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A comparison of lumbar spine and muscle loading between male and female workers during box transfers

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ABSTRACT

There is a clear relationship between lumbar spine loading and back musculoskeletal disorders in manual materials handling. The incidence of back disorders is greater in women than men, and for similar work demands females are functioning closer to their physiological limit. It is crucial to study loading on the spine musculoskeletal system with actual handlers, including females, to better understand the risk of back disorders. Extrapolation from biomechanical studies conducted on unexperienced subjects (mainly males) might not be applicable to actual female workers. For male workers, expertise changes the lumbar spine flexion, passive spine resistance, and active/passive muscle forces. However, experienced females select similar postures to those of novices when spine loading is critical. This study proposes that the techniques adopted by male experts, male novices, and females (with considerable experience but not categorized as experts) impact their lumbar spine musculoskeletal systems differently. Spinal loads, muscle forces, and passive resistance (muscle and ligamentous spine) were predicted by a multi-joint EMG-assisted optimization musculoskeletal model of the lumbar spine. Expert males flexed their lumbar spine less (avg. 21.9° vs 30.3-31.7°) and showed decreased passive internal moments (muscle avg. 8.9 % vs 15.9-16.0 %; spine avg. 4.7 % vs 7.1-7.8 %) and increased active internal moments (avg. 72.9 % vs 62.0-63.9 %), thus producing a different impact on their lumbar spine musculoskeletal systems. Experienced females sustained the highest relative spine loads (compression avg. 7.3 N/BW vs 6.2-6.4 N/BW; shear avg. 2.3 N/BW vs 1.7-1.8 N/BW) in addition to passive muscle and ligamentous spine resistance similar to novices. Combined with smaller body size, less strength, and the sequential lifting technique used by females, this could potentially mean greater risk of back injury. Workers should be trained early to limit excessive and repetitive stretching of their lumbar spine passive tissues.

Keywords: musculoskeletal modeling; expertise; sex; manual materials handling; lifting; lumbar spine; muscle forces; joint forces; EMG; optimization

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1. Introduction

A clear relationship between lumbar spine loading and back musculoskeletal disorders (MSD) in manual materials handling (MMH) is supported by the National Research Council (2001). It is reported that almost 10 % of the workforce in Québec, Canada, experience back MSD interfering with their activities (Stock et al., 2011). According to the same authors, the incidence of back MSD is greater in women (11.3 %) than men (7.6 %). As females are smaller in body size and not as strong as males, they are functioning closer to their physiological limit than males in situations where physical work demands are the same for all employees (Stock et al., 2011; Côté, 2012). Thus, it is clear that the results of MMH studies conducted on males cannot be extrapolated to females.

To better understand the risk of back MSD in this field, it is crucial to study loading on the spine musculoskeletal system with actual handlers of both sexes. Thus far, results obtained from male handlers indicate that expertise or experience in MMH is a critical factor which influences lumbar spine flexion and passive spine resistance, as well as active and passive muscle contributions (Gagnon et al., 2016; Plamondon et al., 2014a). Moreover, Plamondon et al. (2014b) observe different interjoint coordination in female MMH workers. They adopt a sequential motion (knee extension, then hip and back) while expert males show a more synchronized movement. In addition, females select a posture similar to novices at the instant of peak lumbosacral joint moment, and keep the same interjoint (sequential) coordination even when the external load is adjusted to their body size and strength (Plamondon et al., 2017). Indeed, the few other studies comparing males and females for spinal loading during MMH (Marras et al., 2002, 2003) are based on data collected with unexperienced subjects, not representative of actual workers.

The prediction of coherent muscle forces and spinal loads with sufficient biological integrity requires a full multiple-joint musculoskeletal EMG-driven model of the lumbar spine (Arjmand et al., 2007; Gagnon et al., 2011, 2016; Stokes & Gardner-Morse, 1995). The capability of the model to predict individual muscle strategies (including coactivation) while respecting mechanical criteria is crucial, an

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accomplishment which is attainable by EMG-driven approaches (Cholewicki et al., 1995; Gagnon et al., 2001). To our knowledge, there has been no study with such models on actual female handlers.

Based on a series of simulated box transfers, the present study proposes that the techniques adopted by male handlers (experts and novices) and female handlers (with considerable experience but not categorized as experts) impact their lumbar spine musculoskeletal systems differently. A multiple-joint EMG-assisted optimization lumbar spine musculoskeletal model (Gagnon et al., 2011, 2016) was used to predict spinal load, muscle force, and passive spine resistance. It is hypothesized that, when compared to other groups, experts employ safer work techniques requiring (RH_1) less passive muscle force and spine resistance (RH_2) but more active muscle force, thus resulting in (RH_3) smaller relative lumbar spine joint loads.

2. Materials and methods

2.1 Experimental study

Details on data collection and processing are described elsewhere (Plamondon et al., 2010, 2014a,b). Ten males categorized as expert handlers (age 39.1 yr. SD 10.0; mass 71.8 kg SD 9.5; height 1.72 m SD 0.08; experience 15 yr. SD 9.3), 10 females with work experience (age 40.7 yr. SD 9.4; mass 65.6 kg SD 10.1; height 1.63 m SD 0.08; experience 7 yr. SD 2.3), and 10 males categorized as novice handlers (age 23.3 yr. SD 3.2; mass 69.0 kg SD 7.3; height 1.74 m SD 0.05; experience 0.5 yr. SD 0.4) with entire EMG dataset were retained for this study. As explained in Plamondon et al. (2014b), the females selected for this study did not meet the criterion of low lifetime incidence of back injury and so could not be categorized as experts, but they had much more work experience than novices and none had musculoskeletal problems that could have affected their work. Two box transfer tasks were selected (**Fig. 1**) to allow group (expert vs female vs novice), destination height (ground level vs top of the pile), and phase (lift vs carry vs deposit) comparisons. Fifteen-kg boxes were transferred from a conveyor (12 cm

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from the ground) to a hand trolley at H1 (2 cm from the ground) and H4 (98 cm from the ground) in the 90° orientation. A total of 30 subjects performed 2 reps for 2 tasks, resulting in the analysis of 120 trials and 6490 postures (Experts: 2218, avg. 55 per trial; Females: 2264, avg. 57; Novices: 2008, avg. 50). Work technique and speed were selected by the individual participant. A 2-min. rest was allowed after each block of 8 round trips to prevent fatigue. A large in-house-designed force platform recorded ground reaction forces at 1024 Hz. A 4-sensor system (OptoTrak, NDI, Waterloo, Canada) tracked 48 markers attached to 12 rigid clusters at 30 Hz to get 3D kinematics.

2.2 Lumbar spine musculoskeletal model

Analyses involved a 76-muscle lumbar spine musculoskeletal model **(Fig. 2;** Gagnon et al., 2011, 2016; Arjmand et al., 2009, 2010): (1) this model is coherent from a biomechanical standpoint as it resolves muscle forces to satisfy equilibrium at all crossed lumbar joints simultaneously; (2) at the same time, this model is EMG-driven, so it attempts to keep the biological integrity of the system; (3) the model was aligned to each subject in the neutral posture (anatomical position) using the measured location of T12-L1 and L5-S1 joint centers; (4) the kinematics of this model is driven by the measured 3D kinematics of the subject; and (5) the net joint moments and forces of the subject obtained by 3D inverse dynamics are input to this model. In other words, these variables, used as inputs in the EMG-driven model, are allowed to change according to sex differences in motion (Plamondon et al., 2014b) and trunk muscle activation strategies. Superficial muscle activity was measured bilaterally by 6 pairs of active surface electrodes (DelSys, Boston, USA) to serve as input to the model: longissimus pars thoracis (LGPT), iliocostalis pars thoracis (ICPT), multifidus (MUF-L1), rectus abdominis (RA), and both obliques (EO, IO) (Gagnon et al., 2011). Other superficial and deep muscle fascicles (32 pairs) were also part of the model: multifidus (MUF), longissimus pars lumbaris (LGPL), iliocostalis pars lumbaris (ICPL), quadratus lumborum (QLO), and iliopsoas (IP).

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Active (contractile) and passive muscle forces were estimated before optimization (Christophy et al., 2012, based on Schutte, 1993 and Thelen, 2003). Muscle contraction velocity was ignored. Active and passive muscle forces were weighted by standardized coefficients as a function of fascicle length. The active muscle force was converted to active muscle moment (MAmus) using this coefficient and normalized EMG based on maximum voluntary contractions (Arjmand et al., 2010; Larivière et al., 2001). The EMG signal recorded from surface sites initialized the activity of deep local muscles (McGill et al., 1996). Passive muscle force was directly converted to passive muscle moment (MPmus). Maximum allowable stress in muscles was set to 0.7 MPa. The passive resistance of ligamentous spine (MPcol; discs and ligaments) was estimated using relationships between lumbar spine flexion and passive spine moment about each anatomical axis (Shirazi-Adl, 2006).

2.3 Multi-joint EMG-assisted optimization (M-EMGAO)

The M-EMGAO method (Gagnon et al., 2011, 2016) deals with the redundancy in the lumbar spine musculoskeletal model. The procedure partitions in all muscles and tissues simultaneously, the external moments (Mnet) acting about the six lumbar joints (T12-L1 to L5-S1). The approach seeks to minimize the following objective function:

$$min \sum_{i=1}^{176} Mnorm_i \, (1 - g_i)^2 \qquad (1)$$

with:

$$Mnorm_i = \sqrt{\sum_j Mr_{i,j}^2} \qquad (2)$$

and:

$$Mr_{i,j} = \sqrt{M_{L(i,j)}^2 + M_{S(i,j)}^2 + M_{T(i,j)}^2} \qquad (3)$$

The problem is constrained by three equality equations at each joint:

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$$\begin{cases} \sum_{i=1}^{N_j} \left(g_i MAmus_{L(i,j)} + g_i MPmus_{L(i,j)}\right) + g_i MPcol_{L(j)} \pm g_i Merr_{L(j)} - g_i Mnet_{L(j)} = 0 \\ \sum_{i=1}^{N_j} \left(g_i MAmus_{S(i,j)} + g_i MPmus_{S(i,j)}\right) + g_i MPcol_{S(j)} \pm g_i Merr_{S(j)} - g_i Mnet_{S(j)} = 0 \\ \sum_{i=1}^{N_j} \left(g_i MAmus_{T(i,j)} + g_i MPmus_{T(i,j)}\right) + g_i MPcol_{T(j)} \pm g_i Merr_{T(j)} - g_i Mnet_{T(j)} = 0 \end{cases} \quad (4)$$

subjected to the following bounds:

$$\begin{cases} \left(\frac{MAmusMax_i}{MAmus_i}\right) \geq g_i \geq 0.5 \text{ (or 0) for MAmus} \\ 1.05 \geq g_i \geq 0.5 \text{ for MPmus} \\ 1.05 \geq g_i \geq 0.95 \text{ for MPcol} \\ 1.05 \geq g_i \geq 0.95 \text{ for Mnet} \\ 1 \geq g_i \geq 0 \text{ for Merr} \end{cases} \quad (5)$$

Subscript j is for the lumbar joint and subscripts L, S, T indicate the anatomical longitudinal, sagittal and transverse local axes. The longitudinal (L) axis points from the center of the upper endplate of the lower vertebra to the center of the lower endplate of the upper vertebra. The transverse (T) axis is perpendicular to L and points to the left of the subject. The sagittal (S) axis is normal to the plane formed by the two other axes and points anteriorly. Subscript i refers to the following moments: MAmus (1-76), MPmus (77-152), MPcol (153-158), Mnet (159-164) and Merr (165-176). Within their respective bounds (Eq. 5), the least possible adjustment is applied to the initial moments in MAmus, MPmus, MPcol, Mnet and Merr (adjustment error) to minimize the sum of the moment norm (objective function in Eq. 1) acting on the lumbar spine (T12-S1). The problem is constrained by three equalities (Eq. 4) repeated for six joints (18 equations). For each equality equation (Eq. 4), the sum of all internal moments (MAmus, MPmus, MPcol) and adjustment error (±Merr because in each direction) minus the external moment (Mnet) is zero. Thus, to meet each equality constraint, Mnet must be balanced by internal moments (MAmus, MPmus, MPcol) within their bounds (Eq. 5), including some adjustment error (Merr). Each gain g can be adjusted (±5 %) by the optimization, except for Merr, MAmus, and MPmus (lower bound).

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For Merr, an iterative procedure increments its value by 1 Nm until convergence. For MAmus, the 12 EMG-measured fascicles cannot be zeroed because they have a lower bound of 0.5 (Gagnon et al., 2001, 2011; Zheng et al., 1998) to limit modification of these active muscle moments to 50 %. The lower bound is set to zero for the 64 other fascicles so that the optimization can use the full force range (0-100 %) to converge. To be coherent with the MAmus lower bound, the estimated passive muscle moment (MPmus) of any fascicle can be reduced by no more than 50 %. Maximal active muscle moment about a lumbar joint for a given posture is represented by MAmusMax (Eq. 5). Optimization problems were solved by quadratic programming (*quadprog*, MATLAB optimization toolbox, MathWorks, Natick, MA, USA).

2.4 Statistical analyses

Three independent variables were considered: worker group (G: experts vs females vs novices), box destination height (H: H1 ground vs H4 top of the pile), and box transfer phase (P: lift vs carry vs deposit). Data were time normalized to flight time (box completely supported by hands). Normalized time from 0 to 10 % corresponded to the lift phase, 45-55 % to carry, and 90-100 % to deposit. For each subject, the results from two reps per task were averaged out. Three-way between-within-within factorial ANOVA (G × H × P) was used to obtain the main effects, interactions and their effect size (Bakeman, 2005: 0.02 is a small effect, 0.13 is medium, and 0.26 is large). Post hoc Bonferroni adjusted pairwise comparisons were conducted for group (t-test with pooled SD) and phase (paired t-test). Alpha was set to 0.05 and all tests were run in R (R Core Team, 2017).

Statistical analyses involved three collections of dependent variables (DV). Within each phase, the value of each DV was extracted at the instant of maximal *resultant* external moment (Mnet_R) at the L5-S1 joint. The first set of DV contains normalized joint forces at L5-S1: compression (Fcomp), posterior-anterior shear (Fshear PA), and medial-lateral shear force (Fshear ML) acting respectively along local L, S, and T axes. These forces were normalized to body weight (BW) to account for anthropometric differences between males and females (Plamondon et al., 2014b). Additionally, lumbar spine flexion

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angle and external moment at L5-S1 were included in this collection. The second set of DV includes normalized internal moments (MAmus, MPmus, and MPcol) which are acting to counterbalance the flexion-extension (about T axes) external moment (Mnet) at L5-S1. These internal moments were normalized to remove unwanted between-subject variance in the data (**Fig. 3**). The active muscle moment (MAmus) was further subdivided between two functional moments: the agonist active extensor moment (conserved in MAmus) and the antagonist active flexor moment (new variable MAmusa). The third set of DV comprises the normalized muscle moments (individually and grouped). Agonist muscles within MAmus were split up between global (Back G are ICPT and LGPT) and local (Back L are LGPL, MUF, ICPL, and QLO) extensors. Passive moments (MPmus) were detailed in the same way. Similar procedures were utilized for global antagonist flexor muscles within MAmusa (Abdo are RA, EO, and IO).

3. Results

For concision, only statistically significant ($p < 0.05$) and marginally significant ($0.05 \leq p < 0.10$; see Limitations) main effects on the group factor are presented. However, all factors are illustrated on figures to contextualize group differences. Detailed muscle moment results (Electronic Supplementary Material: *Table 2* and *Fig. 7-8*) and all interactions of group with height and phase are presented in the Electronic Supplementary Material.

3.1 Joint forces with flexion angle and external moment (Table 1.A and Fig. 4)

Compression and PA shear forces were different between groups. On average, compression and PA shear were up to 1.1 and 0.6 N/BW greater, respectively, in females. Using average female BW, these values convert to 707 N in compression and 385 N in PA shear. Generalized effect size for group was much stronger for PA shear (0.22), approaching a large effect size. Joint forces were not different between the males (experts and novices) and were noticeably larger at lift time as well as when

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destination height was H1. These observations coincide with smaller lumbar flexion for experts and smaller external moments for females when the box was close to the ground.

3.2 Internal moments (Table 1.B and Fig. 5)

Active internal moment for the agonists (MAmus) differed between experts and others. Experts showed on average up to 11 % more active internal moment. There was a concomitant marginal decrease in both sources of passive resistance (MPmus, MPcol) for experts. Internal moments were not different between females and novices for these variables.

3.3 Muscle moments (Table 1.C and Fig. 6)

Involvement of global extensors was different between groups. On average, experts increased the active extensor moments (Back G_{ac}) of global muscles and reduced their passive contribution (Back G_{pc}). Effect size was stronger for passive (0.17) than active extensors (0.11). The same pattern was observed for local extensors (Back L_{ac}, Back L_{pc}), although these differences were marginal. The magnitude of the gap between experts and the others was always more marked when the box was close to the ground (H1, Fig. 6) except for Back L_{ac} in the carry phase.

3.4 Detailed muscle moments: Global and local extensors (Electronic Supplementary Material: Table 2.D-E and Fig. 7)

One global extensor (LGPT, active and passive) followed the pattern explained above for experts: more active combined with less passive global muscle demand. One local extensor ($ICPL_{ac}$) followed the pattern of the active global extensors, so there was more demand on this active local muscle for experts. Marginal differences were observed for other variables, all showing the same trade-off pattern between active contributions ($ICPT_{ac}$) and passive resistance ($ICPT_{pc}$, $LGPL_{pc}$, $ICPL_{pc}$, MUF_{pc}).

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3.5 Detailed muscle moments: Global flexors (Electronic Supplementary Material: Table 2.F and Fig. 8)

There was no significant group effect but RA contribution was amplified for females when the destination was H4 (Fig. 8, clearly visible during carry and deposit).

4. Discussion

Based on the tasks investigated, our results confirm that the MMH techniques used by the three handler categories impacted their lumbar spine musculoskeletal systems differently by the way internal moments were distributed. Larger normalized joint forces were sustained by females (RH_3 partly supported). Overall, experts exerted more active muscle force (RH_1 supported) than females and novices, who both relied on additional passive resistance from the muscles to counterbalance the external load (RH_2 supported). For females and novices, the significant extra passive resistance came from global and local extensor muscles.

4.1 Joint forces with lumbar flexion angle and external moment

Compression and PA shear forces, once normalized by BW to correct for anthropometric differences, were larger in females. However, as observed in our previous study (Gagnon et al., 2016), joint forces did not differ between the males (experts vs novices). At the same time, the external moments supported by the females were smaller, especially at lift time. To verify that the picture was coherent with absolute force values, average normalized joint forces were converted back to N using the average BW of each group: for females, we get 4673 and 1477 N for compression and shear respectively, for experts 4516 and 1201 N, and for novices 4200 and 1238 N. These absolute force differences are not statistically significant, since the small number of subjects resulted in a lack of sufficient statistical power. This indeed shows that the females, with their smaller body size, still sustained larger absolute forces, putting them at a higher risk of injury. Across all phases of box transfers, females sustained more normalized joint forces, with the largest gap between them and the males observed in the carry phase. During the

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carry phase, the trunk was closest to an upright posture (lumbar flexion < 15° when box destination was H4) and more antagonist activity from RA_{ac} was generated (Electronic Supplementary Material: **Fig. 8**). This additional abdominal activity, also greater for EO_{ac}, IO_{ac} and $Abdo_{ac}$ but not significantly so, might be seen as a way to further increase trunk stability in females at the cost of extra joint force. The need for more trunk stability in females might be related to strength differences between males and females, since all workers were transferring the same 15-kg boxes, as routinely happens in the workplace. It is worth mentioning that the detection of such antagonist muscle activity might not be feasible without an EMG-driven model able to respect both the biological and mechanical integrity of the musculoskeletal system (Reeves & Cholewicki, 2003).

4.2 Internal moments

Observations made previously for novices' internal moments (Gagnon et al., 2016) are corroborated in the present study, but this time for females and novices: they both bent their lumbar spine more than experts and therefore created more demand on the passive portion of the lumbar spine extensors as well as on the passive ligamentous spine. In other words, to counterbalance the net (external) moment, the females and novices squeezed/bent their intervertebral discs more and stretched their extensor muscles and lumbar ligaments more. The most critical spine loading events happened when the box was close to the ground, either at lift or deposit time, thus corroborating (with a subset of the same subjects doing box transfers in the present study) the interpretation of Plamondon et al. (2014b, 2017) for box palletizing.

Global extensors, taken together (Back G) and individually (LGPT and ICPT), respected the additional use of active contributions for experts as well as the extra use of passive contributions for females and novices. The picture for local extensors (Back L) is somewhat different, showing extra passive contribution by most local extensors in females and novices (LGPL, ICPL, and MUF) but less so for experts (ICPL). In other words, experts recruited mainly their active global extensors to balance the

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external moment, while females and novices stretched all their extensors (global and local) more to attain mechanical equilibrium. As emphasized before (Gagnon et al., 2016, Plamondon et al., 2010, Dolan et al., 1994), the strategy adopted by females and novices might be advantageous from the point of view of energy transfer: stretched tissues and squeezed/bent discs store energy during flexion and return some of it during extension, at the cost of added risk of injury. However, cumulative muscle fatigue as well as repeated/sustained passive tissue stretching could degrade the situation (threat to lumbar spine stability) (Solomonow et al., 1999). Moderate lumbar flexion, a strategy followed to some extent by experts who were flexing their lumbar spine less and relying more on active muscle contributions, appears a better way to get some energy return while preventing excessive stress on lumbar discs (Adams et al., 2002). Another benefit of this latter strategy is the preservation of a safety margin for passive tissues.

4.3 Limitations

Issues concerning the subjects (injuries, age, number, expertise), the work context (laboratory vs field work) and biomechanical results (human and instrument errors, use of surrogate EMG) that were addressed in Plamondon et al. (2010, 2014ab, 2017) and Gagnon et al. (2011, 2016) are still relevant in the present study. Model scaling to the female anatomy is an additional issue. In the present study, however, there was no scaling of the model so the same musculoskeletal model of the lumbar spine was used for all subjects. Two main considerations support this decision: (1) sensitivity analyses on the effect of age, sex, body height (BH) and body weight (BW) on spinal loads (Ghezelbash et al., 2016) demonstrate that BW is by far the most influential factor on spinal loads (effect of BW is 98.9% in compression and 96.1% in shear; effect of sex is 0.7 % in compression and 2.1 % in shear); and (2) results of recent studies (Anderson et al., 2012; Ghezelbash et al., 2016) indicate that such scaling would increase spinal loads in females (for identical BH and BW, spinal loads in females are slightly greater than those in males by ~4.7% in compression and ~8.7% in shear), thus reinforcing the differences observed in the present study. Our experimental approach, designed to maximize external validity with the use of fixed

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load and height for all groups, is not intended to explain specifically the role of sex in the lifting technique, which would have required more internal validity in the adjustment of load to subjects' strength and in the adjustment of origin/destination heights to body stature. The consequence of this is that it is difficult to know if the difference observed is strictly due to sex or due to the differences in strength and/or body size of our participants. On the other hand, as males and females differ generally in strength and in body height, the results are a true representation of a real work context. From a measurement standpoint, the distinctive antagonist activity detected in females when compared to males implies the need for an EMG-driven musculoskeletal model to predict muscle forces. Even in a controlled laboratory environment, this requirement becomes a challenge: in the present study, 15 subjects out of 45 were dropped for technical reasons, resulting in a substantial loss of statistical power. Ultimately, this lack of statistical power may have affected a number of variables (Table 1; p = 0.06-0.08 for lumbar flexion, MPmus, MPcol, Back $L_{ac,pc}$, $ICPT_{ac,pc}$, $LGPL_{pc}$, $ICPL_{pc}$, MUF_{pc} and RA_{ac}) which presented trends coherent with the statistically significant variables.

The present results corroborate that expert MMH workers select safer handling techniques than females and novices by partitioning internal moments to reduce the impact on their lumbar spine musculoskeletal systems. Consequently, the experience of experts might be associated with safer handling practices, their limited use of passive tissues being consistent with their good back injury record. Besides, females sustained larger relative spine loads than novices as well as similar additional passive muscle moments and spine resistance. These observations confirmed that the sequential technique of females, combined with smaller body size and strength, point to a greater potential risk of back injury. In any case, MMH workers should be trained early to limit excessive and repetitive stretching of their lumbar spine passive tissues.

Conflict of interest statement

There is no conflict of interest in this study.

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References

1. Adams, M. A., Bogduk, N., Burton, K., Dolan, P., 2002. The biomechanics of back pain. Edinburgh: Churchill Livingstone.
2. Anderson, D. E., D'Agostino, J. M., Bruno, A. G., Manoharan, R. K., Bouxsein, M. L., 2012. Regressions for estimating muscle parameters in the thoracic and lumbar trunk for use in musculoskeletal modeling. Journal of Biomechanics, 45, 66-75.
3. Arjmand, N., Gagnon, D., Plamondon, A., Shirazi-Adl, A., Larivière, C., 2010. A comparative study of two trunk biomechanical models under symmetric and asymmetric loadings. Journal of Biomechanics, 43, 485-491.
4. Arjmand, N., Gagnon, D., Plamondon, A., Shirazi-Adl, A., Lariviere, C., 2009. Comparison of trunk muscle forces and spinal loads estimated by two biomechanical models. Clinical Biomechanics 24, 533-541.
5. Arjmand, N., Shirazi-Adl, A., Parnianpour, M., 2007. Trunk biomechanical models based on equilibrium at a single-level violate equilibrium at other levels. European Spine Journal, 16 (5), 701–709.
6. Bakeman, R., 2005. Recommended effect size statistics for repeated measures designs. Behavior Research Methods, 37(3), 379-384.
7. Cholewicki, J., McGill, S. M., Norman, R. W., 1995. Comparison of muscle forces and joint load from an optimization and EMG assisted lumbar spine model: towards development of a hybrid approach. Journal of Biomechanics, 28(3), 321-332.
8. Christophy, M., Senan, N. A. F, Lotz, J., O'Reilly, O. M., 2012. A musculoskeletal model for the lumbar spine. Biomechanical Modelling and Mechanobiology, 11, 19-34.
9. Coté, J.N., 2012. A critical review on physical factors and functional characteristics that may explain a sex/gender difference in work-related neck/shoulder disorders. Ergonomics 55, 173-182.

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Dépôt institutionnel

10. Dolan, P., Mannion, A. F., Adams, M. A., 1994. Passive tissues help the back muscles to generate extensor moments during lifting. Journal of Biomechanics, 27, 1077–1085.
11. Gagnon, D., Plamondon, A., Larivière, C., 2016. A biomechanical comparison between expert and novice manual materials handlers using a multi-joint EMG-assisted optimization musculoskeletal model of the lumbar spine. Journal of Biomechanics, 49, 2938-2945.
12. Gagnon, D., Arjmand, N., Plamondon, A., Shirazi-Adl, A., & Larivière, C., 2011. An improved multi-joint EMG-assisted optimization approach to estimate joint and muscle forces in a musculoskeletal model of the lumbar spine. Journal of Biomechanics, 44, 1521-1529.
13. Gagnon, D., Lariviere, C., Loisel, P., 2001. Comparative ability of EMG, optimization, and hybrid modelling approaches to predict trunk muscle forces and lumbar spine loading during dynamic sagittal plane lifting. Clinical Biomechanics 16, 359-372.
14. Larivière, C., Gagnon, D., Gravel, D., Arsenault, A. B., Dumas, J. P., Goyette, M., Loisel, P., 2001. A triaxial dynamometer to monitor lateral bending and axial rotation moments during static trunk extension efforts. Clinical Biomechanics 16, 80-83.
15. Marras, W. S., Davis, K. G., Jorgensen, M., 2003. Gender influences on spine loads during complex lifting. The Spine Journal, 3, 93-99.
16. Marras, W. S., Davis, K. G., Jorgensen, M., 2002. Spine loading as a function of gender. Spine, 27 (22), 2514-2520.
17. McGill, S., Juker, D., Kropf, P., 1996. Appropriately placed surface EMG electrodes reflect deep muscle activity (psoas, quadratus lumborum, abdominal wall) in the lumbar spine. Journal of Biomechanics 29 (11), 1503–1507.
18. National Research Council, 2001. Musculoskeletal disorders and the workplace: Low back and upper extremities. National Research Council and Institute of Medicine.

Dépôt institutionnel

19. Plamondon, A., Larivière, C., Denis, D., Mecheri, H., Nastasia, I, IRSST MMH research group, 2017. Difference between male and female workers lifting the same relative load when palletizing boxes. Applied Ergonomics, 60, 93-102.
20. Plamondon, A., Delisle, A., Bellefeuille, S., Denis, D., Gagnon, D., Larvière, C., 2014a. Lifting strategies of expert and novice workers during a repetitive palletizing task. Applied Ergonomics, 45, 471-481.
21. Plamondon, A., Larivière, C., Denis, D., St-Vincent, M., Delisle, A., 2014b. Sex differences in lifting strategies during a repetitive palletizing task. Applied Ergonomics, 45, 1558-1569.
22. Plamondon, A., Denis, D., Delisle, A., Larivière, C., Salazar, E., 2010. Biomechanical differences between expert and novice workers in a manual material handling task. Ergonomics 53, 1239-1253.
23. R Core Team, 2017. R: A language and environment for statistical computing. R Foundation for Statistical Computing, Vienna, Austria. URL https://www.R-project.org/.
24. Reeves, N. P., Cholewicki, J., 2003. Modeling the human lumbar spine for assessing spinal loads, stability, and risk of injury. Critical Reviews in Biomedical Engineering, 31, 73-139.
25. Schutte, L. M., Rodgers, M. M., Zajac, F. E., Glaser, R. M., 1993. Improving the efficacy of electrical stimulation-induced leg cycle ergometry: an analysis based on a dynamic musculoskeletal model. IEEE Transactions on Rehabilitation Engineering, 1 (2), 109-125.
26. Shirazi-Adl, A., 2006. Analysis of large compression loads on lumbar spine in flexion and in torsion using a novel wrapping element. Journal of Biomechanics, 39, 267-275.
27. Solomonow, M., Bing-He, Z., Baratta, R. V., Lu, Y., Harris, M., 1999. Biomechanics of increased exposure to lumbar injury caused by cyclic loading: part 1. Loss of reflexive muscular stabilization. Spine, 24, 2426–2434.
28. Stock, S., Funes, A., Delisle, A., St-Vincent, M., Turcot, A., Messing, K., 2011. Chapitre 7: Troubles musculo-squelettiques, Enquête québécoise sur des conditions de travail, d'emploi et de santé et de

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sécurité du travail (EQCOTESST). Institut de recherche Robert-Sauvé en Santé et en sécurité du travail (IRSST)- Rapport R-691, Montréal, Québec, Canada, pp. 445-530.

29. Stokes, I. A., Gardner-Morse, M., 1995. Lumbar spine maximum efforts and muscle recruitment patterns predicted by a model with multijoint muscle and joints with stiffness. Journal of Biomechanics, 28, 173-186.
30. Thelen, D.G., 2003. Adjustment of muscle mechanics model parameters to simulate dynamic contractions in older adults. ASME Journal of Biomechanical Engineering, 125(1):70–77.
31. Zheng, N., Fleisig, G. S., Escamilla, R. F., Barrentine, S. W., 1998. An analytical model of the knee for estimation of internal forces during exercise. Journal of Biomechanics 31 (10), 963-967.

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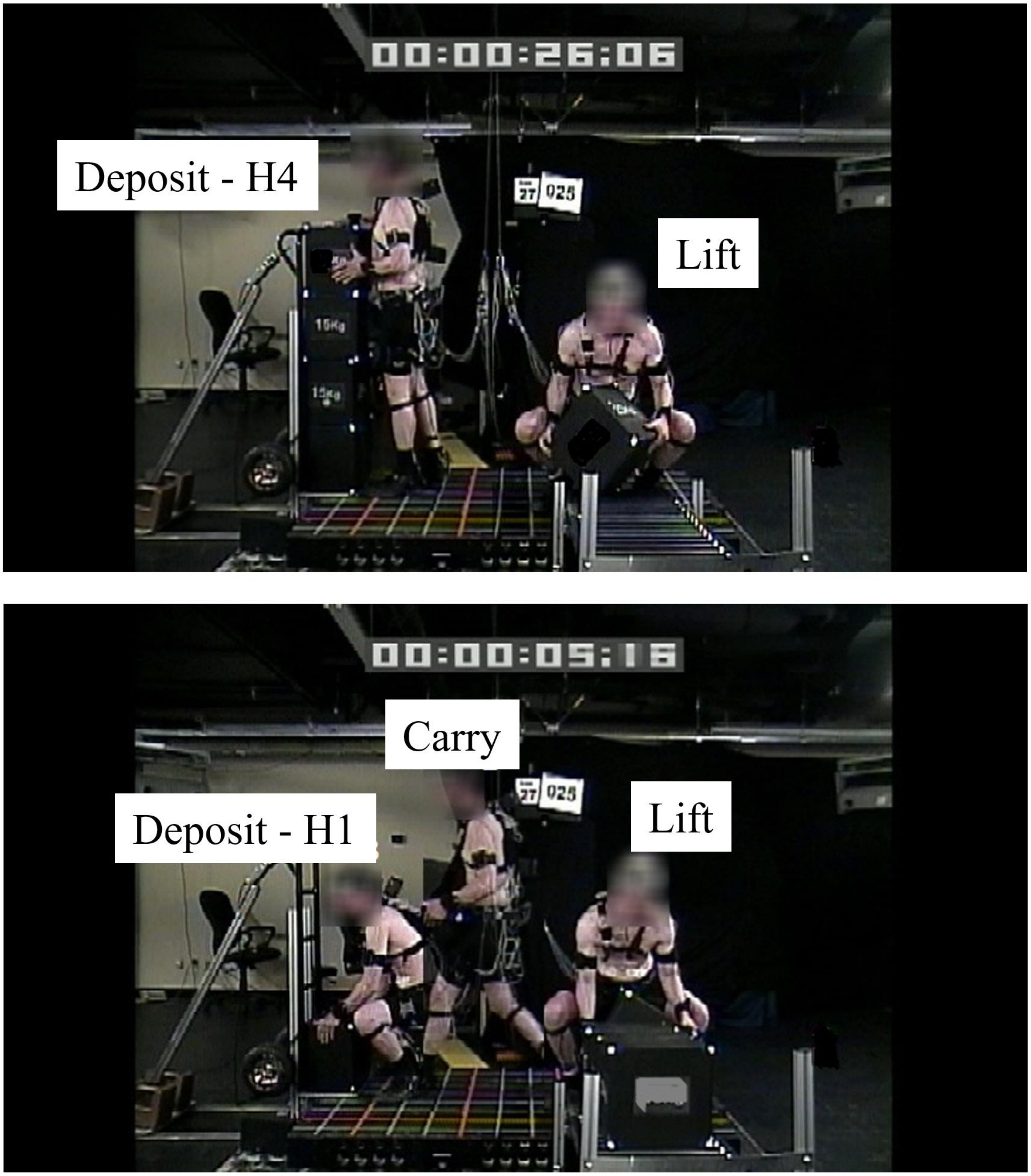

Fig. 1. Subject performing the tasks under study. The bottom image shows superposed snapshots of all phases of the transfer of a 15-kg box taken from the conveyor: Lift, Carry and Deposit on the hand trolley platform (H1). The top image shows only two of these phases: Lift and Deposit at the top of the pile (H4).

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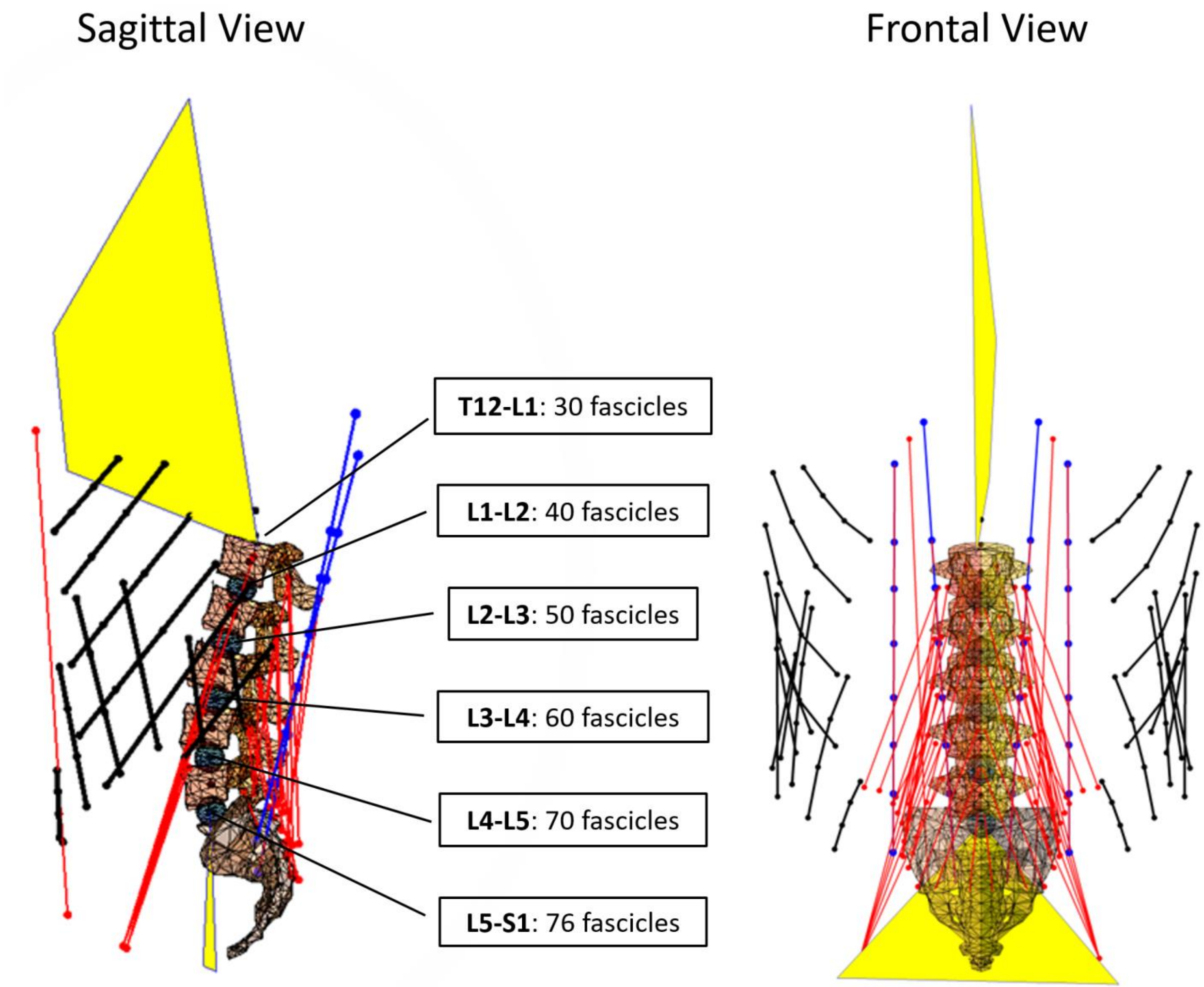

Fig. 2. The lumbar spine musculoskeletal model includes 30 global muscle fascicles crossing all 6 joints (T12-S1). Going down the spine, local muscle fascicles acting across the remaining lumbar joints (L1-S1: 10; L2-S1: 10; L3-S1: 10; L4-S1: 10; L5-S1: 6) are added to these global muscle fascicles until all 76 muscle fascicles intersect the lumbosacral joint (L5-S1).

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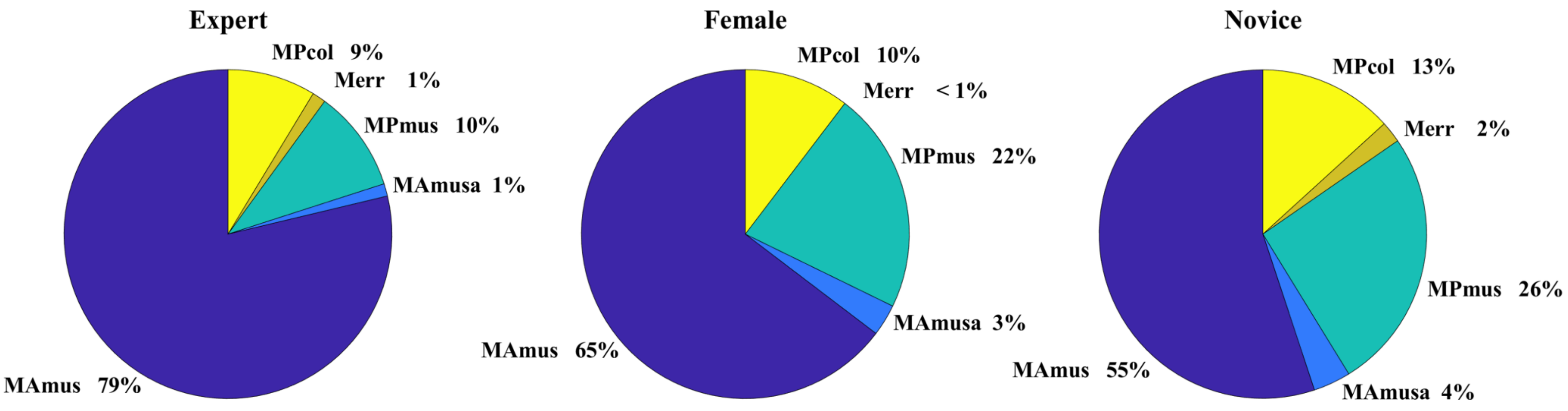

	MAmus	MAmusa	MPmus	Merr	MPcol	Total
Nm	168	3	21	3	19	214
%	79	1	10	1	9	100

	MAmus	MAmusa	MPmus	Merr	MPcol	Total
Nm	136	7	46	0	22	210
%	65	3	22	0	10	100

	MAmus	MAmusa	MPmus	Merr	MPcol	Total
Nm	131	9	62	5	32	238
%	55	4	26	2	13	100

Fig. 3. Comparative pie chart distribution of the L5-S1 normalized internal flexion-extension moments between typical Expert (mass: 74 kg), Female (mass: 68 kg), and Novice (mass: 72 kg) workers during the Lift phase of a 15-kg box. In this illustrative example, the total internal flexion-extension moments are 214 vs 210 vs 238 Nm (Total: sum of absolute values) for Expert vs Female vs Novice, respectively. The active parts of the agonist muscles (MAmus) provide 79 vs 65 vs 55 % of this total moment, the passive ligamentous spine (MPcol) 9 vs 10 vs 13 %, the passive agonist muscles (MPmus) 10 vs 22 vs 26 %, and the active antagonist muscles (MAmusa) 1 vs 3 vs 4 %. The adjustment error (Merr) is less than 2 %. Model output (Nm) and normalized values (%) are provided in a summary table at the bottom of each chart. (For interpretation of the colors in this figure legend, the reader is referred to the web version of this article.)

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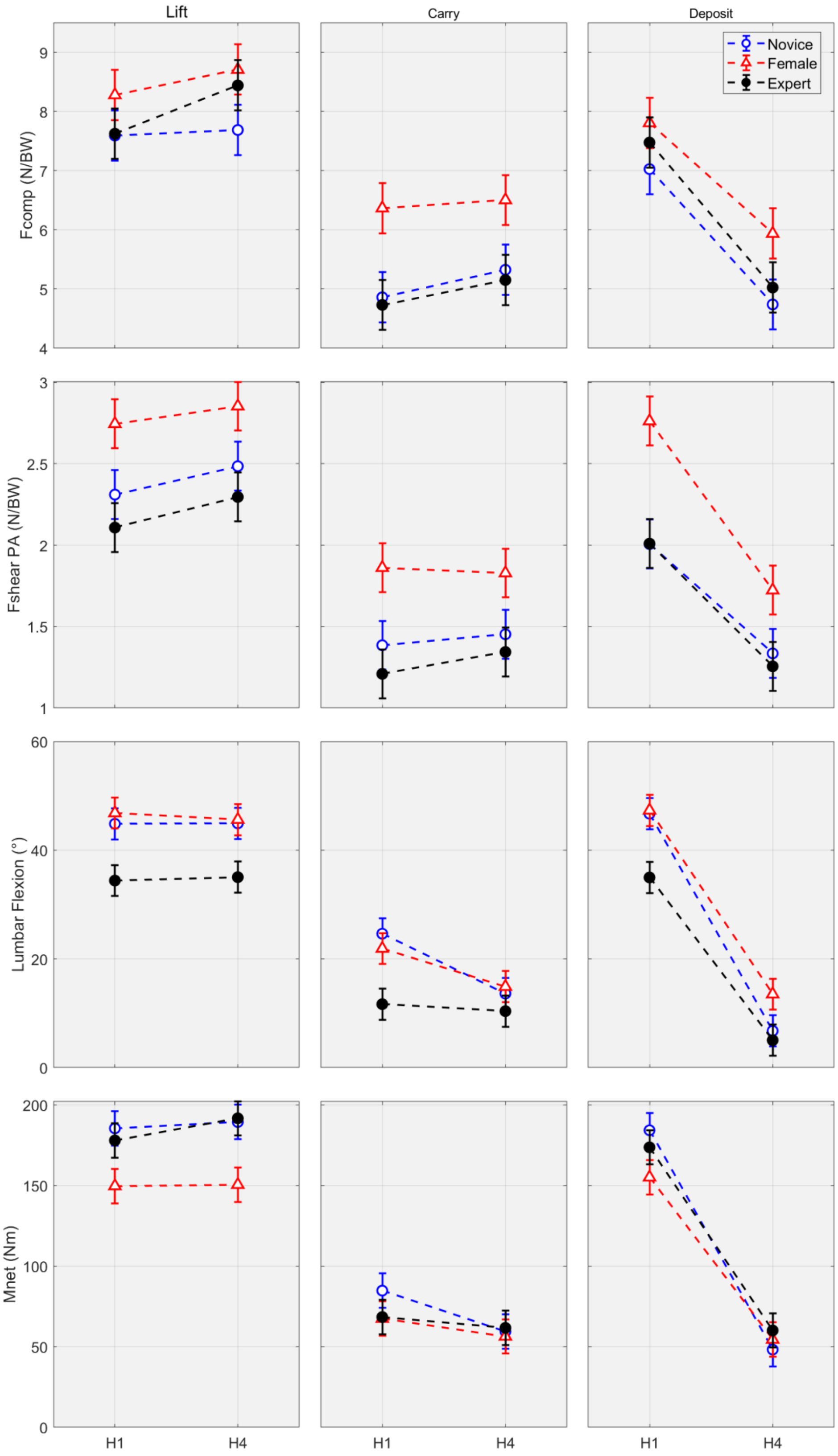

Fig. 4. Graphical illustration of height (H1 vs H4; horizontal axis of each subplot) and phase (Lift vs Carry vs Deposit; one level by column of subplots) factors plotted by group (Expert vs Female vs Novice; dashed lines) for the following variables (from the top down): Fcomp (N/BW), Fshear PA (N/BW), Lumbar Flexion (°), and L5-S1 Mnet (Nm). Error bars represent half Fisher's Least Significant Difference (FLSD) for the plotted effect so that means are different when the

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bars do not overlap. (For interpretation of the colors in this figure legend, the reader is referred to the web version of this article.)

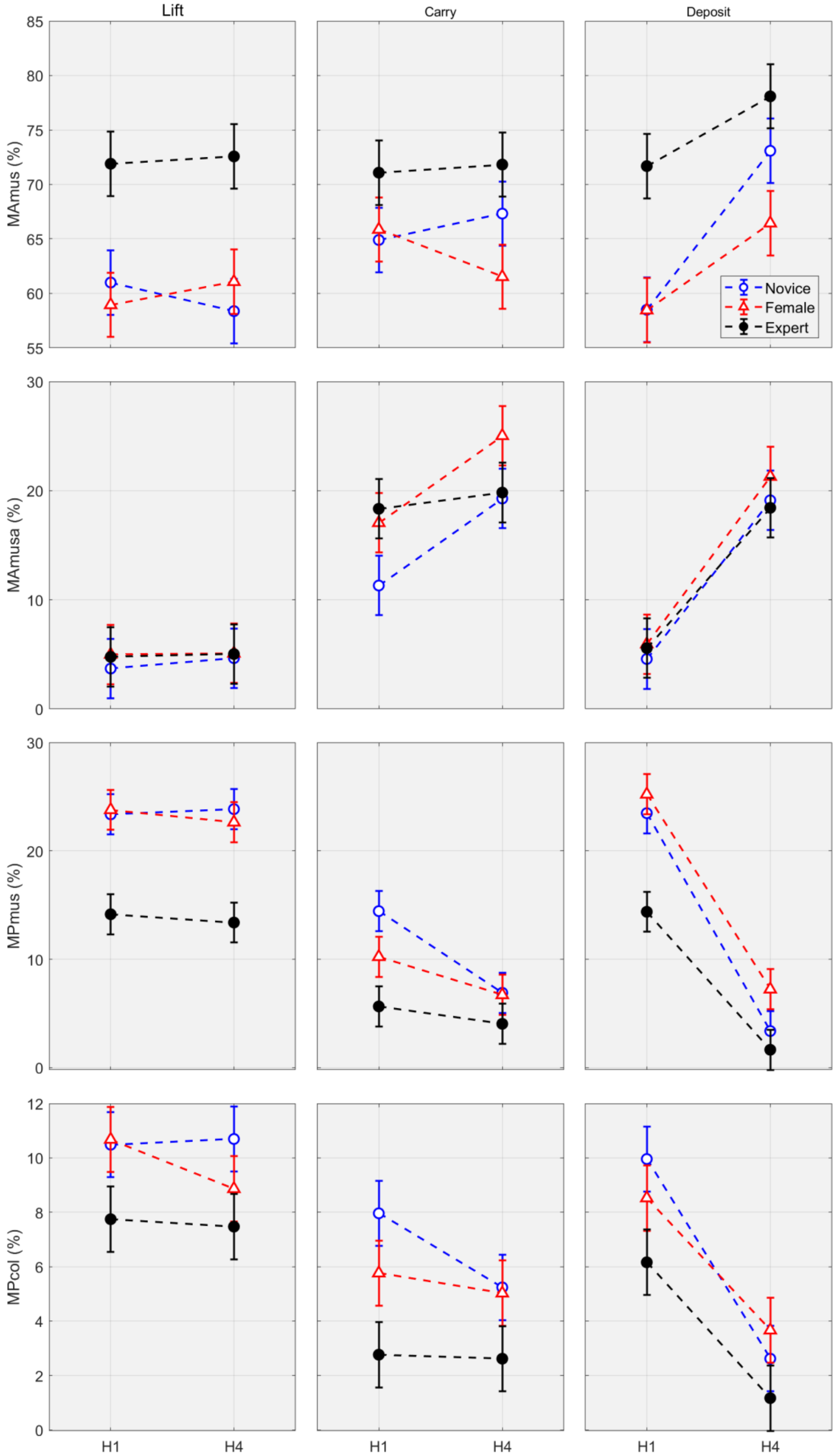

Fig. 5. Graphical illustration of height (H1 vs H4; horizontal axis of each subplot) and phase (Lift vs Carry vs Deposit; one level by column of subplots) factors plotted by group (Expert vs Female vs Novice; dashed lines) for the following variables (from the top down): MAmus (%), MAmusa (%), MPmus (%), and MPcol (%). Error bars represent half Fisher's

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Least Significant Difference (FLSD) for the plotted effect so that means are different when the bars do not overlap. (For interpretation of the colors in this figure legend, the reader is referred to the web version of this article.)

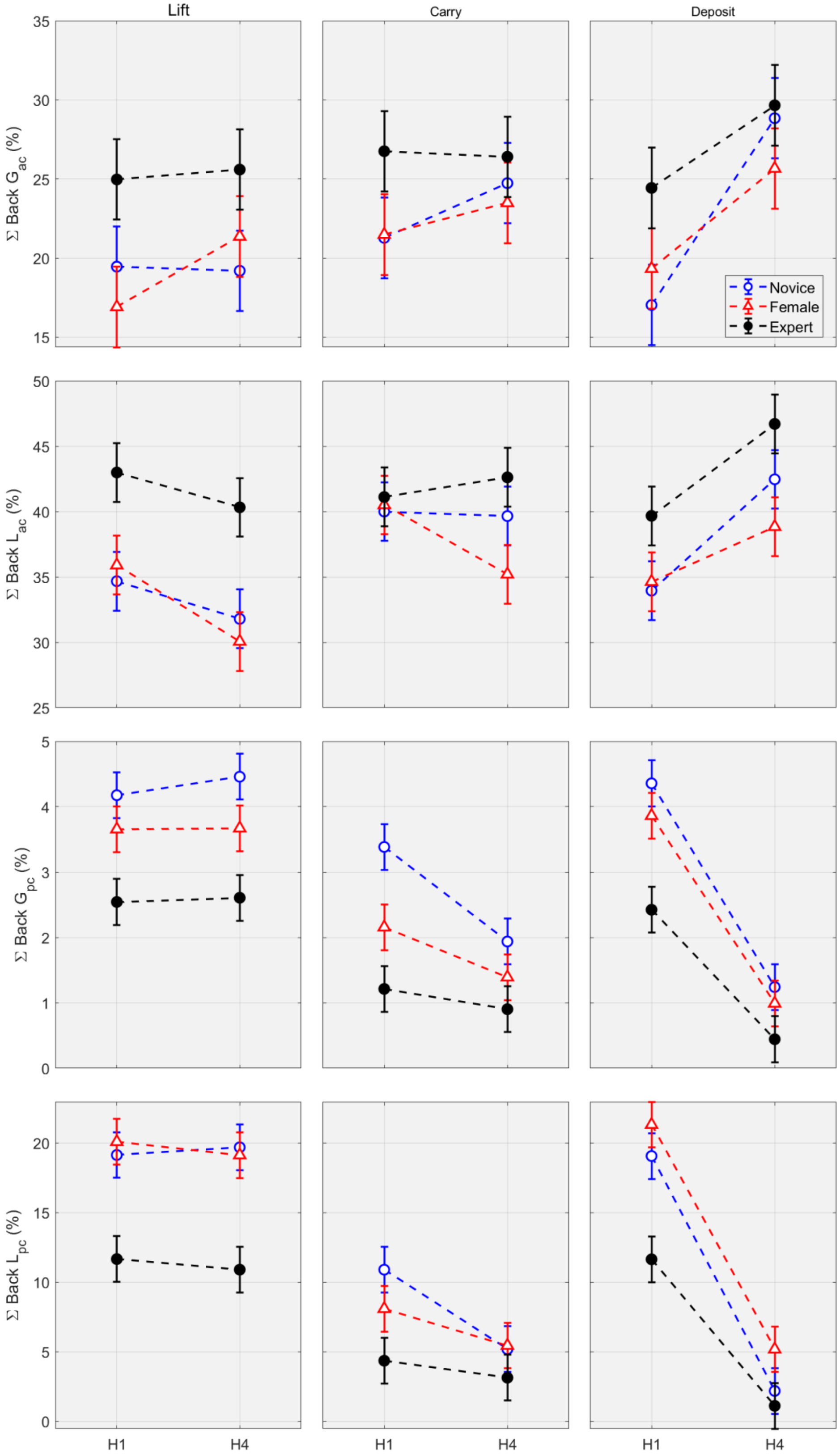

Fig. 6. Graphical illustration of height (H1 vs H4; horizontal axis of each subplot) and phase (Lift vs Carry vs Deposit; one level by column of subplots) factors plotted by group (Expert vs Female vs Novice; dashed lines) for the following

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variables (from the top down): ∑ Back G_{ac} (%), ∑ Back L_{ac} (%), ∑ Back G_{pc} (%), and ∑ Back L_{pc} (%). Error bars represent half Fisher's Least Significant Difference (FLSD) for the plotted effect so that means are different when the bars do not overlap. (For interpretation of the colors in this figure legend, the reader is referred to the web version of this article.)

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Table 1

Marginal means, standard deviations (SD) and factorial ANOVA results for the group (G) factor and its interactions with height (GH) and phase (GP) on main dependent variables (n = 10 subjects per group)

Dependent Variable	**Group (G)**			**Height (H)**		**Phase (P)**			**ANOVA p value**			**ANOVA ges value**			**Post hoc**
	Expert (E)	**Novice (N)**	**Female (F)**	**H1**	**H4**	**Lift**	**Carry**	**Deposit**	**G**	**GH**	**GP**	**G**	**GH**	**GP**	**G**
A. Joint forces, lumbar flexion angle and net joint moment															
Fcomp (N/BW)	6,4	6,2	7,3						**0,03**	0,85	0,50	0,11	0,00	0,02	F > E,N
	(1,9)	(1,8)	(1,8)												
Fshear PA (N/BW)	1,7	1,8	2,3						**0,01**	0,13	0,85	0,22 *	0,01	0,01	F > E,N
	(0,7)	(0,7)	(0,7)												
Lumbar Flexion (°)	21,9	30,3	31,7	34,8	21,1				0,08	**0,02**	0,66	0,13 *	0,01	0,01	E < F,N
	(16,6)	(19,6)	(20,0)	(17,1)	(18,7)										
Mnet (Nm)	122,3	125,3	103,1	138,6	96,9				**0,05**	**0,00**	0,12	0,09	0,02	0,05	F < E,N
	(66,9)	(68,9)	(57,5)	(54,7)	(65,1)										
B. Internal moments															
MAmus (%)	72,9	63,9	62,0						**0,02**	0,73	0,32	0,17 *	0,00	0,02	E > F,N
	(8,3)	(11,6)	(13,3)												
MAmusa (%)	12,0	10,4	13,2						0,32	0,56	0,56	0,02	0,01	0,02	
	(8,7)	(9,6)	(12,4)												
MPmus (%)	8,9	15,9	16,0			20,2	8,0	12,6	0,07	0,12	**0,05**	0,14 *	0,01	0,02	E < F,N
	(7,9)	(11,6)	(13,1)			(10,5)	(7,7)	(12,5)							
MPcol (%)	4,7	7,8	7,1						0,06	0,43	0,86	0,12	0,01	0,00	E < F,N
	(4,2)	(5,3)	(4,3)												
C. Muscle moments: Extensors (global, local) and global flexors															
Σ Back G_{ac} (%)	26,3	21,8	21,4						**0,02**	0,46	0,85	0,11	0,01	0,01	E > F,N
	(6,1)	(7,8)	(7,1)												
Σ Back L_{ac} (%)	42,2	37,1	35,9	38,2	38,6				0,08	0,09	0,27	0,11	0,02	0,02	E > F,N
	(6,5)	(7,6)	(11,0)	(9,2)	(8,8)										
Σ Back G_{pc} (%)	1,7	3,3	2,6	3,1	2,0				**0,04**	0,08	0,64	0,17 *	0,01	0,01	E < F,N
	(1,2)	(2,2)	(1,9)	(1,9)	(1,8)										
Σ Back L_{pc} (%)	7,1	12,7	13,2			16,8	6,2	10,1	0,07	0,17	0,06	0,13 *	0,01	0,02	E < F,N
	(7,1)	(9,9)	(11,2)			(9,4)	(6,2)	(10,5)							
Σ $Abdo_{ac}$ (%)	8,9	7,5	10,3						0,39	0,50	0,65	0,02	0,01	0,01	
	(8,5)	(8,8)	(12,2)												

p value: Probability from the three-way factorial ANOVA (G × H × P), bolded for **$p < 0.05$** and underlined for $0.05 \le p < 0.10$

ges: Generalized eta squared as defined by Bakeman (2015), moderate and above followed by an asterisk (ges ≥ 0.13 *)

Post hoc (G): Pairwise group comparison using t test with pooled SD and Bonferroni adjustement method ($p < 0.05$)

To simplify this table, maginal means (SD) for factors height (H) and phase (P) are only presented when relevant

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Table 2

Marginal means, standard deviations (SD) and factorial ANOVA results for the group (G) factor and its interactions with height (GH) and phase (GP) on selected dependent variables (n = 10 subjects per group)

Dependent Variable	Group (G)			Height (H)		Phase (P)			ANOVA p value			ANOVA ges value			Post hoc
	Expert (E)	Novice (N)	Female (F)	H1	H4	Lift	Carry	Deposit	G	GH	GP	G	GH	GP	G
D. Detailed muscle moments: Global extensors															
Σ $LGPT_{ac}$ (%)	20,2	16,7	16,7						**0,05**	0,74	0,88	0,09	0,00	0,01	E > F,N
	(6,0)	(5,9)	(5,5)												
Σ $LGPT_{pc}$ (%)	1,0	2,1	1,7	2,0	1,3				**0,03**	**0,03**	0,65	0,18 *	0,01	0,00	E < F,N
	(0,7)	(1,5)	(1,3)	(1,2)	(1,2)										
Σ $ICPT_{ac}$ (%)	6,1	5,1	4,6						0,07	0,37	0,35	0,05	0,01	0,03	E > F
	(3,1)	(3,6)	(3,0)												
Σ $ICPT_{pc}$ (%)	0,6	1,2	0,9						0,06	0,37	0,63	0,13 *	0,01	0,01	E < N
	(0,5)	(0,8)	(0,7)												
E. Detailed muscle moments: Local extensors															
Σ $LGPL_{ac}$ (%)	8,4	7,3	7,3						0,24	0,71	0,32	0,06	0,00	0,02	
	(2,0)	(2,1)	(2,6)												
Σ $LGPL_{pc}$ (%)	1,4	2,4	2,5			3,1	1,2	1,9	0,07	0,21	**0,04**	0,13 *	0,01	0,03	E < F,N
	(1,3)	(1,8)	(2,0)			(1,7)	(1,1)	(2,0)							
Σ $ICPL_{ac}$ (%)	12,4	10,1	9,9						**0,04**	0,45	0,50	0,09	0,01	0,02	E > F,N
	(3,3)	(3,3)	(4,1)												
Σ $ICPL_{pc}$ (%)	2,5	4,4	4,6			5,9	2,1	3,5	0,08	0,26	0,08	0,13 *	0,01	0,03	E < F,N
	(2,6)	(3,5)	(4,0)			(3,4)	(2,1)	(3,7)							
Σ MUF_{ac} (%)	17,4	15,9	14,9						0,21	0,17	0,96	0,05	0,02	0,00	
	(5,6)	(5,9)	(5,8)												
Σ MUF_{pc} (%)	2,3	4,2	4,3	4,7	2,6	5,4	2,2	3,3	0,06	0,06	0,07	0,14 *	0,01	0,02	E < F,N
	(2,1)	(3,2)	(3,5)	(3,1)	(2,8)	(2,8)	(2,2)	(3,4)							
Σ QLO_{ac} (%)	4,0	3,8	3,7						0,52	0,70	0,12	0,01	0,00	0,06	
	(1,7)	(1,5)	(1,5)												
Σ QLO_{pc} (%)	1,0	1,7	1,8						0,13	0,28	0,15	0,10	0,01	0,02	
	(1,1)	(1,4)	(1,7)												
F. Detailed muscle moments: Global flexors															
Σ RA_{ac} (%)	1,5	1,2	2,4	1,0	2,4				0,06	**0,00**	0,29	0,07	0,05	0,03	F > N
	(1,4)	(1,5)	(3,2)	(1,2)	(2,7)										
Σ EO_{ac} (%)	7,2	6,3	7,4						0,74	0,76	0,69	0,01	0,00	0,01	
	(7,7)	(7,2)	(9,8)												
Σ IO_{ac} (%)	0,2	0,3	0,6						0,44	0,54	0,35	0,03	0,00	0,04	
	(0,4)	(0,8)	(1,7)												

p value: Probability from the three-way factorial ANOVA (G × H × P), bolded for **$p < 0.05$** and underlined for $0.05 \leq p < 0.10$
ges: Generalized eta squared as defined by Bakeman (2015), moderate and above followed by an asterisk (ges ≥ 0.13 *)

Post hoc (G): Pairwise group comparison using t test with pooled SD and Bonferroni adjustement method ($p < 0.05$)

To simplify this table, maginal means (SD) for factors height (H) and phase (P) are only presented when relevant

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Additional Results: Detailed Muscular Moments

Table 2

Marginal means, standard deviations (SD) and factorial ANOVA results for the group (G) factor and its interactions with height (GH) and phase (GP) on selected dependent variables (n = 10 subjects per group)

Dependent Variable	**Group (G)**			**Height (H)**		**Phase (P)**			**ANOVA p value**			**ANOVA ges value**			**Post hoc**
	Expert (E)	**Novice (N)**	**Female (F)**	**H1**	**H4**	**Lift**	**Carry**	**Deposit**	**G**	**GH**	**GP**	**G**	**GH**	**GP**	**G**
D. Detailed muscle moments: Global extensors															
Σ LGPT$_{ac}$ (%)	20.2	16.7	16.7						**0.05**	0.74	0.88	0.09	0.00	0.01	E > F,N
	(6.0)	(5.9)	(5.5)												
Σ LGPT$_{pc}$ (%)	1.0	2.1	1.7	2.0	1.3				**0.03**	**0.03**	0.65	0.18 *	0.01	0.00	E < F,N
	(0.7)	(1.5)	(1.3)	(1.2)	(1.2)										
Σ ICPT$_{ac}$ (%)	6.1	5.1	4.6						0.07	0.37	0.35	0.05	0.01	0.03	E > F
	(3.1)	(3.6)	(3.0)												
Σ ICPT$_{pc}$ (%)	0.6	1.2	0.9						0.06	0.37	0.63	0.13 *	0.01	0.01	E < N
	(0.5)	(0.8)	(0.7)												
E. Detailed muscle moments: Local extensors															
Σ LGPL$_{ac}$ (%)	8.4	7.3	7.3						0.24	0.71	0.32	0.06	0.00	0.02	
	(2.0)	(2.1)	(2.6)												
Σ LGPL$_{pc}$ (%)	1.4	2.4	2.5			3.1	1.2	1.9	0.07	0.21	**0.04**	0.13 *	0.01	0.03	E < F,N
	(1.3)	(1.8)	(2.0)			(1.7)	(1.1)	(2.0)							
Σ ICPL$_{ac}$ (%)	12.4	10.1	9.9						**0.04**	0.45	0.50	0.09	0.01	0.02	E > F,N
	(3.3)	(3.3)	(4.1)												
Σ ICPL$_{pc}$ (%)	2.5	4.4	4.6			5.9	2.1	3.5	0.08	0.26	0.08	0.13 *	0.01	0.03	E < F,N
	(2.6)	(3.5)	(4.0)			(3.4)	(2.1)	(3.7)							
Σ MUF$_{ac}$ (%)	17.4	15.9	14.9						0.21	0.17	0.96	0.05	0.02	0.00	
	(5.6)	(5.9)	(5.8)												
Σ MUF$_{pc}$ (%)	2.3	4.2	4.3	4.7	2.6	5.4	2.2	3.3	0.06	0.06	0.07	0.14 *	0.01	0.02	E < F,N
	(2.1)	(3.2)	(3.5)	(3.1)	(2.8)	(2.8)	(2.2)	(3.4)							
Σ QLO$_{ac}$ (%)	4.0	3.8	3.7						0.52	0.70	0.12	0.01	0.00	0.06	
	(1.7)	(1.5)	(1.5)												
Σ QLO$_{pc}$ (%)	1.0	1.7	1.8						0.13	0.28	0.15	0.10	0.01	0.02	
	(1.1)	(1.4)	(1.7)												
F. Detailed muscle moments: Global flexors															
Σ RA$_{ac}$ (%)	1.5	1.2	2.4	1.0	2.4				0.06	**0.00**	0.29	0.07	0.05	0.03	F > N
	(1.4)	(1.5)	(3.2)	(1.2)	(2.7)										
Σ EO$_{ac}$ (%)	7.2	6.3	7.4						0.74	0.76	0.69	0.01	0.00	0.01	
	(7.7)	(7.2)	(9.8)												
Σ IO$_{ac}$ (%)	0.2	0.3	0.6						0.44	0.54	0.35	0.03	0.00	0.04	
	(0.4)	(0.8)	(1.7)												

p value: Probability from the three-way factorial ANOVA (G × H × P), bolded for **$p < 0.05$** and underlined for $0.05 \le p < 0.10$

ges: Generalized eta squared as defined by Bakeman (2015), moderate and above followed by an asterisk (ges ≥ 0.13 *)

Post hoc (G): Pairwise group comparison using t test with pooled SD and Bonferroni adjustement method ($p < 0.05$)

To simplify this table, maginal means (SD) for factors height (H) and phase (P) are only presented when relevant

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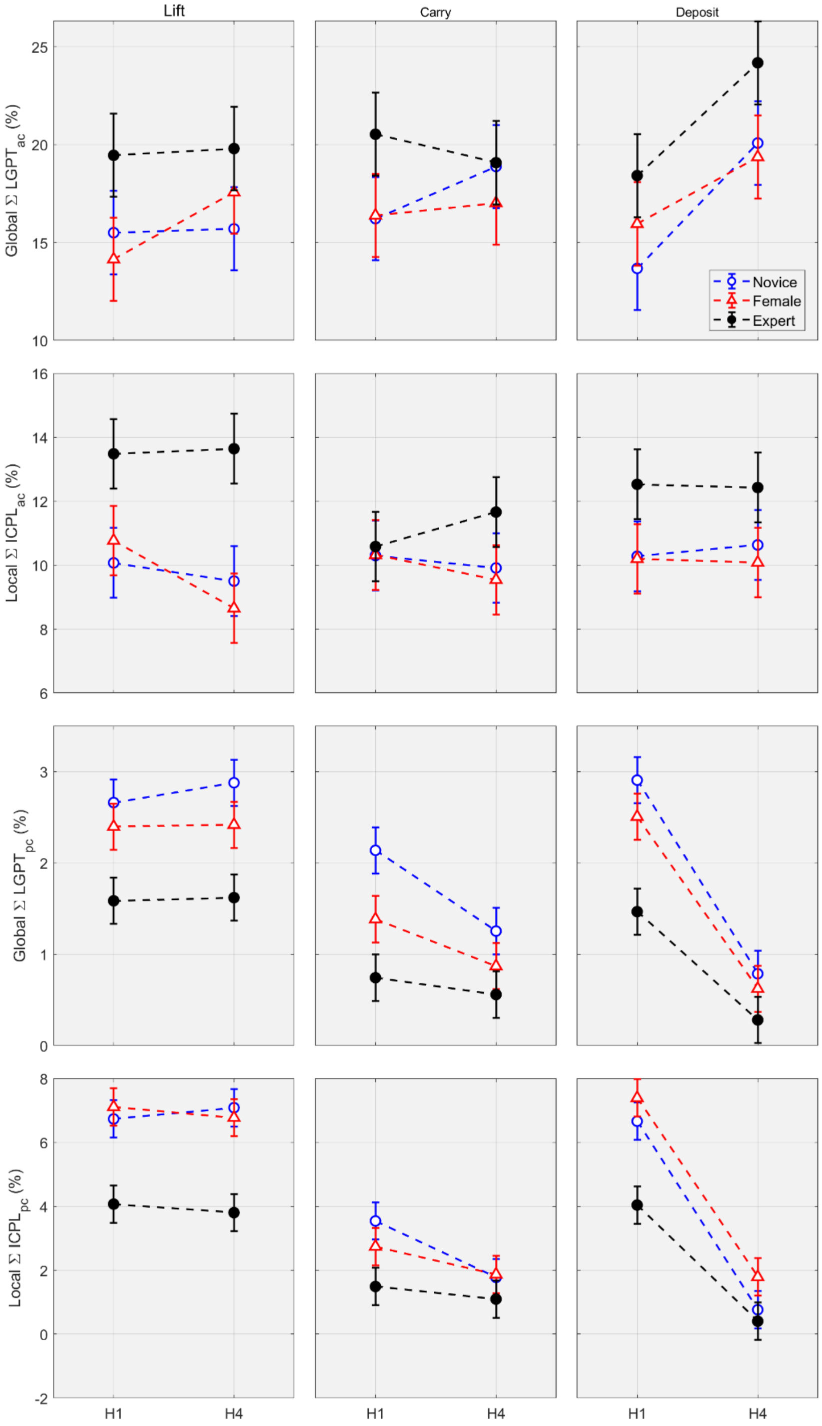

Fig. 7. Graphical illustration of height (H1 vs H4; horizontal axis of each subplot) and phase (Lift vs Carry vs Deposit; one level by column of subplots) factors plotted by group (Expert vs Female vs Novice; dashed lines) for the following variables (from the top down): Global $\sum$ LGPT$_{ac}$ (%), Local $\sum$ ICPL$_{ac}$ (%), Global $\sum$ LGPT$_{pc}$ (%), and Local $\sum$ ICPL$_{pc}$ (%). Error bars represent half Fisher's Least Significant Difference (FLSD) for the plotted effect so that means are different when the bars do not overlap.

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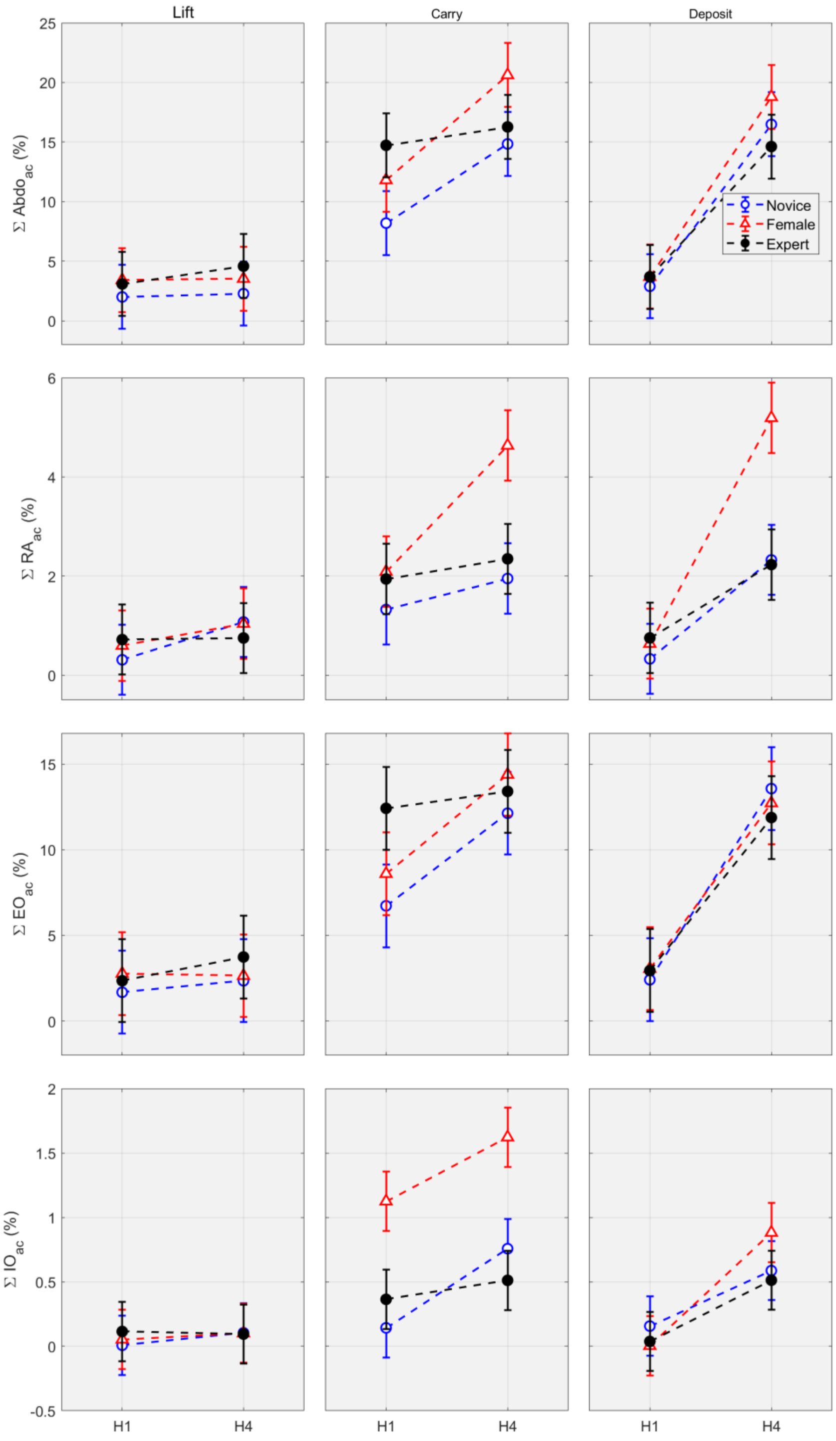

Fig. 8. Graphical illustration of height (H1 vs H4; horizontal axis of each subplot) and phase (Lift vs Carry vs Deposit; one level by column of subplots) factors plotted by group (Expert vs Female vs Novice; dashed lines) for the following variables (from the top down): $\sum$ Abdo$_{ac}$ (%), $\sum$ RA$_{ac}$ (%), $\sum$ EO$_{ac}$ (%), and $\sum$ IO$_{ac}$ (%). Error bars represent half Fisher's Least Significant Difference (FLSD) for the plotted effect so that means are different when the bars do not overlap.

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Interpretation of interactions

Some interactions (Table 1-2) were statistically significant ($p < 0.05$) or marginally significant ($0.05 \leq p < 0.10$), but their generalized effect sizes were all small (ges ≤ 0.06). To stay coherent with the manuscript, section and figure numbers of this supplement are the same as those of the corresponding elements in the manuscript (or in the previous section), but end with the letter S.

3.1S Joint forces with flexion angle and net joint moment (Table 1.A and Fig. 4S)

Statistically significant GH interactions were detected for lumbar flexion angle and net (external) moment. These interactions indicate that novices decreased their lumbar flexion from H1 to H4 more, while experts decreased their net (external) moment less (**Fig. 4S**). On the one hand, novices straighten their lumbar spine more than females but still less than experts from H1 to H4. On the other hand, experts exerted a net moment in between novice and female at H1 but then their moment became the largest at H4.

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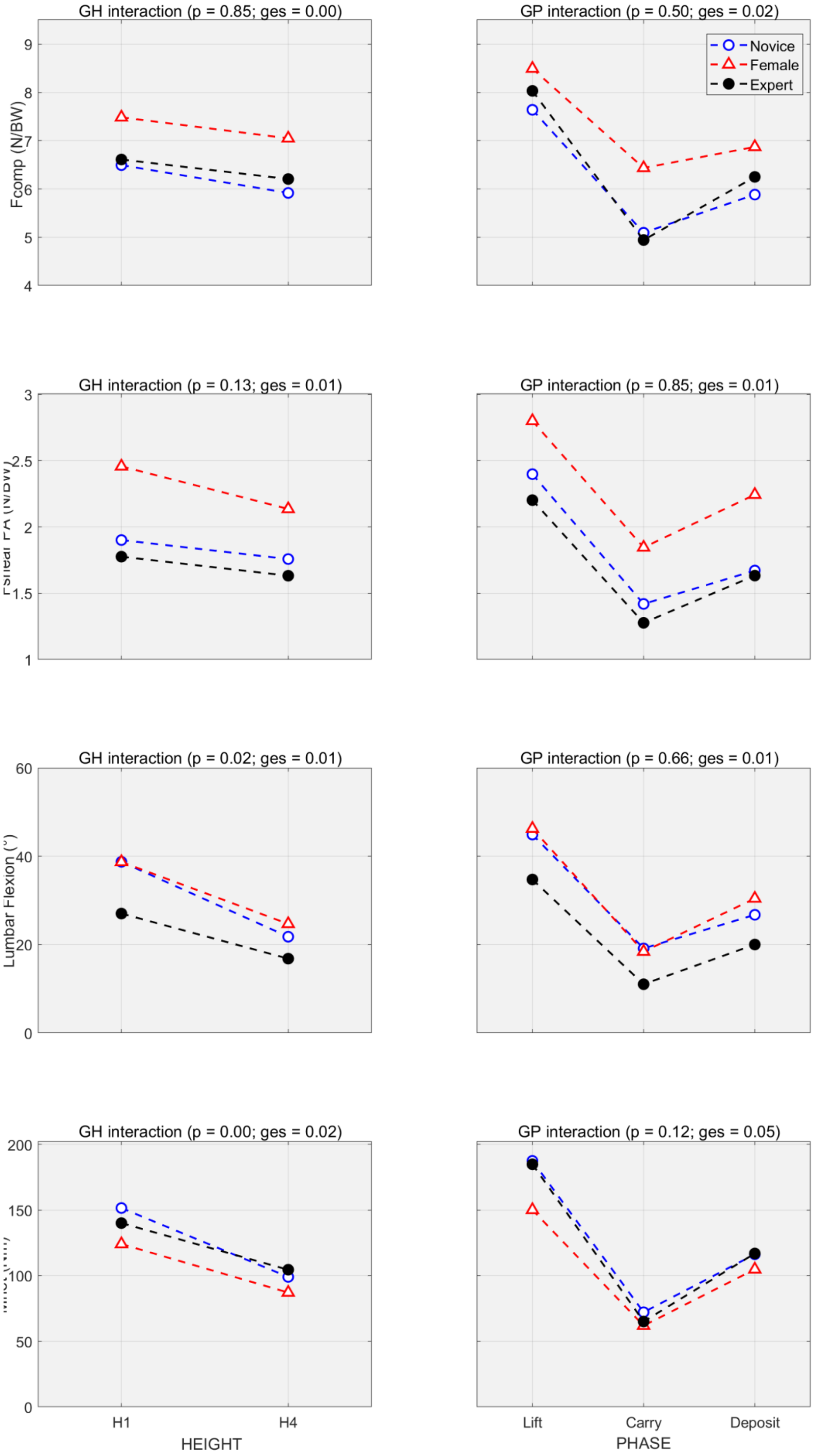

Fig. 4S. Interactions GH and GP for the same variables as **Fig. 4**.

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3.2S Internal moments (Table 1.B and Fig. 5S)

A statistically significant GP interaction for MPmus indicated that groups did not behave the same way across phases: changes in passive muscle moment for females were not following the trend of the males, with females increasing their passive muscle resistance at deposit time more (**Fig. 5S**).

3.3S Muscle moments (Table 1.C and Fig. 6S)

There was no statistically significant interaction for the muscle moments. Marginally significant GH interactions for active local extensors as well as for passive global extensors highlighted some slope of changes from H1 to H4 between the groups: the slope of Back L_{ac} was positive for experts and novices, but negative for females; for Back G_{pc}, slopes are all negative but the incline is slightly less pronounced for experts. One marginally significant GP for Back L_{pc} interaction reproduces the statistically significant GP interaction for MPmus and could be explained the same way. It is worthwhile to note that females and novices had the same overall extensor contributions.

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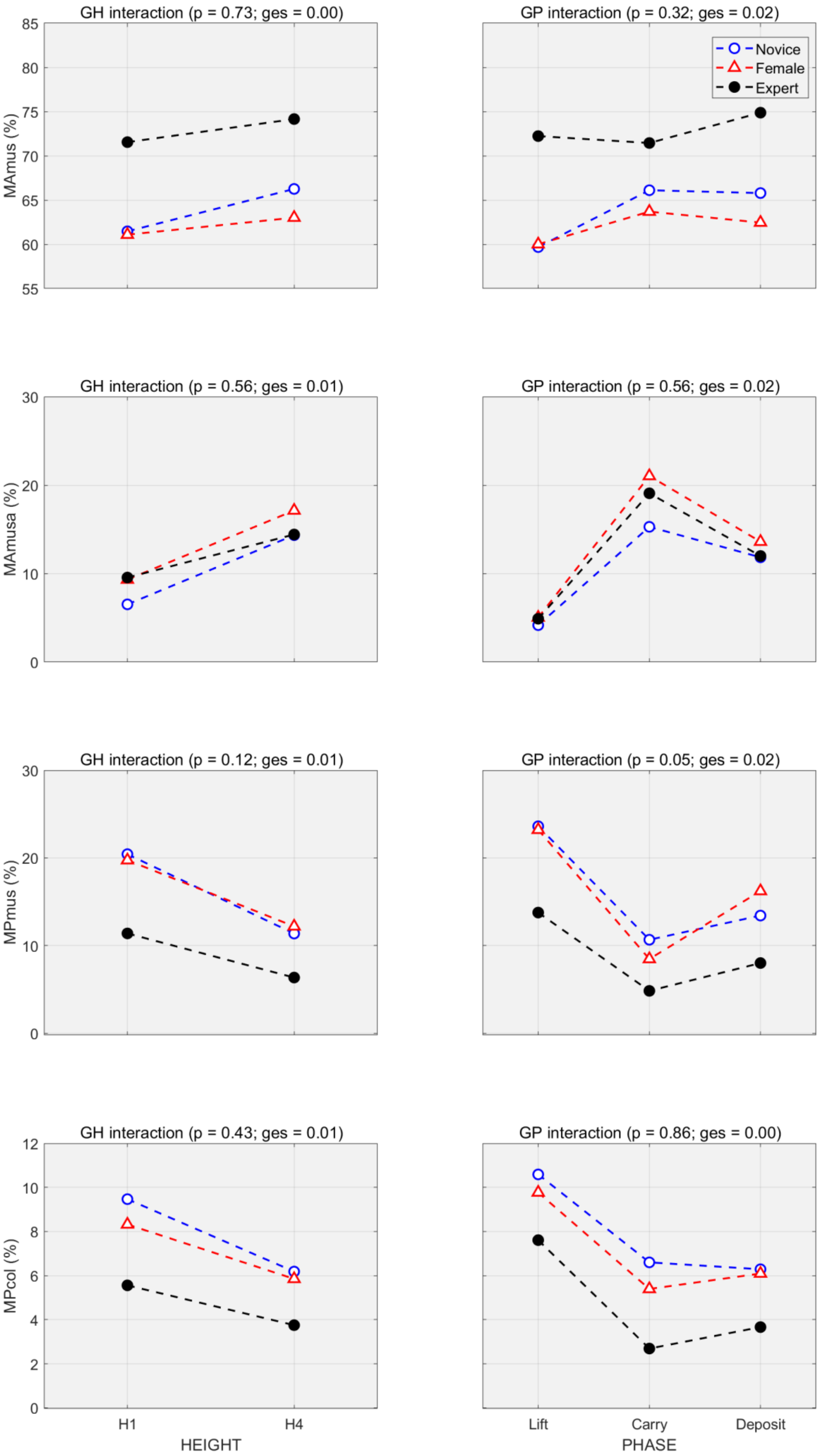

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Fig. 5S. Interactions GH and GP for the same variables as **Fig. 5**.

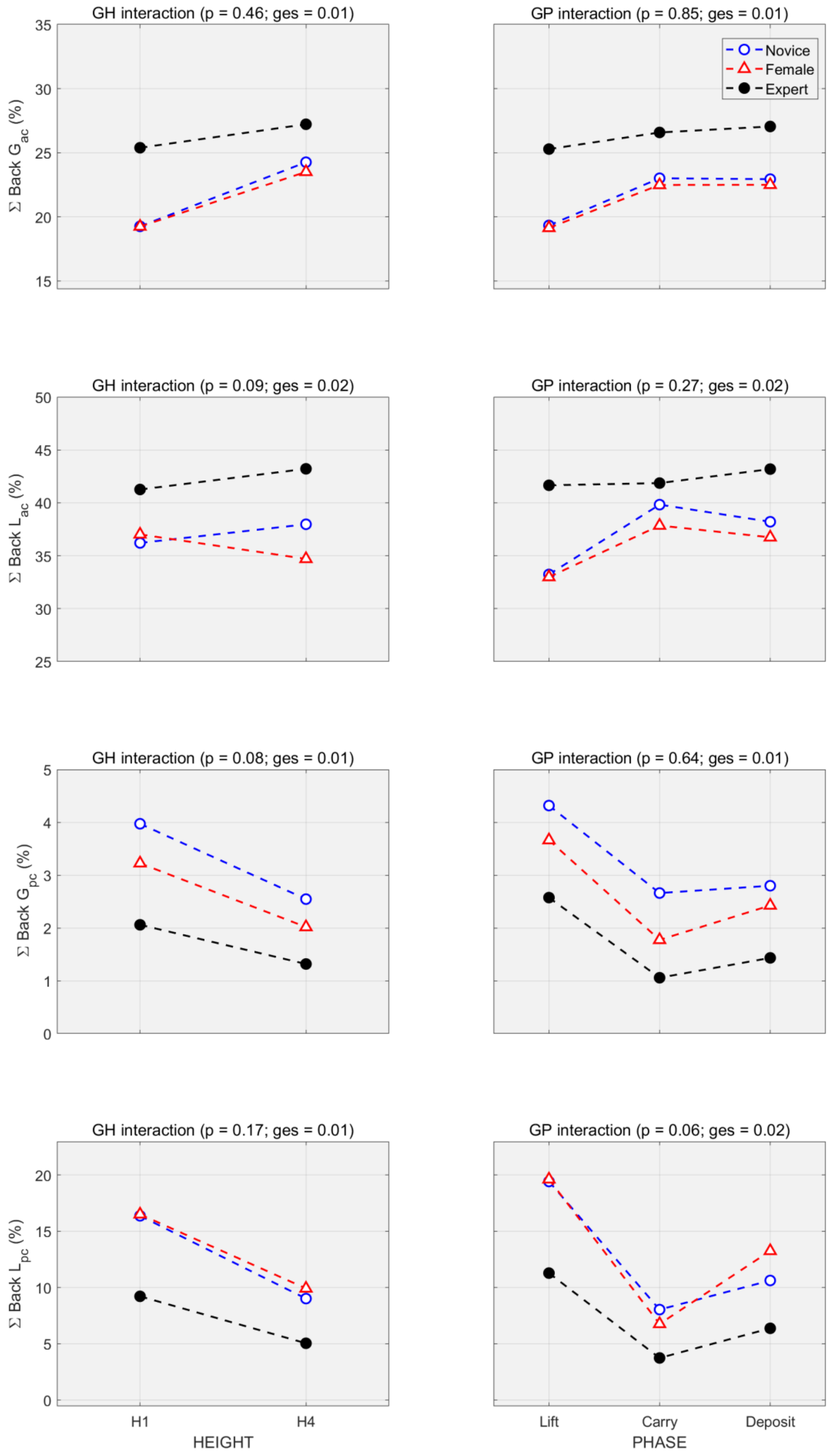

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Fig. 6S. Interactions GH and GP for the same variables as **Fig. 6**.

3.4S Detailed muscle moments: Global and local extensors (Table 2.D-E and Fig. 7-7S)

Statistically significant GH interactions for $LGPT_{pc}$ (marginally significant for MUF_{pc}) indicated that both females and novices decreased the passive resistance of this muscle from H1 to H4 more than experts: thus, changes were greater for females and novices but experts still required less passive tension from these muscles throughout the tasks. Statistically significant GP interactions for local extensor $LGPL_{pc}$ (marginally significant for $ICPL_{pc}$ and MUF_{pc}) indicated a different strategy for females across the phases. For this local extensor, females showed a more marked decline in passive resistance from lift to carry and then a more prominent rise from carry to deposit. These passive contributions occurred in parallel with lumbar spine flexion-extension: because females were more flexed at lift and deposit times (more passive contributions), they had to straighten more to approach an erected trunk posture during the carry phase (steeper increase-decrease slopes).

3.5S Detailed muscle moments: Global flexors (Table 2.F and Fig. 8-8S)

A statistically significant GH interaction indicated that the demand for RA_{ac} was amplified for females when the destination was H4.

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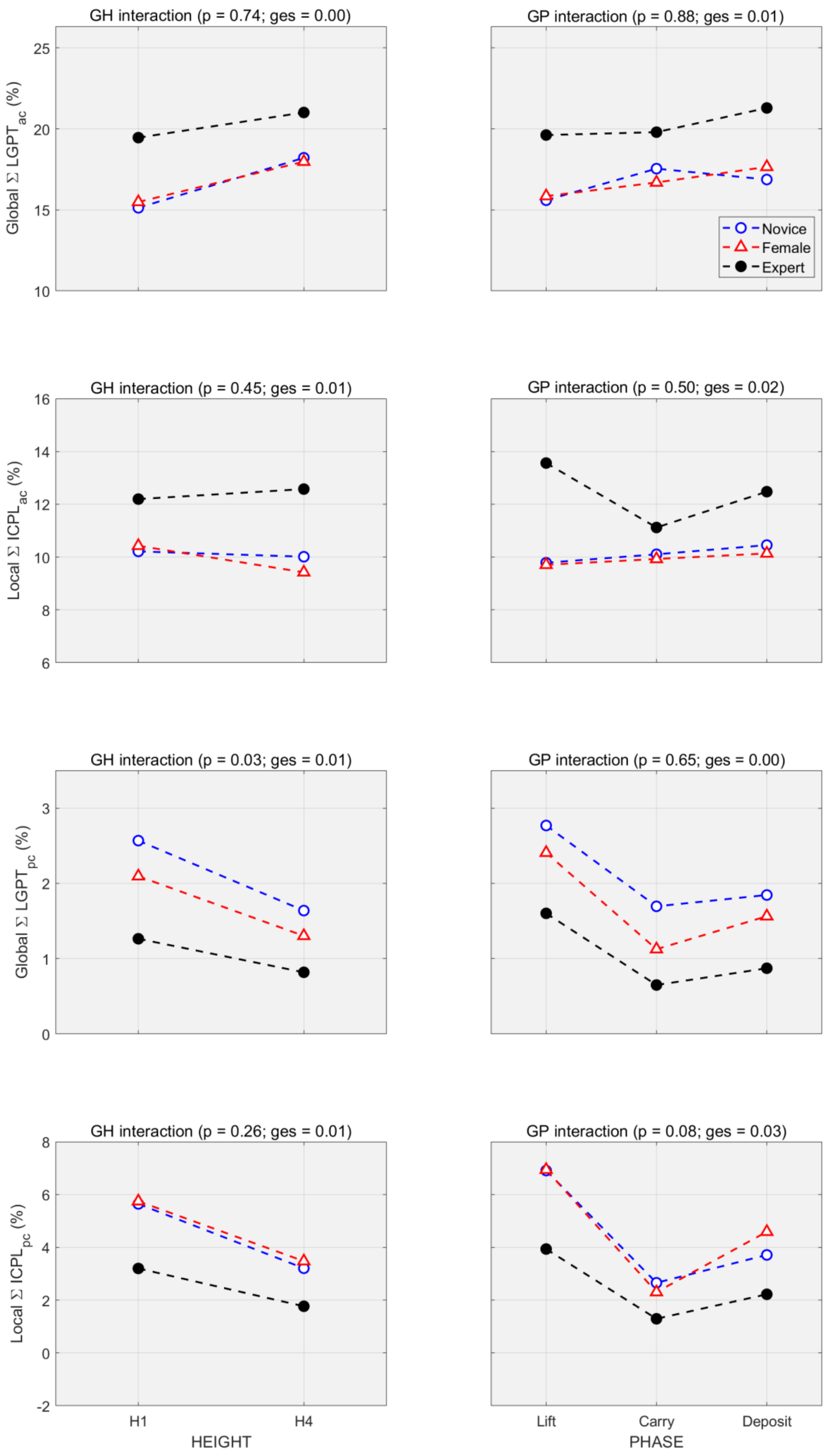

Fig. 7S. Interactions GH and GP for the same variables as **Fig. 7**.

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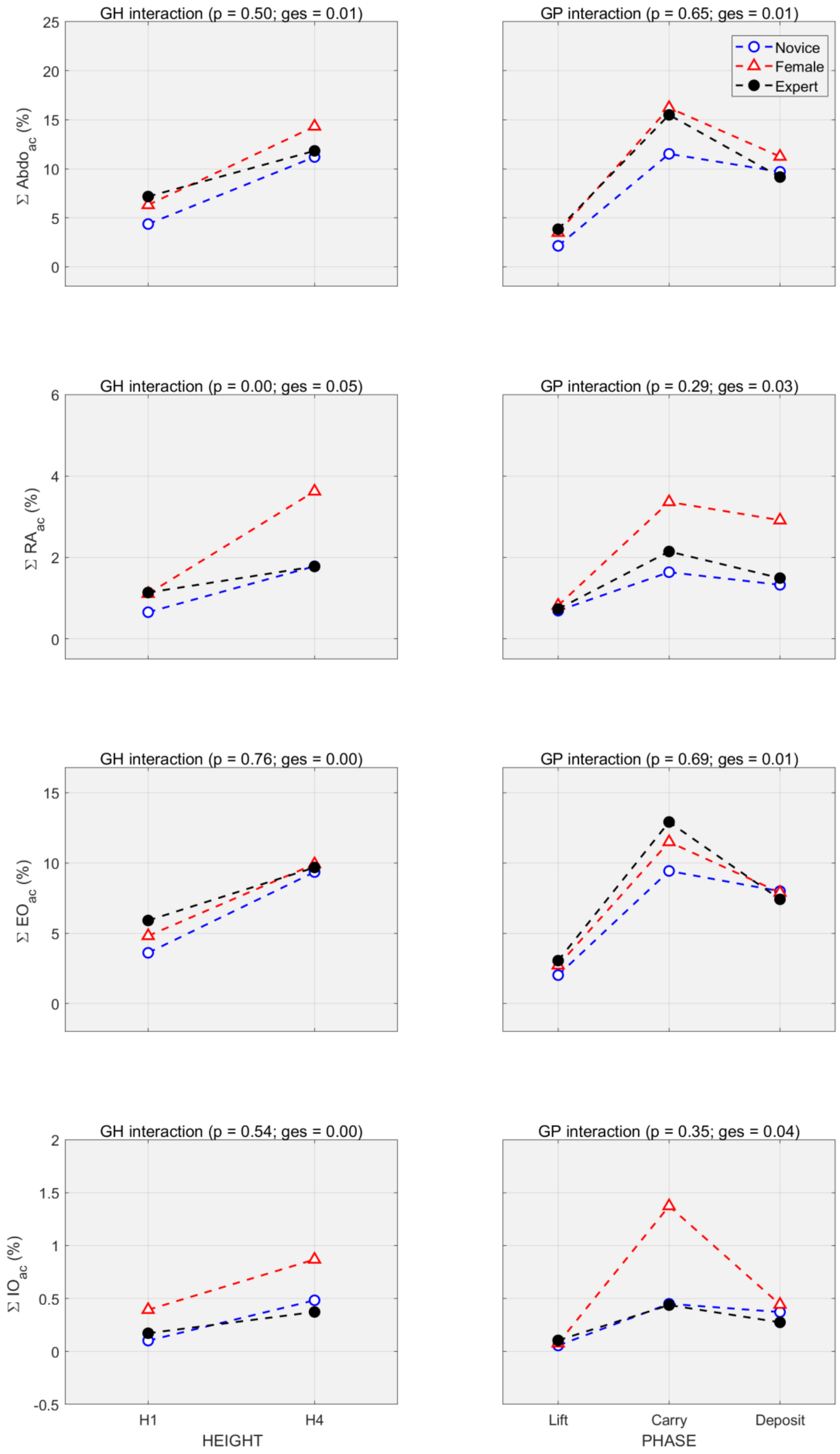

Fig. 8S. Interactions GH and GP for the same variables as **Fig. 8**.

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Abbreviations

Abbreviation	**Definition (units)**
Joint forces	
Fcomp	Joint compression force (N)
Fshear	Joint shear force (N)
Moments	
MAmus	Active muscle moment (Nm)
MAmusMax	Maximal active muscle moment about a lumbar joint for a given posture (Nm)
Merr	Adjustment error moment (Nm)
Mnet	Net (external) joint moment (Nm)
Mnorm	Euclidean norm of the resultant moment(s) about relevant joint(s) (Nm)
MPcol	Passive ligamentous spine (resistance) moment (Nm)
MPmus	Passive (agonist) muscle moment (Nm)
M_L, M_S, M_T, Mr	Longitudinal (L), sagittal (S), transverse (T), and resultant moment about a joint (Nm)
Dependent variables	
Fshear PA, Fshear ML	Postero-anterior and medial-lateral shear joint force (N/Body weight)
MAmus	Normalized agonist muscle moment (%)
MAmusa	Normalized antagonist muscle moment (%)
MPcol	Normalized passive ligamentous spine moment (%)
MPmus	Normalized passive muscle moment (%)
∑Abdo	Sum of global flexor normalized moments: EO, IO, RA (%)
∑Back L	Sum of local extensor normalized moments: LGPL, MUF, ICPL, QLO (%)
∑Back G	Sum of global extensor normalized moments: ICPT and LGPT (%)
∑EO	Sum of external oblique normalized moments (%)
∑ICPL	Sum of iliocostalis pars lumbaris normalized moments (%)
∑ICPT	Sum of iliocostalis pars thoracis normalized moments (%)
∑IO	Sum of internal oblique normalized moments (%)
∑LGPL	Sum of longissimus pars lumbaris normalized moments (%)
∑LGPT	Sum of longissimus pars thoracis normalized moments (%)
∑MUF	Sum of multifidus normalized moments (%)
∑QLO	Sum of quadratus lumborum normalized moments (%)
∑RA	Sum of rectus abdominis normalized moments (%)

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