The effects of a heel wedge on hip, pelvis and trunk biomechanics during squatting in resistance trained individuals

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Abstract:

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 favourably 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 to 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 2.5cm wooden block while 3D 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 ($p<0.05$) compared to 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 utilizing 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). While 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, 27, 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 in order 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). Further, 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 in order 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 bottom of the squatting movement when peak angular displacement
of the trunk and pelvis is 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, 27, 38, 41). As joint moment
differences indicate the change in demand placed on a joint, it is important to understand how
any modification to 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, 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), while 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. Though 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) to
running shoes during back squats. Significant reductions in a forward trunk flexion proxy measurement while wearing weightlifting shoes were reported. In contrast, two more recent studies reported no change in forward trunk flexion, though greater peak knee flexion and ankle dorsiflexion, and an anterior shift in centre of pressure were observed when comparing weightlifting shoes to 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, biceps femoris, 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 utilized 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 rectus
femoris (RF), biceps femoris (BF), gluteus medius (GMed), and gluteus maximus (GMax). It was hypothesized that squatting with a heel wedge 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 rectus femoris, biceps femoris and gluteus medius would not vary between conditions. However, based on decreased hip flexion angles we expected gluteus maximus muscle activity to decrease, when squatting with heel wedges.

METHODS

Experimental Approach to the Problem

A within-subject, repeated measures design was utilized to determine the effects on pelvis, trunk and lower limb biomechanics while performing a minimally loaded barbell back squat under two conditions: with a heel wedge (WHW) and with no heel wedge (NHW). Lower extremity, pelvis and trunk kinematics, hip joint kinetics, and muscle activation patterns of four hip and thigh muscles were measured. The order in which participants performed the two 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 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 in order to reduce the variability in squat kinematics between sexes (27). Descriptive demographic data are displayed in Table 1 for all participants. Inclusion criteria required participants to have a minimum of one year of experience with resistance training including back squat exercises. Interested participants were excluded if they reported a history of musculoskeletal injuries, neurological or cardiopulmonary conditions that could affect regular participation in a resistance-training program. All participants were asked to indicate if they were currently using or previously used Olympic weightlifting shoes and/or heel wedges during barbell back squats; five 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 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 informed consent prior to data collection.

Table 1 about here

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 (27). 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 $\alpha$ 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 one repetition maximum barbell back squat weight, age, height, body mass and BMI (Table 1). Participants were asked to refrain from any lower body resistance training or intense physical activity in the 24 hours preceding testing.

Immediately prior to 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 2.5cm high by 9.0cm wide by 21.5cm long (Figure 1) was utilized. Approximately 2/3 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 5th metatarsal on the floor, providing a 2.5cm lift between these two points, similar to typical weightlifting shoe designs. The NHW condition consisted of barefoot squatting.
A metronome set at 60 beats per minute (bpm) was used to standardize the timing and rhythm of the squat during all trials. During pilot testing, 60 bpm allowed for consistent timing of repeated full depth squats. The squat movement was performed beginning on beat one in the standing position; beat three 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 five. Participants performed five sets of three repetitions for each condition with approximately 20-30 seconds of rest between 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 20kg Olympic barbell (Eleiko Sport, Chicago, IL) in a high-bar position as defined by Wretenberg, et al. (41). Given that variability in load may increase inter-subject 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 employed.

Data Collection

3-D kinematic data were collected using a ten camera motion capture system (Raptor-E, Motion Analysis Corporation, Santa Rosa, CA) sampling at 100 Hz. Forty-three passive retro-reflective markers were placed bilaterally on boney landmarks to create foot, shank, thigh, pelvis
and trunk segments. Additionally, four rigid tracking plates, consisting of four nonlinear retro-
reflective 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 for estimate of
the hip joint center. The pelvis was defined using markers placed bilaterally on the anterior
superior iliac spine (ASIS), iliac crest, posterior superior iliac spine (PSIS) 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, xyphoid 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 prior to the squatting trials. Kinematic and kinetic data were
synchronized as participants stood on two (one per foot) floor mounted force platforms
(Advanced Mechanical Technology Inc. Watertown, MA) sampling at 2000 Hz.

Surface electromyograms (EMG) for four muscles were collected at 2000 Hz during each
squatting trial, and synchronized with kinematic and kinetic data. Wireless bipolar surface
electrodes (Delsys Inc., Natik, M.A.) were placed on the dominant limb’s RF, BF, GMed and
GMax according to SENIAM.org international guidelines (16). Placement was verified using
palpation during a sub-maximal contraction for each muscle. Prior to 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) trials were conducted during four 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 two recorded trials consisting of three seconds each, during which strong verbal encouragement was given. Participants were given opportunity to rest between trials. Following 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/kg) and joint angles (°) of the dominant limb were calculated via inverse dynamics using Visual3D commercial modeling software (C-Motion Inc, Rockville, MD). EMG signals were corrected for resting baseline level, converted to microvolts, 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 0.1s 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) 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 ten degrees on the descent (0%), and less than ten degrees on the ascent (100%). Data from the second of three repetitions for each of the five 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: 1) the absolute peak sagittal trunk angles (TA) and pelvis (PA) in relation to the vertical axis of the lab (Figures 2a and 2b); and 2) the trunk-pelvis angle (TPA), calculated as the difference between the trunk and pelvis angle (Figure 2c). More positive angles represent greater trunk or pelvis flexion. Similar to Escamilla, et al. (13) kinematic data were calculated at peak knee flexion, a proxy of maximum depth achieved by the participant. Additionally, peak external sagittal plane hip joint moments normalized to body weight (Nm/kg) 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 five analyzed trials of each condition for each individual.
Statistical Analysis

Means and standard deviations 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 $p < 0.05$. All analyses were conducted using the Statistical Package for the Social Sciences (SPSS v. 23; IBM Corp., Armonk NY).

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 to without 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 to without 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.

Main kinematic outcome variables are displayed in Table 2. The pelvis angle at PKF (Figure 4a) was significantly less ($p < 0.001$) when squatting with wedges compared to without
wedges, indicating a more vertically oriented pelvis (i.e. less anterior pelvic tilt). Similarly, the trunk angle at PKF (Figure 4b) was significantly decreased (less forward flexion) while squatting with wedges compared to without wedges ($p < 0.001$). However, when the computed difference between the trunk and pelvis angle 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/kg) compared to squatting without (-1.10 (0.18) Nm/kg). 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) compared to without (4.75 (2.20) %MVIC). Overall RMS and peak EMG activity of the four 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 utilize heel wedges when squatting as a means to reduce injury risk associated with these two 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 trunk-pelvis angles are more important in determining injury risk.

Our participants performed the back squat manoeuvre consistent with the expectations of the movement and exhibited kinematic profiles that were similar to previous studies. For example, though peak dorsiflexion angles observed in the present study were similar to research utilizing 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 with a heel wedge, supporting research by Sato, et al. (34), who observed less forward trunk flexion while squatting with weightlifting shoes compared to running shoes. However, this is contradictory to recent findings, which report no significant differences in trunk kinematics between squatting with weightlifting shoes compared to 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
trunk and pelvis angle at PKF in order 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 compared to squatting
without wedges. This was a product of a more vertically oriented trunk and pelvis during 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.
Further, increased hip muscle activity may be required in order 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. In 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 utilized a within-subject design and therefore accounted for some of this inter-individual 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, 28, 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 utilizing 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 (greater 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 (27). 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 utilizing 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 prior to 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 utilized. 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 APPLICATION

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 like 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 2.5cm 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 with a heel wedge, 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 four muscles examined. Lastly, although not the topic of this research, individuals with limited sagittal ankle ROM may benefit from utilizing 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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**ACKNOWLEDGEMENTS**

No funding was received for this research project.
**Figure legends:**

Figure 1: Wedge positioning under a participant’s foot.

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

Figure 3: Ensemble average curves for hip (Figure 3a), knee (Figure 3b), ankle (Figure 3c) 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. An asterisk represents a significant difference in peak values between conditions (p < 0.05).

Figure 4: Ensemble average curves for pelvis (Figure 4a) and trunk (Figure 4b) 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. An asterisk represents a significant difference in values at peak knee flexion between conditions (p < 0.05).
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. An asterisk represents significant difference between conditions at an alpha level of 0.05.
Table 1: Participant descriptive statistics (n=14).

<table>
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<tr>
<th>Descriptive</th>
<th>Mean (SD)</th>
<th>(Min, Max)</th>
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<tbody>
<tr>
<td>Training Experience (yr)</td>
<td>3.9 (1.2)</td>
<td>(2, 5)</td>
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<tr>
<td>Age (yr)</td>
<td>23.9 (2.7)</td>
<td>(20.1, 29.1)</td>
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<tr>
<td>Height (m)</td>
<td>1.8 (0.1)</td>
<td>(1.67, 1.94)</td>
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<td>Mass (kg)</td>
<td>79.0 (13.7)</td>
<td>(67.4, 121.9)</td>
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<td>BMI (kg/m^2)</td>
<td>25.2 (2.5)</td>
<td>(22.2, 32.4)</td>
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<td>Estimated 1RM (%BW)</td>
<td>132.1 (33.4)</td>
<td>(84.0, 179.6)</td>
</tr>
</tbody>
</table>
Table 2: Mean (SD) for kinematic main outcome variables for no wedge and with wedge conditions (N=14). Positive values indicate forward trunk flexion, anterior pelvic tilt, or flexion of the pelvis relative to the trunk.

<table>
<thead>
<tr>
<th>Variable</th>
<th>No Wedge</th>
<th>With Wedge</th>
<th>Mean Difference (95% CI)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Trunk Angle at Peak Knee Flexion (°)</td>
<td>42.80 (6.46)</td>
<td>37.03 (6.38)</td>
<td>5.77 (3.79, 7.76)*</td>
</tr>
<tr>
<td>Pelvis Angle At Peak Knee Flexion (°)</td>
<td>35.05 (11.31)</td>
<td>29.27 (10.12)</td>
<td>5.78 (3.33, 8.23)*</td>
</tr>
<tr>
<td>Trunk-Pelvis Difference Angle (°)</td>
<td>7.74 (11.77)</td>
<td>7.74 (10.14)</td>
<td>0.00 (-2.07, 2.06)</td>
</tr>
</tbody>
</table>

*Significant difference observed between conditions (α=0.05).
Table 3: Mean (SD) for peak and root mean square electromyographic activity of the four muscles while squatting with wedges and without wedges, represented as a %MVIC (N=14).

<table>
<thead>
<tr>
<th>Variable</th>
<th>No Wedge</th>
<th>With Wedge</th>
<th>Mean Difference (95% CI)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Peak</td>
<td>RMS</td>
<td>Peak</td>
</tr>
<tr>
<td>Rectus femoris</td>
<td>52.97</td>
<td>18.46</td>
<td>50.50</td>
</tr>
<tr>
<td></td>
<td>(32.01)</td>
<td>(10.58)</td>
<td>(28.79)</td>
</tr>
<tr>
<td>Biceps femoris</td>
<td>11.29</td>
<td>4.00</td>
<td>12.39</td>
</tr>
<tr>
<td></td>
<td>(6.95)</td>
<td>(2.82)</td>
<td>(7.39)</td>
</tr>
<tr>
<td>Gluteus medius</td>
<td>17.97</td>
<td>4.75</td>
<td>21.11</td>
</tr>
<tr>
<td></td>
<td>(8.44)</td>
<td>(2.20)</td>
<td>(9.14)</td>
</tr>
<tr>
<td>Gluteus maximus</td>
<td>16.96</td>
<td>5.11</td>
<td>18.67</td>
</tr>
<tr>
<td></td>
<td>(7.18)</td>
<td>(2.11)</td>
<td>(9.05)</td>
</tr>
</tbody>
</table>

*Significant difference observed between conditions (α=0.05).