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| Quantifying knee joint kinetics during the overhead squat using different heel heights in male adults: manipulating stance and grip width |

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| A STAFFORDSHIRE UNIVERSITY UNDERGRADUATE DISSERTATION |

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| Submitted in Partial Fulfilment of the Requirements for the D*egree of Sports Therapy* |
| 28/04/2025 |

# Acknowledgements

I would like to thank my Supervisor Rob Needham for all his support and help within this study as well as all the staff and my class of the Sports Therapy Undergraduate Degree for all their help and support throughout the course.

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# Abstract

BACKGROUND: This study quantified the knee joint kinetics in the overhead squat in the FMS stance and Preferred Stance (PS) using different heel height wedges. Different squat techniques have been found to place different stresses and forces onto the patellofemoral joint however no study has quantified these stresses and forces when performing the FMS technique versus a PS using different heel heights. Alternate hypothesis one, FMS results in greater Patellofemoral joint stress (PFJS), force (PFJF) and Knee Extensor Moment (KEM) versus a preferred stance. Alternate hypothesis two, greater heel heights will result in greater PFJS, PFJF and KEM in both techniques. METHODOLOGY: Ten participants (Age-22.9±2.72 years, Height-179.65±6.83cm, Mass-79.45±10.83kg) performed an overhead squats in the FMS and PS techniques in flatfooted (0CM) and using heel wedges of 1CM, 2CM, 3CM and the FMS BOX (4.3cm). RESULTS: PS found significantly greater PFJS (1.81±0.70 MPa, *p* = 0.03) and PFJF (4.63±1.68 N/kg, *p* = 0.022) compared to FMS in the FF condition, however no significance was seen in any other heel height conditions between techniques. Both squat techniques found statically significantly increases in PFJS (PS=5.74±1.12 MPa, *p* = 0.006, FMS=8.52±1.35 MPa, *p* = < .001), PFJF (PS=14.40±2.94 N/kg, *p* = 0.009, FMS=21.18±3.60 N/kg, *p* = 0.002) and KEM (PS=0.31±0.6 N/m/kg, *p* = 0.006, FMS=0.41±0.7 N/m/kg, *p* = 0.003) from flatfooted (0 cm) to the FMS box (4.3cm) with a progressive increase in all results between each heel height.

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# 1.0 Introduction

## 1.1 Background

Movement screening is used clinically to evaluate an individual’s strengths, weaknesses, and movement limitations (Bennett et al., 2020; Chimera & Warren, 2016). These evaluations often include movement patterns such as squats and lunges, during which clinicians assess for compensations, asymmetries, or abnormalities (Chimera & Warren, 2016; Morgan et al., 2023). High-quality movement is typically defined as symmetrical, balanced, and well-coordinated (Bennett et al., 2020). Movement screening tools are widely used to identify deficits that could increase the risk of non-contact injuries and inform targeted training interventions (Chimera & Warren, 2016).

Various clinical screening tools have been developed to assess mobility, stability, coordination, and flexibility. These include the Functional Movement Screen™ (FMS), Y Balance Test (YBT), Star Excursion Balance Test (SEBT), Tuck Jump Assessment (TJA), Drop Jump Screening Test (DJST), and the Landing Error Scoring System (LESS) (Skibski et al., 2024). These tools are cost-effective, require minimal equipment, and are often supported by standardised scoring systems that make them accessible for repeated use (Bennett et al., 2020; Chimera & Warren, 2016). Despite the widespread adoption of these tools, debates remain about their predictive validity, especially regarding their use in injury prevention strategies for athletic and general populations (Bonazza et al., 2017).

The current study focuses on the FMS, developed by Cook et al. (2014), which evaluates an individual’s ability to perform seven movements: deep squat, hurdle step, inline lunge, shoulder mobility, active straight-leg raise, trunk stability push-up, and rotary stability. Each movement is scored on a scale from zero to three. A score of three indicates perfect form, two indicates compensatory movement, one represents an inability to complete the task, and zero reflects pain during the movement (Cook et al., 2014; Morgan et al., 2023). A recent update by Morgan et al. (2023) has recommended extra clearing tests prior to some exercises in which if pain is present, a score of zero is awarded. A total FMS score below 14 has been associated with increased injury risk (Bonazza et al., 2017). The FMS is widely adopted due to its low cost, ease of administration, and good inter-rater reliability, although the deep squat component has shown comparatively lower reliability (ICC = 0.78) (Morgan et al., 2023).

The deep squat is a key screening tool used to evaluate bilateral mobility and control of the hips, knees, and ankles. During this test, participants stand with feet shoulder-width apart, hold a dowel overhead, and descend into a squat while attempting to maintain an upright torso and the dowel aligned over the feet (Cook et al., 2014). A score of three is awarded if the trunk remains upright, the femur reaches or exceeds parallel to the ground, knees track over the feet, and heels remain on the floor. If this is not possible, a 4.3 cm wedge is placed underneath the heel, if then the participant meets the criteria, a score of two is awarded. If the movement cannot be performed adequately or pain is present, the score is reduced to one or zero, respectively (Cook et al., 2014).

Squatting is a foundational movement relevant to daily activities such as stair climbing, sitting, and standing. It is a compound exercise that requires coordinated activation of the hips, knees, and ankles, and involves several muscle groups including the quadriceps, hamstrings, gluteus maximus, and gastrocnemius (Gullett et al., 2009). Due to its ability to strengthen the lower limbs, the squat is commonly used in rehabilitation and sports performance settings (Straub & Powers, 2024). Squats have also been used to examine movement deficits and functional asymmetries in athletes, providing insight into neuromuscular control and dynamic stability (Myer et al., 2014).

As a result of the movement’s adaptability, numerous squat variations exist. Adjustments in foot angle (Lorenzetti et al., 2018), stance width (Sinclair et al., 2022), tibial angle (Monteiro et al., 2022), trunk posture (Escamilla et al., 2009), and depth (Bloomquist et al., 2013) can all influence muscular demand and joint loading (Straub & Powers, 2024). While squats are widely prescribed, the biomechanics of different variations remain under investigation. Clinicians and coaches use a range of styles, including back squats, front squats, single-leg squats, goblet squats, and overhead squats, with each variation altering load distribution and technical demands (Salles & Pascoal, 2025). For example, the front squat generally produces a more upright trunk angle, while the back squat allows for greater hip flexion and posterior chain activation (Swinton et al., 2012).

The overhead squat is biomechanically demanding as it requires lower body strength, core stability, upper body mobility, and postural control (Morgan et al., 2023). As a screening tool, the overhead squat is used to highlight compensatory movements such as excessive trunk flexion, heel elevation, or knee valgus. These compensations may reflect underlying imbalances in mobility or motor control (Cook et al., 2014). Monteiro et al. (2022) found that increased heel height reduces trunk inclination and promotes a more upright posture, confirming the importance of heel position in squat mechanics. This study found trunk flexion to be 7.5° less when using a 50 mm heel wedge compared to flat footed. However, the knee flexion angle was found to be 10.9° greater when using the 50mm heel raise compared to flat footed (Monteiro et al., 2022). Such findings underscore the need to consider how slight modifications in stance influence posture, kinematics and joint loads.

Sato et al. (2012) found a back squat with weightlifting shoes (25mm heel height) would display less trunk lean displacement when compared with that of the running shoe condition. There will always be some anterior bar displacement accompanied by some posterior hip displacement during the back squat, which creates trunk lean. The goal, then, is to minimize these movements to reduce the amount of trunk lean. This information is vital as the tibia and trunk are required to be parallel during squats to reduce compensatory movements. However, when flatfooted often an excessive trunk flexion and hip hinge is performed (Sato et al., 2012), whereas, if the heel wedges are too high, an extended trunk angle can be present, increasing the anterior translation of the knee (List et al., 2013).

In the FMS protocol, participants unable to perform a flat-footed overhead squat to the specific criteria, a 4.3 cm heel wedge is then used as an aid. While this is intended to facilitate squat depth and alignment, there is limited research validating the biomechanical effects of this specific elevation. A prior study has shown that heel elevation during a back squat with in a weightlifting shoe (25mm heel height) increases tibial inclination and knee flexion, thereby altering the knee extensor moment (F (1,28) = 6.93; P = 0.01) (Legg et al., 2018). This however is in a group of novice lifters as no statistical significance was identified in the group of experienced lifters. Charlton et al. (2017) determined the peak knee flexion angles significantly increased (p = 0.004) when squatting with wedges (133.4 ±9.9°) compared with no wedges (129.4±10.7°). These results are due to the increased tibia angle resulting in the anterior translation of the knee beyond the toes. Additionally, Monteiro et al. (2022) identified a trunk flexion of 32.8° and a knee flexion of 115.1° when squats were performed flatfooted, whereas when using a 50mm heel wedge trunk flexion decreased to 25.3° but knee flexion increased to 126.2°. This is vital as a key criteria for the FMS deep squat is maintaining an upright torso but no considerations are given to the greater knee flexion requirement. Although there are no reports on the patellofemoral joint force or stress within these studies.

Patellofemoral joint forces have however been quantified in other studies, with Wallace et al. (2002) performing bodyweight squats and identified an eccentric peak patellofemoral stress of 9.06 MPa and a concentric peak patellofemoral stress of 9.3 MPa with both these being found when the knee was at 90° flexion. Furthermore, when considering different squat stances, Gullett et al. (2009) quantified the back squat to result in a peak knee compressive force of 11.0 ± 2.3 N/kg−1 and the front squat to result in 9.3 ± 1.5 N/kg−1. These differences are due to the knee extensor moment being greater in the back squat (1.0 ± 0.4 N/m/kg−1 ) compared to the front squat (0.7 ± 0.2 N/m/kg−1). However, it would be assumed that the front squat would have had a greater knee extensor moment due to the trunk being more extended compared to the back squat. Since the magnitude of a moment is also influenced by force, greater knee extensor moment for the back squat would have been due to the increase in the amount of weight lifted in the study. A further study has identified the traditional, powerlifting and box squat to result in different knee extensor moment arms, with the box squat resulting in the greatest knee extensor moment arm (15.2±2.8 cm) and moment (229±39 Nm) at 70% of 1RM whereas the powerlifting squat had the smallest moment arm (8.1±0.8) and moment (192±36) again at 70% of 1RM, with all squat being performed to parallel (Swinton et al., 2012).

Escamilla et al. (2009) quantified patellofemoral joint force and stress in wall squats with short and long stances and single leg squats. This study demonstrated that wall squats at short stances generated greater patellofemoral stress compared to the other techniques at 90° knee flexion. Knees translated over the toes at 90° in the wall squat short whereas knees remained over the feet in the long wall squat, supporting clinician’s beliefs that anterior knee translation beyond the toes while performing squatting type exercises increases patellofemoral force and stress compared with maintaining the knees over the feet (Escamilla et al., 2009). Escamilla et al. (2025) found that squats performed on unstable surfaces such as a BOSU ball significantly reduced patellofemoral joint loading compared to bodyweight squats on firm ground, implying that surface and stability also influence patellofemoral joint stress. A patellofemoral joint force of 1414±777 N was recorded at 90° in the normal squat whereas a BOSU squat scored 1328±762 N. Escamilla et al. (2012) investigated the wall squat versus the ball squat in short (knee beyond toes) and long (knee level with toes) stance. This study found across ball squats and wall squats, patellofemoral joint force and stress were greater with a short heel to wall distance (1338±424) than at long heel to wall distance (1208±418) at 90° knee flexion. These studies show how varieties of squats have quantifiable differences on patellofemoral joint stress, therefore the lack of research in the FMS squat technique is concerning.

The FMS overhead squat stance is a restricted stance being narrow with no foot angle and current research has identified a narrow stance with feet at 0° to have the greatest external moment (0.4 N/m/kg) compared to any other set up, with the study stating, a narrow stance width with small foot angle results in increased hip and knee moments in the frontal plane (Lorenzetti et al., 2018). Narrow stances increase demands on the knees by promoting greater flexion, valgus movement, and anterior knee displacement. This is supported by Sinclair et al. (2022) who identified a narrow stance to produce 121.27° of knee flexion compared to a wider stance resulting in 112.09°. Almosnino et al. (2013) noted that narrower stances with minimal foot rotation significantly increase knee joint moments in the frontal plane. These factors may elevate injury risk, particularly when performed repetitively or under load (Myer et al., 2014).

Despite the prevalence of the FMS protocol in athletic and rehabilitation settings, the stance used in the deep squat is quite restricted compared to functional or sport-specific movements. The strict shoulder-width stance with feet at 0° rotation limits the natural variability typically observed in squatting technique. This constraint may contribute to compensatory patterns, particularly in individuals with limited ankle dorsiflexion or hip mobility (Lorenzetti et al., 2018; Sinclair et al., 2022). Research has demonstrated that allowing athletes to self-select stance width and foot position can improve movement efficiency and reduce abnormal joint loading (Lorenzetti et al., 2018). As such, the FMS deep squat’s standardised setup may not accommodate individual anatomical differences, limiting its ecological validity and potentially inflating dysfunction scores.

Increased heel elevation alters the tibial angle and allows for a more upright torso, which meets FMS criteria (Charlton et al., 2017; Monteiro et al., 2022). While moderate elevation (e.g., 25 mm) has been found to aid squat stability (Duan et al., 2025), the use of 4.3 cm heel wedges has not been thoroughly studied. Assumptions from previous research are that a higher heel wedge may increase patellofemoral joint stress and patellofemoral joint force due to anterior translation beyond the feet (Escamilla et al., 2009; Escamilla et al., 2025), a likely more upright torso (Monteiro et al., 2022) and the narrow stance (Lorenzetti et al., 2018).

Given the lack of kinematic and kinetic data for the specific FMS squat setup, there is a clear research gap. This gap is particularly relevant considering how frequently the FMS is used in clinical, athletic, and military populations (Morgan et al., 2023). The FMS overhead squat—especially when performed with heel wedges—may contribute to elevated PFJS, PFJF and increased stress on the knees, yet few studies have directly measured these outcomes. This is significant when considering movement screening as both a diagnostic and predictive tool for injury risk. Understanding the effects of stance and heel elevation can enhance the practical value of the FMS and improve injury prevention efforts across a variety of populations.

The outcomes of this study may have implications for clinical screening practices, rehabilitation program design, and injury prevention strategies by informing practitioners about compensatory movements on the biomechanical demands on the knee joint. Based on current literature, the aims of this research project and hypotheses are stated below.

## 1.2 Objectives

### 1.2.1 Primary Objective(s)

The first aim of this study is to quantify patellofemoral joint kinetics during variations of the overhead squat in accordance with the FMS protocol and an individual’s preferred stance and grip width. The second aim is to quantify the effect of variation in heel height while performing the overhead squat techniques on patellofemoral joint kinetics.

### 1.2.2 Secondary Objective(s)

The second aim is to quantify the effect of variation in heel height while performing the overhead squat techniques on patellofemoral joint kinetics

## 1.3 Trial Hypothesis

### 1.3.1 Null Hypothesis One

PFJS, PFJF and the knee extensor moment will not be significantly greater in the FMS condition versus the preferred stance condition

### 1.3.2 Alternate Hypothesis

PFJS, PFJF and the knee extensor moment will be significantly greater in the FMS condition versus the preferred stance condition.

### 1.3.3 Null Hypothesis Two

PFJS, PFJF and the knee extensor moment will not be significantly greater when utilising the FMS box in comparison to flatfooted and heel raises of 1CM, 2CM and 3CM.

### 1.3.4 Alternate Hypothesis Two

PFJS, PFJF and the knee extensor moment will be significantly greater when utilising the FMS box in comparison to flatfooted and heel raises of 1CM, 2CM and 3CM.

# 2.0 Methodology

## 2.1 Trial Design

An observational study quantifying PFJS, PFJF and KEM in different overhead squat techniques

### 2.1.1 Instrumentation

Kinetic data were collected using two AMTI force plates (Advanced Mechanical Technology Inc., MA, USA), each sampling at 2000 Hz. These force plates are designed to measure ground reaction forces with high precision, making them suitable for dynamic movement analysis. Kinematic data were captured using an 18-camera optoelectronic motion capture system (Vicon, Oxford, UK) operating at 100 Hz.

### 2.1.2 Marker Placement

Reflective markers, each 14 mm in diameter, were affixed bilaterally to specific anatomical landmarks to facilitate accurate motion tracking. Markers were attached using double sided adhesive tape with participants wearing skintight shorts to prevent abnormal marker movement. These landmarks included the first, second, and fifth metatarsal heads; base of the Achilles tendon; medial and lateral malleoli; tibial tuberosity; fibular head; medial and lateral femoral epicondyles; greater trochanter; anterior superior iliac spine (ASIS); posterior superior iliac spine (PSIS); iliac crest; sacrum; spinous processes of L1, L3, L5, T2, T7, and C7; right scapular region (inferior angle); acromion process; lateral humeral epicondyle; midpoint between acromion and lateral epicondyle; radial and ulnar styloid; first knuckle; and midpoint between radial styloid and lateral epicondyle. A marker cluster was affixed to the T3 spinous process, and four Velcro markers were attached to a headband to track head movements. Markers were also placed at each end of the barbell used during squatting tasks. All marker placements were performed following accurate palpation of a highly educated sport therapist, applied by the same therapist to ensure consistency and reliability across participants. Pre squats procedure was completed in accordance with the Plug-in gait upper body model (Vicon, Oxford, UK) presented by Leardini et al. (2011) and Rizzoli lower body protocol (Leardini et al., 2007).

## 2.2 Participant Information

Ten physically active male participants were recruited for this study (Age-22.9±2.72 years, Height-179.65±6.83cm, Mass-79.45±10.83kg). Inclusion criteria required participants to engage in regular physical activity, defined as at least 150 minutes of moderate-intensity exercise per week. Exclusion criteria included any current musculoskeletal pain during squatting movements or a history of lower limb musculoskeletal injury within the preceding six months. Prior to participation, all individuals provided written informed consent in accordance with the Declaration of Helsinki. The study protocol received ethical approval from the University Research Ethics Committee.

### 2.2.2 Trial Location and Setting

All trials were competed in the Biomechanics laboratory situated in the Science Centre of University of Staffordshire.

## 2.3 Outcomes

### 2.3.1 Primary Outcome(s)

Prior to dynamic trials, static calibration trials were conducted to verify and, if necessary, adjust marker placements. The force plates and motion capture system were calibrated according to manufacturer guidelines to ensure accurate data collection. A metronome set at 60 beats per minute was used to standardise squat tempo, guiding participants to perform a 3-second descent followed by a 3-second ascent. Participants were instructed to squat to a depth of 90°knee flexion.

Each participant completed three valid trials for each squat condition. Trials were excluded if the timing was incorrect, the dowel was not held overhead (when required), or if there was excessive movement before or after the squat. Participants were allowed familiarisation attempts before each new condition to ensure comfort and proper execution. Preferred Stance (PS) overhead squats were performed in accordance with UK Strength & Conditioning Association (UKSCA) guidelines (Bishop et al., 2016; Myer et al., 2014), and the FMS overhead squats were performed in accordance with the protocol detailed by Cook et al. (2014). Using these guidelines, participants were informed to descend in a controlled manner, keep both feet flat on the floor or heel wedges throughout, maintain the bar fully pressed overhead, preserve the proper breath control and maintained a constant/stable pattern of motion for each repetition. Each participant was observed by a sports therapist student with overhead squats being rejected if timings or guidelines were breached, but no additional coaching instructions were provided.

Three squat stances were evaluated:

FMS Stance: Feet shoulder-width apart, pointed straight ahead at 0°, arms bent at 90° with the dowel resting on the head, then pressed overhead (Cook et al., 2014)

Bodyweight Squat: Feet hip-width apart, angled at approximately 10°, arms raised forward during the squat.

Preferred Stance: Feet hip-width apart, angled at approximately 10°, dowel grip self-selected (wider than FMS stance) (Bishop et al., 2016; Myer et al., 2014).

Each squat stance was performed under five heel height conditions: flat-footed (0 cm), 1 cm, 2 cm, 3 cm, and 4.3 cm (FMS box). The order of squat trials was as follows: static trial, FMS squat, bodyweight squat, and preferred stance squat. Cameras were re-synchronised, and force plates were zeroed between protocols as needed to maintain data integrity.

### 2.3.2 Secondary Outcomes(s)

Motion capture data were processed using Nexus software (Vicon, OMG, Oxford), where markers for the pelvis, lower limbs, and spine were manually labelled. Marker gaps resulting from deep squat positions, particularly at the ASIS, were addressed by referencing adjacent markers on the iliac crest and PSIS, as recommended by McClelland et al. (2010). The processed C3D files were then exported to Visual3D (C-Motion, Inc., Germantown, MD) for further analysis.

In Visual3D, Ground reaction force and marker data were filtered at 50 Hz and 15 Hz respectively using a low-pass Butterworth 4th order filter, and processed using Visual 3-D (C-Motion, 140 Germantown, MD, USA). Internal moments were computed using Newton-Euler inverse-dynamics, allowing net knee joint moments to be calculated. Angular kinematics of the knee joint were calculated using an YXZ (sagittal, coronal and transverse) sequence of rotations, allowing sagittal flexion angles at 90 degrees to be extracted. Squat phases (start and end) were defined based on the displacement of the lateral femoral epicondyle marker. A custom script was created in Visual 3D to identify 90° of knee flexion. The software was utilised to calculate knee flexion angles, knee extensor moments (KEM), patellofemoral joint force (PFJF), and patellofemoral joint stress (PFJS). Concentric results at the left knee flexion of 90° was used to maintain consistency. These calculations provided quantitative insights into joint loading patterns during various squat conditions. Calculations are as follows; quadriceps muscle effective lever arm (Leff) is calculated. Knee extensor moment obtained during the squat is divided by the calculated Leff, equalling the quadricep muscle force. Coefficient k is calculated. Coefficient k is a constant that defines the relation between quadriceps force and PFJ reaction force as a function of knee flexion angle. Patellofemoral joint contact area is calculated as a function of knee flexion angle. Patellofemoral Joint Force (PFJF) is the calculated by multiplying the quadriceps muscle force by the calculated coefficient k. Patellofemoral Joint Stress (PFJS) is calculated by dividing the PFJF by the patellofemoral joint contact area (Nunes et al., 2018)

## 2.4 Sample Size

Similar designed studies had recruited 16 (Escamilla et al., 2022), 20 (Song et al., 2023) and 20 (Sinclair et al., 2022) therefore 20 recruits are the aim of this study, however this was not achieved with only 10 participants being recruited.

## 2.5 Blinding and Randomisation

Participants were blind to their results during the procedure to prevent them from adapting squats, with no video footage or feedback given during the trial.

## 2.6 Statistical Methods

Statistical analyses were performed using SPSS (IBM Corp., Armonk, NY). Laerd statistics (2015) was followed to support the application of the two-way repeated measures ANOVA. A two-way repeated measures ANOVA was conducted to examine the effects of stance type (FMS vs. preferred) and heel height (0 cm, 1 cm, 2 cm, 3 cm, and 4.3 cm) on PFJS, PFJF, and KEM. Shapiro-Wilk’s test of normality was performed to assess for normal distribution of data. Mauchly’s Test of Sphericity was used to assess the assumption of sphericity.

Post hoc pairwise comparisons were conducted using Bonferroni correction to identify significant differences between heel heights. Descriptive statistics (mean ± SD) were calculated for each dependent variable. A significance level of *p* < .05 was considered statistically significant for all analyses. All data were visually inspected for normality and outliers using boxplots and Q-Q plots. No extreme outliers or violations of normal.

# 3.0 Results

## 3.1 Baseline Data

A total of ten participants were recruited with the average age of 22.9 years, height of 179.65 cm, and mass of 79.45 kg.



**Table One**. Characteristics of participants

## 3.2 Outcomes

### 3.2.1 Patellofemoral Joint Stress (PFJS)

A two-way repeated measures ANOVA was conducted to examine the effects of heel height level (FF, 1CM, 2CM, 3CM, BOX) and movement technique (FMS vs PS) on the mean peak patellofemoral joint stress (PFJS). Analysis of the studentized residuals showed that there was normality, as assessed by no studentized residuals greater than ±3 standard deviations. PFJS was normally distributed, as assessed by Shapiro-Wilk’s test of normality on the studentized residuals (p>.05). There was sphericity for the interaction term, as assessed by Mauchley’s test of sphericity (p<.05). Post hoc pairwise comparisons were conducted using Bonferroni correction to identify significant differences between heel heights.

Post hoc comparisons revealed the mean peak PFJS was significantly greater in the PS compared to FMS when performed at flatfooted (1.81±0.70 MPa, p = 0.030), however, no other statistically significant differences was found between squat techniques at any other heel height (see table three). Within the FMS technique PFJS was significantly higher at the BOX compared to FF (8.52±1.35 MPa, p = < .001), 1CM (6.42±1.19 MPa, p = .004), 2CM (5.19±1.27 MPa, p = .028), but not at 3CM (4.31±1.51 MPa, p = 0.192) (see table five). Whereas with the PS, the BOX was statically significantly greater compared to FF (5.74±1.12 MPa, p = 0.006) and 1CM (4.61±1.17 MPa, p = 0.34) but not at 2CM (3.48±1.24 MPa, p = 0.207) or 3CM (3.15±1.2 MPa, p = 0.287), however theses values are not as significant compared to the FMS. FMS also found statistically significant differences between FF and 3CM (4.21±0.96 MPa, p = 0.17) but no differences were identified between 1CM, 2CM AND 3CM trials (see table five).



**Table Two**. Descriptive statistics of the mean peak PFJS

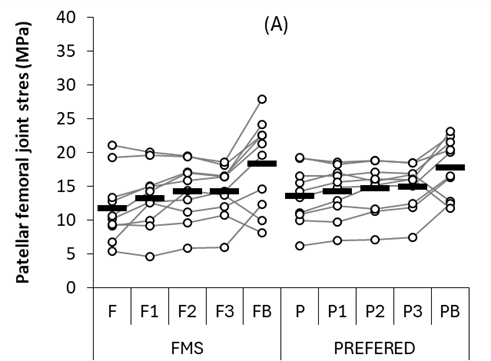
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**Table Three**. Pairwise comparison of PFJS in techniques at different heel heights.

b- Adjustments for multiple comparisons. Bonferroni

\*- Mean difference is statistically significant p = <0.05



**Table Four**. Plot Chart showing PFJS results. Black lines = peak means, grey lines and white circles = individual participants.

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**Table Five**. Pairwise comparisons of PFJS in FMS and PS techniques at different heel heights.

\*. The mean difference is significant at the .05 level.

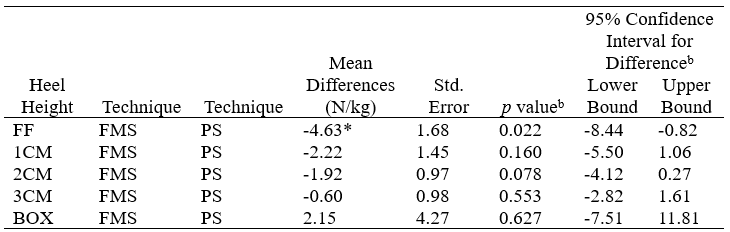
### 3.2.2 Patellofemoral Joint Force (PFJF)

A two-way repeated measures ANOVA was conducted to examine the effects of heel height level (FF, 1CM, 2CM, 3CM, BOX) and movement technique (FMS vs PS) on the mean peak patellofemoral joint force (PFJF). Analysis of the studentized residuals showed that there was normality, as assessed by no studentized residuals greater than ±3 standard deviations. PFJF was normally distributed, as assessed by Shapiro-Wilk’s test of normality on the studentized residuals (p>.05). There was sphericity for the interaction term, as assessed by Mauchley’s test of sphericity (p<.05). Post hoc pairwise comparisons were conducted using Bonferroni correction to identify significant differences between heel heights.

The greatest difference in the mean peak of PFJF was observed in the comparison of FMS FF vs FMS BOX (21.18±3.60 N/kg, p = 0.002). The PS FF vs PS BOX was also statistically significant (p = 0.009) but the mean difference was not as large (14.40±2.94 N/kg) (see table nine). When comparing techniques, a significant difference was noted at FF, with PS having a greater mean peak PFJF by 4.63±1.68 N/kg (p = 0.022) versus FMS but no other statistically significant differences were noted between conditions at any other heel height (see table seven). Both techniques identified FF to have significant differences when compared to 2CM, 3CM and the BOX but not at 1CM. FMS found 1CM to have significant differences compared to 3CM (5.22±1.41 N/kg) and BOX (16.06±3.11 N/kg), whereas PS found 1CM to have statistical differences compared to 2CM (2.74±0.54 N/kg) and BOX (11.69±3.03 N/kg) (see table nine). Similarly to the PFJS, FMS resulted in PFJF being statically significant when comparing BOX to FF (21.18±3.60 N/kg, p = 0.002), 1CM (16.06±3.11 N/kg, p = 0.006) and 2CM (13.01±3.19 N/kg, p = 0.28, but not at 3CM (10.83±3.63, p = 0.154), whereas PS found significant differences only at FF (14.40±2.94 N/kg, p = 0.009) and 1CM (11.69±3.03 N/kg, p = 0.039) when compared to the BOX but not at 2CM (8.94±3.13 N/kg, p = 0.189) and 3CM (8.08±2.93 N/kg, p = 0.221) (see table nine).



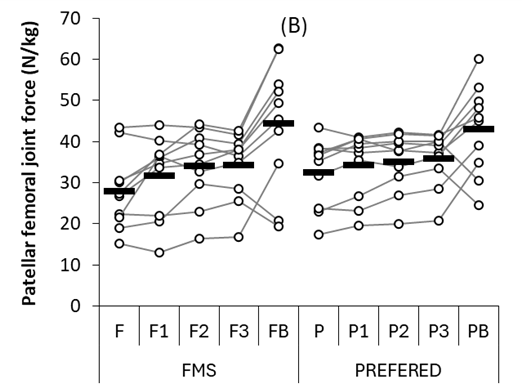
**Table Six**. Descriptive statistics of the mean peak PFJF



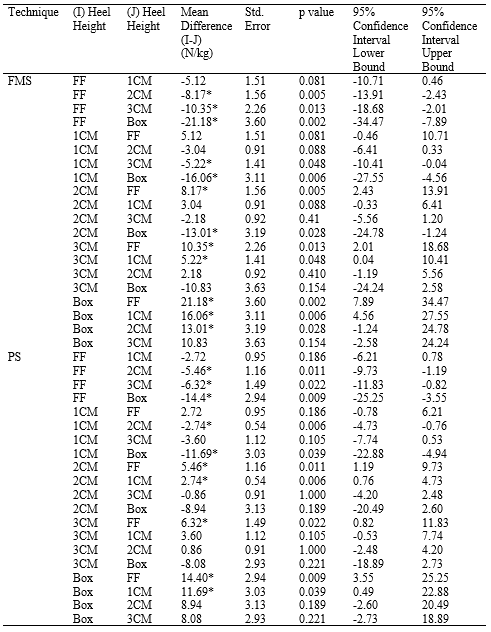
**Table Seven**. Pairwise comparison of PFJF in techniques at different heel heights.

b- Adjustments for multiple comparisons. Bonferroni

\*- Mean difference is statistically significant p = <0.05



**Table Eight**. Plot Chart showing PFJF results. Black lines = peak means, grey lines and white circles = individual participants.



**Table Nine**. Pairwise comparisons of PFJF in FMS and PS techniques at different heel heights.

\*. The mean difference is significant at the .05 level.

### 3.2.3 Knee Extensor Moment (KEM)

A two-way repeated measures ANOVA was conducted to examine the effects of heel height level (FF, 1CM, 2CM, 3CM, BOX) and movement technique (FMS vs PS) on the Knee Extensor Moment (KEM). Analysis of the studentized residuals showed that there was normality, as assessed by no studentized residuals greater than ±3 standard deviations. KEM was normally distributed, as assessed by Shapiro-Wilk’s test of normality on the studentized residuals (p>.05). There was sphericity for the interaction term, as assessed by Mauchley’s test of sphericity (p<.05). Post hoc pairwise comparisons were conducted using Bonferroni correction to identify significant differences between heel heights.

There was no significant main effect of technique (p = .123- p = 0.627), identifying that KEM values did not significantly differ between FMS and PS across all heel heights. FMS at the BOX found the greatest results (+0.48±0.10 N/m/kg) vs PS whereas PS had the greatest KEM when FF (0.05±0.03 N/m/kg), 1CM (0.04±0.03 N/m/kg), 2CM (0.04±0.02 N/m/kg) and 3CM (0.01±0.02 N/m/kg) compared to FMS. However, there was a significant main effect of heel height in FMS FF vs BOX (0.41±0.7 N/m/kg, p = 0.003) and PS FF vs BOX (0.31±0.6 N/m/kg, p = 0.006) indicating KEM increased with increasing heel heights. FMS found FF to be significantly less KEM results compared to 2CM (0.12±0.28 N/m/kg, p = 0.26), 3CM (0.17±0.42 N/m/kg, p = 0.32) and BOX (0.41±0.7 N/m/kg, p = 0.003) but not compared to 1 CM (0.06±0.03 N/m/kg, p = 0.376). The same trend was observed with PS with FF being significantly less than 2CM (0.11±0.26 N/m/kg, p = 0.28), 3CM (0.13±0.31 N/m/kg, p = 0.27) and BOX (0.31±0.6 N/m/kg, p = 0.006) but not compared to 1CM (0.05±0.02 N/m/kg, p = 0.688). No significant differences were observed in FMS between 1CM, 2CM and 3CM however PS identified significant differences between 1CM and 2CM (0.06±0.01 N/m/kg, p = 0.007). Similarly to previous results, FMS found significant differences between BOX and 2CM (0.29±0.72 N/m/kg, p = 0.028) whereas this significance was not present in PS comparing these heel heights (0.20±0.7 N/m/kg, p = 0.183).



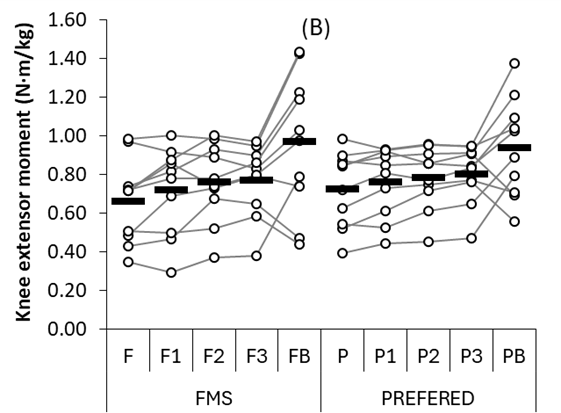
**Table Ten**. Descriptive statistics of the mean peak KEM

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**Table 11.** Pairwise comparison of KEM in techniques at different heel heights.

b- Adjustments for multiple comparisons. Bonferroni



**Table 12**. Plot Chart showing KEM results. Black lines = peak means, grey lines and white circles = individual participants.

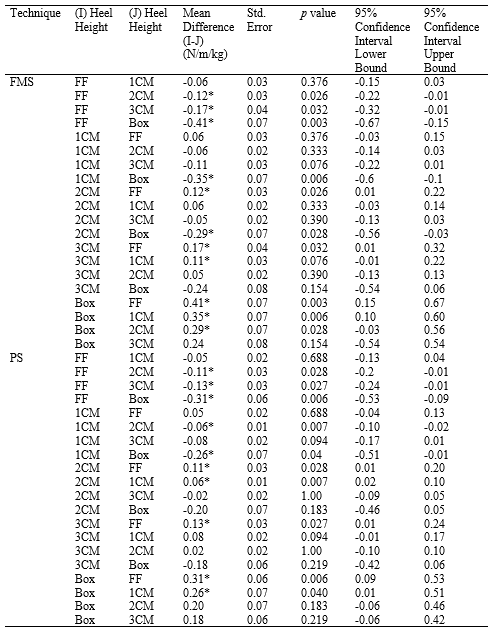


Table 13. Pairwise comparisons of KEM in FMS and PS techniques at different heel heights.

\*. The mean difference is significant at the .05 level.

# 4.0 Discussion

## 4.1 Interpretation

The primary aim of this study was to quantify patellofemoral joint kinetics during variations of the overhead squat, comparing the Functional Movement Screen (FMS) protocol with an individual's preferred stance (PS) and grip width. When analysing the outcomes, the PS technique exhibited significantly greater patellofemoral joint stress (PFJS) (1.81±0.70 MPa, *p* = 0.03) and patellofemoral joint force (PFJF) (4.63±1.68 N/kg, *p* = 0.022) in the flatfoot (FF) condition compared to FMS. In the 1CM, 2CM, and 3CM heel wedge conditions, PS showed higher PFJS, PFJF, and knee extensor moment (KEM) compared to FMS, though these differences were not statistically significant. Conversely, in the BOX condition, the FMS stance demonstrated higher PFJS (0.97±1.8 MPa), PFJF (2.15±4.27 N/kg), and KEM (0.048±0.095 N/m/kg), though again without statistical significance. No statistically significant differences were seen in KEM in any heel height condition comparing the squatting techniques. Given these results, the null hypothesis one was accepted, as no consistent significant differences were found between techniques across all heel heights.

From previous research, it was predicted that due to the FMS technique being narrow and restricted, differences would be found in PFJS, PFJF and KEM however; this study found no statistically significant differences in PFJS, PFJF, or KEM across the heel wedge conditions. However, in the flatfoot (FF) condition, PS exhibited significantly higher PFJS (by 1.81 MPa) and PFJF (by 4.63 N/kg) compared to FMS. These statically significant differences at FF are likely due to the reduced anterior knee displacement within the narrow FMS stance, increasing the demands placed onto the hip. This occurs because ground reaction forces relative to the centre of mass move closer to the knee due to excessive hip flexion. This reduces anterior knee displacement and increases demands on the hip, factors influenced by reduced ankle dorsiflexion. This is shown as an unrestricted back squats produced higher knee moments (150.1 ± 50.8 Nm) than restricted squats (117.3 ± 34.2 Nm) (Fry et al., 2003). The restricted stance limited anterior translation of the knee which resulted in a greater knee flexion angle of 73.4±10.5 (restricted) compared to 66.1±10.0 (unrestricted) but even though the knee moment is greater in the restricted stance, the hip moment was significantly different in the opposing way. The unrestricted squat resulted in 28.2±65.0 Nm whereas the restricted squat resulted in 302.7±71.2, which shows the huge load placed onto the hip when restricted (Fry et al., 2003). A reduced foot angle was also found to have reduced PFJF, with a barbell back squat finding peak patellofemoral joint forces at a foot angle of, 0° to be 32.22±9.32 N/kg, 21° to be 34.32±9.42 N/kg and 42° to be 34.05±8.45 N/kg. Even though there is no statistical significance comparing these, when performed at their own preferred position (foot angle=19.56 ± 6.45°) the force was recorded at 34.71±9.32 N/kg which was statistically significant compared to squats at 0° (Sinclair et al., 2022). These studies explain why the FF condition produced greater demands in the PS position due to its wider stance and unrestrictive nature. However, a key find from this study is the lack of significant differences between techniques when using heel wedges which as stated in the hypotheses, due to previous research identified in the introduction FMS would yield greater PFJS, PFJF and KEM compared to PS.

The second aim was to quantify the effect of heel height variation during overhead squats on patellofemoral joint kinetics. Across both FMS and PS conditions, a progressive increase in PFJS, PFJF, and KEM was observed between each heel height between FF to BOX. Statistically significant differences were present between FF to BOX conditions in PFJS (PS=5.74±1.12 MPa, *p* = 0.006, FMS=8.52±1.35 MPa, *p* = < .001), PFJF (PS=14.40±2.94 N/kg, *p* = 0.009, FMS=21.18±3.60 N/kg, *p* = 0.002) and KEM (PS=0.31±0.6 N/m/kg, *p* = 0.006, FMS=0.41±0.7 N/m/kg, *p* = 0.003) supporting the acceptance of alternate hypothesis two. This outcome suggests that heel elevation consistently increases patellofemoral joint loading, regardless of squat technique.

In examining heel height adaptations, the most notable increases in knee kinetics were consistently observed between 3CM and BOX conditions, with intermediate heel heights also meriting discussion. Although knee kinetics incrementally increased with heel elevation, statistical significance was not always achieved. Nevertheless, the observed trends suggest that even small heel elevations can meaningfully alter squat biomechanics. Both FMS and PS conditions demonstrated gradual increases in PFJS and PFJF from FF to 1CM, with significant increases identified at 2CM relative to FF. This suggests that the 2CM heel height condition may represent the point where patellofemoral joint demands become more substantial. However, these results contradict findings from a systematic review by Duan et al. (2025), which reported that moderate heel elevations of 25 mm provide optimal squat stability in the anterior-posterior direction without compromising ankle dorsiflexion or knee stress. The same review noted that stability reduces when 30mm wedges are used. These discrepancies highlight the need for further clarification regarding heel wedge usage and its effects on biomechanics and joint kinetics. Differences may also stem from variations in squat form, as Duan et al. (2025) did not focus on overhead squats, which is the focus of this study.

Even though our study identifies 2cm heel heights to elicit statistical significance compared to FF there is no consideration of trunk displacement. As identified by Sato et al. (2012), when utilising greater heel heights through weightlifting shoes compared to running shoes, the trunk displacement significantly decreases (216.44±51.04 versus 238.36±52.78 respectively). This is an area that proves to be a limiting factor in this study as the lack of recording trunk displacement may influence our results with assumptions having to be made. However, through our knowledge of PFJF calculations, results are recorded with the knee flexion of 90° in this study meaning the PFJF is highly dependent on the quadriceps muscle. PFJF is the measurement of the compression of the patella against the femur and is dependent on the knee flexion angle and muscle tension (Powers, 1998). This shows that the increase in demand through the increased heel height is not due to the knee flexion angle but due to the increased moment arm of the knee and increased quadriceps contraction.

However, the great significance of the BOX conditions results on PFJS, PFJF and KEM is highlighted by other studies. Wallace et al. (2002) examined the impact of external loading on patellofemoral joint kinetics, identifying peak PFJS and PFJF at 90° of knee flexion in a back squat. A 35% bodyweight external load resulted in a 44% increase in PFJS across all knee flexion angles, with a peak stress of 13.06 MPa in the loaded condition, compared to 9.06 MPa in the unloaded condition. This study found that the FMS BOX condition produced a PFJS 6.94 MPa higher than the loaded condition in Wallace et al.'s study. Similarly, PFJF in FMS and PS FF conditions (27.33 N/kg and 31.96 N/kg, respectively) exceeded those of the unloaded squat but remained below the loaded condition. In the BOX condition, PFJF was greater by an additional 14.1 N/kg (FMS) and 11.9 N/kg (PS) relative to the loaded squat. These results show the significance of how great the PFJS and PFJF was within the overhead squat in all conditions. Wallace et al. (2002) confirmed that deeper squats increased PFJS and PFJF, with peak values recorded at 90° knee flexion. This is crucial, as comparisons cannot attribute results to increased flexion angles, given both studies recorded peak means at 90° knee flexion. Powers et al. (2014) additionally measured PFJS, finding a peak mean of 12.3 ± 1.6 MPa at 90° of knee flexion during bodyweight squats. The elevated joint PFJF and PFJS observed within this study in comparison are likely due to heel height manipulation, which shifts the centre of mass, promotes a more upright trunk, increasing the trunk lean displacement (List et al., 2013) and results in anterior knee translation beyond the toes—factors supported by Escamilla et al. (2009) and Escamilla et al. (2025). These results show the magnitude of stress found in the BOX conditions. However, Wallace et al. (2002) and Powers et al. (2014) recruited mixed-gender samples (nine females, six males) may contribute to result discrepancies, as gender-based anatomical and neuromuscular differences are well-documented (Gaviraghi et al., 2024). Gender-specific differences such as pelvic width, Q-angle, and neuromuscular activation patterns are known to influence knee joint biomechanics, with females often exhibiting higher valgus angles and altered loading patterns compared to males (Gaviraghi et al., 2024). This highlights the need for gender-specific research in patellofemoral joint loading.

## 4.2 Considerations

A consideration must be given in study comparisons and the variation of squat. Escamilla et al. (2012) identified different squat techniques result in different PFJS and PFJF, and the majority of studies that have previous quantified PFJS or PFJF or KEM have typically performed back squats (Wallace et al., 2002; Sinclair et al., 2022; Gullet et al., 2009) or bodyweight squats (Fry et al., 2003; Powers et al., 2014). Comparative studies typically performing back squats is a consideration as the back squat and overhead squat do have different demands. The back squat has been identified to result in significantly greater activity in the gluteus maximus, vastus lateralis, bicep femoris and lateral gastrocnemius in the concentric aspect of the lift (Aspe & Swinton. 2014). The loads however were significantly greater in the back squat compared to the overhead squat, further explaining these differences, likely why differences have been seen in previous studies.

This area of research into squats is heavily performed due to the popularity of the movement within weightlifting, which proved to be a consideration in study comparisons. Whereas our study had no requirements for lifting experience, studies previously stated had inclusion criteria of a year or multiple years’ experience in weightlifting which would likely result in different results due to experiences of the movement resulting in better neuromuscular control and strength within the movement (Sinclair et al., 2022; Sinclair et al., 2022).

Squat tempo in comparisons to certain studies must be considered. Gullett et al. (2009) conducted a biomechanical comparison of back and front squats in trained individuals, identifying KEM values of 1.0 ± 0.4 N/m/kg (back squat) and 0.7 ± 0.2 N/m/kg (front squat). Comparatively, this study found lower KEM in the FF conditions for FMS (0.68 ± 0.24 N/m/kg) and PS (0.73 ± 0.20 N/m/kg). Gullett et al.'s (2009) participants, who had at least one year of resistance training and included mixed genders (nine males, six females), likely exhibited greater KEM due to their training backgrounds and the use of loads within the lifts. Gullett et al. (2009) did not control squat tempo, allowing self-selected pacing. Though squat tempo's impact on joint kinetics remains understudied, resistance training research suggests tempo influences hypertrophy and strength (Wilk et al., 2021), likely affecting joint stress and force. Tempo variations, particularly eccentric loading tempos, can lead to increased time under tension, which has been correlated with higher joint loading in strength-based movements (Wilk et al., 2021). Incorporating controlled tempos into biomechanical studies may provide clearer insights into joint kinetics.

## 4.3 Practical Recommendations

The findings of this study carry notable implications for clinical practice, particularly in rehabilitation, injury prevention, and athletic performance enhancement contexts. The consistent increase in patellofemoral joint stress (PFJS), patellofemoral joint force (PFJF), and knee extensor moment (KEM) with heel elevation across both the FMS and PS conditions underscores the need for individualised squat modifications based on joint loading tolerance. For rehabilitation practitioners, this suggests that prescribing elevated heel squats—especially at higher heel heights such as the BOX condition—must be done cautiously in populations with patellofemoral pain syndrome (PFPS), osteoarthritis, or prior knee injuries (Arazpour et al., 2014). The excessive anterior translation of the knee observed with heel elevation could exacerbate joint stress, potentially aggravating symptoms in these populations.

A 1 cm heel height within the FMS protocol is recommended, as it does not significantly increase PFJS, PFJF, or KEM compared to FF. Integrating our knowledge from previous studies to this study, the 1CM heel wedge height will allow for a reduced trunk displacement due to increasing heel elevation (Sato et al., 2012), whilst not forcing excessive anterior translation of the knee.

For movement screening, particularly using the Functional Movement Screen (FMS), this study raises important concerns about the standardisation of squat assessments without accounting for individual anatomical or mobility differences. The increased variability in joint loading and compensatory patterns in the FMS condition suggests that rigid protocols may yield misleading dysfunction scores. Incorporating self-selected stance widths or adjustable foot angles into screening protocols could enhance ecological validity and provide more actionable insights for practitioners.

Additionally, this study highlights the potential role of heel wedges as a therapeutic tool for modifying squat mechanics. In cases of limited ankle dorsiflexion, heel wedges can facilitate upright trunk posture and deeper squat depth, improving movement quality (Monteiro et al., 2022). However, the corresponding increase in knee joint loading necessitates careful progression, possibly starting at lower wedge heights (1CM to 2CM) before advancing to higher elevations.

## 4.4 Limitations

This study mainly focused on FMS versus PS overhead squat techniques and the influence of heel heights on PFJS, PFJF and KEM, however it is important to report some limiting factors to this study to aid future research in this area. Firstly, the requirement for trunk flexion angles to be recorded, as not doing so has limited our ability to assess its influence on PFJS, PFJF and KEM throughout both squat techniques and may have provided explanation for results. Recording trunk flexion could have clarified the relationship between torso position and patellofemoral forces and stress, as previous research suggests that elevated heel wedges increase trunk extension, shifting stress away from the hips and onto the knees (List et al., 2013). Ankle dorsiflexion is also a limitation in this field as the ankles ability to dorsiflex directly influences the knee ability to anteriorly translate beyond the toes, with reduced ankle dorsiflexion reducing the knees anterior translation. If this was recorded, more justification for results would be been provided.

Asymmetries may also have been present however due to only analysing the left knee, no asymmetries can be identified. Analysing data for both knees would have been beneficial and then performing a t-test on these results would have helped to eliminate any discrepancies. Previous research has identified asymmetry to be common in participants performing back squats and even though this elevating heel raises may not increase asymmetries the study that found this only reported this after using one heel elevating board (Sayers et al., 2020). Potential asymmetries would justify two participants seeing reduced PFJS. PFJF and KEM at BOX conditions compared to 3CM (table four, table eight, table 12).

Finally, in calculating knee joint kinetics, muscle co contractions are not accounted for. This is in terms of the knee flexor moment, considering the hamstrings and gastrocnemius contractions are not accounted for. Different methods have been used to calculate patellofemoral joint stress, with inconsistencies being found in results. This information may also provide reasoning for results differing to other studies (Nunes et al., 2018).

## 4.5 Future Research

Future research should explore KEM thresholds and their relationship to injury mechanisms. Furthermore, integrating electromyography (EMG) analysis of quadriceps and hamstrings activation during squat variations could enhance understanding of muscular contributions to joint kinetics, providing a more comprehensive perspective on injury risk factors.

Joint contact area of the patellofemoral joint increases through the angle of knee flexion, with the total area at 90 being twice as much as 30 (Powers, 1998). Quantifying the stress and loads at different joint angles would serve as a recommendation for future therapeutic exercises where limited knee flexion would be required.

Future research needs to investigate the hip moments, stress and force to evaluate whether FMS does result in greater results, additionally, integrating participants ROM measuring and joint angles during the squats. Finally, the investigation of asymmetries needs to be further researched as this serves to be a limiting factor of this study.

## 4.6 Conclusions

Overall, this is the first study to quantify the PFJS, PFJF and KEM in the overhead squat in different stance setups, using different heel heights. This study has provided vital information to lead clinicians in their application of the overhead squat using different heel wedges. The PS technique was found to have statistically significantly greater PFJS and PFJF when FF compared to the FMS technique. However, no significant differences were present in PFJS, PFJF and KEM between the FMS and PS techniques when utilising heel wedges. Heel heights have a statistically significant effects on an individual’s PFJS, PFJF and KEM, with greater heel heights resulting in greater results. Due to this, the FMS deep squat protocol needs to be further considered due to the significant difference between FF and BOX conditions. Clinicians must consider the use of smaller heel wedges of 1CM initially to prevent the excessive stress placed on the patellofemoral joint. Additionally, FMS may need to rethink its protocol to replace FMS technique squats with preferred stance overhead squats as results will be more consistent between participants.

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# Appendix I

[Consent Form.pdf](https://staffsuniversity-my.sharepoint.com/:b:/g/personal/e021878m_student_staffs_ac_uk/Eakuc2_hOaNCmxLHkEcQaRIB_Cqm06krEJoqoZNQhXQOXw?e=BGP9Nk)

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