies Using Relative Muscular Effort Analysis. Relative muscular effort is a measure of the ratio of a muscle group's NJM during a task relative to the muscle group's contribution to NJM during the maximal voluntary isometric contraction (6,7). Previous findings report that RME of the knee increases with depth more so than load, with the opposite happening at the ankle (6,7). At the hip, RME increase with depth and load (6,7). In a follow up study modelling the role of the hip extensor strategy and its effect of quadriceps effort, the authors found that it is necessary to minimize the activation of the hamstrings because greater hamstring co-contraction would require a greater knee extensor moment, thus, a hip strategy that prioritizes gluteus maximus activation over hamstrings activation is warranted (7). The NJM and RME results in the literature show similarities in that the knee moment increase with increasing depths and the hip moment increases with depth to a given extent.

Squat Studies Using Electromyography Analysis. Electromyography is used to measure muscle activity. Squat research using EMG to study the interaction of depth and load have found varying results related to muscle activity of the lower extremity (11). Muscle activity has been shown to increase when increasing external load (9,39,41,52). Activity in the quadriceps, a large muscle group involved in the squat, tends to peak at approximately 80° to 90° of knee flexion, and remains relatively constant thereafter, suggesting that squatting past 90° might not result in further quadriceps contribution (16,40,45).

To further investigate the influences of depth on muscle activity during the squat, full, front, and parallel squats have been studied to observe the activity in the upper gluteus maximus, lower gluteus maximus, biceps femoris, and vastus lateralis. In the front squat, the barbell is on the anterior side of the body on the clavicles. In a full squat, the squat is considered complete when the hamstrings contact the gastrocnemius muscles. In a parallel squat, the squat is considered complete when the tops of the thighs are parallel to the floor. Front, full, and parallel squats have not been found to significantly differ in electrical activity of these muscles (11).

Another finding that varies in the literature regarding squat depth is the activity of the gluteus maximus, a large muscle involved in the squat (8,11). Gluteus maximus muscle activity has not been found to increase between the partial squat (squats performed above parallel) and parallel squat, but to increase significantly during the full squat (8). These findings suggest that deep squats have the greatest amount of gluteus maximus activity.

The differing results of depth-related gluteus maximus EMG between Caterisano et.al. (2002) and Contreras et. al. (2016) may be due to differences in relative loading, or percent contribution to ones 1 repetition maximum (8,9,11). This variable could have affected the outcome because subjects used greater 10 repetition maximum loads during the parallel squat compared to the full squat (8,9,11). However, some speculate that Contreras's findings may have not found differences in EMG activity, like previous research, because the study was underpowered (11).

To further complicate the EMG findings pertaining to hip extensor muscle activity Silva et. al. (2017) found that the EMG activity of the gluteus maximus, biceps femoris and soleus was greater in the partial back squat compared to the full back squat (32). It is apparent that studies using EMG analysis to estimate the role of muscle during the squat present mixed findings regarding the effects of depth on lower extremity muscle activity.

Squat Studies Using Support Moment Analysis. Support moment is sum of the extensor moment of the knee (M_k), extensor moment of the ankle (M_a), and extensor moment of the hip (M_h) (51). Winter originally defined M_s in the context of lower extremity limb support during stance phase of gait. Winter proposed that the "basic function of the lower limb during stance is to resist collapse and extend sufficiently to achieve the required push-off" (51). This suggests that when the lower extremity collapses, flexion takes place at the three lower extremity joints (ankle, knee and hip). To prevent this collapse, extensor activity is required at these joints. This method investigates the coordination and contribution of the ankle, knee and hip to support the lower extremity from collapsing while performing a given movement.

Previous research has investigated summing kinetic measures such as lower extremity NJM of the ankle, knee and hip into a M_s during the barbell back squat with increasing loads but not depths (19,47). In a study by Flanagan et. al. (2008) contribution to M_s was investigated with increasing loads at a fixed depth, 45% of leg length, and found that the ankle contribution increased with loading to 75% 3RM, the knee contribution decreased when increasing load, and the hip contribution increased with loading to 75% 3RM (47). Overall, the ankle contribution was less than the knee, which was less than the hip in all except the lightest condition (47).

However, there is no current study specifically investigating the contribution to M_s of the lower extremity joints, in the squat, as it pertains to increasing depths. Investigating the extensor moments at each joint during movement may give insight to the net effect of the musculature involvement throughout the movement (51).

Limitations of Previous Methodologies. Overall, the previous literature regarding the performance of the squat exercise present mixed information and, in general, no consensus statement can be made regarding how differing technique is produced or mediated by the muscles and joints of the lower body. This may be in part because while the methods described although useful, are subject to the limitations inherent to each analytical method.

For example, NJM analysis only considers the net of the flexor and extensor moments and resolves the problem to a single moment expressed at each joint. However, in reality, there are numerous muscles being coordinated about each joint that result in the net joint moment which, while represented in the magnitude of the net moment, cannot be individually determined by this type of analysis. Furthermore, NJM doesn't consider, due to dynamical coupling, that muscles can accelerate joints that they do not cross or opposite to its anatomical classification (54).

Relative muscular effort is a relative contribution of the torque produced about a joint during a task compared to the torque produced in a maximum voluntary contraction. This again does not calculate the force that a muscle is producing but rather the net of all muscles acting about a joint and their cumulative torque generated about a joint during a task compared to its "maximum" isometric torque at the same joint angle.

Similarly, EMG is a measure of muscle activity not a measure of muscle force. Electromyography measures if a muscle is "on" and how much it is "on", but it does not measure how much force a muscle is producing or the contraction type (concentric, isometric, eccentric).

Finally, M_s does give insight into the coordination of the hip, knee and ankle to support but it does not calculate muscle force but rather the NJM about each lower extremity joint and the contribution of those NJM to the total M_s. While each of these methods can detail if there is a general shift in the pattern of force production at each joint during the movement, they are neither sensitive enough to describe the movement produced by a given muscle nor its overall contribution to the movement.

Musculoskeletal Modelling
<u>An Induced Acceleration Analysis</u>

Many of the previously described limitations to analytical methods are a result of the inability to directly measure forces produced by muscle. In fact, it is so invasive and difficult to measure muscle forces *in vivo* that it is ethically unreasonable to consider it a valid methodological approach in human subjects. To overcome this limitation, use of musculoskeletal models is required within which muscle forces that cannot be directly measured are constructed within a theoretical framework of a multi-link rigid body model that approximates the human musculoskeletal system. This allows for evaluation of cause and effect relationships between muscle forces (based on muscle and tendon properties such as the force-length-velocity relationships, pennation angle, instantaneous moment arms, tendon slack length and tendon elasticity), and neuromuscular excitation patterns and body motions (14) .

The elements of the musculoskeletal model are sets of differential equations that describe muscle contraction dynamics, musculoskeletal geometry, and body segmental dynamics which account for the time dependent behavior of the musculoskeletal system in response to neuromuscular excitation (14). The inputs into a model can include subject anthropometry and the experimental kinematics (which are used to scale the model and find joint angles) and kinetics.

While it is possible to use an inverse dynamics approach during the back squat, it is difficult to determine how individual muscle forces contribute to motion of the center of mass. Inverse dynamics calculates a summation of all the muscles acting to produce torque about the joint which results in a force-sharing problem. Conceptually, inverse dynamics is subject to substantial and largely unpredictable errors in the computed motion characteristics due to an overdetermined system and experimental error resulting in inaccurate representation of the kinematics and ground reaction force and low model fidelity (29). This is the fundamental inverse dynamics problem (29). This leads to the generation of fictious forces or residuals in the model.

To address the fundamental inverse dynamics problem and to attempt to solve the muscle redundancy problem, an optimization algorithm, such as static optimization, dynamic optimization, or computed muscle control is used to match individual muscles to the net moment about a joint. Static optimization solves for the unknown forces by using the known motion of the model and a set of constraints, such as force-length-velocity properties. There is an infinite amount of muscle coordination strategies that could have led to the measured net joint moment; therefore, the solution is determined using not only an optimization algorithm but also a cost function. Commonly used cost functions minimize metabolic consumption and/or minimize muscle activation required to produce the observed joint moment (13).

To resolve the problems of dynamic inconsistency (where ground reaction forces and marker accelerations don't equate to F=ma) between experimental data and the model a residual reduction algorithm (RRA) is used. Residuals, or "left over forces" are present in the model because of parameters that are either not represented in the model or only represented in part. For example, it is difficult to model how force is dampened by soft tissues such as muscle or cartilage. For this reason a RRA is used to refine the model (14). Residual reduction algorithm is used to minimize residuals, nonphysical forces, associated with errors involved in modelling and data processing. Residual reduction algorithm redistributes the mass or the kinematics of the participants to allow for data to more dynamically consistent with the ground reaction force.

An important consideration with respect to human motion and coordination is that muscles produce forces that can accelerate all segments, joints, and the whole-body COM through intersegmental dynamic coupling (54). This consideration allows muscles forces to be evaluated mathematically within the framework of a musculoskeletal model to determine the contribution to this acceleration (26). This is called an induced acceleration analysis (IAA). An advantage to IAA is that it can determine a plausible answer to the contribution of each muscle force to the acceleration of the body during movement outside of the textbook anatomical classification. Many studies have used IAA to investigate muscle forces during running (26,27), walking (33,37), stair climbing (36), balance recovery (25) and impaired gait (48); however, there are no studies that use IAA during the back squat. Employing an IAA will provide valuable information about the role of individual muscle forces to squat performance.

Thus, the specific aims of this study are to first examine the joint moment contribution of the ankle, knee, and hip to support and second, perform and IAA to evaluate how muscle forces contribute to back squat performance at different depths.

Statement of Purpose

Different methodologies and the limitations inherent to the analytical techniques of NJM, RME and EMG make it difficult to come to a conclusion regarding the coordination of the lower extremity joints and muscles to performing the squat when altering depth and load. Therefore, future investigation is necessary to determine how alteration of the squat performance is produced by the neuromotor system. Two principal analyses appear well suited to this purpose 1) M_s analysis can determine if joint extensor moments differ in their contribution to the M_s when varying squat technique such as load and depth and 2) musculoskeletal modeling can determine differences in the magnitude of individual forces and the contribution of muscle forces to the acceleration of the whole body center of mass. Therefore, the purpose of this study is to examine how lower limb joint and muscle coordination changes when manipulating squat parameters.

CHAPTER 2

SUPPORT MOMENT DISTRIBUTION DURING THE BARBELL BACK SQUAT AT DIFFERENT DEPTHS AND LOADS IN RECREATIONALLY TRAINED FEMALES

Contribution of Authors and Co-Authors

Manuscript(s) in Chapter(s) 2

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Contributions: This author made contributions to the initial study conception and data collection.

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Support Moment Distribution During the Barbell Back Squat at Different Depths and Loads in Recreationally Trained Females

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Running Title: Support Moment Distribution with Squat Depth and Load

Support Moment Distribution During the Barbell Back Squat at Different Depths and Loads in Recreationally Trained Females

Abstract:

The squat exercise is widely used in both athletic and clinical settings. However, the coordination of the lower extremity during the lift is not well understood. The purpose was to compare the peak moments of the lower extremity joints at three squat depths (above, parallel and below parallel) and three depth-specific squat loads (unloaded, 50% 1RM, and 85% 1RM) and find their contribution to support (M_s). Nineteen females performed squats in a randomized fashion. Inverse dynamics and Winter's M_s equation were used to calculate peak moments of the hip, knee and ankle and calculate their contribution to M_s ($\alpha < .05$). Peak hip and ankle extensor moments varied with load but not depth and were greatest when using 85% 1RM. Peak knee extensor moments demonstrated a depth by load interaction. Within each depth as load increased so too did peak knee extensor moments and were highest squatting below parallel when loaded. The hip and knee contribution to M_s demonstrated a depth by load interaction while the ankle was only influenced by load. Within each depth as load increased hip contribution increased whereas the knee decreased contribution. When squatting with load the contribution of the hip decreased at below parallel while the knee increased. There is a complex coordination of the lower extremity to perform the squat when varying parameters. To maximize peak moments squat with load and squat deep for the knee; however, depth and load dosages should be taken into consideration based on the status of the individual.

Key Words: biomechanics, coordination, strength training

INTRODUCTION

The barbell squat is a commonly used strength and conditioning exercise often incorporated into both athletic development and clinical rehabilitation exercise programs (9,45). In athletic development settings, the squat is used broadly to develop hypertrophy, power, and strength of the lower body musculature (9,45). In clinical programs, the squat is used for rehabilitation to assist in the progression back to full lower body function following injury. Importantly, there is no one preferred method by which practitioners perform the squat movement. This has consequences related to user outcomes because modifying squat parameters such as depth and load can influence training effect and lower extremity joint loading (9,45).

Previous research has examined how the joints of the lower limb contribute to squat performance and how variation of squat technique alters joint contributions. This has primarily focused on modification of squat depth and load, and four primary techniques have been used: net joint moment (NJM), relative muscular effort (RME), electromyography (EMG), and support moment distribution (M_s) (6,7,9,12,20,21,45,47,52,53). It has been reported that as squat external load increases net extensor moments, RME, EMG and peak M_s increase (6,7,12,20,38,47,50), however it is less clear as to the effects of varied depth on lower extremity muscle activity and joint loading (6,8,11,12,21,32,52,53). While NJM and RME suggest that the contribution of the knee extensors increase with depth (6,7,12,20,52,53), EMG literature suggests that knee extensor muscle activity peaks around parallel depth (9,45,53). Relative muscular effort findings suggest that contributions from the hip extensors increase with increasing depth (6,7); however, hip NJM have not been found to increase past parallel depth (53). Electromyography data are mixed as well in that some studies find hip extensor activity, especially the gluteus maximus, to be greatest at deep depths (8) while others have found hip extensor activity to be greatest at shallow depths (32) and still others have found that the hip extensor activity peaks at parallel depths and plateaus thereafter (11).

The different conclusions regarding the interaction of depth and load between different methodologies suggests no consensus statement can be made regarding how changing technique, via changes in depth or load, is produced or mediated by the muscles and joints of the lower body. This may be in part because while the methods described although useful, are subject to the limitations inherent within each analytical method. For example, NJM analysis only considers the net of the flexor and extensor moments and resolves the problem to a single moment expressed at each joint and does not consider the coordination of the lower extremity muscles throughout the movement. Relative muscular effort is a relative contribution of the torque produced about a joint during a task compared to the torque produced in a maximum voluntary contraction. This does not consider the contribution of each joint to the entire lift but rather each joint to its own maximum torque producing capabilities. Lastly, EMG only measures if a muscle is “on” and how much it is “on”.

A methodological approach that overcomes many of the aforementioned limitations is the support moment (M_s) which is the sum of the extensor moment of the knee (M_k), extensor moment of the ankle (M_a) and extensor moment of the hip (M_h) (51). M_s is useful because it provides insight into lower extremity coordination by measuring the relative contribution of each joint to support and how that relative contribution changes across activities. Previous research, investigating the contribution to peak M_s with increasing loads at a fixed depth of 45% leg length reported that the ankle contribution increased with loading to 75% 3RM, the knee contribution decreased when increasing load, and the hip contribution increased with loading to 75% 3RM (47). In all but the lightest conditions the hip contributed the most, followed by the knee then ankle (19,47).

While this provides insight that joint contributions to total lower body support change with load, to date it is unknown how joint contributions to lower body support change with depth, or with combinations of increased depth and load. Understanding the changes in coordination will be helpful when creating exercise programs because neuromuscular adaptations are specific to the task at hand. Thus, the purpose of this study is to examine how the ankle, knee, and hip joint moments contributions to M_s change while increasing squat loads and depths. We hypothesized the peak M_s , peak M_a, peak M_k, and peak M_h would increase with load but not depth and that percent contribution of the ankle would decrease with depth but increase with load, percent contribution of the knee would increase with depth but not load, and percent contribution of the hip would increase with depth and load.

METHODS

Experimental Approach to the Problem

A within-subject experimental design was used to investigate the effects of squatting to different depths with different percentages of 1RM. Participants were evaluated on two days: on day one, depth specific 1RM were obtained for above parallel, parallel and below parallel depths. Depths were defined as follows: above parallel was measured with a goniometer at 90 degrees of knee flexion, parallel was reached when the top of the thighs were parallel to the floor, and below parallel was reached when the hamstrings contacted the back of the gastrocnemius muscles. The load conditions were unloaded, 50% 1RM, and 85% 1RM and specific to each depth. Within 24 hours to 1 week later, participants returned to the laboratory for the 2nd session. During the 2nd session motion capture data was collected while two squat repetitions were performed at the at each of the three different depths with the three different load combinations. A total of 9 trials were performed with the order of trials randomized across participants. Whole body kinematics were recorded using a motion capture system while ground reaction force data was recorded using two force plates.

Subjects

Nineteen female participants (age: 25.05 ± 5.76 years; height: 1.58 ± 0.08 m; body mass: 62.51 ± 10.17 kg) with an average of 3.84 ± 2.59 years of squatting experience participated in this study. A Physical Activity Readiness Questionnaire (PAR-Q) used for screening and for exclusion criteria. If subjects marked “Yes” on any of the questions they were excluded from the study. Competitive elite lifters were also excluded from this study. The inclusion criteria were being between 18 - 60-year-old females with a minimum of 1 year of squat training experience. Visual assessment of the squat was also used in the inclusion criteria. If subjects could not perform a full depth squat while keeping heels on the ground, a neutral lumbar spine, and no pain, then they were excluded from the study. Written informed consent was provided by each subject and the protocols of the study were approved by the University Institutional Review Board.

Procedures

Data were collected over two sessions. In the first session, subjects established the appropriate depth for each condition and a 1RM for each depth. A field laser and piezo light switch was used to help establish depth and cue the ascent of the squat (Banner Engineering Corp., Minneapolis, MN)(20). A brief pause was implemented before the squat ascent. An individualized 10-minute warm up, typical of each participant’s usual routine, was implemented prior to squatting. During the 1RM testing, 1 minute of rest was allowed between squat attempts.

Motion capture collection occurred during the second session. Retroreflective markers were placed bilaterally on the following bony landmarks: anterior superior iliac spine, posterior superior iliac spine, medial and lateral femoral epicondyles, medial and lateral malleoli, posterior calcaneus, and head of the 2nd metatarsal. Additional tracking markers were placed on the iliac crests, with clusters of 4 markers on a plastic shell on the lateral thigh and shank, and over the base of the 5th metatarsal. Kinematic data was recorded using a 12- camera motion capture system (Qualisys Inc., Gothenburg, Sweden) sampling at 250 Hz while ground reaction forces were recorded from two force plates (Model 600900-10, Bertec Corporation, Columbus, OH) sampling at 1000 Hz. A 64-channel analog-to-digital board was used to synchronize force and kinematic data.

Data Analysis:

Raw kinematic and kinetic data were exported to Visual 3D (C-Motion, Inc., Rockville, MD) where they were filtered using fourth order zero-lag low pass Butterworth filters with cutoff frequencies of 8 Hz and 20 Hz, respectively. Joint angles were calculated using an XYZ Cardan rotation sequence corresponding to flexion/extension, ab/adduction, and axial rotation. Peak knee joint angles on each trial were then calculated. Net joint moments at the ankle, knee, and hip were calculated using standard inverse dynamics techniques, and expressed as internal joint moments. Total support moment was calculated as described by Winter (51):

$$\mathrm{M_s = M_k - M_a - M_h} \qquad (1)$$

where M_s is the total support moment, M_k is the knee extensor moment, M_a is the ankle plantar flexor moment, and M_h is the hip extensor moment. For each trial the peak joint moments, peak M_s, and the percent contribution from each joint to peak M_s were calculated.

Statistical Analysis

Participant 1RMs at each depth were analyzed using a one-way repeated measures analysis of variance (ANOVA). For all other dependent variables, values from the two squat repetitions in each condition were averaged and differences between conditions were evaluated using 3x3 (depth x load) repeated measures ANOVAs. An omnibus alpha level of $< .05$ was used to indicate statistical significance, and pairwise comparisons were conducted using a Bonferroni correction. Effect sizes (*d*, difference in means divided by pooled standard deviation) were calculated to aid in interpretation of results, with 0.2, 0.5, and 0.8 indicating small, moderate, and large effects, respectively (10). All statistical analyses were performed using Statistical Package for the Social Sciences v. 26 (IBM SPSS Statistics Inc., Chicago, IL, USA).

RESULTS

One Repetition Maximums and Squat Depths

One repetition maximums were significantly different between depths ($F_{1.14,\ 20.47} = 10.26$, $p = .003$). One repetition maximum values were greater at the above parallel depth (72.37 ± 19.33 kg) then either parallel (63.16 ± 17.42 kg, $p = .017$, $d = .50$) or below parallel (58.85 ± 18.52 kg, p = .011, d = .71) depths. 1RM values were also higher at parallel depth than below parallel depth (p = .048, d = .24).

Both depth ($F_{1.48, 26.957}$ = 225.003, p < .01, η^2 = 0.926) and load ($F_{2, 36}$ = 5.522, p < .01, η^2 = 0.235) had significant main effects on peak knee flexion. Participants squatted deeper at the below parallel (127.41 ± 10.26°) depth than in either parallel (103.53 ± 6.58°, p < .001, d = 2.78) or above parallel (87.48 ± 7.41°, p < .001, d = 4.46) depths. Squats at parallel depth were also deeper than squats at above parallel depth (p < .001, d = 2.29). Participants squatted deeper when using 50% 1RM (106.762 ± 7.45°, p = .038, d = .32) compared to 85% 1RM (104.19 ± 8.44°).

Peak Moments During the Squat

Peak M_s differed with load ($F_{1.113, 20.031}$ = 248.212, p < .001, η^2 = 0.932) but not with squat depth ($F_{2, 36}$ = 2.492, p = .0907, η^2 = 0.122). Peak M_s was greater when using 85% 1RM loads than either 50% 1RM (p < .001, d = 1.07) or unloaded (p < .001, d = 3.09, Table 1). Peak M_s was also higher with 50% 1RM than when unloaded (p < .001, d = 2.39, Table 1). Peak hip extensor moments also varied with load ($F_{1.154, 20.764}$ = 167.038, p < .001, η^2 = 0.903) but not with depth ($F_{2, 36}$ = 2.162, p = .130, η^2 = 0.107). Peak hip extensor moments were higher when using 85% 1RM loads than when using either 50% 1RM (p < .001, d = 1.06) or when unloaded (p < .001, d = 3.01). Peak hip extensor moments were also higher when using 50% 1RM than when unloaded (p < .001, d = 2.23, Table 1).

Peak knee extensor moments demonstrated a significant depth by load interaction ($F_{2.04, 36.427}$ = 5.687, p = .007, η^2 = 0.240). Within each depth, as load increased, so too did peak knee extensor moments (Table 1**)**. Within the unloaded condition, there were no differences in peak knee extensor moments between the above parallel and parallel depths (p = 1.000, d = .04), the parallel and below parallel depths (p = 0.162, d = .17), or the above parallel and below parallel depths (p = 0.138, d = .21). Within the 50% 1RM condition, peak knee extensor moments were higher in the below parallel depth than in either the parallel (p = .007, d = .47) or above parallel depths (p = .031, d = .44). Finally, in the 85% 1RM condition, peak knee extensor moments were higher in the below parallel depth than the parallel depth (p = .009, d = .72), but not different between above and below parallel depths (p = .122, d = .43, Table 1).

At the ankle, peak plantar flexor moments varied with load ($F_{1.203, 21.661}$ = 197.819, p < .001, η^2 = 0.917) but not depth ($F_{2, 36}$ = 1.138, p = .332, η^2 = 0.059). Peak ankle plantar flexor moments were higher when using 85% 1RM than when using either 50% 1RM (p < .001, d = 1.25) or when unloaded (p < .001, d = 3.10). Peak plantar flexor moments were also higher when using 50% 1RM then when unloaded (p < .001, d = 2.47, Table 1).

[Insert Table 1 about here]

Joint Contributions to Peak Support Moment

The percent each joint contributed to the support moment during the squat is shown in Figure 1 and Figure 2. At peak M_s, the percent contributed from the hip joint demonstrated a depth by load interaction ($F_{2.418, 43.529} = 3.024$, $p = .05$, $\eta^2 = 0.144$, Figure 1). At the above parallel depth, the hip contributed less to peak M_s when squatting unloaded than when using 50% 1RM ($p < .001$, $d = 1.14$) or 85% 1RM ($p < .001$, $d = 1.47$, Figure 1). When squatting to parallel depths, the contribution of the hip to peak M_s was lower when unloaded than when using 50% 1RM ($p < .001$, $d = 1.09$) or 85% 1RM ($p < .001$, $d = 1.71$, Figure 1). Hip contributions to peak M_s at parallel were also lower when using 50% 1RM than when using 85% 1RM ($p = .003$, $d = .59$, Figure 1). At below parallel depths, results were similar to above parallel, with hip contributions to peak M_s being lower when unloaded than when using 50% 1RM ($p < .001$, $d = 1.03$) or 85% 1RM ($p < .001$, $d = 1.19$), but not different between 50% 1RM or 85% 1RM ($p = .098$, $d = .30$, Figure 1).

When unloaded, there were no differences in hip contributions to peak M_s when squatting to different depths (Figure 2). When using 50% 1RM hip contribution to peak M_s was not different between above parallel and parallel ($p = 1.00$, $d = .06$) or above parallel and below parallel (p=.229, $d = .34$); however, hip contribution was less at below parallel compared to parallel (p=.004, $d = .43$, Figure 2). Similarly, when using 85% 1RM hip contribution to peak M_s was not different between above parallel and parallel (p

= .625, *d* = .16) or above parallel and below parallel (*p*=.294, *d* = .40); however, hip contribution was less at below parallel compared to parallel (*p*=.029, *d* = .61, Figure 2).

Percent knee contribution to M_s had a depth x load interaction ($F_{2.551,\ 45.913}$ = 3.682, $p < .024$, $\eta^2 = 0.170$). At each depth as load increased knee contribution decreased ($p < .001$), however; there was no difference in knee contribution when squatting below parallel when increasing load from 50% to 85% 1RM (*p*=.079, *d* = .45, Figure 1). When squatting without load, knee contribution was greater at above parallel compared to parallel (*p* =.017, *d* = .52, Figure 2). When squatting with 85% 1RM knee contribution was greater at below parallel than parallel (*p* =.033, *d* = .74, Figure 2).

Percent ankle contribution to M_s also had a main effect of load ($F_{1.435,\ 25.836}$= 26.143, $p <.01$, $\eta^2 = 0.592$). Percent ankle contribution increased from unloaded to 50%1RM ($p < .001$, *d* = .89) and unloaded to 85% 1RM ($p < .001$, *d* = 1.27) but did not differ between 50% and 85% 1RM (*p* = .066, *d* = .38, Figure 1 & 2).

[Insert Figure 1 about here]

[Insert Figure 2 about here]

DISCUSSION

This study investigated the effects of load and depth on peak lower extremity moments and contribution of the ankle, knee and hip to M_s. In partial support of our hypothesis our results suggest that lower extremity peak moments are more effected by increasing external loads than by increasing squat depth. Contrary to our hypothesis, increasing depth did not significantly change the contribution of the ankle. In partial support of our hypothesis the hip was influenced by depth and load; however, the below parallel squat resulted in the lowest hip contributions and squatting with greater than 50% 1RM tended not to further increase hip contributions. Contrary to our hypothesis, the contribution of the knee decreased from above parallel to parallel; however, in support of our hypothesis, increased from parallel to below parallel depths. Since this occurred in tandem with a decrease in hip contributions from parallel to below parallel depths, these results suggest that at deep depths when using heavy loads, the knee increases its contribution to support while the hip decreases contribution. Furthermore, contribution of the ankle and the hip increase with loading while the knee decreases contribution, indicating that a peak joint moment can increase while its contribution to support can decrease.

The results of the current study are in agreement with previous research indicating that increasing load increases peak lower extremity moments during the barbell back squat (12,20,47,50) and that peak knee extensor moments are greatest at deep depths (6,12,20,53). Our findings are also in agreement with previous studies evaluating contribution to M_s with increased load as, similar to Flanagan et al., we observed that the contribution of the knee decreased with increasing load and the contribution of the hip and ankle increased with increasing load (47). However, our results differ from previous literature in that we did not find peak hip extensor moments to be influenced by depth (6,53). Our results may differ from previous literature because we used depth specific 1RMs while these studies used the 1RM from the deepest depth to establish test loads. Maximum squat strength is joint angle specific (12,20). If deep squat 1RMs are used to dictate shallower depth loads it is unlikely to represent the appropriate test load. Therefore, it is possible that previous literature found hip moments to increase with depth because the load prescribed was based on the deep squat 1RM and was not enough to challenge the hip musculature at shallower depths.

Direct comparison to studies which have used RME to assess how muscles contribute to squat performance is difficult, as the different approaches lead to different conclusions. Byranton et al reported that hip contributions to the squat increased with both squat depth and load while knee contributions were more effected by depth than load (6). They concluded that heavy deep squats should be performed to train the hip extensors while light deep squats should be performed to train the knee extensors (6). In contrast, the current study observed that peak hip extensor moments were not affected by depths and hip contributions to squat performance were more affected by load than depth suggesting that hip extensors can be trained with heavy loads at relatively shallow depths. However, the peak knee extensor moments were the largest at deep depths with load while the knee contribution to support decreased with higher loads but increased with depth. Thus, in partial agreement with Bryanton et al., this suggests that knee extensors can be trained at deep depths.

Electromyography is frequently used to measure muscle activity when squatting. Muscle activity is related to force production, but it is not a perfect relationship. Several studies using EMG amplitude to assess muscle function during squats have reported that EMG amplitude increases with increasing squat load (5,9,39,41). However, there is disagreement in the literature regarding how squat depth influences muscle activity. Contreras et al. reported no differences in hip muscle activity between parallel and deep squats (11). In contrast, Silva et al. observed that above parallel squats resulted in greater hip muscle activity than deep squats and Caterisano et al. observed lower hip muscle activity during partial and parallel squats compared to deep squats (4, 13). There is less discrepancy at the knee as Contreras et al (11), Silva et al (32), and Caterisano et al (8) all reported no difference in quadriceps muscle activity between partial, parallel, and full depth squats. Our findings are difficult to compare to EMG findings because as mentioned above, muscle activity is related to force production and torque, but it is not the same.

The current study conceptually supports some of the previous EMG findings as we observed that peak M_s and peak hip, knee and ankle extensor moments all increased with load and that peak hip extensor moments were not different with increasing squat depth. However, direct comparison between results from peak moments and EMG activity is not possible as the peak moments represent the peak net moment about the joint and could be influenced by co-contraction. No changes in quadriceps muscle activity with increasing squat depth, as has been reported (4,6,13), if occurring in combination with decreased hamstring muscle activity, could result in higher net knee extensor moments. This might explain the increased peak knee extensor moments with deep heavy squats, despite no differences in EMG.

Furthermore, previous studies are not consistent with the methodology used to established squat loads and the level of training experience among participants. Previous studies have used a 1RM from the deepest depth to prescribe squat loads across all depths (6,53), while others have used depth specific 1RMs to prescribe squat loads (12,20). These differences in methodologies are likely to yield difference results because squat strength in depth specific (12). For example, if only a below parallel squat 1RM is performed and prescribed to shallower depths, it is likely to underestimate the squat loads at those shallower depths, potentially influencing joint moments, lower extremity kinematics, and lower extremity EMG (9). The level of squat experience is also likely to influence the results. Previous literature indicates that co-activation of antagonist muscles is higher in novice lifters than those that are more experienced (1). This could likely influence the joint moments and EMG findings between studies.

There are several limitations to our study which need to be considered when interpreting the results. First, participants squatted slightly deeper at the same depth when using lighter loads, although the differences in depths were only a few degrees. Next, bar position was not controlled. Previous research has suggested that there are kinematic differences between the high bar and low bar back squat techniques (23,24). While most participants used a high bar position, we cannot rule out that differences in bar position influenced the results. Similarly, while each participant's stance width was the same for both testing days, it was not was controlled across participants. Gluteus maximus activity has been shown to increase with increasing stance width (39,41), which could influence the hip moment values. However, these two studies used extreme stance widths not representative of normal squatting technique and that were not used in the current study. Finally, shoe type was not controlled. While many of the participants wore tennis sneakers, several did wear squatting shoes. These shoes reduce forward trunk lean and increase plantar flexion compared to wearing running shoes, and potentially contribute to greater knee extensor activity (44). Squat shoes have also been suggested to reduce ankle dorsiflexion, increase knee flexion, decrease trunk lean and increase knee moments in an unloaded condition compared to an athletic shoe (35).

In conclusion, this study investigated the effects of load and squat depth on peak moments of the lower extremity joints and their contribution to the total support moment during the squat. Lower extremity joint moments and contributions to the support moment were more effected by load than depth. Peak moments at the hip, knee, and ankle increase with load, but not necessarily with depth. However, the contribution to support moment of the knee and hip change with depth and load while the ankle is only influenced by load. The findings from this study indicate that although the peak extensor moment about a joint may increase, the contribution to overall squat performance as determined by the support moment can increase, decrease or stay the same.

PRACTICAL APPLICATION

These findings are important for using the squat exercise in practice because load and depth will have different influences on the peak moments and percent contribution of the hip extensors, knee extensors, and ankle plantar flexors. Specifically, to maximize the contribution and peak moments required of the hip extensors, increasing load is encouraged. Our findings also suggest that to maximize the peak moments required of the knee extensors, loaded deep squats should be performed. At times, it may also be necessary to consider rehabilitation of a lower extremity injury when creating a training program. Our findings suggest that peak moments increase with increasing loads and that knee moments are greatest when squatting to deep depths suggesting appropriate load and depth dosages should be taken into consideration based on the status and goals of the individual.

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FIGURE LEGEND

Figure 1**.** Joint contributions (percent peak support moment) for each of the three squat depths when squatting with different percentages 1RM (mean ± SD).

Figure 2. Joint contributions (percent peak support moment) for each of the three squat loads when squatting to different depths (mean ± SD).

Table 1. Mean and standard deviations for peak total support, hip, knee, and ankle moments at the different depth and load combinations.

		Above Parallel	Parallel	Below Parallel	Load Mean
Peak total support moment (Nm/kg)	Unloaded	113.56 ± 24.74	125.08 ± 23.12	124.71 ± 23.49	121.11 ± 6.54*
	50 % 1RM	190.07 ± 32.07	192.95 ± 38.01	196.08 ± 35.76	193.04 ± 3.01*
	85% 1RM	235.33 ± 45.51	236.25 ± 46.84	242.81 ± 50.64	238.13 ± 4.08*
	Depth mean	179.65 ± 61.55	184.76 ± 56.04	187.87 ± 59.48	
Peak hip moment (Nm/kg)	Unloaded	41.75 ± 14.08	49.18 ± 11.55	48.95 ± 10.26	46.63 ± 4.22*
	50 % 1RM	87.07 ± 23.06	89.96 ± 25.05	88.50 ± 22.84	88.51 ± 1.44*
	85% 1RM	116.24 ± 31.09	119.79 ± 31.10	117.45 ± 31.70	117.83 ± 1.80*
	Depth mean	81.69 ± 37.54	86.31 ± 35.45	84.96 ± 34.39	
Peak knee moment (Nm/kg)	Unloaded	59.42 ± 14.32	59.99 ± 13.71	62.38 ± 14.38	60.60 ± 1.57
	50 % 1RM	71.67 ± 15.41	71.03 ± 15.67	79.06 ± 18.21#	73.92 ± 4.46
	85% 1RM	80.93 ± 18.37	75.80 ± 16.51	89.33 ± 20.78#	82.02 ± 6.83
	Depth mean	70.67 ± 10.78	68.93 ± 8.11	76.83 ± 13.59	
Peak ankle moment (Nm/kg)	Unloaded	18.88 ± 5.54	22.34 ± 6.65	21.77 ± 6.81	21.00 ± 1.86*
	50 % 1RM	37.25 ± 7.63	39.61 ± 9.14	41.23 ± 8.47	39.36 ± 2.00*
	85% 1RM	53.92 ± 13.35	52.87 ± 13.69	52.80 ± 12.71	53.20 ± 0.63*
	Depth mean	36.68 ± 17.53	38.28 ± 15.31	38.60 ± 15.68	

Note: * indicates main effect of load while # indicated depth by load interaction.

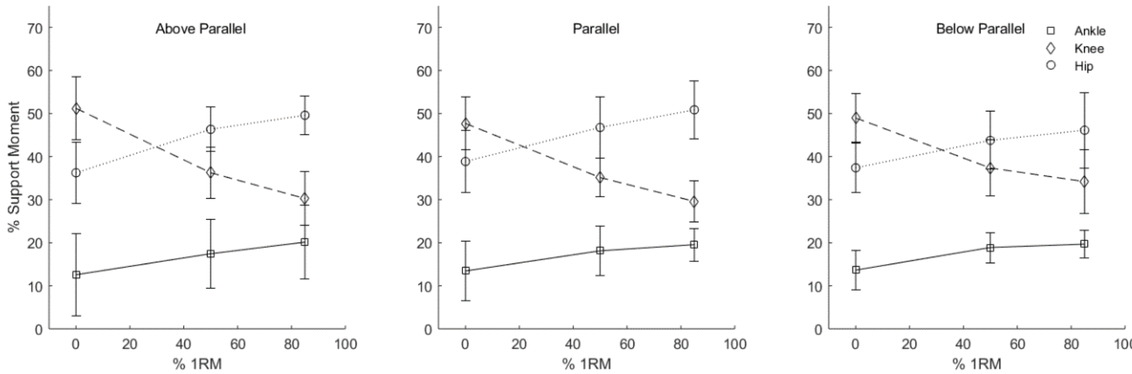

Figure 1; TIF file

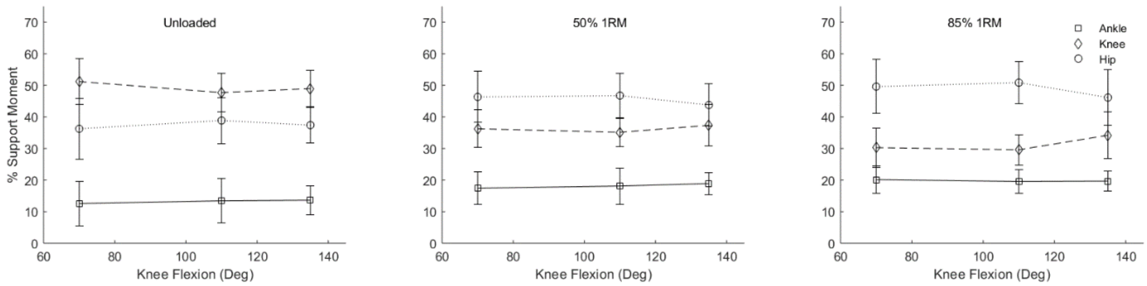

Figure 2; TIF file

CHAPTER 3

AN INDUCED ACCELERATION ANALYSIS OF THE BARBELL BACK SQUAT AT DIFFERENT DEPTHS IN TRAINED FEMALES

Contribution of Authors and Co-Authors

Manuscript(s) in Chapter(s) 2

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An Induced Acceleration Analysis of the Barbell Back Squat at Different Depths in Trained Females

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Running Title: Induced Acceleration Analysis Between Squat Depths

An Induced Acceleration Analysis of the Barbell Back Squat at Different Depths in Trained Females

Abstract:

The squat is used to enhance performance and rehabilitate the lower body. However, muscle forces and how muscles accelerate the center on mass (CoM) is not well understood. The purpose was to calculate peak muscle forces and peak induced acceleration (pIA) of the CoM when squatting to above parallel, parallel, and below parallel using 85% 1RM. Thirteen females performed squats in a randomized fashion. Musculoskeletal modelling was used to obtain peak muscle forces and pIA. The gluteus maximus and adductors increased peak force at parallel while the hamstrings and rectus femoris increased to below parallel. At deep depths, the vasti decreased peak force and force at the transition between descent and ascent while the hamstrings and rectus femoris increased peak force. The pIA and the acceleration at transition induced by the vasti decreased with depth while the pIA of the hamstrings and rectus femoris increased. Due to dynamical coupling a muscle can accelerate a joint opposite of its anatomical classification or a joint that it does not span. It is possible that at transition the hamstrings accelerated the hip and knee into extension while the rectus femoris also accelerated the knee and hip into extension while the soleus accelerated the ankle and knee into extension. Results indicate that coordination of the lower extremity is depth specific.

Key Words: musculoskeletal modelling, biomechanics, motor control, coordination

INTRODUCTION

The barbell squat is a commonly used strength and conditioning exercise often incorporated into both athletic development and clinical rehabilitation exercise programs (9,45). In athletic development settings, the squat is used broadly to develop hypertrophy, power, and strength of muscles of the lower body (9,45). In clinical programs, the squat is used for rehabilitation to assist in the progression back to full lower body function following common injury. Importantly, neuromuscular adaptations are specific to the task at hand. This has consequences related to user outcomes because modifying squat parameters such as depth and load can influence coordination, training adaptations and lower extremity joint loading (9,45).

Several previous studies examine how joints of the lower limb contribute to performance of the squat exercise and how varying both load and depth of squat may alter how the joints and muscles of the lower limb contribute to performance. Generally the methods used in these studies fall into one of 4 categories: evaluation of net joint moments during squatting (NJM), evaluation of relative muscular effort during squatting (RME), evaluation of muscle activity during squatting using electromyography (EMG), and evaluation of the support moment through the relative contribution by lower limb joints (M_s) (6,7,9,12,20,21,45,47,52,53). Generally as squat external load increases NJMs, RME, muscle activity and M_s all increase (6,7,12,20,38,47,50), however it is less clear as to the effects of varied depth on lower extremity muscle activity and joint loading (6,8,11,12,21,32,52,53).

Examinations of NJM and RME suggest that the contribution of the knee extensors increase with depth (6,7,12,20,52,53), although knee extensor muscle activity peaks around parallel depth (9,45,53). Outcomes from studies employing RME suggest that the hip extensors increase with increasing depth (6,7), however, hip NJM have not been found to increase past parallel depth (53). Overall studies that included evaluation of muscle activity data lack consensus in that some studies find hip extensor activity, especially the gluteus maximus, to be greatest at deep depths (8) while others have found hip extensor activity to be greatest at shallow depths (32) and others have found that the hip extensor activity peaks at parallel depths and plateaus thereafter (11). When considering the contribution to M_s with increasing loads at a fixed depth, Flanagan et al found that the ankle contribution increased with loading, the knee contribution decreased when increasing load, and the hip contribution increased with loading (47), and overall, the ankle contribution was less than the knee, which was less than the hip in all except the lightest condition (19,47). The diverse nature in both methodical approach of the studies mentioned above make it difficult to conclude how variations in joint moments, muscular effort or muscle activity related to varied squat load and depth,

The lack of consensus may be in part explained by the limitations inherent to each of the analytical methods described. For example, NJM analysis only considers the net of the flexor and extensor moments and resolves the problem to a single moment expressed at each joint, in doing so it does not consider the instantaneous coordination of the lower extremity muscles throughout the movement. Relative muscular effort is a relative contribution of the torque produced about a joint during a task compared to the torque produced in a maximum voluntary contraction. This does not consider the coordination of each joint to the lift but rather each joint to its own maximum torque producing capabilities. Electromyography measures if a muscle is "on" and how much it is "on", but it does not measure how much force is produced. Lastly, M_s does give insight into the coordination of the hip, knee and ankle to support but it does not calculate muscle force but rather the NJM and the contribution of that NJM to the M_s. Additionally, none of these methods consider that muscles can produce forces that accelerate other segments of the body, even if the muscle does not cross that segment, joints, or the whole-body CoM through intersegmental dynamic coupling. While each of these methods can detail if there is a general shift in the pattern of force production at each joint during the movement, they are not sensitive enough to describe the force produced by a given muscle nor its overall contribution to the movement.

In vivo, it is difficult to measure individual muscle forces and how those forces contribute to a given movement. Due to this difficulty, a musculoskeletal model can be used to generate a plausible solution. An advantage to induced acceleration analysis is the determination of how each muscle force contributes to the acceleration of the whole-body center of mass (CoM) during movement regardless of its anatomical location or joint that it spans. Previously, studies have used IAA to investigate the role of muscle during running (26,27), walking (33,37), stair climbing (36), balance recovery (25) and impaired gait (48); however, there are no studies that use IAA to measure the muscle forces during the back squat. To address the aforementioned limitations, we employed an induced acceleration analysis. Therefore, the purpose of this study is to determine differences in the magnitude of individual muscle forces and the contribution of muscle forces to the acceleration of the whole-body center of mass.

METHODS

Experimental Approach to the Problem

A within subject experimental design was used to investigate the effects of squatting to different depths with 85% of depth specific one repetition maximum (1RM). Participants were evaluated on two days: on day one, depth specific 1RM were obtained for above parallel, parallel and below parallel depths. Depths were defined as follows: above parallel was measured with a goniometer at 90 degrees of knee flexion, parallel was reached when the top of the thighs were parallel to the floor, and below parallel was reached when the hamstrings contacted the gastrocnemius muscles. Within 48hrs to 1 week later, participants returned to the laboratory for the 2nd session. During the 2nd session motion capture data was collected. In this session, 3 squat repetitions were randomly performed at the 3 different depths for a total of 3 trials. Whole body kinematics were recorded using a 10-camera motion capture system and ground reaction force (GRF) data was recorded using 2 1000 Hz AMTI force plates (Advanced Mechanical Technology, Inc., Watertown, MA, USA).

Subjects:

A total of 13 healthy female participants (age: 24 ± 3.37 years; height: 161.46 ± 8.43 cm; body mass: 62.65 ± 10.19 kg) had an average of 5.77 ± 3.59 years of squatting experience during the time of the study. The inclusion criteria required participants be between 18-45-year-old and be female with a minimum of 1 year of squat training experience. Participants were excluded if they had a lower extremity injury within the preceding 3 months of the study. Visual assessment of the squat was also used in the inclusion criteria. If subjects failed to meet the requirements of a full squat, due to technical breakdown, they were excluded from the study. All participants met physical requirements. Written informed consent was granted by each subject and approved by the Institutional Review Board (IRB).

Procedures:

Two separate sessions were conducted. In the first session, subjects established 1RMs and depths which were recorded for later use in the actual trial. 1RM were established for each depth. A safety spotter arm was used to help establish depth and cue the ascent of the squat. A non-standardized warm up, typical of each participant's usual routine, was implemented prior to squatting. During the 1RM testing, 5 minutes of rest was allowed between squat attempts. Motion capture collection occurred during the second session. Motion capture was collected 48 hours to 1-week post 1RM testing. In this session, 3 squat repetitions were randomly performed at the 3 different depths for a total of 3 trials. A modified 65 marker Helen Hayes marker set was used. 3-D kinematic data was recorded using 10 cameras sampling at 200 hz (Motion Analysis, Rohnert Park, CA, USA). GRF and Moments were calculated using AMTI force plates each sampling at 1000 hz (Advanced Mechanical Technology, Inc., Watertown, MA, USA). A 64 channel A/D board used to synchronize data collection.

Data Analysis:

Kinetic and kinematics were filtered using fourth order zero-lag low pass Butterworth filters with cutoff frequencies of 6 Hz and 20 Hz, respectively. Data was cleaned in Cortex (Motion Analysis, Rohnert Park, CA, USA) and exported as a C3D file. All data analysis were performed using OpenSim (Version 4.0)(14) along with custom MATLAB scripts (The MathWorks Inc., USA) . A scalable model consisting of 17 bodies, 17 joints, 94 muscle actuators and 36 degrees-of-freedom (27), was used as the initial generic model for analysis. A wrap object scaled uniformly with the pelvis was embedded in the generic model to maintain a realistic erector spinae muscles moment arms during trunk flexion (25). Model scaling and inverse kinematic analysis were performed by fitting the anatomical model to measured 3D marker positions with a high weighting on virtual markers which defined the joint center of the hip, knee and ankle. The mass of the barbell was added to the model following scaling as a lumped mass to the torso. Joint centers were estimated from experimental marker trajectories: the regression equations of Harrington et al. (2007) were used for the hip joint (as suggested by Kainz et al. (2015)), while the knee and ankle joint centers were identified as the midpoints of the femoral condyles and the medial and lateral malleoli respectively

(28,34). Residual Reduction Analysis (RRA) was subsequently performed to improve the dynamic consistency between measured ground reaction forces and the mass-acceleration product of the model (14). The contribution to total force by reserve actuators was small. Mean peak reserve actuator forces for all participants across all degrees of freedom was within the tolerances suggested by Hicks et al. (31). Passive muscle forces were also checked for each simulation and found to be negligible (i.e., muscles tended to operate on the ascending limb and plateau region of the force– length relationship). Muscle forces were computed using the static optimization tool in OpenSim by minimizing the sum of squared muscle activations within the force–length–velocity constraints of each muscle. Induced acceleration analysis was subsequently performed to determine the contribution of each muscle force to the acceleration of the body COM (3,42,54). For reporting purposes muscle actuators were grouped as follows: Erector spinae, gluteus maximus, iliopsoas, rectus femoris, glute medius/minimus, vasti (vastus lateralis, intermedius and medius), adductors, hamstrings (biceps femoris, semi-membranosus and semi- tendonosus), abdominals, gastrocnemius (medial and lateral gastrocnemius), soleus and tibialis anterior.

Statistical Analysis:

Variables included in the statistical analysis included individual peak muscle forces produced at each depth and individual peak muscle induced accelerations about the COM in the vertical direction at each depth. The squat phase was normalized (0 to 100%) starting at onset of knee flexion and ending at return to starting position and further separated into descent and ascent. The first 50% of trial time was descent and the last 50% of trial time was ascent. Participant peak muscle forces and peak induced accelerations at each depth were analyzed using a one-way repeated measures analysis of variance (ANOVA) with an alpha level of $\alpha < .05$ being used to indicate statistical significance. In the event of significant omnibus tests, pairwise comparisons were conducted using a Bonferroni correction. Effect sizes (*d*, difference in means divided by pooled standard deviation) were calculated to aid in interpretation of results, with 0.2, 0.5, and 0.8 indicating small, moderate, and large effects, respectively (10). All statistical analyses were performed using Statistical Package for the Social Sciences (SPSS, v. 26, IBM Corp., Armonk, NY).

RESULTS

1RM Between Depths

Participants lifted 94.01 ± 20.11 kg when squatting above parallel, 76.92 ± 20.79 kg when squatting to parallel and 70.98 ± 20.65 kg when squatting to below parallel.

Peak Joint Angles Between Depths

At above parallel, participants squatted to 99 ± 4.3°of knee flexion, 77.5 ± 10.9° of hip flexion and 28.2 ± 5.4° of ankle flexion. At parallel, participants squatted to 121 ± 7.7° of knee flexion, 92.1 ± 10.4° of hip flexion and 33.7 ± 4.1° of ankle flexion. At below parallel, participants squatted to 142.7 ± 5.7° of knee flexion, 99.4 ± 10° of hip flexion and 37.3 ± 4.7° of ankle flexion.

Peak Joint Moments Between Depths

At above parallel, the peak knee moment was 114.1 ± 27.4 Nm, the peak hip moment was 140.3 ± 28.43 Nm and the peak ankle moment was 56.9 ± 14.8 Nm. At parallel, the peak knee moment was 110.5 ± 25.3 Nm, the peak hip moment was 138.8 ± 24.7 Nm and the peak ankle moment was 60.7 ± 14.2 Nm. At below parallel, the peak knee moment was 136.3 ± 54.6 Nm, the peak hip moment was 133.8 ± 23.6 Nm and the peak ankle moment was 60.9 ± 22.2 Nm.

Peak Muscle Forces During Squatting

Qualitatively, the largest forces for all conditions were, in descending order, produced by the vasti, erector spinae, hamstrings, glute max, soleus, rectus femoris, adductors, gastrocnemius and glute medius/minimus (Table 1).

Peak Muscle Forces During Squat Descent

Differences were observed in the hamstrings during descent ($F_{1.239, 14.871} = 43.614$, p = .000, η^2 =.784) with the hamstrings producing greater peak force when squatting to parallel compared to when squatting above parallel (p <. 001, d = 1.60) and when squatting below parallel compared to parallel (p = .000, d = 1.72) (Table 1). The hamstrings also produced greater peak muscle force when squatting below parallel compared to above parallel (p = .000, d = 2.75) (Table 1). The rectus femoris produced different peak forces between depths during descent ($F_{1.204,\ 14.443}$ = 10.157, p = .000, η^2 =.458). The rectus femoris produced greater forces when squatting below parallel compared to parallel (p = .018, d = .92) (Table 1). The adductors produced different peak forces between depths during descent ($F_{1.187,\ 14.240}$ = 12.247, p = .002, η^2 =.505). The adductors produced greater peak force when squatting to parallel compared to above parallel (p = .022, d = 1.01) and when squatting to below parallel compared to above parallel (p = .008, d = 1.23) (Table 1).

Peak Muscle Forces During Squat Ascent

The vasti produced different forces between depths during ascent ($F_{1.278,\ 15.335}$ = 6.656, p = .016, η^2 =.357) (Table 1). The vasti produced less peak force when squatting below parallel compared to parallel (p = .001, d = .57). Differences were observed in the hamstrings during ascent ($F_{2,\ 24}$ = 63.7, p =.000, η^2 =.841) with the hamstrings producing greater peak force when squatting to parallel compared to when squatting above parallel (p < .001, d = 1.67). The hamstrings produced greater peak muscle force when squatting below parallel compared to parallel (p = .000, d = 1.68) and greater peak muscle force when squatting below parallel compared to above parallel (p = .000, d = 3.17) (Table 1). The rectus femoris produced different peak forces between parallel and below parallel depths during ascent ($F_{2,\ 24}$ = 10.588, p = .001, η^2 =.468). The rectus femoris produced greater forces when squatting below parallel compared to parallel (p = .027, d = .98) and greater forces when squatting below parallel compared to above parallel (p = .010, d = 1.47) (Table 1). The adductors produced different peak forces between depths during ascent ($F_{1.373, 16.471}$ = 11.173, p = .000, η^2 =.482). The adductors produced greater peak force when squatting to parallel compared to above parallel (p = .018, d = 1.10) and produced greater peak forces when squatting to below parallel compared to above parallel (p = .009, d = 1.26) (Table 1). The gluteus maximus produced different peak forces between depths during ascent ($F_{2, 24}$ = 7.069, p = .004, η^2 =.371). The gluteus maximus produced greater peak forces when squatting to parallel compared to above parallel (p =.009, d = .43) and greater peak force when squatting below parallel compared to above parallel during ascent (p = .035, d = .53) (Table 1).

[Insert Table 1 about here]

[Insert Table 2 about here]

[Insert Figure 1 about here]

Peak Muscle IAA

Qualitatively, the soleus, vasti and gastronemius produced the greatest peak induced accelerations of the CoM in the vertical direction (pIA) of all muscles (Table 3&4). The hamstrings, erector spinae, rectus femoris, gluteus maximus, glute medius/minimus and adductors produced far lower pIA compared to the three main contributors listed above (Table 3&4).

Peak IAA During Squat Descent

No significant differences in muscle induced acceleration were observed when comparing between squatting from above parallel to parallel. The hamstrings produced different pIA between depths during descent ($F_{1.050, 12.598} = 53.681$, $p = .000$, $\eta^2 = .871$) (Table 3). The hamstrings produced a greater pIA when squatting below parallel compared to parallel ($p = .000$, $d = 2.48$) and greater pIA when squatting below parallel compared to above parallel ($p = .000$, $d = 2.61$) (Table 3).

Peak IAA During Squat Ascent

No significant differences in muscle induced acceleration were observed when comparing between squatting from above parallel to parallel. The hamstrings produced different pIA between depths during ascent ($F_{2, 24} = 26.234$, $p = .000$, $\eta^2 = .686$) (Table 4). The hamstrings produced a greater pIA when squatting below parallel compared to parallel ($p = .001$, $d = 1.79$) and greater pIA when squatting below parallel compared to above parallel ($p = .000$, $d = 2.19$) (Table 4). The rectus femoris produced different pIA between depths during ascent ($F_{1.306, 15.667} = 9.650$, $p = .004$, $\eta^2 = .446$) (Table 4). The rectus femoris produced a greater pIA when squatting below parallel compared to parallel ($p = .025$, d =

1.05) and greater pIA when squatting below parallel compared to above parallel (p = .019, d = 1.40).

[Insert Table 3 about here]

[Insert Table 4 about here]

[Insert Figure 2 about here]

DISCUSSION

The purpose of this study was to examine the individual muscle forces generated throughout the squat at differing depths and how these forces contributed to the acceleration at the CoM in the vertical direction. Qualitatively, the muscles that produced the greatest peak forces across all depths were the vasti, erector spinae, hamstrings, gluteus maximus and soleus. When comparing above parallel and parallel squats, we observed differences in peak muscle forces produced by the hamstrings, gluteus maximus, and adductors. However, when squatting at depths past parallel only the hamstrings continued to increase peak force. When comparing parallel to below parallel squats the vasti demonstrated decreased peak muscle forces while the hamstrings and rectus femoris increased peak muscle forces. Furthermore, when examining the time course of forces, there was a decrease in force production of the vasti when squatting below parallel. At the transition between descent and ascent the vasti decrease force while the hamstrings and rectus femoris increase force production. This finding demonstrates the coordination of the vasti, hamstrings and rectus femoris to accomplish the lift at deep depths.

Overall, the muscles that produced the greatest peak induced acceleration about the CoM in the vertical direction were the soleus, vasti and gastrocnemius. Importantly though while the gastrocnemii have high peak accelerations they are primarily produced at the start and end of the movement when the knee is near full extension and the ankle is in mid dorsi/plantar flexion, and they rapidly decrease their peak induced acceleration as the knee and ankle begin to flex during descent. All the muscles in our model produced positive peak accelerations about the CoM except the tibialis anterior at above parallel and below parallel depths. There were no differences in peak induced acceleration of any muscle when squatting from above parallel to parallel; however, when comparing parallel and below parallel squats the hamstrings and rectus femoris produced greater peak induced acceleration at below parallel. Qualitatively, the vasti decreased induced acceleration about the CoM at the transition between descent and ascent at deep depths. While the vasti decrease acceleration at the transition between descent and ascent, the hamstrings and rectus femoris drastically increase their peak induced accelerations.

It is evident that as depth of the squat increases many muscles of the lower limb increase their respective force production. However, the change in each muscles force production did not uniformly translate into a different contribution to the vertical acceleration of the CoM. Most notably the acceleration induced by the vasti, hamstrings, and rectus femoris occurs in a coordinated fashion with the vasti reducing their contribution as depth increases while the hamstrings and the rectus femoris simultaneously increase their contribution. Fundamentally, an induced acceleration analysis decomposes the sum of internal muscle forces into their contribution to the external ground reaction force (GRF) (15). Using this framework, we theorize that as depth increases the moment arm of the vasti decreases reducing its contribution to the vertical GRF while the hamstrings may increase their contribution to the GRF (22,30). To enable the hamstrings to contribute a greater percentage of its force in the vertical dimension requires the simultaneous contribution of the rectus femoris. Using this framework, our results may be explained by dynamical coupling where a biarticular muscle can accelerate one of the joints that it crosses in the opposite direction of its typical torque production (54). Importantly, a muscle cannot accelerate both joints that it crosses simultaneously in directions opposite to its typical torque production. It is important to consider that muscles will simultaneously contribute to joint, segment and CoM accelerations. Therefore, muscles may produce force and the force may be represented as a joint acceleration, segment acceleration, or CoM acceleration but not necessarily all three at the same time. This is likely to explain why the general increase in muscle force that was observed did not translate to a uniform increase in muscle induced accelerations at the CoM. While, based on theoretical perspectives, we can suggest the consequences of the muscle force changes we observed a joint level induced acceleration analysis is required to further investigate this possible explanation.

We suggest that at the transition between descent and ascent at deep depths it is possible that the hamstrings are producing an extensor torque about the hip acting to accelerate the hip into extension in the same direction that it is producing torque; however, as a consequence of how it is acting at the hip, it may accelerate the knee into extension which is opposite to the direction of which it would typically apply torque about the knee. Similarly, the rectus femoris may be applying an extensor torque about the knee which is accelerating the knee into extension which is in the same direction of torque production about the knee. As a consequence of how it is acting about the knee, it may be acting to accelerate the hip into extension which is opposite to the direction of its typical torque production about the hip. Furthermore, while the uniarticular soleus can only produce a moment at the joint that it crosses, the ankle, due to dynamical coupling, it can accelerate a joint that it does not span, possibly the knee, into extension. This dynamical coupling of the hip and knee accelerating into extension may offer a further explanation as to the combined contribution hamstrings and rectus femoris to the increase in acceleration of the CoM in the vertical direction, particularly at the transition between descent and ascent. Similarly, the extensor torque produced by the soleus about the ankle may lead to an acceleration of the knee into extension which may explain how the soleus contributes to the acceleration of the CoM in the vertical direction. There are many muscles in our model that produce large forces but do not induce large accelerations about COM in the vertical direction. Some of these large force producing muscles may induce acceleration about the CoM in different directions.

There are several considerations when evaluating the results of this study. Firstly, while we used the same relative maximum, we used three different absolute loads. The participants in our study completed depth specific 1RMs and then used 85% of the depth specific 1RM at the corresponding depth. This means that the relative intensity was equated between depths, but the absolute intensity was not. The absolute barbell load was not the same between depths. This difference in barbell load may have influenced the peak induced acceleration and muscle forces at the same joint angles between squat conditions. This limitation makes it difficult to distinguish whether absolute external load or depth led to the observed differences in peak induced acceleration and muscle forces between depths. In the future an additional study is required to examine the differing muscle contributions due to both load and depth independently. Secondly, we used static optimization to determine the force produced by individual muscles to equilibrate each joint moment. Importantly static optimization solves for a specific time frame without consideration of previous or future time frames. It is possible this may lead to non-physiological solutions and minimize co-contraction. However, previous research investigating gait found no differences between static and dynamic optimization (2). Because similar solutions using both static and dynamic optimization techniques are observed during cyclical movements, such as gait, with better computational efficiency we chose to utilize static optimization. However, in the future it is of interest to evaluate different force estimation techniques such as computed muscle control (49) and/or EMG informed modeling (18).

In conclusion only some peak muscle forces increase as the depth of squat increased. However, not all changes in muscle force production result in differing contribution to the vertical acceleration of the CoM. There is a coordination of vasti, hamstrings and rectus femoris at deep depths, indicating a codependence of key muscles to maintain performance while squatting to different depths .The vasti decreases peak force production, peak induced accelerations and force at the transition between ascent and descent while the hamstrings and rectus femoris increase peak force production and peak induced acceleration to compensate for the vasti. Due to dynamical coupling, and the foot being fixed to the ground in the squat, the uniarticular soleus can accelerate the knee into extension and, at deep depths, the biarticular hamstrings may accelerate the knee into extension and the rectus femoris may accelerate the hip into extension which may lead to an increase in the acceleration of the CoM in the vertical direction.

PRACTICAL APPLICATION

It is important to consider the coordination and codependence of key lower limb muscles in the performance of the squat. We suggest that depth specificity is an important consideration for lifters who intend to squat at deep depths, that is if you compete in a sport such as weightlifting or powerlifting, you need to squat to the depth required by your sport, and lifters may potentially be limited in their ability to squat at deep depths without practicing this movement pattern. Additionally, due to the role mechanical tension plays in hypertrophy and strength, if the goal is to grow or strengthen the hamstrings, gluteus maximus, or adductors it is recommended to squat to at least parallel because, in our model, this depth maximized their force production. For lifters with deficits in performance it is possible that strengthening the vasti, hamstrings, rectus femoris and the soleus may help to further accelerate the CoM in the vertical direction. In particular, strengthening the hamstrings and rectus femoris may help in accelerating out of the bottom of the deep squat. However, it is important to consider the coordination of these muscles and learn to use the available strength of muscles within the framework of the specific movement pattern required.

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FIGURE LEGEND

Figure 1. Muscle forces throughout the squat comparing above parallel, parallel, and below parallel depths.

Figure 2. Muscle induced accelerations about the center of mass throughout the squat comparing above parallel, parallel and below parallel depths.

Table 1: Mean and standard deviation of peak muscle forces in the vertical direction during descent

Muscles	Above Parallel (N)	Parallel (N)	Below Parallel (N)
Gluteus Maximus	1283.48 ± 599.45	1338.80 ± 723.46	1342.172 ± 604.89
Gluteus Medius	703.90 ± 188.17	624.49 ± 205.91	680.01 ± 259.99
Iliopsoas	179.87 ± 147.75	331.81 ± 595.07	327.69 ± 571.10
Rectus Femoris	84.48 ± 81.33	344.39 ± 504.84	1185.89 ± 1184.18$^{\#+}$
Vasti	5152.23 ± 1035.56	5381.56 ± 977.54	5113.97 ± 1022.93
Adductors	443.51 ± 176.47	834.99 ± 521.34*	971.28 ± 578.07^{+}
Gastrocnemius	588.86 ± 382.34	460.21 ± 288.51	547.40 ± 339.70
Hamstrings	1635.18 ± 455.53	2717.29 ± 841.50*	4967.94 ± 1650.43$^{\#+}$
Abdominals	439.36 ± 277.06	384.42 ± 207.60	330.78 ± 215.92
Erector Spinae	2815.41 ± 1129.13	3143.94 ± 1019.25	2915.89 ± 1028.38
Soleus	1455.20 ± 344.41	1693.82 ± 518.17	1823.58 ± 771.53
Tibialis Anterior	89.99 ± 86.01	213.26 ± 345.06	174.86 ± 309.28

Note: *indicates greater force at parallel than above parallel, #indicates greater force at below parallel than parallel, +indicates greater force at below parallel than above parallel, ^indicates lower force at below parallel compared to parallel (at $p < .05$).

Table 2: Mean and standard deviation of peak muscle forces in the vertical direction during ascent

Muscles	Above Parallel (N)	Parallel (N)	Below Parallel (N)
Gluteus Maximus	1546.74 ± 596.74	1852.39 ± 806.13*	1905.26 ± 745.06^{+}
Gluteus Medius	822.42 ± 240.44	797.86 ± 233.06	831.68 ± 269.38
Iliopsoas	108.43 ± 4.54	108.52 ± 2.66	107.41 ± 1.84
Rectus Femoris	82.37 ± 42.74	294.83 ± 442.58	949.82 ± 831.15$^{\#+}$
Vasti	5128.74 ± 902.18	5227.09 ± 991.36	4668.87 ± 954.14$^{\wedge}$
Adductors	460.51 ± 152.23	883.21 ± 519.32*	945.94 ± 524.29^{+}
Gastrocnemius	662.22 ± 331.31	534.70 ± 355.04	492.21 ± 376.87
Hamstrings	1838.08 ± 431.22	3067.81 ± 947.14*	5040.29 ± 1362.27$^{\#+}$
Abdominals	476.23 ± 389.94	419.10 ± 255.09	348.02 ± 275.36
Erector Spinae	3026.31 ± 1000.41	3135.46 ± 1090.10	3116.44 ± 793.01
Soleus	1499.21 ± 312.60	1613.70 ± 295.54	1709.96 ± 631.48
Tibialis Anterior	142.20 ± 152.64	115.73 ± 124.67	98.85 ± 83.35

Note: *indicates greater force at parallel than above parallel, #indicates greater force at below parallel than parallel, +indicates greater force at below parallel than above parallel, ^indicates lower force at below parallel compared to parallel (at $p < .05$).

Table 3: Mean and standard deviation of COM peak induced acceleration in the vertical direction during descent

Muscles	Above Parallel (m/s^2)	Parallel (m/s^2)	Below Parallel (m/s^2)
Gluteus Maximus	0.46 ± .21	0.54 ± .27	0.55 ± .25
Gluteus Medius	0.22 ± .11	0.31 ± .23	0.29 ± .19
Iliopsoas	0.08 ± .16	0.28 ± .56	0.08 ± .12
Rectus Femoris	0.13 ± .17	0.37 ± .56	0.44 ± .31
Vasti	2.71 ± .55	2.88 ± .55	2.76 ± .51
Adductors	0.09 ± .05	0.13 ± .05	0.14 ± .09
Gastrocnemius	2.69 ± 1.70	1.98 ± 1.27	2.75 ± 1.47
Hamstrings	0.04 ± .13	0.09 ± .11	1.03 ± .52$^{\#+}$
Abdominals	0.08 ± .07	0.11 ± .08	0.13 ± .12
Erector Spinae	0.60 ± .80	0.90 ± 1.07	1.34 ± 1.74
Soleus	3.24 ± .42	3.62 ± .69	3.41 ± .55
Tibialis Anterior	-0.04 ± .01	0.18 ± .78	-0.02 ± .01

Note: #indicates greater acceleration at below parallel than parallel, +indicates statistically greater force at below parallel than above parallel (at $p < .05$).

Table 4: Mean and standard deviation of COM peak induced acceleration in the vertical direction during ascent

Muscles	Above Parallel (m/s^2)	Parallel (m/s^2)	Below Parallel (m/s^2)
Gluteus Maximus	0.60 ± .45	0.77 ± .47	0.90 ± .55
Gluteus Medius	0.19 ± .12	0.23 ± .17	0.25 ± .14
Iliopsoas	0.01 ± .02	0.05 ± .11	0.02 ± .03
Rectus Femoris	0.07 ± .07	0.13 ± .12	0.36 ± .29$^{\#+}$
Vasti	2.94 ± .80	2.97 ± .86	3.21 ± .85
Adductors	0.12 ± .11	0.16 ± .13	0.17 ± .05
Gastrocnemius	2.52 ± 1.24	2.05 ± 1.15	1.87 ± 1.14
Hamstrings	0.02 ± .14	0.11 ± .31	0.96 ± .60$^{\#+}$
Abdominals	0.10 ± .11	0.12 ± .11	0.09 ± .07
Erector Spinae	0.29 ± .41	0.66 ± .89	0.84 ± .88
Soleus	3.59 ± .52	3.65 ± .63	3.77 ± .71
Tibialis Anterior	-0.05 ± .02	0.04 ± .27	-0.03 ± .01

Note: #indicates greater acceleration at below parallel than parallel, +indicates statistically greater force at below parallel than above parallel (at $p < .05$).

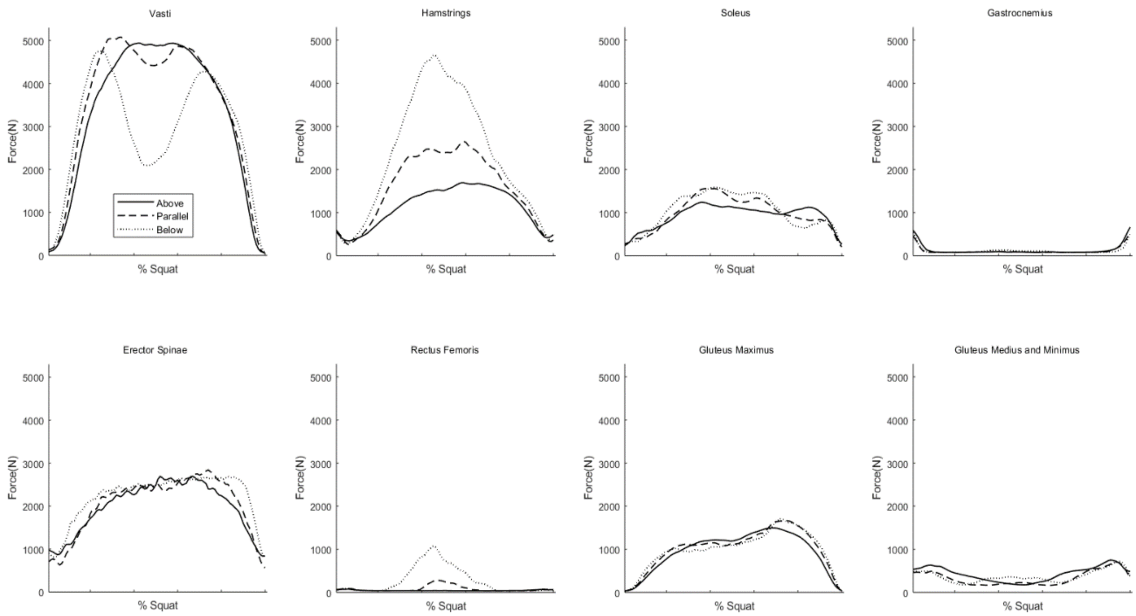

Figure 3, TIF file

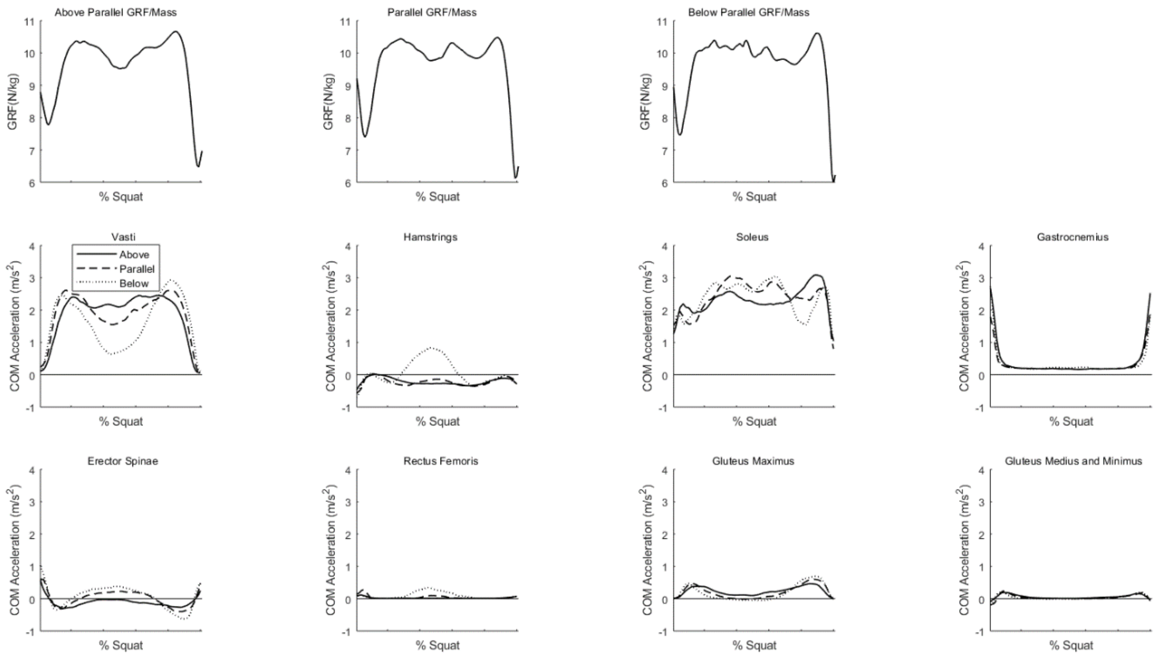

Figure 4, TIF file

CHAPTER 4

CONCLUSION

The purpose of this thesis was to build upon previous squat analysis (NJM, RME, EMG) because the influence of changing squat techniques on lower extremity muscle forces and coordination could not be made due to previous analytical limitations. For example, NJM only looks at the net moment acting about a given joint. This alone does not consider the coordination of the joints to produce movement, what each individual muscle is doing or that a muscle can accelerate joints that it does not cross. Relative muscular effort is a comparison of the maximum torque that is produced during an isometric task at a given joint angle which is compared to the torque produced at the same joint angle during a dynamic movement such as the squat. Relative muscular effort is a good measure of the effort that a joint is producing compared to its maximum but it doesn't consider what individual muscles are doing, how muscles can accelerate joints that they don't cross, and doesn't consider the coordination of the lower extremity joints to supporting the body. Electromyography is a measure of muscle activity and provides information about whether a muscle is "on" and how much it is "on". It provides neither the force a muscle is producing nor the type of contraction.

These methods are all valuable, but they are limited in the scope to which they can inform about muscle force and coordination. It is too invasive to gather individual muscle force data *in vivo*; therefore, a model must be used. A musculoskeletal model can take many inputs such as the anthropometry, kinematics, kinetics, and EMG gathered during data collection and physiology principles of muscle such as, but not limited to, the force-length-velocity relationship and together with a cost function, minimization of muscle activation squared, calculate a plausible solution of muscle forces throughout the squat. Furthermore, by calculating individual muscle forces it is possible to calculate how individual muscles accelerate joints that they do not span and the CoM. Musculoskeletal modelling therefore provides a plausible answer to a question that is impossible to calculate using previous analytical techniques.

To address the limitations of NJM and RME we conducted our first study using M_s. Support moment is the summation of the extensor moments of the hip, knee and ankle. This analytical technique describes the coordination of the lower extremity to supporting the body. The purpose was to investigate the effects of squat load and depth on lower extremity moments and contribution of the hip, knee and ankle to support.

Although M_s considers the coordination of the lower extremity to support it still does not provide a muscle level analysis. The net moments used as inputs to calculate M_s is the total moment about a joint. This does not provide answers to which muscles and how much force each muscle is producing contributing to that net joint moment. Support moment also does not consider that muscles can accelerate joints that they do not cross.

To address the limitations of NJM, RME, EMG and M_s we conducted our second study using an induced acceleration analysis. This analytical technique requires a musculoskeletal model and calculates plausible muscle forces generated throughout the squat based on experimental inputs, physiological principles, an optimization algorithm and a cost function. This analysis addresses the limitations of previous analytical techniques and provides valuable information at the muscle level and how those muscles contribute to movement. The purpose was to investigate the effects of squat depth on lower extremity muscle forces and contribution of those muscles to the acceleration of the CoM in the vertical direction.

Study 1

This study investigated the effects of load and depth on peak lower extremity moments and contribution of the ankle, knee and hip to M_s. The results suggest that, in general, lower extremity peak moments are more effected by increasing external loads than by squat depth. However, the peak knee moments were greatest at the below parallel depth when loaded.

The contribution of the hip within each depth increased with load and was lowest when squatting below parallel. The contribution of the knee within each depth decreased with load. The contribution of the ankle increased with load but was unchanged by depth. The decreased contribution of the hip occurred in tandem with the increased contribution of the knee at deep depths with load suggesting that at deep depths with load the knee increases it contribution to the squat while the hips contribution decreases.

Study 2

This study investigated how individual muscle forces generated throughout the squat at differing depths contributed to the acceleration of the CoM in the vertical direction. Qualitatively, the muscles that produced the greatest peak forces were the vasti, erector spinae, hamstrings, gluteus maximus and soleus. The hamstrings increased peak force production with increasing depth and the gluteus maximus and adductors increased peak force production to parallel. The vasti decreased peak force production and force production at transition between decent and ascent with increasing depth while the hamstrings and rectus femoris increased peak force production demonstrating a coordination strategy of the lower extremity.

Qualitatively, the muscles that produced the greatest peak induced accelerations were the soleus, vasti and gastrocnemius; however, the gastrocnemius drops off quickly during decent as the knee and ankle begin to flex and increases quickly towards lockout as the knee is near full extension and the ankle are near mid extension. Like what was observed with peak force production, peak induced acceleration decreased in the vasti at deep depths and at transition between descent and ascent while the hamstrings and rectus femoris peak accelerations increased.

To explain these findings it is possible that at the transition between descent and ascent the hamstrings produce an extensor torque about the hip which may accelerate both the hip and knee into extension while the rectus femoris may produce an extensor torque about the knee accelerating both the knee and hip into extension. Furthermore, the soleus may produce and extensor torque about the ankle which may accelerate the ankle and knee into extension.

General Conclusion

When comparing the two studies qualitatively, we observed similarity in peak moments of the lower extremity between depths; in both studies the hip and ankle joint moments did not change substantially with depth but the knee joint moment was the greatest at below parallel.

Study 1 indicated a change in the contribution of joint moments to the overall M_s, leading to the summation that there is likely a coordinative difference required to produce the movement at varying depths. However, and most importantly, the change in joint moment and a change in that joint's contribution to M_s does not allow the assignment of causality to a particular muscle in the creation of that movement. This is because joint moments are produced by a combination of many muscles acting about a joint. It is an oversimplification to associate only the muscles acting in the direction of the net joint moment as the muscles responsible. Therefore, considering joint moments alone cannot explain the muscle forces contribution to function with respect to whole body support. To further this understanding required a musculoskeletal model. Using such a model indicated that some muscles forces act in such a way as to contradict the function that would be intuitively assigned based on their anatomical arrangement. For example, in study 2 we observed an increase in the peak force production of adductors, gluteus maximus, and hamstrings muscles when squatting to parallel and a further increase in the hamstrings below parallel. The aforementioned muscles are anatomical hip extensors; however, there was little to no change in peak moments about the hip regarding depth in both studies. All these muscles aid in a hip extensor torque so it would need to be counteracted by flexor moment if the hip extensor moment was not influenced by depth. The rectus femoris also increased force production to below parallel which may have contributed to a hip flexor moment prior to transition which may explain why there was little to no change in peak hip moments with depth.

In another example, when squatting below parallel the vasti decrease force production while the knee extensor moment increases with depth. This is odd because the vasti is typically thought of as a primary knee extensor. The rectus femoris and hamstrings also increase force production at deep depths. Because biarticular muscles can act about either or both joints they cross it is possible that the increase in the knee extensor moment observed can be explained by the increase in rectus femoris force which may be directed about the knee during transition while the increase in hamstrings force in directed about the hip during transition. This would explain how it is possible for the net knee joint moment to increase even with co-contraction of an anatomical antagonist.

A further explanation regarding changes in moment without corresponding change in muscle force productions can be explained as follows: either a muscles force and/or moment arm need to increase to increase a joint moment. Therefore, it is possible that a muscle can increase its force production while its moment arm reduces thus the joint moment is unaltered. Similarly, a muscle can increase its moment arm while its force is decreasing not increasing the joint moment. It is also possible that the muscle forces and moment arms of an anatomical agonist are increasing while the forces and moment arms of an anatomical antagonist are also increasing leading to a small or non-existent net joint moment. Therefore, the total joint moment may stay the same while the muscle force, moment arms, and coordination changes. For example, a potential reason the knee extensor moment was not different between above parallel and below parallel depths whereas it was different between parallel and below parallel depths is, in part, the difference in muscle coordination. At below parallel the vasti drops in force while the rectus femoris and hamstring increases force production. At above parallel the vasti does not drop off in force and the hamstrings and rectus femoris force production is far less. The maintenance of vasti force production at above parallel without the increase in rectus femoris and hamstrings force may explain why net knee extensor moments did not differ between above and below parallel depths.

The concepts described above indicate how important it is to consider that the many muscles crossing a joint can be coordinated in an infinite number of ways to equilibrate the joint moment. When considering whole body support it is also important to consider that muscles are coordinated in a complex interaction throughout the entire musculoskeletal system and at times may function opposite of its anatomical classification by accelerating joints in an opposite direction or accelerating joints that they do not span. For example, the soleus was a large contributor to the pIA in the vertical direction about the CoM. The soleus is not typically considered an important muscle in squat performance; however, it contributed substantially. It is possible that the soleus is contributing to the GRF in such a way as to accelerate both the ankle and knee into extension which is driving the CoM upwards.

While this thesis presents two studies that address a gap in the current literature, it is not without its own limitations. Firstly, to more fully describe the performance of the squat exercise it is necessary to examine the influence of load as well as depth in a systematic fashion. Secondly, the estimation of muscle forces and coordinative structure is highly dependent on the methodology used. While static optimization produces a plausible and reliable result it would be of interest to extend the analysis to use estimations that may be more physiologically relevant such as EMG informed models. The benefit of an EMG informed estimation of muscle force is the inclusion of a biological signal in the optimization. This in turn constrains the computation of muscle force production within a smaller solution space that must include the measured muscle activity. This produces a solution that tends to have greater co-contraction about each respective joint. Thirdly, the IAA presented here only considers the contribution of muscles to vertical support. To develop a more complete understanding requires an analysis of muscle contributions to the CoM in all three dimensions as well as muscle contributions to individual lower limb joints.

Overall the outcome of the two studies indicate that 1) moments produced at the respective lower limb joints do change and contribute differently to the overall M_s during squatting and 2) a complex interaction of lower limb muscle coordination is specific to the depth of squat, particularly at depth below parallel. An important outcome when considering these two studies and the respective analytical techniques is the observation that interpreting variations of this movement by considering net joint moments alone should be taken with a “grain of salt” because joint moments, moments arms, muscle forces, muscle induced accelerations and the coordination of muscles are time-varying and not only effect each other but also the joints that they do and do not cross along with the whole body CoM. These findings lead to implications that are consequential for squat performance. It is important to consider the principle of specificity, that is, even within a class of movement such as the squat there are fundamental coordinative differences between different movement depths. Fundamentally this indicates that without squatting to the depth required by the athletes sport they may not learn the coordinative structure to perform the lift effectively or efficiently.

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