<u>Title:</u> Sex Differences in Glenohumeral Muscle Activation and Coactivation During a Box Lifting Task

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ABSTRACT

Manual material handling is associated with shoulder musculoskeletal disorders, especially for women. Sex differences in glenohumeral muscle activity may contribute to women's higher injury risk by affecting shoulder load and stability. We assessed the effects of sex (25 women vs 26 men) and lifting load (6 kg vs 12 kg) on muscle activation during box lifting from hip to eye level. Surface and intramuscular electromyography were recorded from 10 glenohumeral muscles. Most muscles were more activated for the heavier box and for women. These effects were larger for "prime movers" than for stabilisers and antagonists. Despite their apparently heterogeneous effects on muscle activity, sex and mass did not affect Muscle Focus, a metric of coactivation. This may be partly related to the limited sensitivity of the Muscle Focus. Nevertheless, sex differences in strength, more than in coactivation patterns, may contribute to the sex imbalance in the prevalence of musculoskeletal disorders.

KEYWORDS

Shoulder, manual material handling, occupational biomechanics, electromyography, gender

PRACTITIONER SUMMARY

We studied sex differences in glenohumeral muscle activity in a lifting task to eye level. Women lifting a 6 kg box activated their muscles similarly to men lifting a 12 kg box, i.e. up to 48% of their maximum capacity. Interventions minimising shoulder load should be implemented, especially for women.

INTRODUCTION

Neck and shoulder pain is experienced by 40.7% of workers in various parts of the world and frequently leads to functional limitations, medical consultations, and work absenteeism (Sarquis *et al.* 2016). Multiple systematic reviews of epidemiological studies identified physical work-related factors associated with shoulder disorders including highly repetitive work, forceful exertions, awkward postures and high psychosocial job demands (van Rijn *et al.* 2010, Mayer *et al.* 2012). Mayer *et al.* (2012) found strong evidence that exposure to manual material handling (MMH) tasks, such as box lifting, is associated with future neck and/or shoulder complaints. Previous efforts for understanding the motor control strategies and biomechanical impacts of MMH mostly focused on the lumbar region (reviewed in Potvin (2008)). Indeed, very few studies assessed shoulder kinematics, kinetics and muscle activity using electromyography (EMG). Given that most of the glenohumeral joint stability is provided by muscle contractions (Veeger and van der Helm 2007), a detailed investigation of EMG at this joint is necessary to understand shoulder biomechanical risks during MMH. The amount of coactivation between agonist and antagonist muscles during motor activities is particularly important for motor control and joint stability. Indeed, excessive antagonistic muscle activation during movements may lead to a decreased motor efficiency. Meanwhile, a lack of coactivation may result in joint instability and may be detrimental to motor control, especially in the ability of the motor system to react to unexpected perturbations (Latash 2018). Coactivation might have an important role in the prevention of glenohumeral joint luxation by increasing compression forces and counteracting shear forces produced by the prime movers (Ackland and Pandy 2009). Most coactivation metrics compare the normalized activation level of two or multiple opposing muscles defined as agonists and antagonists (reviewed in Le *et al.* (2017)). Such methods are not suitable for the shoulder given that the large number of muscles crossing the glenohumeral joint can hardly be defined as agonist-antagonist pairs. This is even more true during high-amplitude movements, such as MMH, as muscles' lines of action change with the glenohumeral elevation angle (Ackland *et al.* 2008).

The Muscle Focus is a coactivation metric that has been developed to address these challenges in complex joints (Yao *et al.* 2004). Instead of computing coactivation ratios between arbitrarily identified agonist-antagonist pairs, the Muscle Focus takes into consideration each muscle's instantaneous orientation on a continuous scale. It returns a value between 0 and 1: the former meaning that muscles are similarly activated no matter their action, and the latter indicating that only muscles rotating the joint in the same direction are activated.

Some authors examined the influence of different lifting task characteristics, such as the load, lift height and movement velocity, on shoulder muscle action (Yoon *et al.* 2012, Blache *et al.* 2015a, Blache *et al.* 2015b). In these studies, anterior and lateral deltoids were the most active muscles and provided the greatest amount of force during box-lifting to shoulder or eye level. Lifting heavier boxes resulted in a relatively homogeneous increase in activity in those muscles as well as in the surrounding ones, presumably to maintain joint stability (Yoon *et al.* 2012, Blache *et al.* 2015a). Indeed, Blache et al. (2015a) showed no changes in Muscle Focus, even with a three-fold increase in the load lifted (6 kg to 18 kg). Yet, using static optimisation for estimating muscle forces during the same lifting task, the same authors also showed that the mechanical work increased with box mass in the anterior deltoid and subscapularis muscles, but not in the supraspinatus and infraspinatus (Blache *et al.* 2015b). As studies of muscle activation patterns have not recorded rotator cuff muscles, it is impossible to state if these muscles increase their activity in a uniform way with heavier boxes, as do recorded surface muscles (Yoon *et al.* 2012, Blache *et al.* 2015a), or if their activity remains constant, as suggested by the modelling study (Blache *et al.* 2015b).

Few studies assessed the influence of individual factors on the level and coordination of muscle activity exerted during MMH. Among individual factors, sex needs to be considered for clinical and biomechanical reasons. First, epidemiological studies showed that more women than men live with shoulder pain (Sarquis *et al.* 2016, Bodin *et al.* 2017). Many anthropometrical, biomechanical and motor control-related differences could affect motor behaviour and contribute to this higher risk of shoulder pain in women (Cote 2012). Among them, women are on average smaller (163.04 cm) and lighter (70.94 kg) than men

(178.18 cm, 87.06 kg) (Statistics Canada 2015), and they are able to generate 30% to 60% less force (Faber *et al.* 2006, Harbo *et al.* 2012, Douma *et al.* 2014). Considering this, women are, on average, working closer to their maximal muscle capacity when they lift the same load as men. In such equal conditions, studies have shown that the posture and the kinematic coordination of the lower limbs and trunk are different between men and women (Plamondon *et al.* 2014). However, even when the load is adjusted relative to men and women's maximal capacity, small differences persist (Lindbeck and Kjellberg 2001, Sheppard *et al.* 2016, Plamondon *et al.* 2017). Since those studies focused on biomechanical variables related to low back musculoskeletal disorders, all conditions studied involved MMH below shoulder height. We recently showed that women significantly changed their inter-joint coordination with increasing box mass during a lifting task to eye level (Martinez *et al.* 2019). However, such compensations were not observed in men. To our knowledge, no studies have investigated sex differences in shoulder muscle activity when lifting boxes above shoulder level. The objective of the present study was, consequently, to evaluate sex differences in EMG activity of muscles acting at the glenohumeral joint when lifting boxes from hip to eye level. The effects of the mass of the box (6 *kg versus* 12 kg) on sex differences were also investigated. **Hypothesis 1:** When lifting the same absolute load, it was expected that women would generate more EMG activity than men to compensate for their lower strength. It was expected that women would need as much EMG activity relative to their maximal capacity when they lifted a 6 kg box than men lifting a box twice heavier given that their muscle strength is about 50% lower in many upper limb muscle groups (Harbo *et al.* 2012). **Hypothesis 2:** Also, we hypothesized that men and women would increase their EMG activity with increasing load. **Hypothesis 3** For men, we hypothesized that this increase would be uniformly distributed across muscles, leading to an unchanged Muscle Focus (Blache *et al.* 2015a). **Hypothesis 4:** The kinematic compensations previously observed in women when they lifted the 12 kg box may suggest that they were close to their maximal capacity (Martinez *et al.* 2019). Thus, in the present study we expected that women lifting 12 kg boxes would decrease their antagonist muscle coactivation (i.e. increase the Muscle Focus) to maximise the contribution of prime movers towards effective torques.

METHODS

Fifty-one healthy students (26 men [age: 25.5 ±5.8, weight: 73.1 ±10.8 kg, height: 178.0 ±7.4 cm], 25 women [age: 21.9 ±2.4, weight: 59.6 ±6.1 kg, height: 166.9 ±6.6 cm]) participated in this study. Participants were free of self-reported musculoskeletal injury and scored <3 on the Quebec Back Pain Disability Scale (Kopec *et al.* 1995) and >23 on the Disabilities of Arm, Shoulder and Hand questionnaires (Hudak *et al.* 1996, Durand *et al.* 2005). All could safely perform physical activity based on the Physical Activity Readiness Questionnaire (Thomas *et al.* 1992). No participants had more than six months of experience in manual material handling work. The research protocol was approved by the ethic committee of the University of Montreal (No. 15-016-CERES-P), and all participants gave their written informed consent before the experiment.

Procedure

Participants moved a box ($height * width * length$: $8 * 35 * 50$ cm) of 6 kg or 12 kg between three shelves, up and down. The box was equipped with two cylindrical handles ($diameter \times length$:$2.9 * 13$ cm) containing a force sensor used to detect the beginning and end of each trial. Six lifting and lowering movements (i.e. lifting: hips → shoulders, hips → eyes, shoulders → eyes; lowering: shoulders → hips, eyes → hips, eyes → shoulders) were performed with each box (6 kg and 12 kg) for a total of 12 conditions. Three trials were performed for each condition, for a total of 36 trials (Figure 1, see (Blache *et al.* 2015a, Blache *et al.* 2015b) for more details). Movements were performed in a randomised order, with 30 s rest periods in-between. In the present study, only the six trials when participants lifted the box from the hip to the eye-level shelves were analysed. Three of these trials were performed with the 6 kg box and 3 with the 12 kg one. Participants stood at a comfortable distance from the shelves so they could reach the box without moving their feet and used a natural lifting technique at a self-selected speed.

Data Acquisition

Hand contact force, EMG, and kinematic signals were synchronously recorded with Vicon Nexus software at 2000 Hz, 2000 Hz and 200 Hz, respectively. Force signals were acquired through the right instrumented handle of the box (Sensix SH2653-1106B3, Poitiers, France) and were solely used to detect the beginning and the end of each trial in the present study. Surface and intramuscular EMG signals (Trigno EMG Wireless System, Delsys, USA) were recorded from different muscles for different subgroups of participants, as indicated in Table 1. Surface EMG was recorded from the following 7 muscles crossing the dominant glenohumeral joint: anterior deltoid (DeltA), lateral deltoid (DeltL), posterior deltoid (DeltP), pectoralis major (Pect), latissimus dorsi (Lat), biceps brachii (BB) and triceps brachii long head (TB). Electrodes were placed according to the SENIAM recommendations (Hermens *et al.* 2000), after shaving and cleaning the skin with alcohol. Intramuscular EMG was recorded for the infraspinatus (Infra), supraspinatus (Supra), and subscapularis (Subscap) muscles (Kadaba *et al.* 1992, Perotto and Delagi 2005). A series of 12 submaximal voluntary contractions were performed to validate electrode placement. Then, the same muscle contractions were performed twice at maximal voluntary intensity for normalisation purposes, in line with the recommendations of Dal Maso *et al.* (2016).

Pelvis, trunk and upper-limb kinematic data was acquired with an 18-camera Vicon motion analysis system (Oxford Metrics Ltd, Oxford, UK). Thirty-five markers were placed on participants' skin over the pelvis (4), trunk (6), clavicle (5), scapula (9), upper arm (7) and forearm (4), in line with the Jackson *et al.* (2012) kinematic model. After marker placements, subjects maintained a static position in an anatomical stance, in order to scale the kinematic model.

Data Analysis

Each EMG signal was first rebased by subtracting its mean value. Rebased EMG data were then filtered using a second-order, zero-lag 20-425 Hz Butterworth band-pass filter (Merletti 1999). Filtered signals were then rectified, and EMG envelops were extracted with a second order zero-lag 5 Hz Butterworth low-pass filter. Finally, the EMG envelops were normalised to each muscle's maximal voluntary activation (MVA).

As for kinematics, soft tissue artefacts were first reduced using a multibody kinematic optimisation model based on an extended Kalman filter (Jackson *et al.* 2012). Corrected marker positions were then introduced into Wu's shoulder model (Wu *et al.* 2016) in OpenSim (Delp *et al.* 2007) for further analysis. The model was composed of 6 joints and 16 degrees of freedom (DoF; trunk [6 DoF], sterno-clavicular [2 DoF], acromio-clavicular [3 DoF], glenohumeral [3 DoF], elbow [1 DoF] and radio-ulnar [1 DoF]). The model was scaled using the static trial data. Generalised coordinates were then computed using inverse kinematics in OpenSim. Finally, muscle attachment sites and lines of action were extracted with the Opensim *muscle direction* toolbox for the Muscle Focus analysis (van Arkel *et al.* 2013). When multiple lines of action were available for one muscle, the one which best fitted with the position of EMG electrodes was selected. EMG and kinematic data were time-normalised from 0 to 100% of movement duration.

The outcomes were the time histories of the muscle activity, the sum of normalised EMG signals (*SumEMG,* ***Hypotheses 1 and 2***), and the Muscle Focus ***(Hypotheses 3 and 4)***. The latter is an indicator of the selectivity of muscle activation (Yao *et al.* 2004) and is calculated as follows:

$$MF = \frac{\sum_{i=1}^{M} \left\| EMG_i \, \overrightarrow{d_i} \right\|}{\sum_{i=1}^{M} EMG_i}, \tag{1}$$

where EMG_i corresponds to the EMG normalised amplitude of the muscle i, and $\overrightarrow{d_i}$ is the muscle unit moment (i.e. the cross product between the line of action and the moment arm of the muscle). *SumEMG* and *MF* outcome measures were computed for three sets of muscles for practical and biomechanical reasons, as described in Table 2. Since the long heads of BB and TB have no direct insertions on the humerus, those muscles were not included in this analysis (Blache *et al.* 2015a).

[Insert Tables 1 and 2 here]

After averaging each participant's time histories for the same box mass, the statistical non-parametric mapping (Nichols and Holmes 2002) procedure, implemented by Pataky (2012) in the spm1D Matlab toolbox for biomechanical data, was used to compare the entire time histories of our outcomes. This method prevents information loss associated with standard methods that reduce time series into a single data point, while controlling for type α errors due to multiple comparisons. Non-parametric testing was chosen as it leads to qualitatively identical results than parametric testing, while being robust to non-normal and non-spherical data (Pataky *et al.* 2015). We thus tested the effects of *sex* (men *versus* women) and *mass* (6 kg *versus* 12 kg), and the interaction between these factors, on each muscle's EMG activity, Muscle Focus and *SumEMG* variables. When significant interactions were observed, men and women's muscle activity at the same absolute (6 kg and 12 kg) and relative loads (women 6 kg *versus* men 12 kg) were compared using two samples t-tests. Post-hoc paired t-tests (6 kg *versus* 12 kg) were also computed for each sex. All post-hoc analyses were tested using the region of interest approach (Pataky *et al.* 2016) with a Bonferroni correction ($p = 0.0083$). To simplify the presentation of the results, isolated clusters of statistical significance lasting less than 5% of the movement duration were discarded, while clusters separated by less than 5% were merged (Belaise *et al.* 2018a). For each significant main effect and post-hoc test, Cohen's d effect size (ES) statistic was computed across significant clusters (Cohen 1988, Lakens 2013):

$$ES = \frac{\overline{X_2} - \overline{X_1}}{pooledSD}$$

$$pooledSD = \sqrt{\frac{{SD_{X1}}^2 * (n_1 - 1) + {SD_{X2}}^2 * (n_2 - 1)}{n_1 + n_2 - 2}}$$

In this equation, X_1 and X_2 are the within-movement averages calculated for each subject and each condition by considering only the data points within significant clusters. The variables n_1 and n_2 are the sample sizes for each condition. The total duration of significant clusters (i.e. effect duration [ED]) was also extracted. ES were qualitatively interpreted as large ($ES > 0.8$), moderate ($0.8 > ES > 0.5$) or small/absent ($ES < 0.5$), as suggested by Cohen (Cohen 1988). Figures were generated with the gramm Matlab toolbox (Morel 2018).

RESULTS

The general profile of muscle activity during the lifting task was characterised by mostly two bursts of activity (Figure 1). The first one, from the beginning of the movement to ~25% of the movement duration, was small in most muscles, but large in the BB and, to a lesser extent, in the Pect. The second (dominant) EMG burst started between 25 and 50% of the movement duration and finished at the end of the movement. The anterior and lateral deltoid muscles were the most active, followed by the supraspinatus (Figure S1). The least active muscles were the subscapularis, the posterior deltoid and the latissimus dorsi.

Effects of Sex and Mass on Individual Muscle Activity

The activity of all recorded muscles was significantly influenced by the mass of the box; the heavier box resulted in greater EMG activity (Figure 2 and Figure 3). This effect was more important during the lifting and transfer phases of the movement than during the deposit phase. Moreover, EMG increased more between the 6 and 12 kg boxes for "prime movers" (DeltA, DeltL, Pect and BB : ES range = 0.96-1.62; ED range: 59-91% of the movement duration) than for stabilisers (Supra, Infra and Subscap: ES range = 0.45-0.76; ED range: 26-39% of the movement duration) and antagonists (DeltP, Lat and TB: ES range = 0.39-0.57; Effect duration range: 34-65% of the movement duration) muscles. A similar pattern was observed regarding the effect of sex on muscle activity. Women needed proportionally more muscle activation to lift a box than men in their "prime mover" muscles (ES range = 1.26 -1.95; ED range: 45-100% of the movement duration) and, to a lesser extent, in their antagonists (ES range = 0 -1.37; ED range: 0-94% of the movement duration). Moreover, while the increase in EMG activity with the box mass in those muscles was present for both sexes, it was more important for women (sex x mass interaction: Figure 3, post-hoc tests: Figure 2). No sex differences were observed in the activity of the rotator cuff muscles.

[Insert Figure 2 and Figure 3]

Effects of Sex and Mass on composite measures of muscle activity amplitude (SumEMG) and coactivation (Muscle Focus)

No matter which subgroup of muscles was analysed (Table 2), main effects of mass (ES range: 1.14-1.55; ED range: 55-72%) and sex (ES range: 1.72-1.95; ED range: 44-90%) were observed for the *SumEMG* variables during most phases of the movement (Figure 3: $SumEMG_{delt}$, Figure S2: $SumEMG_{1joint}$, Figure S3: $SumEMG_{sEMG}$). Sex x mass interactions were observed for the $SumEMG_{delt}$ and $SumEMG_{sEMG}$ variables during a small portion of the transfer phase, but not for the $SumEMG_{1joint}$. Like individual muscle measurements, this interaction indicates that the increase in *SumEMG* variables from 6 to 12 kg was greater for women than for men (Figure 3).

As for the Muscle Focus variables, no main effects of sex or mass lasting more than 5% of the movement duration were observed (Figure 3, Figure S2 and Figure S3). A short sex x mass interaction was observed for the MF_{delt} variable. Nevertheless, no significant post-hoc test explained this interaction.

Finally, none of the post-hoc tests comparing men (12 kg) and women (6 kg) lifting a similar mass relative to their maximal force were significant for individual muscles or composite variables.

DISCUSSION

This study assessed the activity and coactivation of glenohumeral muscles when men and women lifted 6 and 12 kg boxes from hip to eye level. As expected, women generated more muscle activity than men for a similar absolute load, and both men and women increased their muscle activity when lifting the heavier box. This increase in muscle activity was more important for prime mover muscles and for women. Despite these heterogeneous changes in muscle activity, sex and mass had minimal effects on the coactivation of glenohumeral muscles.

General Profile of Muscle Activity

Only few studies previously assessed the profile of upper limb muscle activity during a box lifting task (Yoon *et al.* 2012, Blache *et al.* 2015a, Blache *et al.* 2017), and none included rotator cuff muscles. The large burst of activity in the BB is probably more related to its action at the elbow joint, as participants flexed their elbows to bring the box close to their glenohumeral centre of rotation at the beginning of the movement (Martinez *et al.* 2019). As for the Pect muscle, its action early in the movement might serve to stabilise the box through horizontal adduction (Anders *et al.* 2004). The second burst of activity contains the peak activity of most muscles. As previously reported, prime mover muscles during the lifting task are the DeltA and DeltL muscles (Blache *et al.* 2015b, Blache *et al.* 2017). As for rotator cuff muscles, the infraspinatus was the most activated, and the subscapularis the least. These results are opposite to what was expected from our recent musculoskeletal modelling studies based on static optimisation (Blache *et al.* 2015b, Blache *et al.* 2017). Many factors may influence force-sharing and muscle activation estimated by static optimisation, such as the cost function optimised, the model parameters, and the computed joint kinematics (Bolsterlee *et al.* 2013). An important amount of coactivation was present during most of the movement, with the mean Muscle Focus variables ranging from 0.39 ± 0.09 (MF_{sEMG}) to 0.66 ± 0.08 (MF_{delt}). Such coactivation levels are hardly reproducible by standard static and forward optimisation cost functions (Morrow *et al.* 2014, Belaise *et al.* 2018b).

Effects of Mass and Sex on Muscle Activity

As expected ***(Hypothesis 2)***, participants produced more muscle activity to lift the 12 kg than the 6 kg box. Furthermore, women's muscles were more activated ***(Hypothesis 1)***, relative to their maximal activation, than men's, when they lifted boxes of similar mass. Indeed, women lifting the 6 kg box needed muscle activity levels similar to those of men lifting the 12 kg box, reaching activity levels up to 48 % of MVA in the anterior deltoid. This result is consistent with several studies that showed that women's maximal force is 30 to 60% lower than men's for various upper limb muscle groups (Faber *et al.* 2006, Harbo *et al.* 2012, Douma *et al.* 2014). Multiple biological factors can explain the lower strength in women than in men such as their lower muscle cross sectional areas and lever arms, as well as a their smaller absolute and relative number of fast-twitch muscle fibers (Cote 2012). Although the effects of mass and sex were more important for agonist than antagonist muscles, this imbalance was not sufficient to induce significant changes in Muscle Focus variables. In a previous paper from our group focusing only on male participants (Blache *et al.* 2015a), the Muscle Focus (MF_{sEMG}) was not affected by the mass of the box lifted, which is in line with the current results ***(Hypothesis 3)***. We were, however, expecting that women would limit the coactivation of their antagonist muscles, leading to a higher Muscle Focus than in men, especially for the heavier box ***(Hypothesis 4)***. This motor strategy would have minimised the production of antagonist torques, thereby decreasing the total muscle activity exerted and the associated metabolic cost of each effort. Meanwhile, insufficient muscle coactivation may compromise joint stability (Veeger and van der Helm 2007, Blache *et al.* 2017) and contribute to the risk of shoulder disorders (Sarquis *et al.* 2016, Bodin *et al.* 2017). Yet, sex did not influence the Muscle Focus. The lack of effects of mass and sex may indicate that the glenohumeral coactivation patterns are robust to these factors, although there is also the possibility that the Muscle Focus is not sensitive enough to capture real differences in coactivation.

To the best of our knowledge, no studies assessed the sex differences in glenohumeral muscles coactivation during a dynamic task. Anders *et al.* (2004) mapped the activation of 12 glenohumeral and scapular muscles during isometric contractions performed in multiple directions and postures. Their results suggest that women produced more coactivation of stabilising muscles during most of the studied isometric contractions. However, such sex differences in glenohumeral muscle coactivation were not observed in the current study. Differences between those results and ours could reflect the task-specificity of the sex differences in muscle coactivation patterns and/or may be related to the metrics used in each study.

Very few studies used the Muscle Focus metric to document differences in muscle coactivation between groups of participants or motor tasks (Yao *et al.* 2004, Yao *et al.* 2006, Blache *et al.* 2015a). It is therefore still difficult to interpret the sensitivity of this variable. Yao et al. (2004, 2006) showed a lower Muscle

Focus between shoulder and elbow muscles in stroke participants than in healthy controls during isometric shoulder abduction and elbow flexion. In Blache et al. (2015a), we showed a trend for a decrement in Muscle Focus (equivalent to MF_{sEMG}) when male participants lifted a box to their eye level compared with lifts to their shoulder level. Those studies provide evidence that the Muscle Focus has some sensitivity, although they did not report the effect sizes of their results. Still, the fact that, in our study, the Muscle Focus was not able to capture the heterogeneity in individual muscles' EMG differences with sex and mass (i.e. greater differences in prime movers than in antagonists or stabilisers) suggests that its sensitivity may be limited. Also, in a previous article, we showed that only women changed their kinematic strategy when they lifted a heavier box (Martinez *et al.* 2019). It appears unlikely that these changes in coordination observed at the joint level are not related to an altered muscle coordination pattern. Further validation studies are needed.

Impacts for Musculoskeletal Risks and Prevention

Our results suggest that variables related to the amplitude of muscle activity, rather than to the patterns of muscle coactivation, may better explain the higher risk of shoulder musculoskeletal disorders in women. According to the revised NIOSH lifting equation, it would be expected that more than 90% of women and almost all men would be able to safely lift a 12 kg box from hip to eye height twice every minute during an 8-hour shift (revised lifting index = 0.97, Figure S4) (Waters *et al.* 1993). A recent review of studies investigating the relationship between the revised lifting index and low back disorders found a range of lifting indices between 0 and 9.37 for various groups of workers, with within-studies mean values ≥ 1.5 (Lu *et al.* 2016). It is therefore likely that the lifting task used in our laboratory study was similar to many occupational situations. Still, the 90th percentile of the anterior deltoid activity observed in the current study while participants lifted the 12 kg box was 62.7 ± 8.5% of MVA for women and 50.4 ± 8.4% MVA for men (Figure S1). These muscular efforts appear considerable to sustain in an occupational setting. In a previous study, we showed that women, but not men, significantly changed their inter-joint coordination between a 6 kg and a 12 kg box (Martinez *et al.* 2019). Taken together, the high level of muscle activation and the kinematic compensations observed in women suggest that lifting 12 kg boxes may be excessive for the average woman included in our study. It must be noted that a significant amount of overlap between men and women can be observed in many of our outcome measures (Figure S1). This overlap suggests that jobs involving lifting such as the task in the present study should be as biomechanically safe for many men and women, arguing against systematic job segregation. In addition to excluding women from the workforce, such practice may even lead to unexpected health injury risks for women doing "light work" (Messing 2017). However, a greater proportion of women than men might need workplace interventions to decrease the shoulder loads relative to their maximal capacity and remain productive and safe in MMH jobs. These interventions may include physical activities for improving their maximal capacity (Christensen *et al.* 2016) and/or workplace adaptations and technological supports to decrease the MMH-related absolute load (de Looze *et al.* 2016, Lavender *et al.* 2017). Unfortunately, research conducted on technological solutions targeting MMH tasks, such as exoskeletons, are almost exclusively performed with male participants (de Looze *et al.* 2016). Technological developments guided by these studies may not consider some characteristics and needs specific to women.

Limitations and Future Directions

The study reported in this article presents some limitations. First, some muscles have been collected only for a subset of participants. Therefore, the statistical power may have been an issue for some of the variables, especially those involving rotator cuff muscles. Indeed, although no significant sex differences were observed for individual rotator cuff and posterior deltoid muscles, a closer inspection of Figure 1 showed that women's mean values were always higher than those of men. However, the effect sizes (mean ES during the entire movement = 0.57 to 0.63) for these muscles were much lower than for the DeltA, Pect, BB and TB (mean ES during the entire movement = 1.32 to 1.90). Therefore, although a greater sample size could have changed some statistical results, the heterogeneity of the effects of sex and mass between muscles would have remained apparent. As discussed previously, the Muscle Focus may not be sensitive enough to detect changes in coactivation patterns with sex and mass. Its value depends on each muscle's line of action, which in turn are affected by the choice of the musculoskeletal model, and by the estimated joint kinematics (Bolsterlee *et al.* 2013). Moreover, the choice of muscles included in Muscle Focus computations can affect the results. In the current study, the Muscle Focus was lower when it was computed from both uniarticular and multiarticular muscles (MF_{sEMG}), compared to solely uniarticular ones (MF_{delt} and MF_{1joint}). Moreover, the muscles not directly attached to the humerus (e.g. long heads of biceps and triceps brachii) were not implemented. An optimal approach would be to integrate all muscles acting at the glenohumeral joint when calculating the Muscle Focus, which is very challenging as it involves the consideration of a substantial number of EMG recordings. However, the combination of measured EMG (e.g. for surface muscles) and muscle activation estimated from musculoskeletal modelling (e.g. for deep muscles) may offer a suitable alternative. Moreover, such an approach may improve the validity of the modelled muscle activation and forces, as using only static optimisation procedures underestimates muscle coactivation (Morrow *et al.* 2014). Furthermore, while we are confident that, in our sample, women had less strength than men given our results (Figure 2 lower panel) and the published literature (Faber *et al.* 2006, Harbo *et al.* 2012, Douma *et al.* 2014), we did not directly measure this parameter. Indeed, each individual's shoulder strength would have been very interesting to assess as it could have allowed us to quantify the contribution of sex differences in strength to the sex differences observed on EMG activation during the lifting task.

CONCLUSION

Women generate significantly more EMG in most of the recorded upper limb muscles, relative to their maximum, compared to men ,when lifting a box from hip to eye level. This is even more apparent when participants lifted the heavier box. Yet, muscle coactivation during box-lifting, which is much higher than predicted by musculoskeletal modelling studies, is not affected by the load lifted, or by the participant's sex. Although the Muscle Focus may lack some sensitivity, our data suggests that sex differences in strength, more than in glenohumeral coactivation patterns, may contribute significantly to the higher risk of shoulder musculoskeletal disorder in women during MMH.

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Table 1: Sample size for each variable

Muscle	**N women**	**N men**
Anterior deltoid	24	26
Lateral deltoid	25	26
Posterior deltoid	25	26
Pectoralis major	15	17
Latissimus dorsi	16	16
Supraspinatus	11	10
Infraspinatus	11	10
Subscapularis	11	9
Biceps brachii	25	26
Triceps brachii	24	24
Composite variables		
$SumEMG_{delt}$	24	26
Muscle $Focus_{delt}$	22	24
$SumEMG_{sEMG}$	14	16
Muscle $Focus_{sEMG}$	13	16
$SumEMG_{1joint}$	11	9
Muscle $Focus_{1joint}$	9	8

Table 2: Composite variable descriptions

	Muscles	**Justification**
$SumEMG_{delt}$, MF_{delt}	DeltA, DeltL, DeltP	- Uniarticular muscles accessible by surface EMG; - Greatest sample size.
$SumEMG_{sEMG}$, MF_{sEMG}	DeltA, DeltL, DeltP, Pect, Lat	- Uniarticular and multiarticular muscles accessible by surface EMG.
$SumEMG_{1joint}$, MF_{1joint}	DeltA, DeltL, DeltP, Supra, Infra, Subscap	- Uniarticular muscles accessible by surface and intramuscular EMG.

MF : Muscle Focus; SumEMG : Sum of EMG signals; DeltA: Anterior deltoid; DeltL: Lateral deltoid; DeltP: Posterior deltoid; Pect: Pectoralis major; Lat: Latissimus dorsi; Supra: Supraspinatus; Infra : Infraspinatus; Subscap: Subscapularis.

Figures

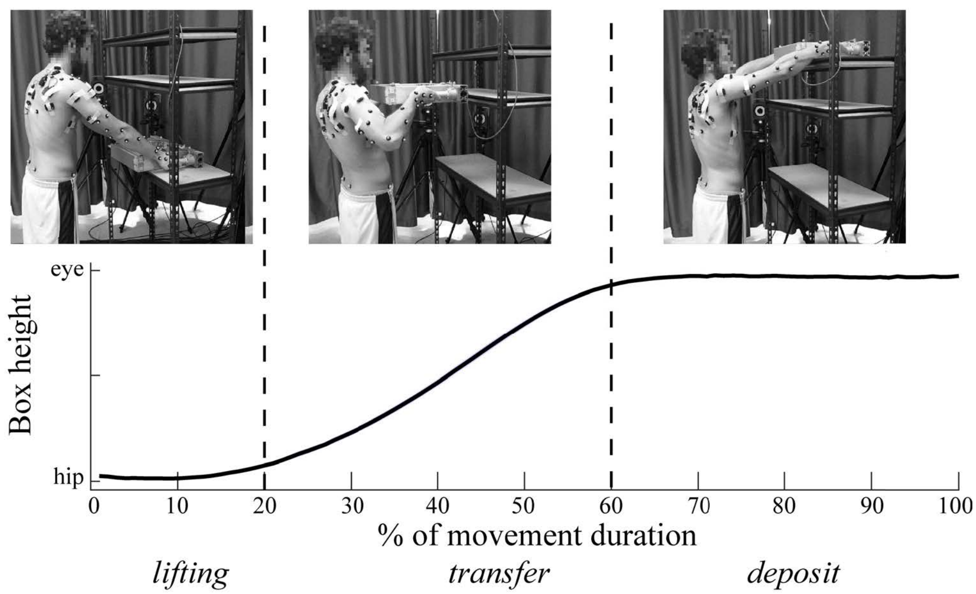

Figure 1: Experimental set up and representative subject during the box lifting task.

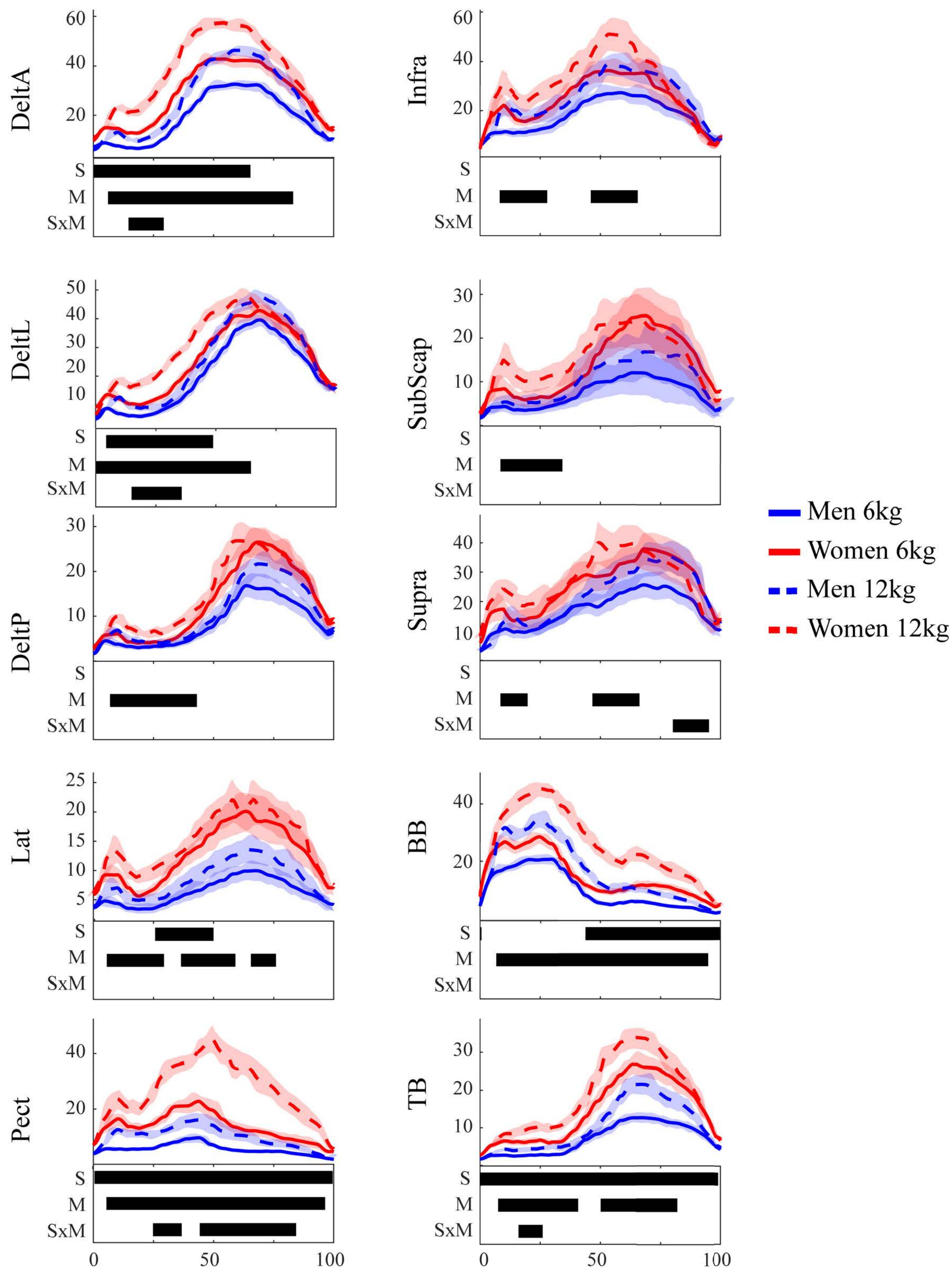

Figure 2: Individual muscle activations when men and women lifted a 6 or a 12 kg box. The dark bands

in the panel under each EMG waveform present clusters of statistically significant main effects of sex (S), mass (M) or sex x mass interactions (SxM).

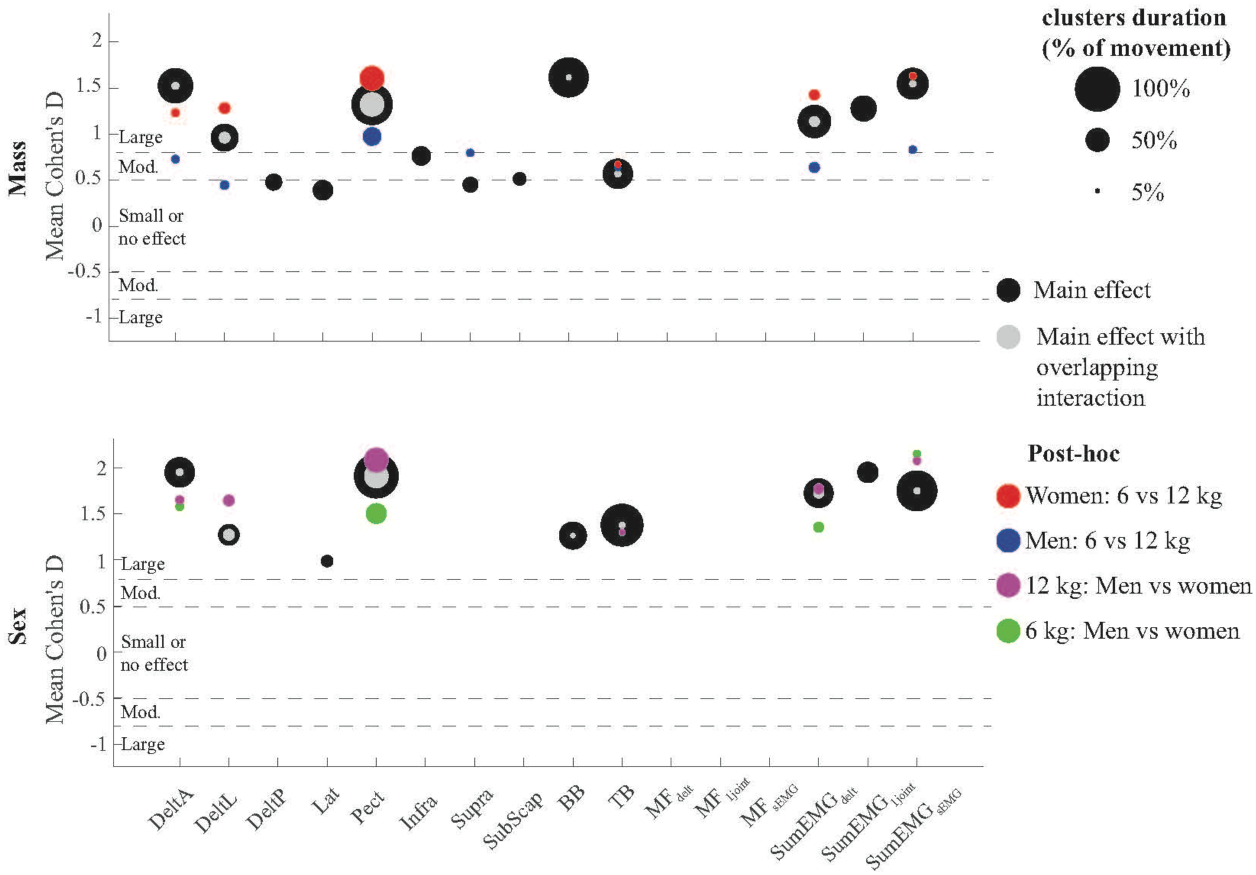

Figure 3: Effect size averaged across significant clusters and effect duration for each statistical comparison. For main effects of sex and mass, SD was pooled across all conditions. For post-hoc tests, SD was pooled across conditions compared.

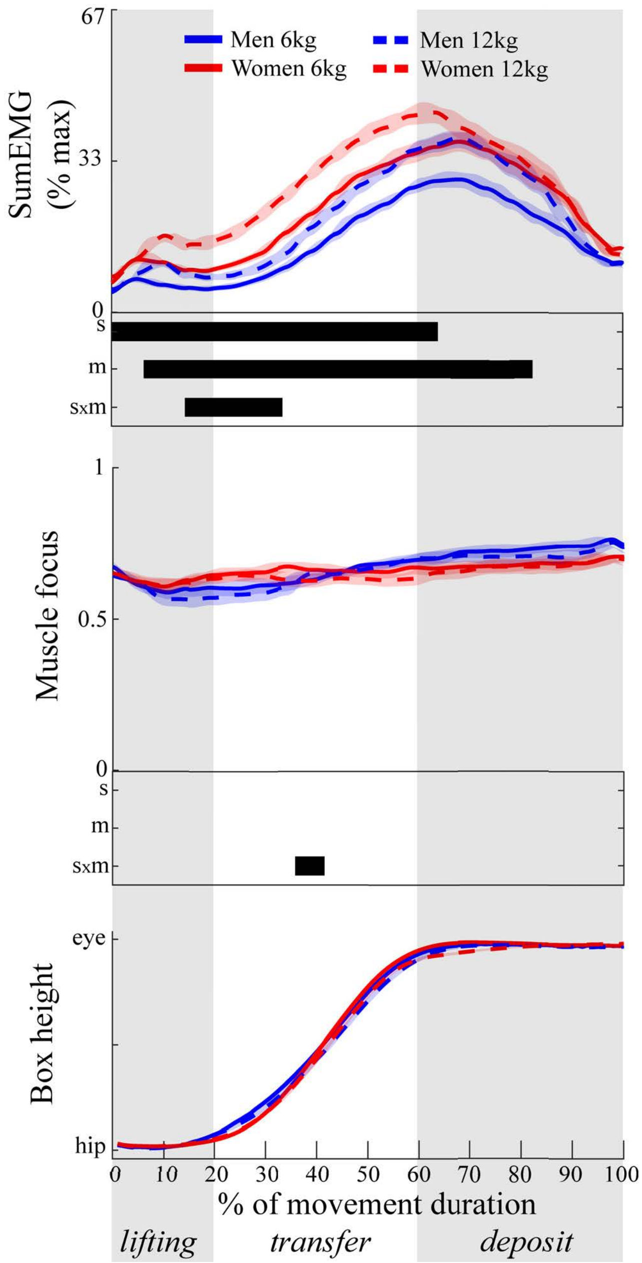

Figure 4: SumEMG$_{delt}$ and MF$_{delt}$ when men and women lifted a 6 or a 12 kg box. The dark bands in the lower panel under each EMG waveform present clusters of statistically significant main effects of sex (S), mass (M) or sex x mass interactions (SxM).

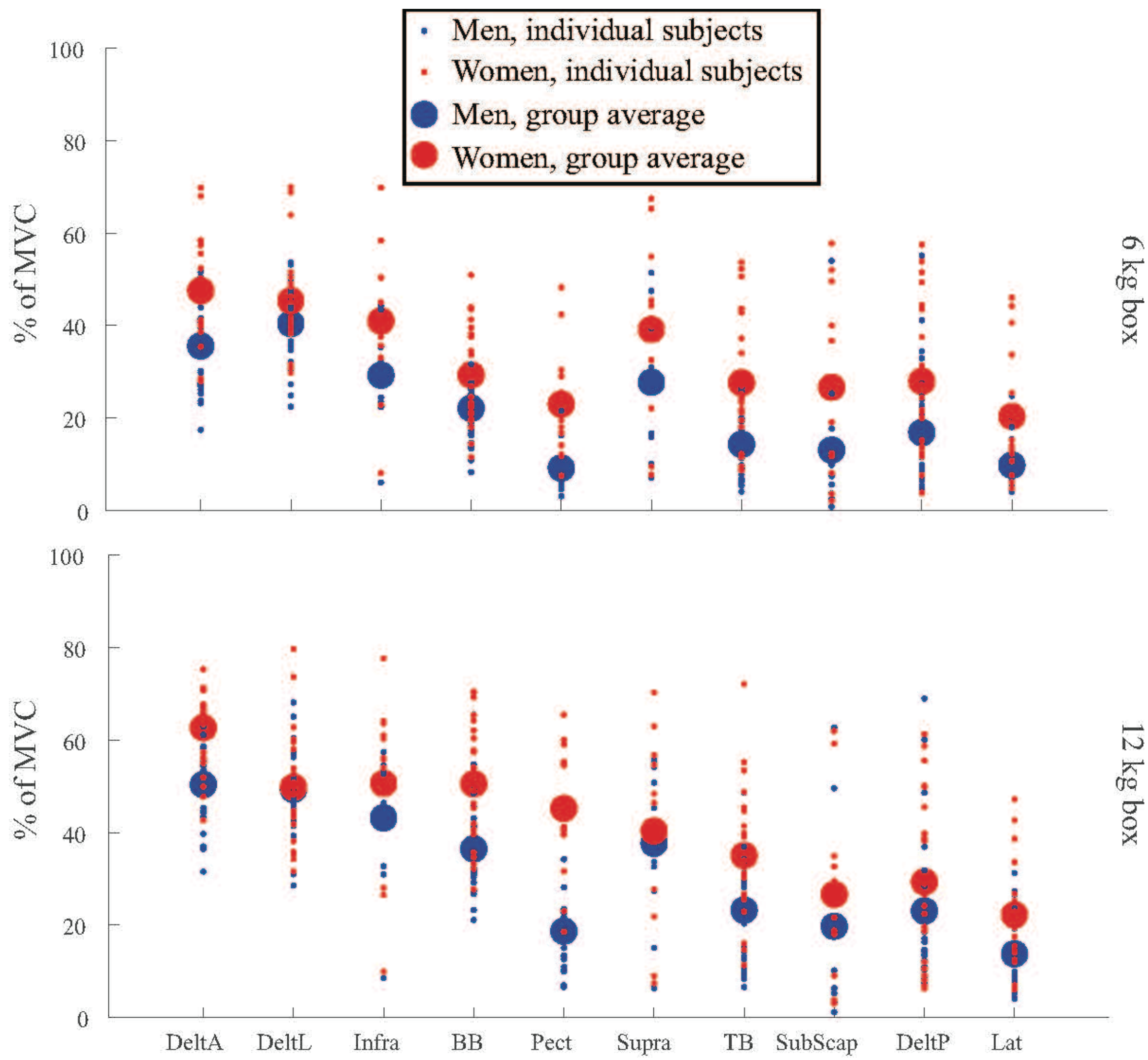

Figure S1: Individual muscles' 90th percentile activation.

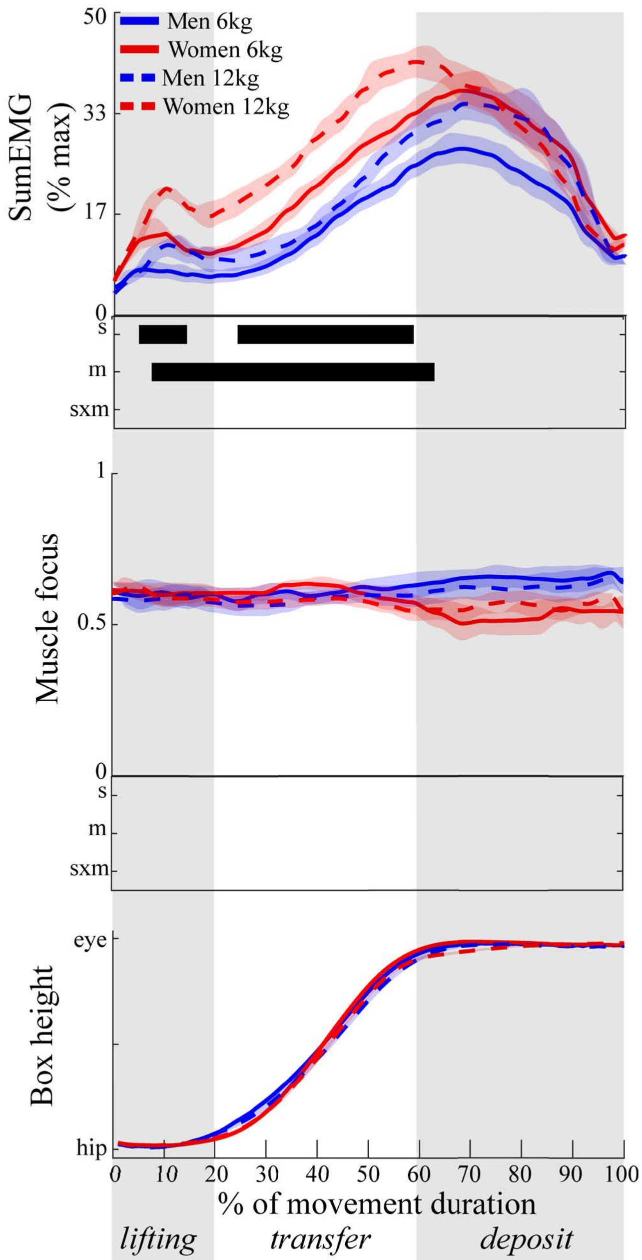

Figure S2: SumEMG$_{1joint}$ and MF$_{1joint}$ when men and women lifted a 6 or a 12 kg box. The dark bands in the lower panel under each EMG waveform present clusters of statistically significant main effects of sex (S), mass (M) or sex x mass interactions (SxM).

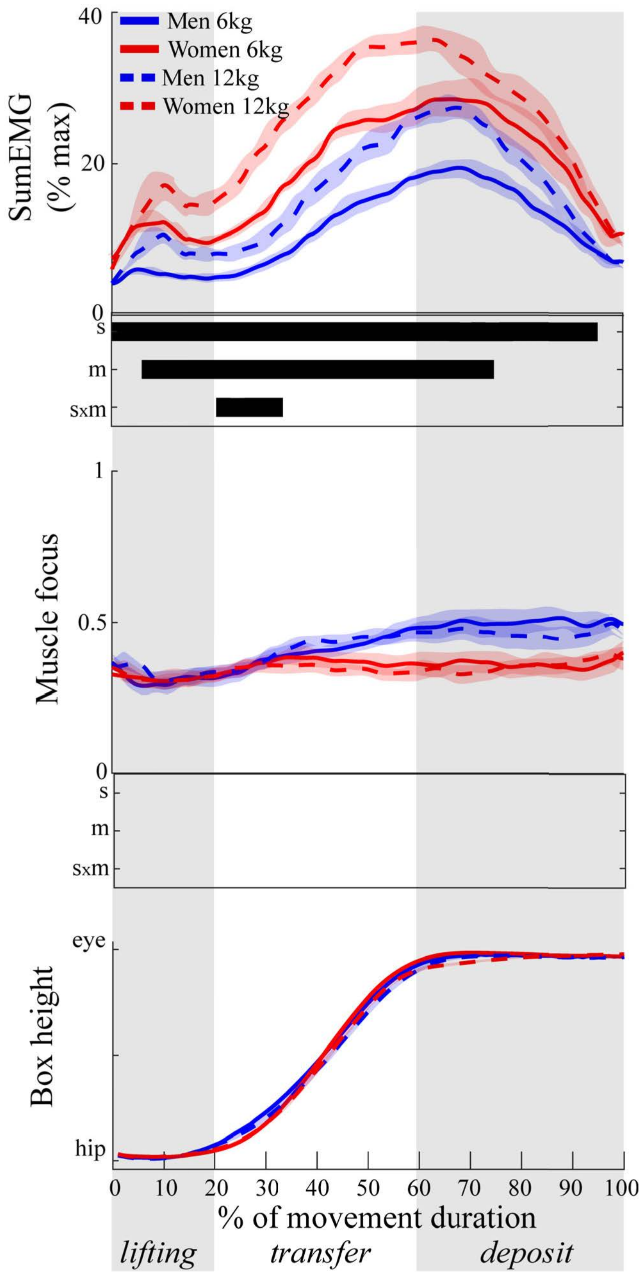

Figure S3: SumEMG$_{sEMG}$ and MF$_{sEMG}$ when men and women lifted a 6 or a 12 kg box. The dark bands in the lower panel under each EMG waveform present clusters of statistically significant main effects of sex (S), mass (M) or sex x mass interactions (SxM).

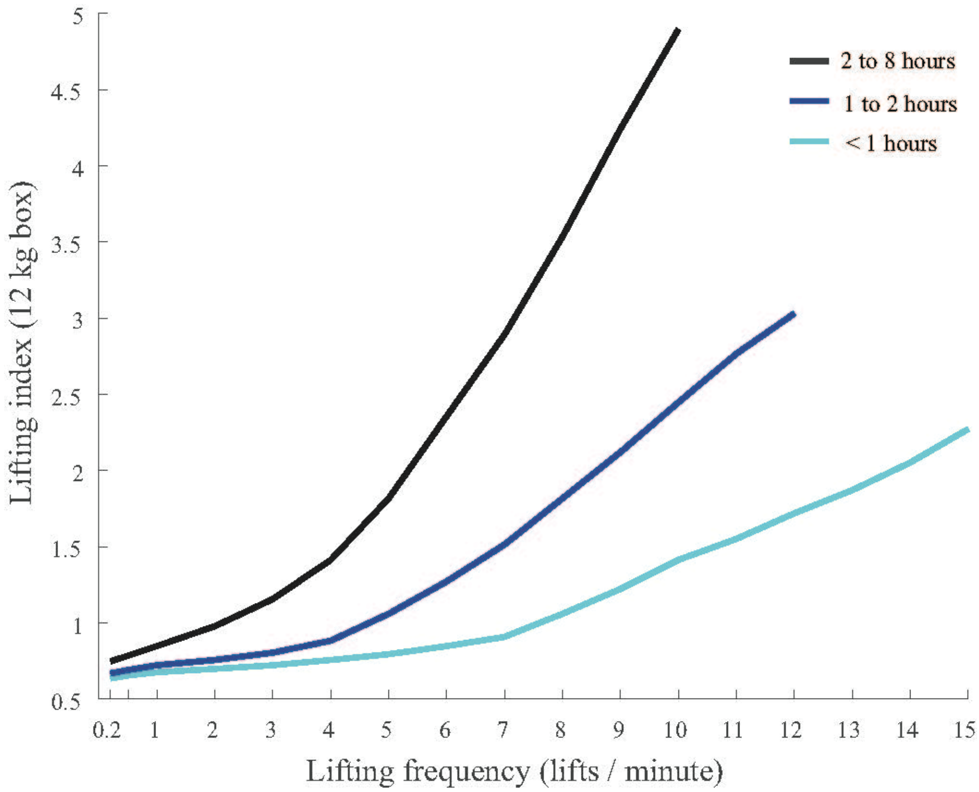

Figure S4: NIOSH revised lifting index for the 12 kg box as a function of lifting frequency and duration. The following parameters were used: Vertical distance from floor (origin) = 100 cm, travel distance (group average) = 78 cm, horizontal distance from center of mass = 25 cm, angle of asymmetry = 0°, coupling factor = 1.