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*Original Research*

# Rear Leg-derived Moment Contributes to Resistance Against Hip Extension in Bulgarian Split Squats

Hiroshi Arakawa‡[1,3], Hikaru Nakashima#[2], Xuanrui Li*[3], Michiya Tanimoto‡[4]

[1]Department of Physical Education, International Budo University, Chiba, Japan; [2]Okada Denki Sangyo Co., Ltd., Fukushima, Japan; [3]Graduate School of Budo and Sports, International Budo University, Chiba, Japan; [4]Faculty of Health and Sports Science, Juntendo University, Chiba, Japan

*Denotes student investigator, #Denotes Corporate Employee, ‡Denotes established investigator

## Abstract

***International Journal of Exercise Science 18(7): 881-894, 2025.*** https://doi.org/10.70252/NEXQ5666 The Bulgarian split squat (BSS) is a unilateral exercise that emphasizes hip extension more than knee extension, compared to other squat variations. This study aimed to (1) empirically verify the existence of the rear leg-derived moment ($M_{RL}$)—a theoretically plausible but previously untested external resistive hip moment acting against the net hip extension moment ($M_{HE}$) of the front leg—and (2) examine how stance width and forward trunk-leaning angles affect $M_{RL}$ during the BSS. Nine trained male participants performed bodyweight BSS under two stance conditions (wide and narrow) and three trunk-leaning conditions (additional, natural, and reduced forward lean). A motion capture system and force platforms were used to calculate $M_{RL}$, head–arm–trunk segment-derived gravitational moment ($M_{HAT}$), and $M_{HE}$. $M_{RL}$ substantially contributed to the total external resistance acting against the front hip extensors, ranging from 76 to 86 Nm in the wide stance and 49 to 71 Nm in the narrow stance, accounting for 70–97% and 62–98% of the total resistance ($M_{HAT}$ + $M_{RL}$), respectively. In the narrow stance, $M_{RL}$ increased significantly as the trunk became more upright. The combined $M_{RL}$ and $M_{HAT}$ closely matched $M_{HE}$, supporting the validity of the proposed mechanical model. These findings provide the first experimental evidence of $M_{RL}$ as a key resistance factor in the BSS. Moreover, $M_{RL}$ may enable practitioners to increase mechanical loading on the hip extensors while maintaining a more upright trunk posture, offering a potential advantage for strength training programs aiming to target the hip extensors with minimal forward trunk inclination.

Keywords: Bulgarian squat, unilateral squat, glute, low back pain

## Introduction

The Bulgarian split squat (BSS), a unilateral squat variation,[1–10] places more emphasis on hip extension than the conventional back squat (BS).[6,11–17] Indeed, several studies have shown that the BSS is more hip-dominant than knee-dominant compared to conventional BS in terms of muscle activity[6,11,13] and joint kinetics.[14,18] For athletes in different sports, building hip extension strength is crucial for enhancing performance.[19,20] Therefore, understanding the factors that lead to hip dominance during the BSS will help practitioners to design training programs that target specific muscle groups for improving athletic performance.

A key biomechanical factor contributing to the hip dominance of the BSS is its inherent unilateral structure. In squatting, the relative demand on the hip extensors increases as forward trunk leaning is augmented, and the forward displacement of the knees is simultaneously reduced.[21-24] However, maintaining substantial trunk leaning under external loads requires a significant low back moment, which can limit the hip extension moment ($M_{HE}$) exertion, although it does not limit the knee extension moment. In the BSS, $M_{HE}$ is twice that of the BS on a per-leg basis while maintaining the same magnitude of the low back moment.[14] Therefore, the BSS can reduce spinal moment by 50% compared to the BS for the same hip extensor demand level. This unilateral structure allows for overcoming back-strength limitations, enabling greater $M_{HE}$ exertion per leg and making the BSS more hip-dominant than the BS.

Another possible factor contributing to the specific hip dominance in the BSS is the role of the rear leg. From a biomechanical modeling perspective, the resistive hip flexion moment opposing front hip extension during the BSS may be conceptualized as comprising two components, based on the framework proposed in this study: the head-arm-trunk (HAT) segment-derived gravitational moment ($M_{HAT}$) and the rear leg-derived moment ($M_{RL}$). $M_{RL}$ refers to the resistive hip flexion moment generated by the rear leg, which opposes front hip extension during the BSS. Studies on split squats have estimated that both BSS and front lunge, a similar exercise, place considerable load on the rear leg, accounting for approximately 15% and 25–45%, respectively, of the total load distribution (i.e., rear leg GRF divided by total GRF).[25] These substantial rear-leg loads suggest that $M_{RL}$ contributes to the hip-dominant nature of the BSS. Although theoretically plausible and biomechanically reasonable, the mechanical contribution of $M_{RL}$ to the front hip's $M_{HE}$ in the BSS has remained empirically unverified.

Therefore, the first objective of this study was to empirically verify the existence of $M_{RL}$ as a resistance factor for the net hip extension moment ($M_{HE}$) of the front leg during the BSS, and to quantify the extent of its contribution. The second objective of this study was to explore how $M_{RL}$ changes with variations in stance widths and forward trunk-leaning angles. Previous studies have shown that stance width and trunk leaning significantly affect hip and knee dominance in split-squat variations.[26,27] Therefore, understanding their effects on $M_{RL}$ will help practitioners optimize training effectiveness to strengthen the hip extensors by adjusting the stance width and forward trunk-leaning angle. We hypothesized that the contribution of $M_{RL}$ would increase with a wider stance width and a more upright trunk during the BSS.

## Methods

*Participants*

This study was conducted in accordance with the Declaration of Helsinki and approved by the Ethics Committee. Before informed consent was obtained, all procedures and potential risks were thoroughly explained to the participants, both orally and in writing. This research was carried out fully in accordance to the ethical standards of the *International Journal of Exercise Science*.[28]

In this study, the contribution of $M_{RL}$ was evaluated by calculating biomechanical variables during the bodyweight BSS using a motion capture system and force platforms (Figure 1). A within-subject crossover design was used to compare the kinematics and kinetics between two stance widths (wide and narrow) and three different trunk-leaning angles (additional, natural, and reduced) during the bodyweight BSS. Prior to data collection, participants completed a 4-week familiarization period to become accustomed to the BSS under each condition. The experiment was conducted on a single day, during which participants performed the BSS with their right leg forward. This was because most individuals are right-leg dominant, and using the right leg as the front leg helped to minimize potential variability due to limb dominance. Moreover, investigating side-to-side differences was beyond the scope of this study.

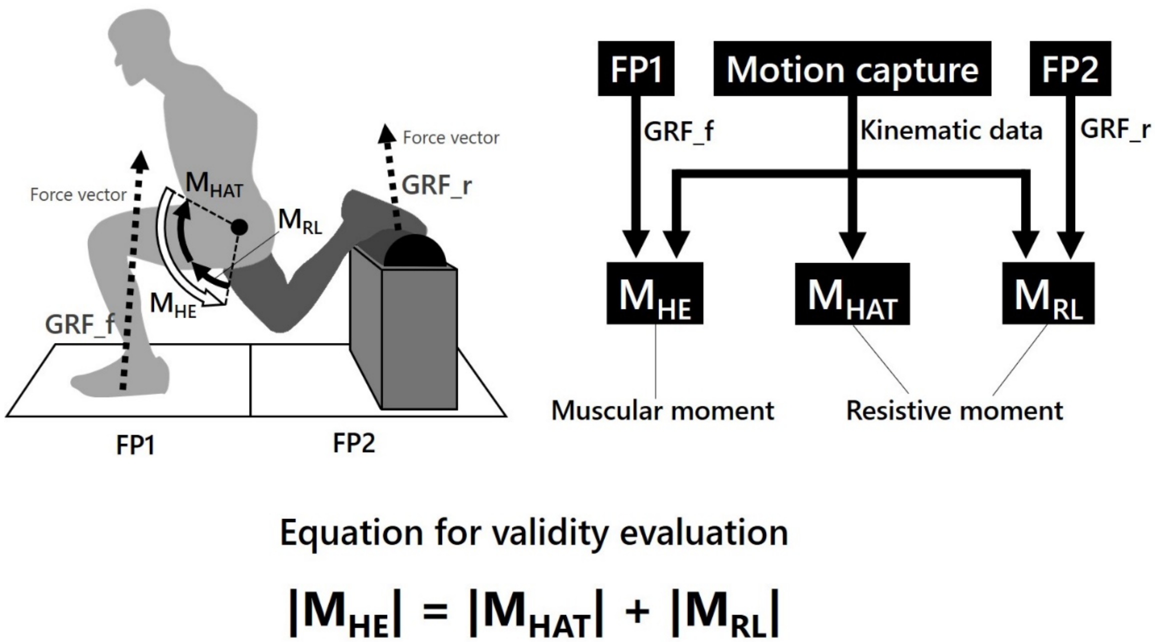


**Figure 1.** Schematic representation of the experimental setup, data acquisition, and biomechanical model used in this study. Ground reaction forces (GRF_f and GRF_r) were measured using force platforms under the front foot (FP1) and rear foot (FP2). Kinematic data were recorded via motion capture. $M_{HE}$ represents the net hip extension moment to the front leg (muscular moment), primarily generated by the front hip extensors. $M_{HAT}$ and $M_{RL}$ represent external resistive hip flexion moments acting against $M_{HE}$, derived from the head–arm–trunk (HAT) segment and the rear leg, respectively. This study specifically focused on the bottom position of the BSS (vinicity if the lowest COM), where $M_{HE}$ typically peaks, regardless of the movement phase (concentric or eccentric). In this study, $M_{HE}$ was conceptualized as the sum of $M_{HAT}$ and $M_{RL}$ at the bottom position of the BSS, representing the total external mechanical demand on the front hip extensors. The validity of the method was assessed by verifying the equation: $|M_{HE}| = |M_{HAT}| + |M_{RL}|$.

Ten male college students participated in this study (age: 19.9 ± 1.7 years; height: 169.7 ± 1.7 cm; weight: 67.9 ± 7.3 kg). The inclusion criteria required participants to have at least 1 year of lower extremity resistance training experience (minimum twice weekly), no orthopedic injuries or pain affecting experimental tasks, and a height of 170 ± 3 cm. This height criterion was implemented to reduce inter-subject variability in the relative bench height used during the BSS and to ensure that the fixed bench height would have a comparable biomechanical effect across participants. According to previous studies on lunge, a movement similar to the BSS, hip joint moments change dramatically depending on trunk inclination and stance width. [26,27] Given the exploratory nature of the first objective, which aimed to empirically examine the existence and magnitude of $M_{RL}$, a power analysis was not conducted. For the second objective, which involved statistical comparisons of $M_{RL}$ between different stance widths and trunk-leaning angles, a pilot experiment estimated an effect size of $d$ = 1.5 for the difference in $M_{HE}$ between stance widths. Based on this estimate, a sample size of 8 participants was calculated to achieve a power of 0.8 with an alpha level of 0.05. To allow for potential dropouts, 10 participants were recruited.

*Protocol*

Each BSS was performed with the rear leg positioned on a customized box (Figures 1 and 2). This box was constructed of wood and designed to fit within a force platform (0.20 m in length, 0.35 m in width, and 0.40 m in height). The top surface of the box featured a cushioned semicylindrical pole (R = 0.07 m). Approximately 20 kg of cement was placed inside the box to ensure stability.

The participants performed the bodyweight BSS with two stance widths (wide and narrow stances) and three trunk-leaning angles (additional, natural, and reduced forward leaning), as shown in Figure 2. The stance width was defined as the horizontal distance between the front leg toe and a vertical line passing through the apex of the semicylindrical pole fixed on the customized box (Figure 2). Wide and narrow stances were set at 130% and 100% of the individual's leg length (distance from the great trochanter to the lateral malleolus of the front leg), respectively. Based on a preliminary pilot study, the trunk angle at the bottom position in the natural forward-lean condition exhibited limited interindividual variability. This natural trunk angle was not determined by participants' arbitrary preferences but was established through instructor-guided practice during a 4-week familiarization period, in which participants were taught a trunk posture that is cnventionally used in strength training practice. Considering this, we set the additional and reduced trunk-leaning conditions by adding and subtracting 10°, respectively, from each participant's baseline angle (Figure 2). This ensured that the manipulation was sufficiently large to produce distinct postural changes while remaining within a natural and biomechanically reasonable range.

Before data collection, participants underwent a 4-week training period focused on bodyweight BSS. Each session comprised six sets of five repetitions of the bodyweight BSS for each condition. The natural forward-lean condition was performed at the beginning of each session for each stance and served as a reference for setting the other trunk angles. The additional and reduced forward-lean conditions were then performed randomly. Participants received feedback on their trunk angles after each set through image analysis using a smartphone application. To ensure accurate body modeling, the participants wore tight-fitting clothing. A thin plastic string was placed parallel to the forehead axis at knee height to control the forward displacement of the knee (Figure 2). The amount of forward knee displacement was standardized based on the natural forward-lean condition within each stance. The squat depth was adjusted such that the lower surface of the thigh was parallel to the floor in the sagittal plane. Both hands were placed on the iliac crests during the squat. A minimum of 5 min of rest was provided between sets. These controls were maintained throughout the familiarization period.

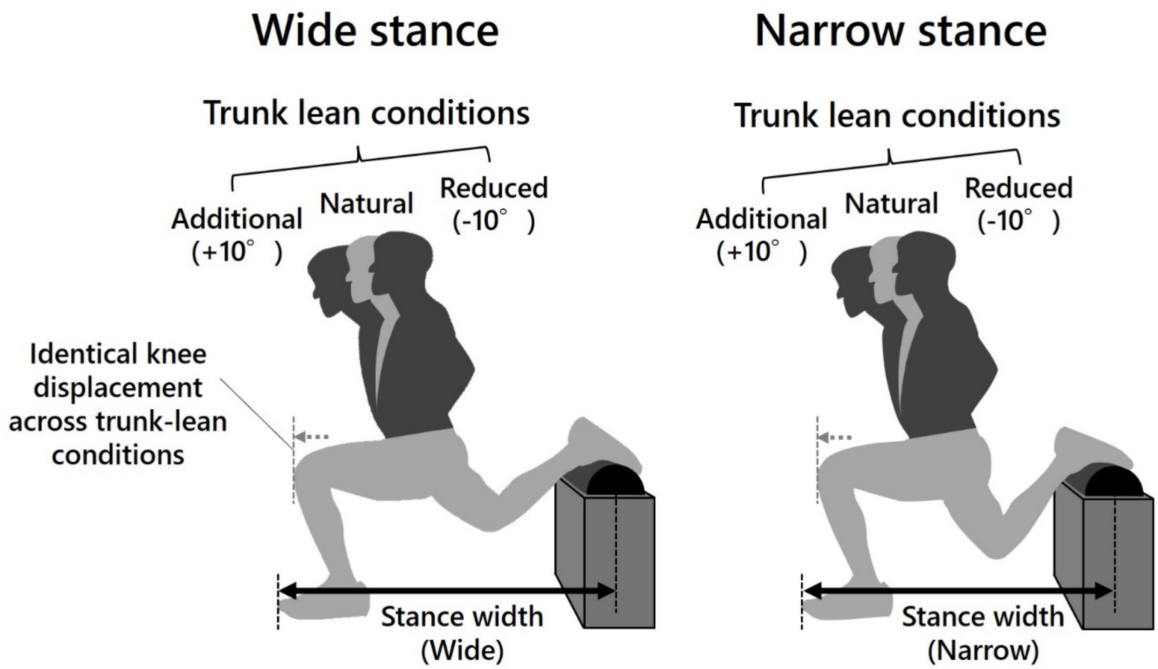


**Figure 2.** Stance width definitions and trunk-lean conditions during the BSS. The stance width was defined as the horizontal distance between the front leg toe and a vertical line passing through the apex of the semicylindrical pole fixed on the customized box. Wide and narrow stances were set at 130% and 100% of the individual's leg length (distance from the great trochanter to the lateral malleolus of the front leg), respectively. Trunk-lean conditions include additional lean (+10°), natural lean, and reduced lean (−10°). Knee displacement was controlled to remain consistent across all trunk-lean conditions within each stance width.

During the experiment, the participants wore tight-fitting clothing on the lower body and remained bare-chested. Following a warm-up session consisting of a few practice sets of BSS and stretching, 22 reflective markers were attached to the participants' bodies. Each participant was instructed to perform the tasks with their dominant leg forward and the nondominant leg placed behind on a customized box positioned on the force platform. The squatting posture, number of repetitions, and rest intervals between sets were identical to those during the familiarization period. The cadence of the squatting movement was controlled by a metronome, with participants instructed to descend in 1.5 s and ascend in 1.5 s.

The two stance conditions (Figure 2) were performed in a random order. The natural forward-lean condition was performed first in each stance condition, and trunk angles were immediately calculated as they were during the familiarization period. The trunk angles for the other two conditions (i.e., additional and reduced forward lean) were individually determined. Knee displacement was controlled using a plastic string, as done during the familiarization period. The trial was repeated if the trunk angle or knee displacement deviated by more than ±3° or ±3 cm, respectively, from the target values. The trunk angle used for the biomechanical analysis was re-evaluated after the experiment using data from the motion capture system.

Movements during the exercises were recorded and analyzed using a three-dimensional motion capture system (VICON MX, Oxford, United Kingdom) with a sampling frequency of 100 Hz and 10 cameras. A total of 22 reflective markers (diameter: 25.4 mm) were affixed to the participant's body to create a link segment model. Markers were placed bilaterally on the auricular points, anterior–superior iliac spines (ASIS), posterior–superior iliac spines (PSIS), greater trochanters (GT), medial and lateral femoral condyles, and medial and lateral malleoli and unilaterally on the upper end of the sternum (US) and the seventh cervical vertebra (C7).

Participants wore shoes with markers affixed to their first and fifth metacarpals. The ASIS markers were positioned 1 cm laterally from the palpated location to ensure camera visibility during deep hip flexion in the BSS. Any potential sagittal errors due to ASIS marker displacement were expected to be mitigated using larger markers.

The vertical and horizontal components of the ground reaction force (GRF) and the position of the center of pressure (COP) were measured separately for the front and rear legs (Figure 1) using two force platforms (length: 0.6 m, width: 0.4 m, Kistler, Winterthur, Switzerland). GRF and COP data were recorded at a frequency of 1,000 Hz and synchronized with the motion capture position data.

The raw coordinates of the markers were smoothed using a fourth-order Butterworth digital low-pass filter with a cut-off frequency of 5 Hz. A two-dimensional link segment model consisting of seven segments (head–arm–trunk [HAT], right foot, right shank, right thigh, left foot, left shank, and left thigh) was constructed in the sagittal plane from the kinematic data. The positions of the left and right hip joint centers were determined based on previously reported methods.[29] The knee and ankle joint centers were defined as the midpoints between the corresponding lateral and medial markers. The kinematics of the HAT segment were evaluated using a line segment connecting the midpoint between the US and C7 markers to the midpoint between the right and left GT. The hip and knee joint angles in the anatomical position were defined as 180°, with hip and knee joint extensions defined as positive.

The mass, center of mass (COM) position, and inertial parameters of each segment were determined based on a previous study[30] and were personalized for each subject's anthropometry. The hip and knee extension moments of the front leg ($M_{HE}$ and $M_{KE}$) were calculated by sequentially solving the equations of motion from the foot to the upper segments using data from the front force platform. $M_{RL}$, which acts as resistance on the front leg, was calculated by solving the equations of motion from the rear foot to the more proximal segments using data from the rear force platform. Additionally, $M_{HAT}$, which also contributes as resistance on the front hip extensors, was calculated by solving the equations of motion for the HAT segment based on the kinematic data. In this study, $M_{HE}$ was conceptualized as the sum of two independent resistive components acting in the hip flexion direction: $M_{HAT}$, generated by the gravitational moment of the HAT segment, and $M_{RL}$, generated by the rear leg. These two components do not directly interact with each other but jointly represent the total external resistance that must be overcome by the front hip extensors. The COP of the GRF acting on the rear foot was assumed to coincide with the apex of the semicylindrical pole in the sagittal plane. Notably, the hip joint moments were calculated based on the coordinate system of the pelvic segment rather than that of the trunk segment.

According to the fundamental mechanical principle, the following relationship was assumed to hold for the three calculated moments:

$$|M_{HE}| = |M_{HAT}| + |M_{RL}|$$

This equation was used to validate the accuracy of the joint moment calculations in this study.

*Statistical Analysis*

For statistical analyses, the trial with the trunk-leaning angle closest to the mean of five repetitions of the natural forward-lean condition for each stance width was selected. In the additional and reduced conditions, the trials closest to ± 10° relative to the natural condition were selected. This method ensured that the selected trial was the most representative of each condition. All trials were conducted under strict control, and those deviating more than ± 3° or ± 3 cm from the target were discarded and repeated, minimizing intra-individual variability.

One-way analysis of variance (ANOVA) was performed to analyze the differences in each variable across the trunk-leaning conditions, followed by Tukey's post-hoc test for pairwise comparisons. Before performing ANOVA, the normality of data was confirmed using the Shapiro-Wilk test. Additionally, the homogeneity of variances across conditions was verified using Bartlett's test. For stance condition comparisons, a two-tailed paired t-test was used to analyze the difference between wide and narrow stances under the natural trunk-lean condition. Statistical analysis was not performed for the additional and reduced trunk-lean conditions because the trials in the wide-stance and narrow-stance conditions did not necessarily correspond to each other. The significance level was set at $P < 0.05$. All statistical analyses were conducted using R software (R version 3.3.3; R Core Team, Vienna, Austria). Cohen's $d$ and partial eta squared ($\eta^2$) were calculated as indices of effect size. The magnitude of the effect size was interpreted as small ($d = 0.2$, $\eta^2 = 0.01$), medium ($d = 0.5$, $\eta^2 = 0.06$), and large ($d = 0.8$, $\eta^2 = 0.14$).

## Results

Data were successfully obtained from 9 of 10 participants, excluding one participant who dropped out during the familiarization period. In this study, the biomechanical variables were analyzed at the bottom position of the BSS, defined as the vicinity of the lowest COM position, where the hip extension moment ($M_{HE}$) of the front leg is typically maximized. According to our model, $M_{HE}$ was conceptualized as the sum of two resistive components acting in the hip flexion direction: $M_{HAT}$ and $M_{RL}$.

**Table 1.** Kinematic and kinetic results of each condition

| Variables | | | Wide stance: Trunk lean Additional | Natural | Reduced | $P$ | $\eta^2$ | Narrow stance: Trunk lean Additional | Natural | Reduced | $P$ | $\eta^2$ | Wide vs. Narrow (Natural trunk lean) $P$ | $d$ |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| COM height (m) | Minimum | | 0.51 ± 0.04 | 0.50 ± 0.04 | 0.50 ± 0.05 | - | - | 0.55 ± 0.04 | 0.52 ± 0.03 | 0.52 ± 0.02 | - | - | - | - |
| | Maximum | | 0.89 ± 0.05 | 0.90 ± 0.04 | 0.89 ± 0.06 | - | - | 0.91 ± 0.03 | 0.91 ± 0.03 | 0.90 ± 0.03 | - | - | - | - |
| | Displacement | | 0.38 ± 0.06 | 0.39 ± 0.04 | 0.38 ± 0.07 | - | - | 0.35 ± 0.06 | 0.39 ± 0.04 | 0.38 ± 0.04 | - | - | - | - |
| Hip joint angle (deg) | Maximum ext. | | 143.6 ± 11.4 | 149.1 ± 5.1 | 148.8 ± 9.3 | - | - | 156.2 ± 7.4 | 162.5 ± 6.5 | 162.3 ± 5.4 | - | - | - | - |
| | Maximum flex. | | 87.2 ± 11.5 | 89.1 ± 8.9 | 92.4 ± 10.2 | - | - | 93.6 ± 11.3 | 96.5 ± 8.2 | 99.8 ± 8.4 | - | - | - | - |
| | Bottom of the COM | | 91.6 ± 15.0 | 89.2 ± 8.8 | 92.5 ± 10.2 | - | - | 93.7 ± 11.3 | 96.1 ± 8.4 | 100.0 ± 8.7 | - | - | - | - |
| Trunk forward lean angle (deg) | | | 66.9 ± 3.8 | 76.6 ± 4.0 | 86.9 ± 4.5 | 0.000 | 0.99 | 66.5 ± 5.1 | 77.0 ± 5.5 | 86.8 ± 4.7 | 0.000 | 0.99 | 0.837 | 0.08 |
| Shank forward lean angle (deg) | | | 30.2 ± 7.2 | 28.6 ± 6.7 | 29.0 ± 5.2 | 0.256 | 0.03 | 32.7 ± 5.5 | 33.3 ± 3.6 | 32.6 ± 5.1 | 0.591 | 0.14 | 0.017 | 0.87 |
| GRF at the bottom of the COM (N) | Front | Horizontal | 1 ± 28 | -6 ± 23 | -8 ± 21 | - | - | -5 ± 16 | -7 ± 16 | -14 ± 18 | - | - | - | - |
| | | Vertical | 420 ± 57 | 391 ± 54 | 370 ± 31 | - | - | 427 ± 49 | 390 ± 35 | 364 ± 71 | - | - | - | - |
| | Rear | Horizontal | 14 ± 28 | 21 ± 26 | 23 ± 16 | - | - | 20 ± 24 | 30 ± 20 | 34 ± 23 | - | - | - | - |
| | | Vertical | 175 ± 40 | 187 ± 42 | 196 ± 42 | - | - | 169 ± 42 | 211 ± 49 | 234 ± 30 | - | - | - | - |
| Joint moment at the bottom of the COM (Nm) | Knee | $M_{KE}$ | 27 ± 13 | 25 ± 11 | 26 ± 9 | 0.840 | 0.02 | 40 ± 11 | 39 ± 9 | 38 ± 13 | 0.756 | 0.01 | 0.003 | 1.43 |
| | Hip | $M_{HE}$ | 102 ± 17** | 94 ± 16 | 87 ± 8 | 0.001 | 0.55 | 78 ± 15** | 73 ± 10 | 63 ± 13* | 0.000 | 0.59 | 0.000 | 1.44 |
| | | $M_{HAT}$ + $M_{RL}$ | 109 ± 16** | 99 ± 16 | 90 ± 16 | 0.004 | 0.42 | 83 ± 19 | 86 ± 13 | 74 ± 23 | 0.169 | 0.18 | 0.024 | 0.74 |
| | | $M_{HAT}$ | 33 ± 5** | 20 ± 5 | 4 ± 7** | 0.000 | 0.96 | 33 ± 6** | 19 ± 8 | 3 ± 7** | 0.000 | 0.96 | 0.740 | 0.14 |
| | | $M_{RL}$ | 76 ± 19 | 79 ± 17 | 86 ± 20 | 0.151 | 0.21 | 49 ± 20** | 67 ± 18 | 71 ± 24 | 0.006 | 0.73 | 0.116 | 0.54 |
| | | %$M_{RL}$ | 70% | 81% | 97% | - | - | 62% | 79% | 98% | - | - | - | - |

$M_{KE}$, extension moment of the knee joint; $M_{HE}$, extension moment of the hip joint; $M_{HAT}$, the HAT (head–arm–trunk) segment-derived gravitational moment; $M_{RL}$, rear leg-derived moment; %$M_{RL}$, percentage contribution of $M_{RL}$ vs. $M_{HAT}$ + $M_{RL}$. Values indicate mean ± standard deviation for each variable. $P$ values are the results of the one-way analysis of variance. *Significant difference compared to the natural trunk-lean condition at $P < 0.05$. **Significant difference compared to the natural trunk-lean condition at $P < 0.01$.

As shown in Table 1, the standard deviation of the trunk forward lean angle in the natural condition was relatively small—4.0° in the wide stance and 5.5° in the narrow stance—indicating low inter-individual variability. In addition, COM height (minimum, maximum, and displacement), shank forward lean angle, and $M_{KE}$ were nearly identical across the three trunk-leaning conditions. Similarly, COM height and trunk forward lean angle were comparable between the wide and narrow stance conditions. These results suggest that neither trunk-leaning angle nor stance width had a substantial effect on BSS kinematics.

Figure 3 presents the overall averages of the time course for $M_{HE}$, $M_{RL}$, and $M_{HAT}$ synchronized with the time of the lowest COM. The stacked areas of $M_{RL}$ and $M_{HAT}$ generally matched the patterns of $M_{HE}$ (solid line), except for the natural and reduced trunk-lean conditions in the narrow stance (Figure 3). In these two cases, the summed curves of $M_{RL}$ and $M_{HAT}$ exceeded that of $M_{HE}$ around the time of the lowest COM (i.e., 0 s).

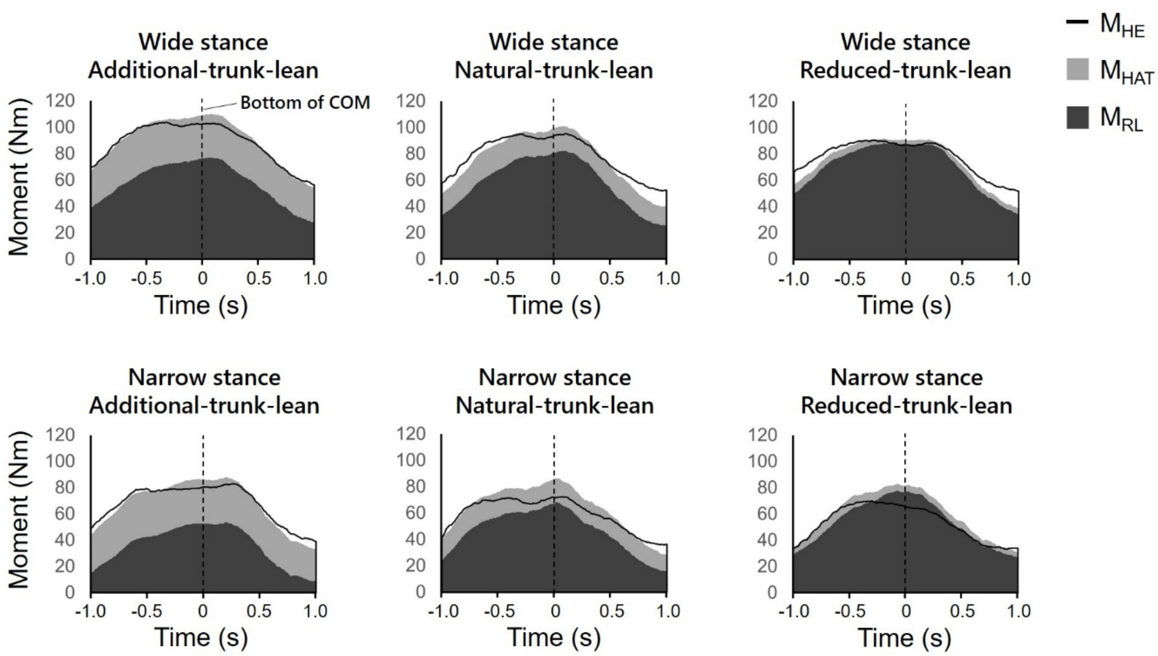


**Figure 3.** Time-series data of the hip extension moment ($M_{HE}$), head–arm–trunk (HAT) segment-derived gravitational moment ($M_{HAT}$), and rear leg-derived moment ($M_{RL}$) during the BSS under different stance width and trunk-lean conditions. $M_{HAT}$ and $M_{RL}$ are presented as stacked time-series areas, with their sum compared to $M_{HE}$ (solid line) to validate the method. The dashed vertical line (t = 0) indicates the bottom position of the center of mass (COM). Moment values are expressed in Nm.

The mean $M_{HE}$ at the lowest COM ranged from 102 to 87 Nm in the wide stance and from 78 to 63 Nm in the narrow stance (Table 1). A significant difference in $M_{HE}$ at the lowest COM was observed between the wide and narrow stances (wide: 94 ± 16 Nm vs. narrow: 73 ± 10 Nm in the natural trunk-lean condition, $P < 0.01$). Our data revealed a significant effect of trunk-leaning conditions on $M_{HE}$ at the lowest COM in both the wide ($P < 0.01$) and narrow ($P < 0.01$) stance conditions. Post-hoc multiple comparisons indicated that $M_{HE}$ at the lowest COM was significantly greater in the additional trunk-lean condition than in the natural trunk-lean condition in both stances ($P < 0.01$ for both stances) and significantly lower in the reduced trunk-lean condition than in the natural trunk-lean condition in the narrow stance ($P < 0.05$).

$M_{RL}$ at the lowest COM ranged from 76 to 86 Nm in the wide stance and from 49 to 71 Nm in the narrow stance. Its contribution to the total resistance (i.e., $M_{RL} + M_{HAT}$) against front hip extensors ranged from 70% to 97% in the wide stance and from 62% to 98% in the narrow stance. No significant difference in $M_{RL}$ at the lowest COM was observed between the wide-stance and narrow-stance conditions within the natural trunk-lean condition (wide: 79 ± 17 Nm vs. narrow: 67 ± 18 Nm, $P = 0.116$). Our data revealed a significant main effect of trunk-leaning conditions on the $M_{RL}$ at the lowest COM only in the narrow stance ($P < 0.01$), with no significant main effect observed in the wide stance ($P = 0.151$). Post-hoc multiple comparisons for the narrow-stance condition indicated that the $M_{RL}$ at the lowest COM was significantly lower in the additional trunk-lean condition than in the natural and reduced trunk-lean conditions ($P < 0.01$).

## Discussion

This study had two primary objectives. The first objective was to empirically verify the existence of the rear leg-derived moment ($M_{RL}$), a theoretically plausible but empirically untested external resistive hip flexion moment that acts against the front hip extensors during the bodyweight BSS. The second was to examine how $M_{RL}$ changes with variations in forward trunk leaning and stance width. Notably, our analysis focused on the bottom position of the BSS, where mechanical demand of the hip extensors is maximized.

Our findings provide the first experimental evidence that $M_{RL}$ constitutes a substantial portion of the total external resistance acting on the front hip extensors during the BSS. Specifically, $M_{RL}$ accounted for 70–97% of the total resistance in the wide stance and 62–98% in the narrow stance. This novel result empirically validates the mechanical model that conceptualizes $M_{HE}$ as the sum of $M_{RL}$ and $M_{HAT}$. Moreover, $M_{RL}$ was significantly greater when the trunk was more upright, particularly in the narrow stance—highlighting its role in enabling practitioners to maintain an upright trunk posture while still loading the hip extensors.

Importantly, this dual function of $M_{RL}$—augmenting hip extensor loading while promoting trunk verticality—offers a unique biomechanical advantage. In traditional squats, increasing hip extensor loading typically necessitates greater forward trunk lean, which elevates lumbar spine stress. In contrast, our findings suggest that the BSS, through $M_{RL}$, can decouple this trade-off, allowing for high hip loading without proportional increases in spinal demand. This characteristic makes the BSS an effective exercise for athletes and rehabilitation settings where spinal load minimization is essential.

It should be noted, however, that this study examined the BSS under bodyweight conditions. Given that $M_{RL}$ is influenced primarily by rear-leg posture and passive tissue tension—rather than by loading magnitude—it may not increase proportionally with added external loads. While $M_{HE}$ would rise with barbell or dumbbell use, $M_{RL}$ may remain relatively constant, thereby reducing its relative contribution. Nevertheless, the absolute $M_{RL}$ observed in this study (49–86 Nm) indicates that even under loaded conditions, $M_{RL}$ likely provides meaningful resistance. For example, Arakawa et al. (2023) reported an $M_{HE}$ of ~220 Nm under 10RM load[14], suggesting that %$M_{RL}$ may still exceed 25% in loaded conditions—a non-negligible portion of total resistance.

The posture-dependent nature of $M_{RL}$ was further supported by the finding that a more extended rear-hip position (seen in upright trunk posture) led to greater $M_{RL}$ in the narrow stance (Table 1). This is likely due to the contribution of the passive hip moment of the rear leg. Prior data have shown that the passive hip flexion moment increases in a hip extended posture, especially beyond the anatomical position.[31] In the present study, the average peak hip extension angle of the rear leg was 183.8 ± 10.6° in the additional trunk-lean condition and 194.1 ± 6.0° in the reduced trunk-lean condition, indicating that the hip extended by approximately 10° more in the upright-trunk condition. This difference in the hip extension angle likely caused the observed difference in the passive hip flexion moment between the conditions, which in turn contributed to the observed difference in $M_{RL}$. Based on this inference, individuals with lower hip flexibility may benefit more from this passive moment, leading to the hip-dominant nature of the BSS. Clarifying this influence would be valuable for developing training prescriptions that are tailored to individual flexibility profiles; future investigations are warranted.

While split-squat variations include exercises such as single-leg squats, step-ups, front and back lunges, and walking lunges, the BSS is distinct in that the rear leg is elevated on a bench, placing the hip joint in a more extended position. This positioning likely increases the contribution of the passive hip moment, resulting in a greater $M_{RL}$ and foreleg $M_{HE}$ in the BSS than in other split-squat variations. Indeed, previous studies using electromyography (EMG) analysis[6,32] reported greater EMG activity in the hip extensors of the front leg during the BSS than during other split-squat variations. However, it should be noted that EMG activity reflects neural activation levels, not muscle force production itself. Nevertheless, these results indirectly support the notion that the BSS places a higher mechanical demand on the hip extensors than other split-squat variations.

This study has some limitations. First, although the trunk lean and stance width varied, the forward displacement of the knee was kept constant. It is important to consider the effect of knee displacement during squatting, as it affects the relative contribution of the hip extensors.[24] Second, the participants performed the bodyweight BSS without external loads. When dumbbells or barbells are used, both the absolute value of $M_{RL}$ and its percentage contribution to the total resistance are likely to differ. Third, in some cases, the sum of $M_{RL}$ and $M_{HAT}$ exceeded $M_{HE}$, suggesting a potential overestimation of $M_{RL}$. This discrepancy was particularly notable under narrow-stance conditions with natural and reduced trunk leaning. When the stance width is narrow, the rear knee is flexed more, causing the actual COP on the customized box to shift forward from the semicylindrical pole. However, in the present study, the COP was assumed to be fixed at the apex of the semicylindrical pole. This assumption might have led to an overestimation of the moment arm length around the hip when calculating $M_{RL}$. Fourth, the study had a small sample size, and all participants were male college students. Thus, the findings of this study may not be applicable to individuals with different profiles, such as sex, body size, body composition, physical strength, or flexibility. Furthermore, the rear-leg support height was fixed at 40 cm to reflect common practice. Although participants varied in stature, a

±6% difference in relative height (e.g., due to 2–3 cm variation for a leg length of ~80 cm) would alter the rear hip angle by only ~2°, suggesting minimal mechanical impact on $M_{RL}$. Therefore, despite individual differences, the findings are likely to be applicable across a typical range of anthropometrics in practice, though caution is advised for extreme cases.

In conclusion, this study provides the first empirical evidence that the rear leg in the Bulgarian split squat (BSS) generates a mechanically significant resistive moment ($M_{RL}$) acting on the front hip extensors. This resistive moment contributes substantially to the hip-dominant nature of the BSS and facilitates the maintenance of a more upright trunk posture during the exercise. While this study did not directly assess spinal loading, these findings suggest that the BSS may offer a unique advantage in strength training: the ability to increase mechanical demand on the hip extensors without necessarily requiring greater forward trunk lean. This characteristic may have practical implications for designing lower-body resistance training programs, particularly in athletes or individuals seeking to target the hip extensors while potentially minimizing trunk inclination — a factor that could help reduce stress on the lower back.

## References

1. Appleby BB, Newton RU, Cormack SJ. Kinetics and kinematics of the squat and step-up in well-trained rugby players. *J Strength Cond Res.* 2019;33(Suppl 1):S36–S44. https://doi.org/10.1519/JSC.0000000000003055

2. Appleby BB, Cormack SJ, Newton RU. Specificity and transfer of lower-body strength: influence of bilateral or unilateral lower-body resistance training. *J Strength Cond Res.* 2019;33(2):318–326. https://doi.org/10.1519/JSC.0000000000002923

3. Comfort P, Jones PA, Smith LC, Herrington L. Joint kinetics and kinematics during common lower limb rehabilitation exercises. *J Athl Train.* 2015;50(10):1011–1018. https://doi.org/10.4085/1062-6050-50.9.05

4. Eliassen W, Saeterbakken AH, van den Tillaar R. Comparison of bilateral and unilateral squat exercises on barbell kinematics and muscle activation. *Int J Sports Phys Ther.* 2018;13(5):871–881. https://doi.org/10.26603/ijspt20180871

5. Krause DA, Elliott JJ, Fraboni DF, et al. Electromyography of the hip and thigh muscles during two variations of the lunge exercise: a cross-sectional study. *Int J Sports Phys Ther.* 2018;13(2):137–142. https://doi.org/10.26603/ijspt20180137

6. DeForest BA, Cantrell GS, Schilling BK. Muscle activity in single- vs. double-leg squats. *Int J Exerc Sci.* 2014;7(4):302–310. https://doi.org/10.70252/MXVZ7653

7. Núñez FJ, Santalla A, Carrasquila I, et al. The effects of unilateral and bilateral eccentric overload training on hypertrophy, muscle power and COD performance, and its determinants, in team sport players. *PLoS One.* 2018;13(3):e0193841. https://doi.org/10.1371/journal.pone.0193841

8. Schellenberg F, Taylor WR, Lorenzetti S. Towards evidence-based strength training: a comparison of muscle forces during deadlifts, good mornings and split squats. *BMC Sports Sci Med Rehabil.* 2017;9(1):1–10. https://doi.org/10.1186/s13102-017-0077-x

9. Secomb JL, Tran TT, Lundgren L, Farley ORL, Sheppard JM. Single-leg squat progressions. *Strength Cond J.* 2014;36(5):68–71. https://doi.org/10.1519/SSC.0000000000000090

10. Speirs DE, Bennett MA, Finn CV, Turner AP. Unilateral vs. bilateral squat training for strength, sprints, and agility in academy rugby players. *J Strength Cond Res.* 2016;30(2):386–392. https://doi.org/10.1519/JSC.0000000000001096

11. Andersen V, Fimland MS, Brennset O, et al. Muscle activation and strength in squat and Bulgarian squat on stable and unstable surfaces. *Int J Sports Med.* 2014;35(14):1196–1202. https://doi.org/10.1055/s-0034-1382016

12. Knoll MG, Davidge M, Wraspir C, Korak JA. Comparisons of single leg squat variations on lower limb muscle activation and center of pressure alterations. *Int J Exerc Sci.* 2019;12(1):950–959. https://doi.org/10.70252/AVXZ2000

13. McCurdy K, O'Kelley E, Kutz M, Langford G, Ernest J, Torres M. Comparison of lower extremity EMG between the 2-leg squat and modified single-leg squat in female athletes. *J Sport Rehabil.* 2010;19(1):57–70. https://doi.org/10.1123/jsr.19.1.57

14. Arakawa H, Mori M, Tanimoto M. Greater hip moments in rear-foot-elevated split squats than in conventional back squats with the same relative intensity of loads. *J Strength Cond Res.* 2023;37(5):1009–1016. https://doi.org/10.1519/JSC.0000000000004351

15. Mackey ER, Riemann BL. Biomechanical differences between the Bulgarian split-squat and back squat. *Int J Exerc Sci.* 2021;14(1):533–543. https://doi.org/10.70252/CIYT8956

16. Schütz P, List R, Zemp R, Schellenberg F, Taylor WR, Lorenzetti S. Joint angles of the ankle, knee, and hip and loading conditions during split squats. *J Appl Biomech.* 2014;30(3):373–380. https://doi.org/10.1123/jab.2013-0175

17. Aguilera-Castells J, Buscà B, Morales J, et al. Muscle activity of Bulgarian squat: effects of additional vibration, suspension and unstable surface. *PLoS One.* 2019;14(8):e0221710. https://doi.org/10.1371/journal.pone.0221710

18. Meyer B, Dapena J. A comparison of hip extension torques in conventional and split squat exercises. *Kinesiology.* 2005;IV:32–42.

19. González-García J, Morencos E, Balsalobre-Fernández C, Cuéllar-Rayo Á, Romero-Moraleda B. Effects of 7-week hip thrust versus back squat resistance training on performance in adolescent female soccer players. *Sports (Basel).* 2019;7(4):80. https://doi.org/10.3390/sports7040080

20. Takahashi K, Kamibayashi K, Wakahara T. Muscle size of individual hip extensors in sprint runners: its relation to spatiotemporal variables and sprint velocity during maximal velocity sprinting. *PLoS One.* 2021;16(4):e0249670. https://doi.org/10.1371/journal.pone.0249670

21. Fry AC, Smith JC, Schilling BK. Effect of knee position on hip and knee torques during the barbell squat. *J Strength Cond Res.* 2003;17(4):629–633. https://doi.org/10.1519/1533-4287(2003)017<0629:EOKPOH>2.0.CO;2

22. Straub RK, Barrack AJ, Cannon J, Powers CM. Trunk inclination during squatting is a better predictor of the knee-extensor moment than shank inclination. *J Sport Rehabil.* 2021;30(6):899–904. https://doi.org/10.1123/jsr.2020-0397

23. Straub RK, Powers CM. A biomechanical review of the squat exercise: implications for clinical practice. *Int J Sports Phys Ther.* 2024;19(4):490–501. https://doi.org/10.26603/001c.94600

24. Barrack AJ, Straub RK, Cannon J, Powers CM. The relative orientation of the trunk and tibia can be used to estimate the demands on the hip and knee extensors during the barbell back squat. *Int J Sports Sci Coach.* 2021;16(4):1004–1010. https://doi.org/10.1177/1747954121997957

25. Hefzy MS, al Khazim M, Harrison L. Co-activation of the hamstrings and quadriceps during the lunge exercise. *Biomed Sci Instrum.* 1997;33:360–365.

26. Song Q, Ma M, Liu H, Wei X, Chen X. Effects of step lengths on biomechanical characteristics of lower extremity during split squat movement. *Front Bioeng Biotechnol.* 2023;11:1277493. https://doi.org/10.3389/fbioe.2023.1277493

27. Farrokhi S, Pollard CD, Souza RB, Chen YJ, Reischl S, Powers CM. Trunk position influences the kinematics, kinetics, and muscle activity of the lead lower extremity during the forward lunge exercise. *J Orthop Sports Phys Ther.* 2008;38(7):403–409. https://doi.org/10.2519/jospt.2008.2634

28. Navalta JW, Stone WJ, Lyons TS. Ethical issues relating to scientific discovery in exercise science. *Int J Exerc Sci.* 2019;2(1):1–8. https://doi.org/10.70252/LVDF8860

29. Kurabayashi J, Mochimaru M, Kouchi M. Validation of the estimation methods for the hip joint center. *J Soc Biomech.* 2003;27(1):29–36. https://doi.org/10.3951/sobim.27.29

30. Winter DA. *Biomechanics and Motor Control of Human Movement.* John Wiley & Sons; 2009. https://doi.org/10.1002/9780470549148

31. Silder A, Whittington B, Heiderscheit B, Thelen DG. Identification of passive elastic joint moment-angle relationships in the lower extremity. *J Biomech.* 2007;40(12):2628–2635. https://doi.org/10.1016/j.jbiomech.2006.12.017

32. Mausehund L, Skard AE, Krosshaug T. Muscle activation in unilateral barbell exercises: implications for strength training and rehabilitation. *J Strength Cond Res.* 2019;33(Suppl 1):S85–S94. https://doi.org/10.1519/JSC.0000000000002617

Corresponding author: Hiroshi Arakawa, PhD; arakawa@budo-u.ac.jp