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# Impact of gender, posture and phase on back and lower extremity moments and muscle excitation during lifting task: a cross-sectional study

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*Purpose*: The aim of this work was to investigate and compare back and lower extremity joint moments and muscle excitation during stoop and squat postures by incorporating gender-based differences and analyzing lifting phases. *Methods*: 18 healthy adults (9 males and 9 females; age: 24.44 ± 4.96 years, body mass: 66.00 ± 12.10 kg, height: 170.11 ± 9.20 cm, lean body mass: 48.46 ± 7.66 kg) lifted an object 30% of their lean body mass using squat and stoop postures. Marker-based motion capture, force plate, and surface electromyography were synchronously used to acquire joint moments and muscle excitation. A 3-way mixed model analysis was performed to determine the effect of gender, posture, and phase on internal joint moments and muscle excitation of the lower back and extremities. *Results*: Significant differences were observed in the interaction of lifting posture and phase on lower extremity moments and excitation of rectus femoris and medial gastrocnemius. Individual effects of posture were significant for peak internal joint moments of the lower extremities only. Anterior lower extremity muscles showed significantly increased excitation during squat, whereas medial gastrocnemius was higher in stoop. Joint moments and muscle excitations were all higher during the lifting than the bending phase. Gender differences were found only in the peak lumbosacral sagittal plane moment and rectus femoris muscle excitation. *Conclusions*: The study identified significant variations in the joint moments and muscle excitation in lifting, influenced by gender, posture, and phase, highlighting its complex nature. Overall interactions were lacking, however individual effects were evident, necessitating larger future studies.

*Key words*: *lifting biomechanics, sex differences, squat, stoop, joint loading, EMG*

## 1. Introduction

Lifting is an essential task in day-to-day life. Despite its simplicity, the underlying mechanisms of lifting movements are complex and multi-factorial. Prior lifting studies [4], [14] have focused on the lower back and less on the lower extremity joints. However, the lower back and lower extremities work together as a kinematic chain. Therefore, it is essential to consider not only the lumbar spine, but also the entire lower extremity. The two common lifting postures are the squat and stoop. During squat lifting, the knee joint alone absorbs the power, while the hip and ankle contribute mostly to the support moment. Meanwhile, in stoop lifting, both the knee and ankle joints absorb the power, while the hip, together with the ankle, contributed mostly to the support moment [19].

Joint moments are typically used to indirectly measure the loading in the joints [6]. The tensile force applied to the skin, joint capsule and ligaments, or the muscular pull within the joint, can result in internal moments to a body segment. Previous studies [19], [33] have focused solely on the sagittal plane moments during lifting. However, recognizing lifting as a comprehensive activity, it is imperative to extend the analysis to

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encompass total joint moments, spanning all planes of motion of the lower back and lower extremity regions. For instance, greater external knee abduction (internal knee adduction) moment is linked with anterior cruciate ligament (ACL) injuries and patellofemoral pain [26]. On the other hand, greater external knee adduction (internal knee abduction) has been associated with an increased force distribution in the medial knee compartment, which is a common site of cartilage degeneration and osteoarthritis [17], [24]. Combined with an elevated internal knee extensor moment, this adds a substantial 22% variance to the medial knee compartment loading prediction, wherein greater internal knee extensor moment was associated with greater joint compressive forces [24]. Given that muscles can modify the loading environment of the joint they are acting on, optimal contraction of the lower extremity muscles can promote efficient and safe lifting.

Aside from these, most studies [4], [19] have only shed light on the main lifting execution and less on the differences in the mechanics from bending to reach the object to lift it. However, this is important as forward bending is just as biomechanically and physiologically demanding as lifting. Additionally, research has shown that individuals perform approximately 58 minutes of work-related forward bending per day [21]. Thus, making the bending phase of lifting as equally important as the lifting itself.

Furthermore, due to anatomical differences, gender also influences the biomechanics of movement. Females are, on average, shorter and lighter than men and generate 30 to 60% less force, with females reportedly having by 40–73% lower lifting strength and lower maximum leg muscle strength than males [9], [11], [16], [23]. Given these factors, females tend to approach their maximum muscle capacity more closely when lifting the same load as men. Even when the load is adjusted to account for the sex differences in maximum capacity, minor disparities still exist [22], [28], [31]. Females are also known to have a broader pelvis, greater Q-angle, and lesser strength of the knee extensor muscles [25]. These characteristics further add up to their risk of lower extremity pain and injury. Hence, female gender is a significant risk factor for most lower extremity musculoskeletal conditions [5], [20].

There is still a continuous debate on the proper lifting posture due to the task complexity and individual anatomical differences [35]. Recent evidences [10], [29] have shown that squatting does not protect the back, and increased lumbar flexion during lifting, a characteristic of stoop, was not associated with low back pain. In addition, most people still prefer stoop lifting due to its lower energy consumption and because it is subjectively reported to be easier and has better balance control [4]. Therefore, this study aimed to investigate and compare the lower back, lower extremity joint moments, muscle excitation during stoop and squat postures, incorporating gender-based differences and a comprehensive analysis of the phases of the lifting task. Based on the above literature, the researchers hypothesized that there will be differences present in the internal joint moments of the lumbosacral, hip, knee and ankle joints, as well as on the muscle excitation of the rectus femoris, biceps femoris, tibialis anterior, and medial gastrocnemius during different lifting postures, and phases, and between genders. The results of the study could contribute to the development of tailored interventions and ergonomic recommendations, ultimately promoting safer and more efficient lifting practices in various settings.

## 2. Materials and methods

A cross-sectional study design was conducted among healthy adults to investigate and compare the back and lower extremity joint moments and muscle excitation during stoop and squat postures, incorporating gender differences and phase analysis. Recruitment and experimental data collection were conducted from February to March 2023. All experimental procedures and data analysis was done at the Biomechanics and Movement Rehabilitation Laboratory of the Department of Public Health Science of Korea University. This study followed the Declaration of Helsinki and was approved by the Korea University Bioethics Review Committee (IRB-2022-0355).

### 2.1. Participants

Based on previous literature [1], [13] and an *a priori* power analysis calculation using G*Power version 3.1.9.7 (Heinrich-Heine-Universität, Düsseldorf, Germany) with an *F*-test: repeated-measures within-between interaction, a statistical power of 0.8, an effect size of 0.5, $\alpha$ level of 0.05, a minimum of 16 participants were needed.

Participants were recruited within the campus and surrounding areas of Korea University through posters and word of mouth. For an individual to be successfully enrolled and participate in the study, he or she must be 19 to 40 years old, generally healthy and can perform lifting without any limitations. Anyone with lifting restrictions or neuromusculoskeletal conditions as well as expert or professional lifters were excluded.

Written informed consents were obtained prior to participation. In the end, a total of 18 healthy adults (9 males and 9 females, age: 24.44 ± 4.96 years, body mass: 66.00 ± 12.10 kg, height: 170.11 ± 9.20 cm, lean body mass (LBM): 48.46 ± 7.66 kg) participated and successfully completed the study.

## 2.2. Procedures

Each participant lifted a load equal to 30% of their computed LBM [18], using both squat and stoop lifting postures. The sequence for the lifting posture was randomized for each participant, with half starting with squat first and the other starting with stoop first. For the squat lift, participants were asked to fully bend their knees to grab the load, whereas the stoop lift was performed while keeping the knees as straight as possible with greater flexion on the hips. All participants lifted the load from floor-to-waist level. Participants started by standing with feet shoulder-width apart and arms on the side, followed by bending to reach the load on the floor (bending phase) using either squat or stoop posture. The participant then grabs the load with both hands and lifts it (lifting phase) until waist level. The end of one lifting cycle is identified once the knees were fully extended. Prior to the actual data acquisition, participants were oriented regarding the method of execution of each lifting posture and the flow of the experiment. Participants were asked to practice the lifting movement, without any presence of weights, to ensure proper performance prior to the actual trial. Three repetitions were completed per lifting posture for the actual data acquisition, and 1-minute rest periods were provided in between.

The motion and muscle data of all the participants were collected by synchronizing a 6-Vicon camera motion capture system (Vicon Motion Systems, Ltd., Oxford, UK) sampling at 100 Hz with a laboratory-embedded AMTI force plate sampling at 1000 Hz (Advanced Medical Technology Inc. Watertown, MA, USA) and wireless surface electromyography (EMG) sampling at 2000 Hz (Delsys Inc., Natick, MA, USA). Synchronization was done using a trigger, with the EMG acting as the master, together with its text-to-speech cueing to guide the participant and the Vicon with the connected force plate as slaves. A complete workflow of the study, from participant recruitment to data analysis, as well as the laboratory set-up were visualized in Figs. 1 and 2, respectively.

The acquisition and initial processing of the marker and force plate data were done using Vicon Nexus version 1.8.5 (Vicon Motion Systems, Ltd., Oxford, UK). A total of 39 markers were attached on specific anatomical landmarks of the participant, based on the Plug-In Gait lower-limb model with extra markers on the upper body, medial knees and medial malleoli. Prior to experimental data acquisition, a static A-pose trial was captured to define the marker trajectories (Fig. 3).

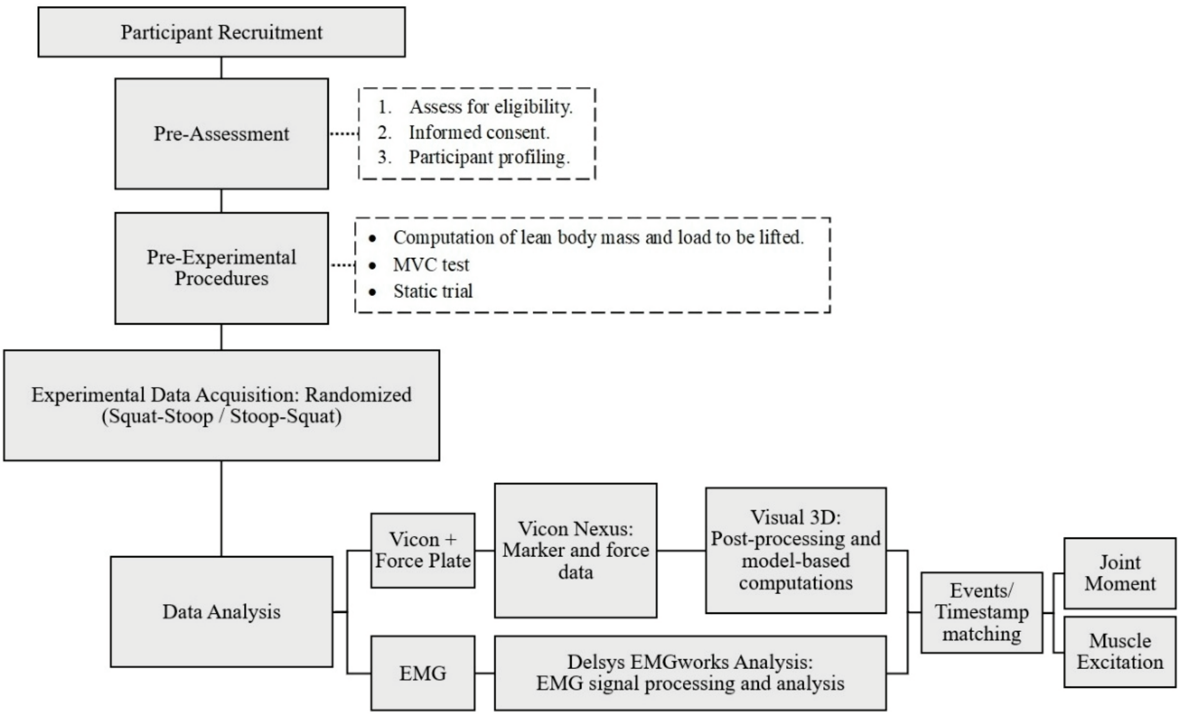



Fig. 1. Study workflow

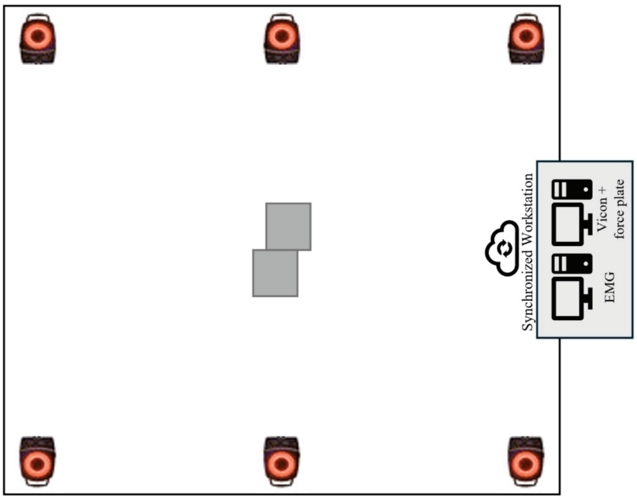
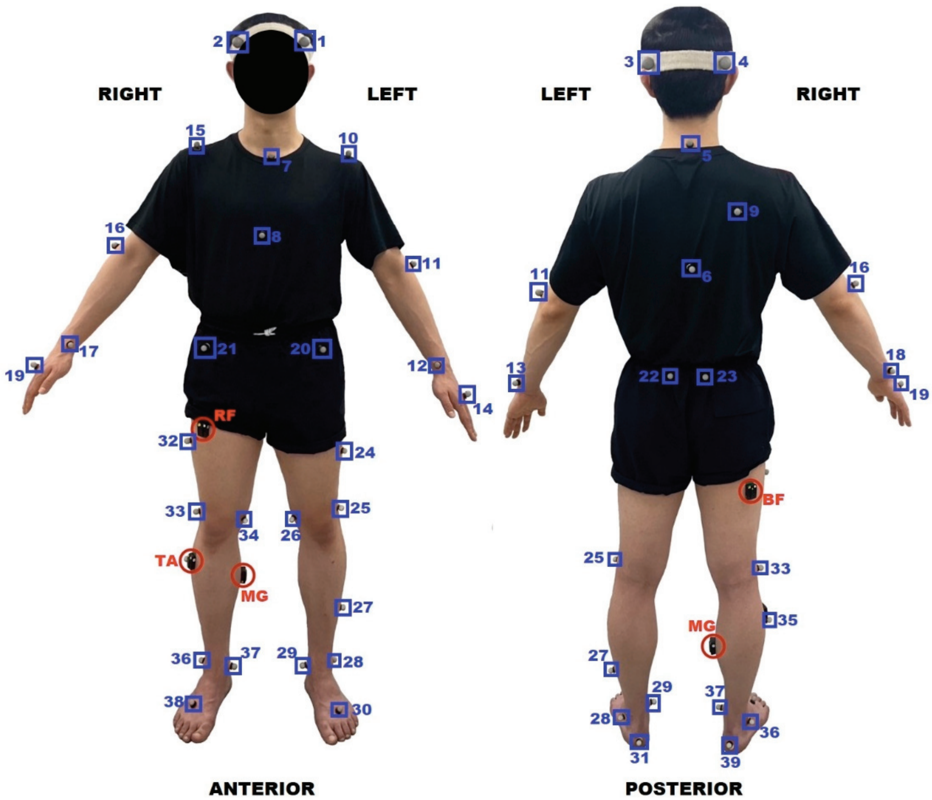



Fig. 2. Laboratory set-up

Delsys Trigno Wireless Surface EMG (Delsys, Inc., Natick, MA, USA) hardware and EMGworks Acquisition (Delsys, Inc., Natick, MA, USA) software were used to obtain and record the EMG signals during lifting. Since this is a symmetric form of lifting, and to prevent inconvenience to the participants, sensors were attached on the right lower extremity only, specifically in the rectus femoris (RF), biceps femoris (BF), tibialis anterior (TA), and medial gastrocnemius (MG). The EMG sensors were positioned following both the manufacturer's guidelines and the recommendations of SENIAM (Surface Electromyography for the Non-Invasive Assessment of Muscles) [32]. To normalize the muscle activity among participants, an isolated manual muscle test [3] of each muscle was performed prior to the experiment. Each muscle test




| Marker Name | | Marker Name | | Marker Name | | EMG Sensors | |
|---|---|---|---|---|---|---|---|
| 1 | Left front head | 14 | Left finger | 27 | Left tibia | RF | Rectus Femoris |
| 2 | Right front head | 15 | Right shoulder | 28 | Left ankle | BF | Biceps Femoris |
| 3 | Left back head | 16 | Right elbow | 29 | Left medial ankle | TA | Tibialis Anterior |
| 4 | Right back head | 17 | Right wrist marker A (thumb side) | 30 | Left toe | MG | Medial Gastrocnemius |
| 5 | 7th cervical vertebrae | 18 | Right wrist marker B (little finger side) | 31 | Left heel | | |
| 6 | 10th thoracic vertebrae | 19 | Right shoulder | 32 | Right thigh | | |
| 7 | Clavicle | 20 | Left ASIS | 33 | Right knee | | |
| 8 | Sternum | 21 | Right ASIS | 34 | Right medial knee | | |
| 9 | Right back | 22 | Left PSIS | 35 | Right tibia | | |
| 10 | Left shoulder | 23 | Right PSIS | 36 | Right ankle | | |
| 11 | Left elbow | 24 | Left thigh | 37 | Right medial ankle | | |
| 12 | Left wrist marker A (thumb side) | 25 | Left knee | 38 | Right toe | | |
| 13 | Left wrist marker B (little finger side) | 26 | Left medial knee | 39 | Right heel | | |

Fig. 3. Placement of Vicon markers and EMG sensors on a model in an A-pose position

was repeated thrice and the maximal contraction was acquired and used [15].

## 2.3. Data processing and analysis

C3D files from Vicon Nexus were imported and were further processed in Visual 3D v.2022.08 (C-Motion, Inc., Boyds, MD, USA). A static trial of each participant was used for calibration and modeling. Data were filtered using a fourth-order Butterworth low-pass (6 Hz) filter. Model-based computations using inverse dynamics were done. Events based on joint angle and force data were created to establish the start and end of each phase. The bending phase starts from full knee extension until the knees are fully bent (for squat posture) or until the hips are maximally flexed (for stoop posture). Meanwhile, the lifting phase starts when the participant has fully grabbed and stabilized the load from the specific lifting posture until the load is lifted with the knees in full extension. Sagittal, frontal and transverse planes were represented by the Cardan sequence (*X*–*Y*–*Z*), which was used to describe joint moments [30]. These moments were then normalized to the subject's body mass in kilograms [Nm/kg]. The peak internal joint moment among the three repetitions per phase and posture were acquired and analyzed.

EMG signals were processed and analyzed using Delsys EMGworks Analysis (Delsys, Inc., Natick, MA, USA) software. A fourth-order Butterworth low-pass filter with a cutoff frequency of 10 Hz was used to rectify and filter the raw EMG signals. Lower extremity muscle excitation was computed using amplitude analysis through root mean square (RMS) calculation and was expressed as a percentage of the maximum voluntary contraction (% of MVC). The EMG data were matched with the events' timestamps from the motion data, and the mean EMG amplitude within each phase and posture was used for analysis. The average of three repetitions per participant per posture and phase was calculated.

## 2.4. Outcomes

Two main measurable outcomes studied were joint moments and muscle excitation. To explore joint loading differences between sexes, postures and phases, joint moments in the lumbosacral, hip, knee and ankle joints were analysed, as these were commonly used as indirect measures of joint loading [6]. In this study, all moment data presented were internal joint moments, as Visual 3D computes the internal moment, which is equal to and opposite of the external moment. The positive and negative signs indicated the direction of the internal moment acting on the joint, following the right-hand rule. For the lumbosacral sagittal plane moment, positive values mean flexion while negative values mean extension. Frontal and transverse plane moments equate to lateral bending and rotation moments, respectively. As for the lower extremities, positive values in the hip sagittal, frontal, and transverse planes describe flexion, adduction and internal rotation joint moments, respectively, while for the knees, positive values represent extension, adduction and internal rotation joint moments in the sagittal, frontal, and transverse planes, respectively. Lastly, for the ankle joints, positive values in the sagittal, frontal, and transverse planes are indicative of dorsiflexion, adduction, and eversion moments. Negative values indicate the opposite of the said motions.

In addition, to gain a better understanding of the extent the major lower extremity muscles are recruited during lifting and compare these across postures, sexes, and phases, the muscle excitation was acquired and analysed through EMG amplitude analysis. At face value, EMG amplitudes reflect and measure muscle excitation, which is the depolarization of muscle fiber's sarcolemma that leads to muscle activation and production of muscle force [34].

## 2.5. Statistical analysis

A three-way mixed model (1 between- and 2 within--factor), general linear repeated measures ($2 \times 2 \times 2$) were used to determine effect of gender, lifting posture (squat or stoop) and phase (bending and lifting phase) on 16 identified measurements, namely, mean muscle excitation of rectus femoris, biceps femoris, tibialis anterior, and medial gastrocnemius, along with the lumbosacral, hip, knee, and ankle peak internal moments in the sagittal, frontal, and transverse plane. Data were tested for normality prior to analysis by checking skewness and kurtosis. All values fall within the acceptable range of −3 to +3 for skewness and –10 to +10 for kurtosis [7], meeting the assumption of normality. All statistical analyses were conducted in IBM SPSS version 27 (IBM Corp., Chicago, IL, USA). A full factorial model with Type III sum-of-squares method was employed. For within-subject factors, posture and phase, the default polynomial contrast was selected. The significance level was set at $p < 0.05$. Main effects were compared. Bonferroni adjustments were used. The magnitude of changes associated with significant main effects was indicated by partial eta-squared ($p\eta^2$) values, which were reported as small (<0.06), medium (0.06–0.14), and large (>0.14) [8].

# 3. Results

The statistical results for the joint moment and muscle excitation data are summarized in Table 1. Significant differences were only seen within the interaction of lifting posture and phase, specifically on the hip transverse plane moment ($p\eta^2 = 0.269$), knee sagittal ($p\eta^2 = 0.453$), frontal ($p\eta^2 = 0.276$), and transverse ($p\eta^2 = 0.327$) planes moment; ankle frontal plane moment ($p\eta^2 = 0.370$), rectus femoris ($p\eta^2 = 0.599$) and medial gastrocnemius ($p\eta^2 = 0.610$). There was no statistically significant gender × posture × phase, gender × posture, nor gender × phase interactions.

ment, with squat (1.384 ± 0.204 Nm/kg) higher than stoop (0.481 ± 0.077 Nm/kg). The knee sagittal plane moment was statistically significantly different, with squat exhibiting a high internal extension moment (2.112 ± 0.306 Nm/kg) while stoop exhibited an internal flexion moment (–0.330 ± 0.208 Nm/kg). For frontal and transverse plane moments of the knee, both postures showed internal abduction and external rotation moments, respectively. Stoop has higher internal abduction moments (–1.283 ± 0.180 Nm/kg) than squat (–1.141 ± 0.166 Nm/kg). While higher external rotation moments in squat (–0.922 ± 0.141 Nm/kg) than in stoop (–0.233 ± 0.580 Nm/kg). Ankle plantarflexion (–0.969 ± 0.138 Nm/kg) and eversion (1.400 ± 0.197 Nm/kg) moments were higher in stoop than in squat (plantarflexion: –0.612 ± 0.130 Nm/kg; eversion: 1.334 ± 0.188 Nm/kg). In contrast, squat (–0.553 ± 0.096 Nm/kg) had internal ankle abduction moments higher than stoop (–0.397 ± 0.057 Nm/kg).

Table 1. Statistical findings for different joint moments and muscle excitation during the lifting task

| | | Sex | Posture | Phase | Gender × posture | Gender × phase | Posture × phase | Gender × posture × phase |
|---|---|---|---|---|---|---|---|---|
| Lumbo-sacral | sagittal | $p =$ 0.001* | $p =$ 0.434 | $p =$ 0.009* | $p =$ 0.274 | $p =$ 0.251 | $p =$ 0.239 | $p =$ 0.683 |
| | frontal | $p =$ 0.344 | $p =$ 0.690 | $p =$ 0.867 | $p =$ 0.682 | $p =$ 0.119 | $p =$ 0.106 | $p =$ 0.528 |
| | transverse | $p =$ 0.202 | $p =$ 0.219 | $p =$ 0.179 | $p =$ 0.543 | $p =$ 0.991 | $p =$ 0.817 | $p =$ 0.911 |
| Hip | sagittal | $p =$ 0.474 | $p =$ 0.191 | $p =$ 0.001* | $p =$ 0.471 | $p =$ 0.526 | $p =$ 0.580 | $p =$ 0.628 |
| | frontal | $p =$ 0.907 | $p =$ 0.115 | $p =$ 0.003* | $p =$ 0.977 | $p =$ 0.351 | $p =$ 0.798 | $p =$ 0.225 |
| | transverse | $p =$ 0.511 | $p =$ <0.001* | $p =$ <0.001* | $p =$ 0.663 | $p =$ 0.160 | $p =$ 0.027* | $p =$ 0.767 |
| Knee | sagittal | $p =$ 0.391 | $p =$ <0.001* | $p =$ 0.119 | $p =$ 0.992 | $p =$ 0.693 | $p =$ 0.002* | $p =$ 0.486 |
| | frontal | $p =$ 0.816 | $p =$ 0.039* | $p =$ <0.001* | $p =$ 0.961 | $p =$ 0.477 | $p =$ 0.025* | $p =$ 0.973 |
| | transverse | $p =$ 0.887 | $p =$ <0.001* | $p =$ 0.002* | $p =$ 0.365 | $p =$ 0.076 | $p =$ 0.013* | $p =$ 0.492 |
| Ankle | sagittal | $p =$ 0.652 | $p =$ 0.004* | $p =$ <0.001* | $p =$ 0.905 | $p =$ 0.117 | $p =$ 0.411 | $p =$ 0.723 |
| | frontal | $p =$ 0.854 | $p =$ 0.022* | $p =$ <0.001* | $p =$ 0.102 | $p =$ 0.171 | $p =$ 0.007* | $p =$ 0.786 |
| | transverse | $p =$ 0.933 | $p =$ 0.029* | $p =$ <0.001* | $p =$ 0.077 | $p =$ 0.299 | $p =$ 0.970 | $p =$ 0.745 |
| Rectus femoris | | $p =$ 0.041* | $p =$ <0.001* | $p =$ 0.104 | $p =$ 1.000 | $p =$ 0.834 | $p =$ <0.001* | $p =$ 0.125 |
| Biceps femoris | | $p =$ 0.216 | $p =$ 0.379 | $p =$ <0.001* | $p =$ 0.825 | $p =$ 0.153 | $p =$ 0.353 | $p =$ 0.890 |
| Tibialis anterior | | $p =$ 0.286 | $p =$ <0.001* | $p =$ 0.001* | $p =$ 0.424 | $p =$ 0.849 | $p =$ 0.130 | $p =$ 0.627 |
| Medial gastrocnemius | | $p =$ 0.367 | $p =$ <0.001* | $p =$ <0.001* | $p =$ 0.913 | $p =$ 0.200 | $p =$ <0.001* | $p =$ 0.278 |

Statistically significant differences were highlighted and marked with (*).

## 3.1. Joint moments

The joint moments for each posture during bending and lifting phases are shown in Fig. 4. The individual effect of posture was significant for the peak internal joint moments of the hip in transverse plane and in all planes of the knee and ankle joints. Both lifting postures exhibited a hip internal rotation mo-

Peak internal joint moments were all higher during lifting compared to bending phases; however, significant values were only observed in the lumbosacral sagittal plane, knee frontal and transverse planes and in all planes of the hip and ankle joints. The lifting phase exhibited high lumbosacral (–1.679 ± 0.199 Nm/kg) and hip (–2.531 ± 0.424 Nm/kg) extension moments. There was also higher hip (–1.470 ± 0.162 Nm/kg) and knee (–1.356 ± 0.196 Nm/kg) abduction moment,

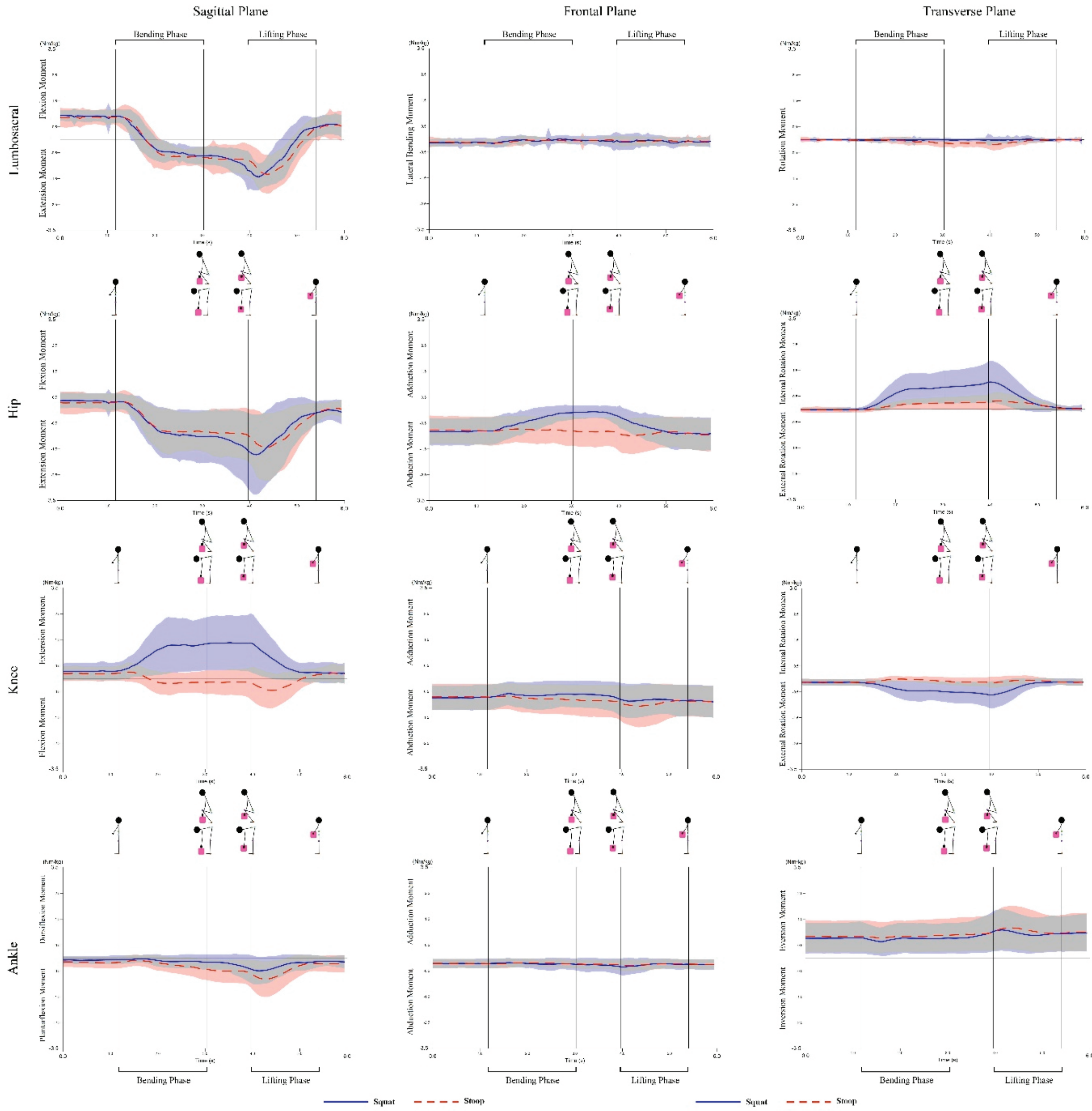



Fig. 4. Joint moments between different lifting postures during the whole lifting cycle, in sagittal, frontal and transverse planes

higher hip (1.065 ± 0.157 Nm/kg) internal rotation moment, while for the knee there was higher (–0.655 ± 0.094 Nm/kg) external rotation moment. The ankle showed higher plantarflexion (–1.058 ± 0.164 Nm/kg), abduction (–0.533 ± 0.081 Nm/kg) and eversion (1.574 ± 0.219 Nm/kg) moments during lifting compared to bending phase.

Significant differences between males and females were only found in the peak lumbosacral sagittal plane moment, with males exhibiting a higher extension moment (–1.690 ± 0.338 Nm/kg) and females exhibiting more of a flexion moment (0.61 ± 0.338 Nm/kg).

## 3.2. Lower extremity muscle excitation

The mean muscle excitations during the different postures (Fig. 5) and phases (Fig. 6) as well as the normalized muscle excitation waveforms during the whole lifting task (Fig. 7) were plotted. Excitation of the anterior lower extremity muscles, specifically the RF and TA, were significantly higher during squat. In comparison, a statistically significant increase in MG muscle excitation was observed during stoop. All muscle

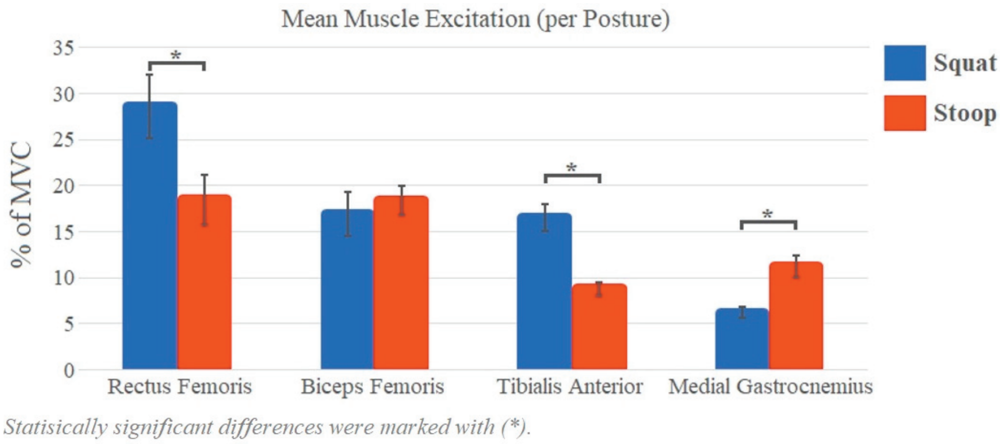



Fig. 5. Mean muscle excitation of lower extremity muscles during different postures of lifting

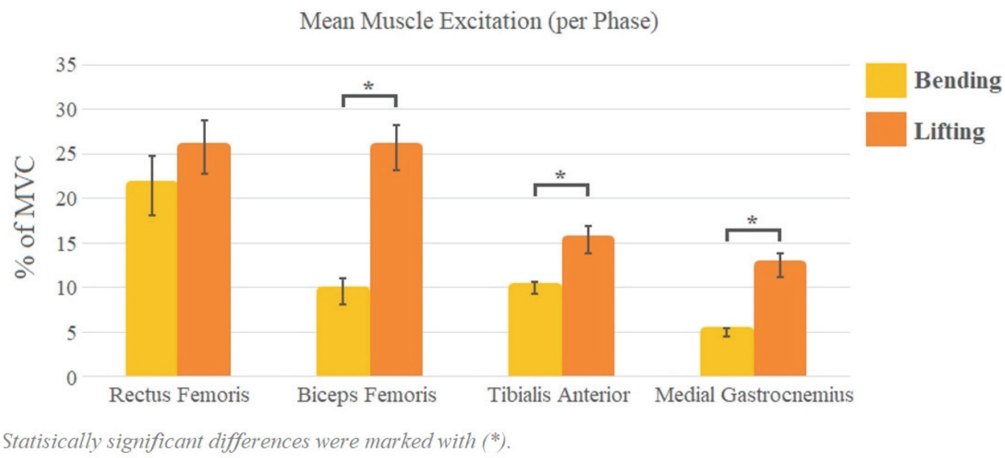



Fig. 6. Mean muscle excitation of lower extremity muscles during different phases of lifting

excitations were higher during the lifting phase than in the bending phase. However, significant differences were only present for the BF, TA and MG. Gender differences were found only in the excitation of RF ($p = 0.041$), with females having higher muscle excitation (30.056 ± 4.131 % of MVC) than males (17.064 ± 4.131 % of MVC).

## 4. Discussion

The present study included a comprehensive biomechanical analysis of the lower back and lower extremity joint moments and muscle excitation, performed during different lifting phases and postures in healthy adults to address significant gaps in the relevant literature and provide a holistic understanding of the lifting process. Significant interaction effects between lifting postures and phases were seen, specifically in the hip internal rotation moment, ankle abduction moments, and all directions of knee moments as well as on RF and MG excitation. This emphasizes that the dynamic adjustments needed to accommodate the varying demands during low-lying lifting tasks primarily occur in the lower extremities and less on the back. This is in accordance with the studies by Hwang et al. [19] and van der Have et al. [33], which showed no significant changes in the lumbar spine moments between squat and stoop postures and that the lower extremity played most of the parts in generating, supporting, and absorbing forces during lifting.

In contrast to previous research [22], [28], the present study's findings showed no significant interaction between sex and lifting posture nor sex and lifting phase, despite the use of an adjusted load. Previous research used a constant mass for males (15 kg) and females (10 kg) to standardize the relative load to be lifted, however, the current study utilized a more definitive computation of the load through the percentage of LBM using Hume's formula [18], which considers each participant's body mass, height and gender, thus

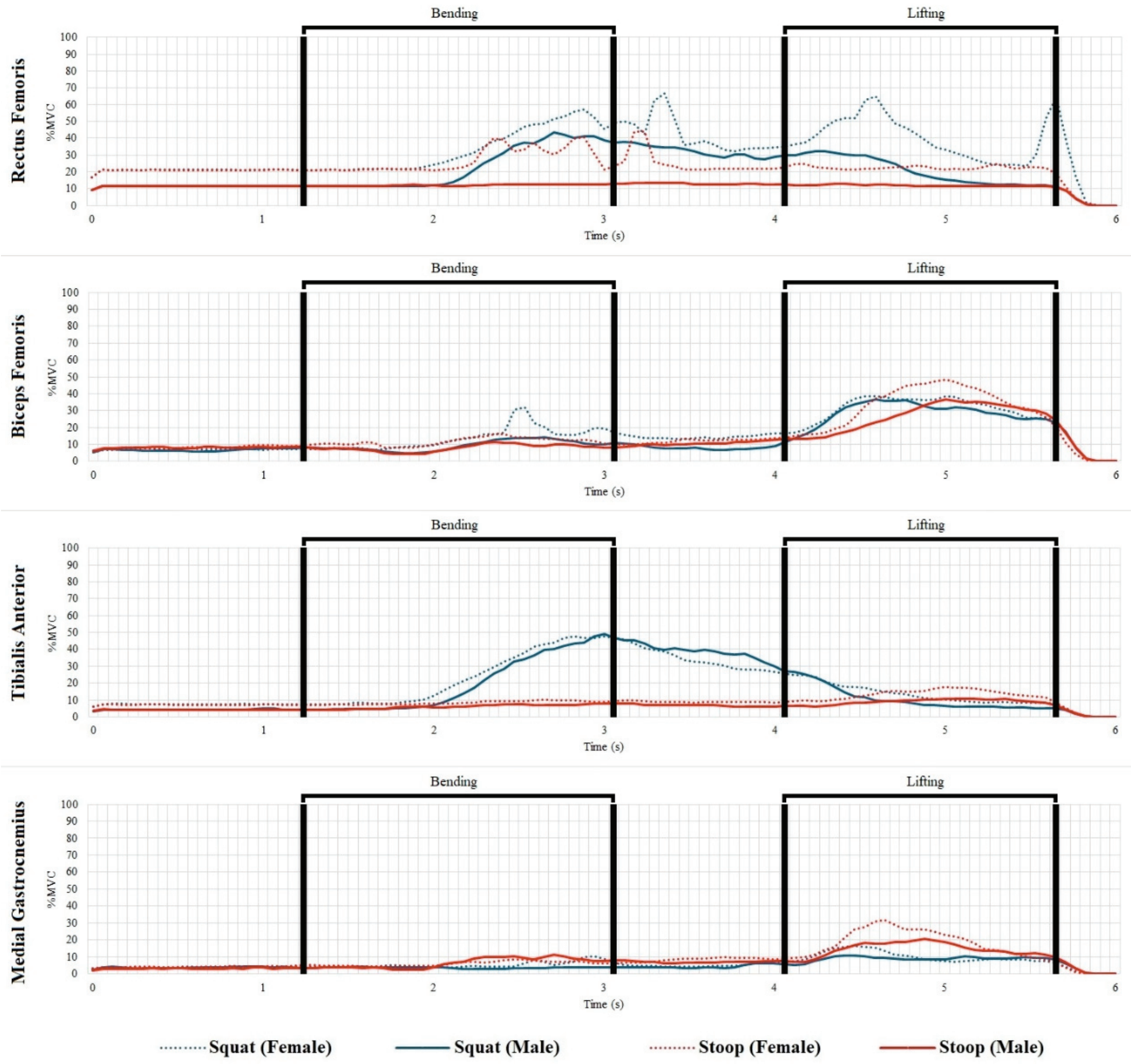



Fig. 7. Normalized muscle excitation waveforms during the whole lifting task

yielding a more individualized amount of load to be lifted. This indicates that when the load was adjusted with a more specific calculation, sex has no impact on the lifting posture or phase.

Despite the absence of significant interaction between gender, posture and phase, there was, however, a statistically significant individual effect of gender, posture, and phase on the lower back and lower extremity joint moments and muscle excitation.

## 4.1. Gender

The lumbosacral sagittal plane moment showed a significant difference between the sexes. Females are known to have a wider pelvis and greater Q-angle. During lifting, females tend to rely more on the pelvis and flex their hips more [23], predisposing them to adopt an internal flexion moment in the lumbosacral spine. In addition, females have reduced extensor muscle strength [23], yet increased flexibility and lower center of mass [2]. These result in females flexing their spine to stretch their back extensors to gain mechanical advantage for energy transfer, while males primarily use their back extensors, as evidenced by high internal extension moment in the lumbosacral joint [14]. Despite being mechanically advantageous, females' strategy can increase the risk for muscular and inert tissue injuries in the back.

Females are also known to have lower knee extensor strength, which requires greater neuromuscular excitation of the quadriceps, making it reach its maximal capacity faster than males. The study's results showed that despite the gender-adjusted load, females still showed higher excitation of RF. This must be carefully considered in injury prevention strategies and ergonomic design, particularly in heavy or repetitive lifting tasks. This heightened neuromuscular demand in females makes them susceptible to fatigue and injury.

## 4.2. Lifting posture

Unlike previous research [19], [33], which were limited to sagittal plane moments only, the current study also investigated the frontal and transverse plane moments. This was important to get a complete understanding of the biomechanics of a lifting task. In line with the hypothesis, there were indeed significant differences not only in the sagittal but also in the frontal and transverse plane joint moments, particularly in the lower extremities.

Contrary to the conventional belief that squat lifting protects the back, the findings of the present study showed no significant difference in the lumbosacral joint moments between the two postures. This aligns with more recent evidence [29], [33] negating the idea that squat is the "proper posture" when lifting low-lying objects. Considering the findings of this study and current literature [19], [33], squat lifting demonstrates higher hip internal rotation, knee extension and external rotation, and ankle abduction moments, which further proves that squat lifting just merely transfers more load to the lower extremity, without making significant unloading in the lower back. However, it is also important to note that squatting has lesser internal knee abduction moments than stooping. A higher knee flexion angle, which is a requirement for the squat posture, can reduce the internal knee abduction moment [27], which may be beneficial for people who repetitively lift or for individuals with an increased risk of developing knee osteoarthritis.

Despite the decrease in the hip and knee sagittal plane moments during stoop, and the insignificant difference in the lumbosacral joints, the load seemed to be redirected to the ankle, as observed by the relatively higher internal plantar flexion and eversion moments. Increased ankle plantarflexion moments during stoop occurs to counteract the ground to prevent from falling forward during bending and to stabilize oneself with the added mass of the object during lifting. Overall, in either posture, moments are just re-distributed to the different joints of the lower extremity and none on the lower back.

RF excitation being higher in squat and medial gastrocnemius excitation being higher in stoop, were consistent with previous literature [19], [33]. However, contrary to previous research [19], TA showed higher excitation when squatting than when stooping and BF excitation was not significantly different between postures. Balanced activation of the anterior and posterior muscle groups is important for stabilization, movement efficiency, and injury prevention. Quadriceps activation helps with knee joint stability, especially during weight-bearing activities [12]. However, overemphasis on the RF without adequate recruitment and flexibility of the BF can lead to injury [25], [26]. Additionally, the absence of adequate hamstring and gastrocnemius activation to counteract the increased internal knee extensor moment observed during squatting predisposes individuals to higher knee joint loading, which can lead to injury [12], [25], [26]. Furthermore, the demand for higher muscular excitations leads to faster fatiguability. Therefore, the interplay of balance and stability as well as muscle excitation and fatigue, explains why people still choose to stoop rather than squat.

## 4.3. Phases of lifting task

Distinguishing the bending and lifting phases provided valuable insights into the dynamic nature of joint moments and muscle excitation. As expected, the lifting phase exhibited higher joint loading and muscle excitation due to the mass of the lifted object. In addition, transitioning from a lower to a higher position requires greater mechanical work and force generation for the joints and muscles.

Lifting is a complex interplay between mobility and stability in which the muscles surrounding the joints play a part. The bending phase requires more flexibility than strength to reach the object and consists predominantly of flexion movements. Meanwhile, the lifting phase requires strength and stability to ensure safe execution and is predominantly an extension movement.

During weight-bearing tasks, knee extension, primarily performed by the quadriceps, induces internal rotation of the tibia in relation to the femur, thereby increasing the tension in the ACL. The hamstrings, particularly the BF, actively counteract this internal rotation to mitigate the risk of ligament injury. Meanwhile, MG contributes to knee flexion and is significantly associated with external knee rotation moments. Flaxman et al. [12] found that gastrocnemius was commonly the first to be activated during rotational perturbations. As observed, there was a significant difference in the excitation of BF and MG during the bending and lifting phases. On the other hand, the observed difference in the excitation of TA is due to its transition from concentric to eccentric contraction during the bending to lifting phase, respectively. Concentric contraction is crucial during the bending phase to initiate movement and move the body closer to the object. Whereas eccentric contraction during the lifting phase counteracts the gastrocnemius muscle to maintain stability as the body moves from a bent to an upright position. No significant difference was observed in RF excitation between phases due to its biarticular nature, serving as both hip flexor and knee extensor.

## 4.4. Limitations

The authors recognize that there are limitations to the present study. First, the study did not include the trunk muscles, particularly the erector spinae and rectus abdominis, due to the technical limitations of the surface EMG in capturing deep muscles and muscles covered by visceral fat. Additionally, other factors, such as the position of the feet, load, and grip on the object to be lifted, were not controlled due to technical difficulties. All participants were asked to stand and lift with feet shoulder-width apart, however, the angle of out-toeing was not controlled. The object to be lifted was also impossible to be too close to the body, as it obstructs the reflective markers. Owing to these technical limitations, future studies may utilize different equipment, such as wearable inertial measurement units, that can capture the motion more freely. Finally, it should be noted that the participants selected for this study may not adequately represent the demographic characteristics of the manual labor workforce engaged in routine lifting tasks. This is because lifting is deemed a daily functional activity and is not exclusive to manual laborers. Consequently, the intention was to encompass a broader representation of the general population. Future studies, however, could further explore the effect of the level of expertise and its interaction with sex, lifting postures, and lifting phases.

Nevertheless, through the inclusion of moments in sagittal, frontal, and transverse planes, along with a detailed exploration of gender, posture and phase, the present study offers a more holistic understanding of the factors influencing joint moments, and muscle excitation during lifting. This study emphasizes the complex nature of lifting and contributes to the ongoing discourse on optimal clinical and ergonomic practices, as well as in shaping future research endeavours. Additionally, the present study calls for the use of a more individualized computation of lifting load, in research and in practice, to minimize bias within gender.

# 5. Conclusions

The study revealed significant differences in joint moments and muscle excitation during lifting, between genders and across postures and phases, particularly in the lower extremities. The findings underscore the complexity of lifting biomechanics and emphasize the need for larger, methodologically higher quality studies with more representative samples to better inform tailored interventions and ergonomic recommendations for promoting safer lifting practices.

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