The Effects of a Heel Wedge on Hip, Pelvis and Trunk Biomechanics During Squatting in Resistance Trained Individuals

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Abstract

Charlton, JM, Hammond, CA, Cochrane, CK, Hatfield, GL, and Hunt, MA. The effects of a heel wedge on hip, pelvis, and trunk biomechanics during squatting in resistance trained individuals. *J Strength Cond Res* 31(6): 1678–1687, 2017—Barbell back squats are a popular exercise for developing lower extremity strength and power. However, this exercise has potential injury risks, particularly to the lumbar spine, pelvis, and hip joint. Previous literature suggests heel wedges as a means of favorably adjusting trunk and pelvis kinematics with the intention of reducing such injury risks. Yet no direct biomechanical research exists to support these recommendations. Therefore, the purpose of this study was to examine the effects of heel wedges compared with barefoot on minimally loaded barbell back squats. Fourteen trained male participants performed a barbell back squat in bare feet or with their feet raised bilaterally with a 2.5-cm wooden block while 3-dimensional kinematics, kinetics, and electromyograms were collected. The heel wedge condition elicited significantly less forward trunk flexion angles at peak knee flexion, and peak external hip joint moments ($p \leq 0.05$) compared with barefoot conditions. However, no significant differences were observed between conditions for trunk and pelvis angle differences at peak knee flexion ($p > 0.05$). Lastly, no peak or root mean square differences in muscle activity were elicited between conditions ($p > 0.05$). Our results lend support for the suggestions provided in literature aimed at using heel wedges as a means of reducing excessive forward trunk flexion. However, the maintenance of a neutral spine, another important safety factor, is not affected by the use of heel wedges. Therefore, heel wedges may be a viable modification for reduction of excessive forward trunk flexion but not for reduction in relative trunk-pelvis flexion during barbell back squats.

Key Words heel wedges, trunk flexion, back squat

Introduction

Barbell back squats are a fundamental strength exercise in fitness, rehabilitation, and athletic training. The biomechanics of this popular movement and its variations have been well researched (4,14,15,22,29,31,38,41). Although previous research has focused mainly on lower extremity biomechanics, few studies have been published regarding the relationship between pelvis and trunk motion during barbell squatting exercises (21,28,39). The motion of the trunk and pelvis may be particularly important as barbell back squatting-type movements inherently require loading through the axial skeleton, and loading of this part of the body has been suggested to be a risk factor for injury (1,23,25,33).

A commonly prescribed technique to mitigate this risk of injury is to maintain the trunk and pelvis as vertically oriented as possible through the duration of the barbell back squat movement (8,10,26,30,35). Any deviation from the natural curvature of the spine and pelvis during the lifting tasks is indicative of a loss of neutrality within the spine, as pelvis and low back motion are closely related (20). Furthermore, it is suggested that while remaining close to neutral overall, the pelvis should remain in slight relative extension to the trunk, assisting in the maintenance of the natural low back curvature (11). The relationship between pelvis and low back motion supports the need for a similar pelvis position relative to the trunk throughout the squat. Doing so may mitigate the chance of errant low back motion, particularly flexion of the pelvis relative to the trunk, and reduce the risk of injury. Accordingly, technical guidelines agree with the need for a neutral spine and pelvis void of any relative motion during squatting movement to reduce injury risk (23,26,35). Therefore, methods to maintain proper trunk and pelvis position are an important consideration when performing these types of movements, especially at the

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bottom of the squatting movement when peak angular displacement of the trunk and pelvis is the greatest.

It has been observed that certain modifications to the barbell back squat technique, such as bar position, stance width, and anterior knee translation, have had varied effects on trunk and pelvis positions, in addition to hip joint moments (14,21,28,38,41). As joint moment differences indicate the change in demand placed on a joint, it is important to understand how any modification to the squatting technique can affect joint moments. Additionally, stance width modifications have been shown to also alter lower extremity muscular activity, particularly of the gluteus maximus (GMax), adductor longus, and gastrocnemius (24,31), which may have an impact on movement efficiency, neuromuscular control, or resultant joint loads. Importantly, anterior knee translation modifications require significant gastrocnemius strength and ankle dorsiflexion range of motion (ROM) (2), whereas stance width modifications may demand increased frontal plane flexibility in the hip joint, possibly precluding their use for some individuals.

An alternative method to the above modifications is to place a heel wedge under the foot, positioning the ankle in slight plantar flexion. Although little direct biomechanical data exist to support the use of heel wedges, some authors have suggested that heel wedges may improve trunk and pelvis positioning during squatting movements by decreasing forward trunk flexion (6,9,12). Instead, there is indirect support for heel wedges from a biomechanical study by Sato et al. (34) that compared weightlifting shoes (a shoe with an equivalent heel wedge design) with running shoes during back squats. Significant reductions in a forward trunk flexion proxy measurement while wearing weightlifting shoes were reported. By contrast, 2 more recent studies reported no change in forward trunk flexion, although greater peak knee flexion and ankle dorsiflexion, and an anterior shift in the center of pressure were observed when comparing weightlifting shoes with running shoes and barefoot conditions (36,40). This indicates that changing heel height may influence some lower extremity kinematic outcomes during squatting.

Importantly, no previous studies examining heel wedges or Olympic weightlifting shoes have reported pelvis kinematics or joint moments, and the available data on muscle activation with raised heels during squatting are limited to a single study that observed no differences in erector spinae, rectus femoris (RF), biceps femoris (BF), tibialis anterior, and gastrocnemius activity between Olympic weightlifting shoes and barefoot conditions (36). Therefore, little is known about the muscle and joint requirements during squatting-type movements with commonly used heel wedges, despite their recommendation in the literature. Without a more thorough understanding of the changes that occur as a result of using heel wedges, recommendations may be inappropriately made.

More research is needed to determine if the use of a heel wedge will elicit changes in trunk and pelvis kinematics that may or may not have an indirect benefit on minimizing injury risk during squatting movements. As such, the purpose of this study was to examine the effects of performing a barbell back squat with and without heel wedges on pelvis and trunk kinematics in addition to hip and thigh muscle activation. Specifically, we investigated sagittal plane trunk, pelvis and lower extremity kinematics, hip joint kinetics, and the muscle activity of RF, BF, gluteus medius (GMed), and GMax. It was hypothesized that squatting with a heel wedge (WHW) would elicit a more vertically oriented trunk and pelvis and a reduced difference in the relative angle between the trunk and pelvis (i.e., more closely aligned segments), representing less deviation from the neutral positions. Consequently, we expected a reduced sagittal plane external hip joint flexion moment. Additionally based on previous research, we hypothesized muscle activity of the RF, BF, and GMed would not vary between conditions. However, based on decreased hip flexion angles, we expected GMax muscle activity to decrease when squatting with heel wedges.

Methods

Experimental Approach to the Problem

A within-subject, repeated measures design was used to determine the effects on pelvis, trunk, and lower limb biomechanics while performing a minimally loaded barbell back squat under 2 conditions: WHW and with no heel wedge (NHW). Lower extremity, pelvis and trunk kinematics, hip joint kinetics, and muscle activation patterns of 4 hip and thigh muscles were measured. The order in which participants performed the 2 conditions was randomized. Participants performed barbell back squats barefoot to a self-selected depth under the instruction to descend to the lowest comfortable position possible according to the National Strength and Conditioning Association technical guidelines (8).

Subjects

Male participants aged between 19 and 35 years were recruited from local weight training facilities to participate in the study. Male participants were recruited to reduce the variability in squat kinematics between sexes (28). Descriptive demographic data are displayed in Table 1 for all participants. Inclusion criteria required participants to have a minimum of 1 year of experience with resistance training including back squat exercises. Interested participants were excluded if they reported a history of musculoskeletal injuries, or neurological or cardiorespiratory conditions that could affect regular participation in a resistance training program. All participants were asked to indicate if they were currently using or had previously used Olympic weightlifting shoes and/or heel wedges during barbell back squatting; 5 of the 20 participants indicated this to be true. Additionally, each participant was asked to select all of the categories of training that best described their resistance training experience, these included: Olympic weightlifting ($n = 4$, 29% of

Table 1. Participant descriptive statistics ($n = 14$).*

Descriptive	Mean (*SD*)	(Min, max)
Training experience (y)	3.9 (1.2)	(2, 5)
Age (y)	23.9 (2.7)	(20.1, 29.1)
Height (m)	1.8 (0.1)	(1.67, 1.94)
Mass (kg)	79.0 (13.7)	(67.4, 121.9)
BMI ($kg \cdot m^{-2}$)	25.2 (2.5)	(22.2, 32.4)
Estimated 1RM (%BW)	132.1 (33.4)	(84.0, 179.6)

*min = minimum; max = maximum; BMI = body mass index; 1RM = 1 repetition maximum; BW = body weight.

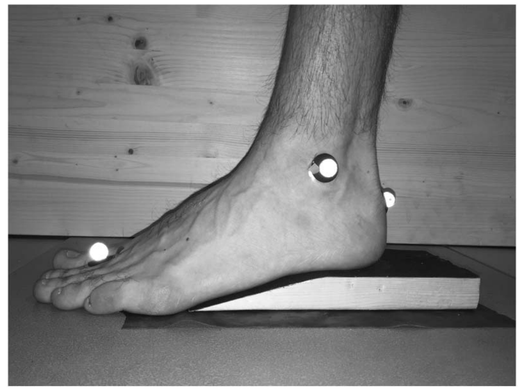

Figure 1. Wedge positioning under a participant's foot.

participants), powerlifting ($n = 3$, 21%), athletic strength and conditioning ($n = 6$, 43%), bodybuilding ($n = 2$, 14%), general fitness ($n = 6$, 43%), and other (0%). The institutional clinical research ethics board approved the study, and all participants provided written informed consent before data collection.

Previous research on pelvis and trunk biomechanics during barbell back squats indicated that differences in sacrum and lumbar peak angles elicited effect sizes between 1.25 and 3.36 (28). For the purposes of this study, a conservative effect size estimate of 1.20 was selected. This effect size, along with a statistical power of 0.8 and an α level of 0.05 was used to determine a minimum requirement of 12 participants (32).

Procedures

All participants attended a single testing session, where demographic and biomechanical data were collected. Demographic and training data included: number of years of training experience, self-reported 1 repetition maximum (1RM) barbell back squat weight, age, height, body mass, and body mass index (Table 1). Participants were asked to refrain from any lower body resistance training or intense physical activity in the 24 hours preceding testing.

Immediately before data collection, participants were provided with 5–10 minutes to perform a brief warm-up consisting of bodyweight movements and dynamic stretching followed by practice of the squatting movement. Preferred stance width and foot position were determined during the first randomly assigned condition. Foot positioning was marked with tape on the floor to ensure consistent positioning across both conditions and all trials. For the WHW condition, a wooden block 2.5-cm high by 9.0-cm wide by 21.5-cm long (Figure 1) was used. Approximately, two-thirds of the block consisted of a sloped edge to support the lateral aspect of the foot. Participants placed their calcaneus on the block and the fifth metatarsal on the floor, providing a 2.5-cm lift between these 2 points, similar to typical weightlifting shoe designs. The NHW condition consisted of barefoot squatting.

A metronome set at 60 $b \cdot min^{-1}$ was used to standardize the timing and rhythm of the squat during all trials. During pilot testing, 60 $b \cdot min^{-1}$ allowed for consistent timing of repeated full-depth squats. The squat movement was performed beginning on beat 1 in the standing position; beat 3 occurred during the deepest position the participant could achieve with comfort; and the participant was instructed to return to the upright standing position on beat 5. Participants performed 5 sets of 3 repetitions for each condition with approximately 20–30 seconds of rest between the sets and 2–5 minutes between conditions. If the participant required longer rest periods, it was provided as needed. The trials were performed with a 20-kg Olympic barbell (Eleiko Sport, Chicago, IL, USA) in a high-bar position as defined by Wretenberg et al. (41). Given that variability in load may increase intersubject variability of performance and increased load may promote fatigue which would have a confounding effect on results, an unloaded bar condition was chosen for all participants. Participants were asked to squat to the deepest comfortable position that they could achieve while keeping both of their heels in contact with the floor. A certified strength and conditioning specialist was present at all testing sessions to ensure safe techniques were being used.

Data Collection

Three-dimensional kinematic data were collected using a 10 camera motion capture system (Raptor-E; Motion Analysis Corporation, Santa Rosa, CA, USA) sampling at 100 Hz. Forty-three passive retroreflective markers were placed bilaterally on boney landmarks to create foot, shank, thigh, pelvis, and trunk segments. Additionally, 4 rigid tracking plates, consisting of 4 nonlinear retroreflective markers, were placed bilaterally on the lateral aspect of the thigh and shank. The segment of the foot was defined by markers on the calcaneus, second metatarsal head and the medial and lateral malleoli. The shank was defined using markers on the medial and lateral malleoli, medial and lateral femoral epicondyles, and the anterior aspect of the tibia. The thigh segment was defined with markers on the medial and lateral femoral epicondyles, anterior aspect of the thigh, right and left greater trochanter, and the anterior superior iliac spine (ASIS) for estimate of the hip joint center. The pelvis was defined using markers placed bilaterally on the ASIS, iliac crest, posterior superior iliac spine, and a single marker on the sacrum. The trunk was defined by placing markers bilaterally on the acromion processes with single markers on the C7 and T10 vertebrae, xiphoid process, and the right inferior aspect of the scapulae. Four additional markers were placed on the medial femoral epicondyles and medial malleoli during static standing calibration trials to determine joint centers and marker orientations but were removed before the squatting trials. Kinematic and kinetic data were synchronized as participants stood on 2 (1 per foot) floor-mounted force platforms (Advanced Mechanical Technology, Inc. Watertown, MA, USA) sampling at 2000 Hz.

Surface electromyograms (EMGs) for 4 muscles were collected at 2000 Hz during each squatting trial and synchronized with kinematic and kinetic data. Wireless bipolar surface electrodes (Delsys, Inc., Natik, MA, USA) were placed on the dominant limb's RF, BF, GMed, and GMax according to the SENIAM.org international guidelines (16). The placement was verified using palpation during a submaximal contraction for each muscle. Before electrode application, the area over the muscle bellies were shaved and prepared with an ethanol wipe to reduce electrical impedance. Maximum voluntary isometric contraction (MVIC)

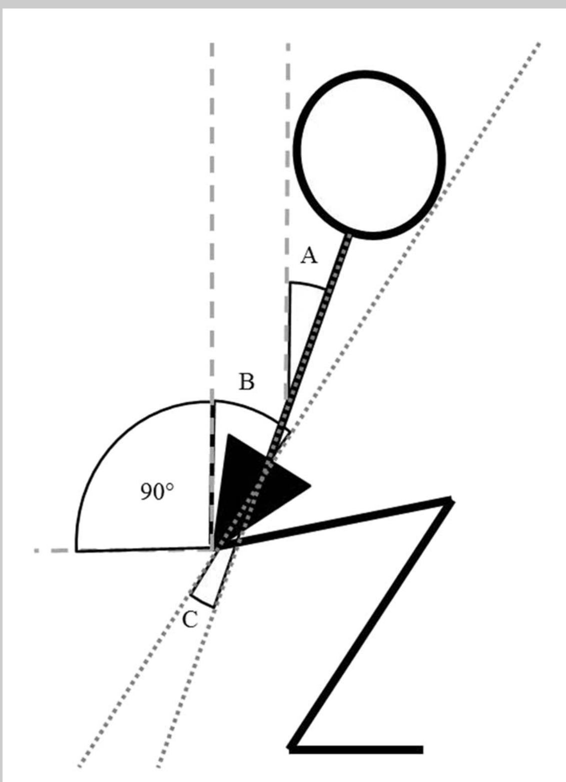

Figure 2. Illustration of the trunk and pelvis outcome measures; trunk forward flexion angle (A), pelvis angle (B), and the difference between these angles (C). More positive angles represent greater forward flexion or anterior pelvic tilt.

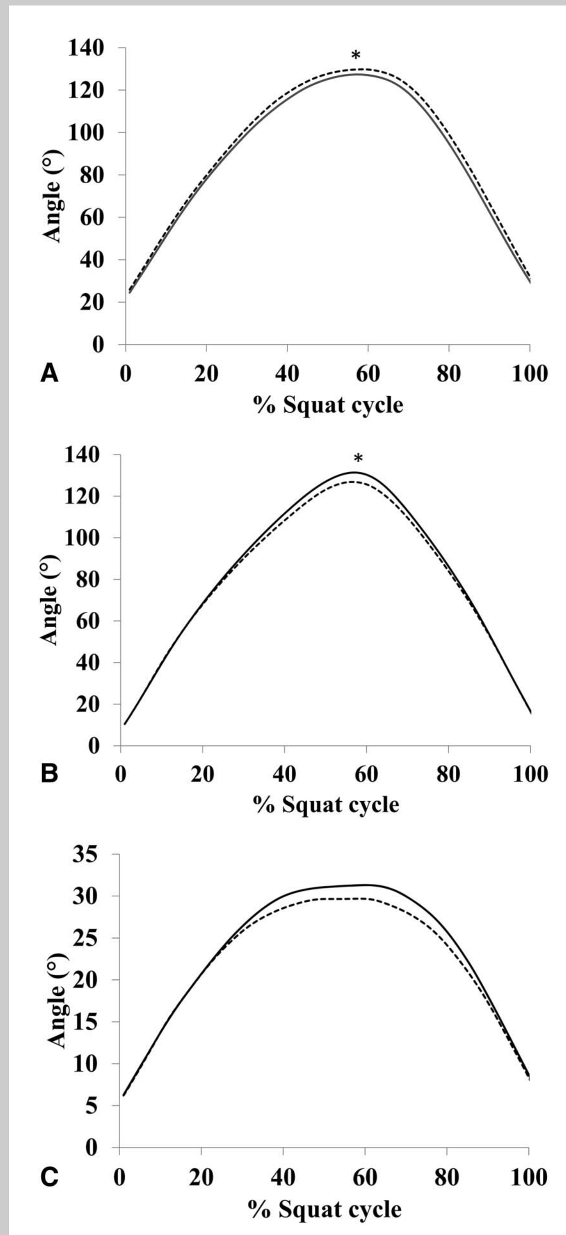

Figure 3. Ensemble average curves for hip (A), knee (B), ankle (C) sagittal plane angles for the with wedges (solid line) and without wedges (dashed line) conditions normalized to 100% of the squat cycle. Larger values represent more flexion at the hip and knee or dorsiflexion at the ankle joint. The asterisks represent a significant difference in peak values between conditions ($p \leq 0.05$).

Table 2. Mean (*SD*) for kinematic main outcome variables for no wedge and with wedge conditions ($N = 14$).*

Variable	No wedge	With wedge	Mean difference (95% confidence interval)
Trunk angle at peak knee flexion (°)	42.80 (6.46)	37.03 (6.38)	5.77 (3.79 to 7.76)†
Pelvis angle at peak knee flexion (°)	35.05 (11.31)	29.27 (10.12)	5.78 (3.33 to 8.23)†
Trunk-pelvis difference angle (°)	7.74 (11.77)	7.74 (10.14)	0.00 (−2.07 to 2.06)

*Positive values indicate forward trunk flexion, anterior pelvic tilt, or flexion of the pelvis relative to the trunk.
†Significant difference observed between conditions ($\alpha = 0.05$).

trials were conducted during 4 different movements to permit amplitude normalization of the resultant squat trial signals. For hip extension, participants were placed in approximately 10° of hip extension and resistance was applied to the posterior knee. During hip abduction, participants were placed with the hip in approximately 25° of abduction and slight extension. Knee extension and flexion MVIC trials were conducted with the participant in high sitting and the knee positioned at 60° using a goniometer. For each MVIC exercise, participants were given a practice trial followed by 2 recorded trials consisting of 3 seconds each, during which strong verbal encouragement was given. Participants were given the opportunity to rest between trials. After the MVIC exercises, a trial was recorded with the participant lying supine. This was used to obtain a baseline, resting level of muscle activity for each muscle.

Data Analysis

External joint moments ($Nm \cdot kg^{-1}$) and joint angles (°) of the dominant limb were calculated via inverse dynamics using Visual3D commercial modeling software (C-Motion, Inc., Rockville, MD, USA). Electromyograms signals were corrected for resting baseline level, converted to microvolts, and full wave rectified and filtered using a second-order Butterworth bandpass filter at 20–500 Hz, followed by a fourth-order Butterworth low pass filter at 25 Hz to create a linear envelope of the signal. All processed EMG data were then normalized to maximum values obtained from the MVIC trials (3,5,19). Specifically, maximum EMG amplitudes were identified using 0.1-second moving average windows of the MVIC trials. The highest EMG amplitude, regardless of the normalization movement, was used for amplitude normalization of the squat trials (17). The EMG signals from each muscle were processed using custom-written MATLAB (The MathWorks, Inc., Natick, MA, USA) programming. To account for alterations in starting and finishing posture during each individual repetition, the squat cycle was defined for analysis purposes as the time period between the instant that knee flexion was greater than 10° on the descent (0%) and less than 10° on the ascent (100%). Data from the second of the 3 repetitions for each of the 5 sets were used in data analysis (15).

Kinematic data examined included peak sagittal hip, knee, and ankle joint angles to quantify the back squat movement performed. Additionally, the following main kinematic outcomes were identified: (a) the absolute peak sagittal trunk angle (TA) and pelvis angle (PA) in relation to the vertical axis of the laboratory (Figures 2A, B) and (b) the trunk-pelvis angle (TPA), calculated as the difference between the TA and PA (Figure 2C). More positive angles represent greater

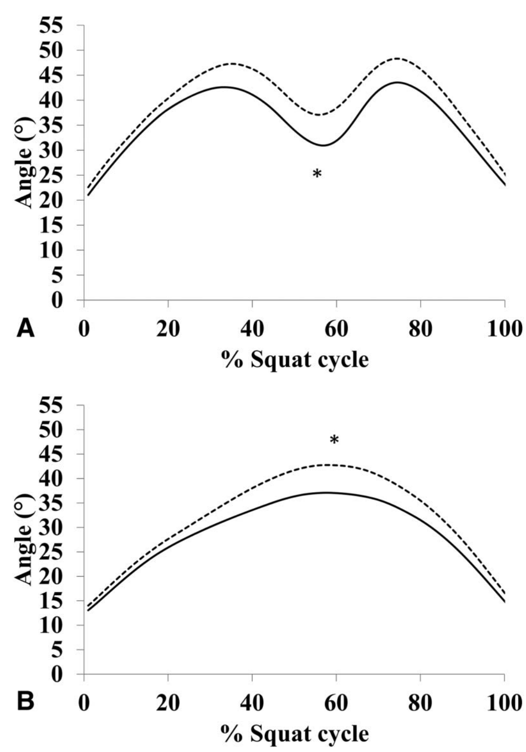

Figure 4. Ensemble average curves for pelvis (A) and trunk (B) sagittal plane angles for squatting with heel wedges (solid line) and without heel wedges (dashed line) normalized to 100% of the squat cycle. Larger values represent more forward trunk flexion or anterior pelvic tilt. The asterisks represent a significant difference in values at peak knee flexion between conditions ($p \leq 0.05$).

trunk or pelvis flexion. Similar to the study by Escamilla et al. (13), kinematic data were calculated at peak knee flexion, a proxy of maximum squat depth achieved by the participant. Additionally, peak external sagittal plane hip joint moments normalized to body weight ($Nm \cdot kg^{-1}$) were calculated where flexion moments are negative. Lastly, peak and root mean square (RMS) muscle activity for each muscle over the duration of the movement was calculated. All outcomes were averaged over the 5 analyzed trials of each condition for each individual.

Statistical Analyses

Mean and *SD* were calculated for all outcome variables. Assumptions of normality and homogeneity of variance were assessed using histograms and skewness statistics. Differences between conditions for each outcome variable were examined using paired *t*-tests. Differences were considered significant if $p \leq 0.05$. All analyses were conducted using the Statistical Package for the Social Sciences (SPSS v. 23; IBM Corp., Armonk, NY, USA).

Results

Fourteen individuals participated. The peak sagittal plane joint angles for the hip, knee, and ankle joints were examined to characterize the movements and ensemble average curves are displayed in Figure 3. Participants exhibited a significant decrease in peak hip flexion angle ($p = 0.002$) when performing the squat with wedges (128.1 [6.2]°) compared with no wedges (130.7 [5.3]°). Conversely, 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]°). No significant differences ($p > 0.05$) were found between squatting with wedges (32.3 [7.1]°) and without wedges (30.6 [6.1]°) for peak ankle dorsiflexion angles.

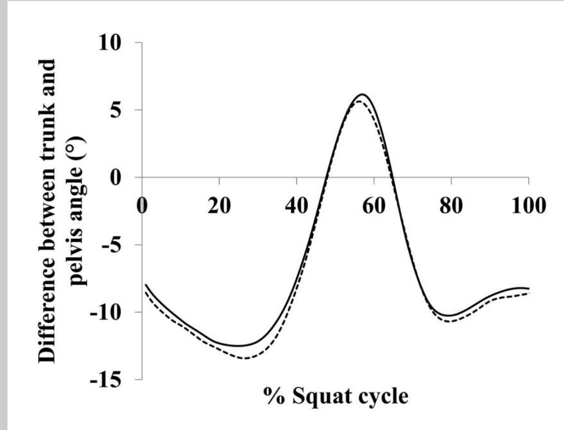

Figure 5. Relative angle between the trunk and pelvis during squatting with heel wedges (solid line) and without heel wedges (dashed line). Positive values indicate flexion of the pelvis relative to the trunk, and a negative value indicates extension of the pelvis relative to the trunk.

Main kinematic outcome variables are displayed in Table 2. The PA at PKF (Figure 4A) was significantly less ($p < 0.001$) when squatting with wedges compared with no wedges, indicating a more vertically oriented pelvis (i.e., less anterior pelvic tilt). Similarly, the TA at PKF (Figure 4B) was significantly decreased (less forward flexion) while squatting with wedges compared with no wedges ($p < 0.001$). However, when the computed difference between the TA and PA at PKF was compared between conditions, no significant difference was found ($p = 0.71$; Figure 5).

The peak external hip flexion moment was found to significantly decrease ($p < 0.001$) while squatting with wedges (−0.94 [0.16] $Nm \cdot kg^{-1}$) compared with squatting

Table 3. Mean (*SD*) for peak and root mean square (RMS) electromyographic activity of the 4 muscles while squatting with wedges and without wedges, represented as a percentage maximum voluntary isometric contraction (%MVIC) ($N = 14$).

Variable (%MVIC)	No wedge		With wedge		Mean difference (95% confidence interval)	
	Peak	RMS	Peak	RMS	Peak	RMS
Rectus femoris	52.97 (32.01)	18.46 (10.58)	50.50 (28.79)	17.68 (10.84)	2.47 (−4.50 to 9.44)	0.78 (−1.05 to 2.60)
Biceps femoris	11.29 (6.95)	4.00 (2.82)	12.39 (7.39)	4.01 (2.77)	−1.10 (−2.60 to 0.40)	−0.013 (−0.42 to 0.40)
Gluteus medius	17.97 (8.44)	4.75 (2.20)	21.11 (9.14)	5.74 (2.76)	−3.15 (−6.80 to 0.47)	−0.99* (−1.89 to −0.093)
Gluteus maximus	16.96 (7.18)	5.11 (2.11)	18.67 (9.05)	5.74 (2.71)	−1.71 (−4.93 to 1.51)	−0.62 (−1.50 to 0.25)

*Significant difference observed between conditions ($\alpha = 0.05$).

without (-1.10 [0.18] $Nm\cdot kg^{-1}$). Additionally, the RF, BF, and GMax muscles elicited no significant between-condition differences ($p > 0.14$) for peak or RMS muscle activity. However, RMS GMed activity was significantly higher ($p = 0.033$) when squatting with wedges (5.74 [2.76] % MVIC) than without (4.75 [2.20] %MVIC). Overall RMS and peak EMG activity of the 4 muscles are presented in Table 3.

Discussion

This study examined the effects of heel wedges on pelvis and trunk kinematics, in addition to hip joint kinetics and muscle activity, during barbell back squats. To our knowledge, no studies have investigated these outcomes while performing barbell squats using heel wedges. In general, participants exhibited similar lower extremity kinematics between the conditions with only slightly less hip flexion, slightly greater knee flexion, and no change in ankle dorsiflexion when squatting with the wedges. However, squatting with wedges resulted in less forward trunk flexion and anterior pelvic tilt, which supports calls in the literature to use heel wedges when squatting as a means to reduce injury risk associated with these 2 parameters during this movement. That said, due to these concurrent changes, the relative angle between the segments remained unchanged. However, it is unclear whether absolute segmental angles (trunk or pelvis with respect to the vertical) or relative TPAs are more important in determining injury risk.

Our participants performed the back squat maneuvre consistent with the expectations of the movement and exhibited kinematic profiles that were similar to previous studies. For example, although peak dorsiflexion angles observed in the present study were similar to research using Olympic weightlifting shoes (36,40) and the knee and hip kinematic curves were similar, the peak joint angles of the present study were larger in magnitude than those reported in related work (36,40). This is likely due to the requirement of the current study's participants to perform the squat to the deepest position possible. The aim of this instruction was to demand maximum lower extremity joint ROM from participants and thus provide the greatest opportunity to observe compensations such as those at the trunk and pelvis segments.

The primary findings of this study indicated that participants exhibited significantly less forward trunk flexion and anterior pelvic tilt at PKF when squatting with the heel wedges. This suggests that participants exhibited a more vertically oriented trunk and pelvis position when squatting WHW, supporting research by Sato et al. (34), who observed less forward trunk flexion while squatting with weightlifting shoes than with running shoes. However, this is contradictory to recent findings, which report no significant differences in trunk kinematics between squatting with weightlifting shoes compared with barefoot or running shoes (36,40). The conflicting results may be due to differences in methodologies including the use of various shoes, squat depth, and foot positions. In an attempt to control for these factors, a single pair of wedges was used for all participants, which mimics the heel-to-toe height difference experienced while using a weightlifting shoe. A more vertically oriented trunk is preferred during squatting as the spine, particularly the lumbar vertebrae, undergoes significantly increased shear forces during excessive forward trunk flexion (33). Accordingly, heel wedges have been recommended for individuals displaying excessive forward trunk flexion (6,9,12) as a means of reducing injury risk to the low back. Our primary finding lends support for this recommendation.

Performing the barbell squat with maintenance of neutral posture throughout the spinal column is another safety factor for this exercise. Potvin et al. (33) postulated that reducing lumbar flexion may be a more important factor than the technique used in reducing trunk lean in lifting task safety. In support, it has been suggested that an anterior pelvic tilt assists in preserving a neutral spinal position (11). The present study measured the difference between the TA and PA at PKF to infer relative spine positions, as the lumbar spine and pelvic kinematics are closely related (20). Our findings indicate the use of heel wedges elicited no significant difference in TPA difference when squatting with or without heel wedges. Thus, if the relative angle between the trunk and pelvis is deemed to be more important than the absolute angles of these segments, this would suggest that the heel wedges do not reduce injury risks associated with performing squats. However, which outcome is more important in injury risk during barbell back squats is unknown. More research in this area is needed. Regardless, wedges should not be recommended as a standalone option to resolve lumbar flexion issues during barbell back squatting.

Under conditions resulting in greater forward trunk flexion, moment forces increase about the hip (14). Consistent with this, when performing the barbell back squat with wedges, participants had significantly smaller external sagittal hip joint moments than squatting without wedges. This was a product of a more vertically oriented trunk and pelvis while squatting with wedges, resulting in a reduced moment arm length between the center of mass and axis of rotation at the hip (14). These reduced joint moments would be beneficial as they would require less internal moments being generated, which would reduce joint reaction forces. This relationship would support the contraindication of squatting with more forward trunk flexion in populations undergoing hip rehabilitation as a means to protect the hip joint.

Furthermore, increased hip muscle activity may be required to balance increased external sagittal hip moments. However, this was not directly observed in the present study. No differences in peak muscle activity were observed between the conditions when examining muscle groups responsible for movements in the sagittal plane. By contrast, we observed a statistically significant difference in GMed activity when squatting with wedges. However, RMS data only increased by 0.99% MVIC, thus questioning the clinical or biomechanical relevance of this difference. Our findings support those of Sinclair et al. (36) who reported no differences in peak or RMS muscle activity for any of the measured lower extremity muscles when comparing a weightlifting shoe and barefoot conditions. However, Caterisano et al. (7) found that GMax provided a greater proportion of the overall muscle activity with increasing squat depth. Our finding of a lack of differences is likely because the resulting knee flexion angle differences (an indicator of squat depth) between conditions in the current study were smaller than those in the Caterisano et al. (7) study.

Additional factors may influence individual response to heel wedges. Notably, individuals with shorter leg lengths and comparatively longer trunk lengths will likely perform squatting motions with a more upright trunk and pelvis despite voluntary modifications. Furthermore, an individual's training history may influence their chosen squatting technique; for example, those who participate in Olympic weightlifting likely aim to squat with a more vertical trunk and pelvis due to the demands of the sport. These individuals likely possess greater dorsiflexion ROM which could reduce the effect of a heel wedge on more proximal joint biomechanics. The current study used a within-subject design and therefore accounted for some of this interindividual variability. However, future research would benefit from quantifying these factors; with enough statistical power a subgroup analysis could illuminate those who would and would not benefit from a heel wedge or Olympic weightlifting shoe. Furthermore, external load, in the form of a loaded barbell, has been reported to influence sagittal plane joint angles and trunk forward flexion to varying degrees, both in studies with and without comparison of a raised heel shoes (18,21,27,40). However, with variable responses to load and inconsistent shoe comparisons, the inclusion of training load conditions was not advisable when aiming to initially determine the effect of raised heels on trunk and pelvis kinematics during squatting. Future research using set resistance levels (either as absolute values or as a percentage of each participant's 1RM) is warranted.

This study is not without limitations. Our sample consisted of individuals who were experienced (more than 2 years) in the barbell back squat exercise. Experienced resistance trained participants were selected to ensure the movement was repeated with consistency; however, these findings may not be generalizable to novice users. Furthermore, our sample was restricted to males only, as previous work has indicated sex differences exist in squatting kinematics, particularly at the lumbosacral joints (28). Thus, it must be noted that these findings may not be generalizable to females. Possible differences between sex and experience should be analyzed in future research to further develop how heel wedges are recommended. Another limitation of this study is the use of a wooden heel wedge, as opposed to Olympic weightlifting shoes. This may reduce the ability to compare findings with research using these shoes; however, the wooden heel wedge allowed the analysis of a single characteristic of weightlifting shoes: the heel-to-toe height difference rather than the structure of the shoe. Issues of familiarity with squatting barefoot and with wedges may be present; however, participants were provided adequate time to practice squatting under both conditions before data collection. As indicated above, an additional limitation was that load was not normalized to each participant's strength level or body weight. Instead, a 20-kg barbell was used, similar to methods in past research (37). Minimizing external load was preferred as it allowed participants to perform the squat with minimal fatigue or deviation in technique across trials. Additionally with minimal external load, participants could perform squats to a maximally comfortable depth requiring the greatest joint excursions possible. However, forward trunk lean has been reported to increase as greater loads are used (40). Importantly, the minimization of independent variables and confounding factors provided the best opportunity to observe results due to the heel wedges alone. Therein, future research can now aim to increase the ecological validity by determining whether the effect of heel wedges remains as typical training loads are used. Lastly, we modelled the spine as a single segment, and thus cannot make conclusions about kinematics of specific vertebral segments. Although our methods do not allow the determination of local vertebral kinematics, they do provide a functional view of technical changes that can occur with the use of heel wedges. The trunk and pelvis positions reported in this study may be more readily observed in real world situations and allow practitioners to intervene before possible injury occurs.

Practical Applications

This research was designed to test the recommendations of heel wedges for individuals who perform barbell squats without a neutral spine or with excessive forward trunk flexion. Heel wedges provide a similar modification to Olympic weightlifting shoes, the likes of which are becoming significantly more popular despite limited research. Furthermore, the use of compound barbell exercises such as the back squat is rising, lending the need to clearly determine the effects that common modifications have on such exercises.

Our findings indicate the use of a heel wedge consisting of a 2.5-cm heel-to-toe height difference may be a viable modification for the barbell squat. The current study lends support to the recommendations that using a wedge will allow the user to squat with a more vertically oriented spine position. Excessive forward trunk flexion is typically contraindicated as it increases shear forces on the lumbar spine and possibly injury potential. Although forward trunk flexion was reduced, no improvement in pelvis and trunk neutrality occurred as a result of using the wedge. Deviations from a neutral spine position while performing lifting activities can increase injury risk, particularly to the lumbar spine. This is an important finding to be aware of when prescribing wedges in practical settings, as a lack of a neutral spine may require interventions other than a heel wedge. Furthermore, a reduction in external torque about the hip joint occurred when squatting WHW, likely a product of a more vertically oriented trunk and pelvis. As a result, wedges could be an appropriate modification for individuals undergoing rehabilitation for hip conditions. The reduction in torque may allow the individual to perform full ROM squats without stressing the hip joint to the same degree as squatting barefoot. Furthermore, no clinically significant muscle activation differences were observed when comparing heel wedges and barefoot conditions for the 4 muscles examined. Lastly, although not the topic of this research, individuals with limited sagittal ankle ROM may benefit from using a heel wedge during squatting exercises where a vertical torso is indicated. However, heel wedges should be further investigated in this population, although frequently recommended, before substantiated claims can be made.

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