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Interface Stability Influences Torso Muscle Recruitment and Spinal Load During Pushing Tasks

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Abstract

Handle or interface design can influence torso muscle recruitment and spinal load during pushing tasks. The objective of the study was to provide insight into the role of interface stability with regard to torso muscle recruitment and biomechanical loads on the spine. Fourteen subjects generated voluntary isometric trunk flexion force against a rigid interface and similar flexion exertions against an unstable interface, which simulated handle design in a cart pushing task. Normalized electromyographic (EMG) activity in the rectus abdominus, external oblique and internal oblique muscles increased with exertion effort. When using the unstable interface, EMG activity in the internal and external oblique muscle groups was greater than when using the rigid interface. Results agreed with trends from a biomechanical model implemented to predict the muscle activation necessary to generate isometric pushing forces and maintain spinal stability when using the two different interface designs. The co-contraction contributed to increased spinal load when using the unstable interface. It was concluded that handle or interface design and stability may influence spinal load and associated risk of musculoskeletal injury during manual materials tasks that involve pushing exertions.

Keywords

Low-back; Stability; Co-contraction; Push-Pull; Handle interface

1. Introduction

Pushing and pulling exertions account for half of manual material handling (MMH) tasks and 20% of low-back injury claims (Damkot *et al.* 1984, Baril-Gingras and Lortie 1995, Hoozemans *et al.* 1998). To assess MMH tasks, torso muscle recruitment and associated spinal loads are often used as markers of risk related to musculoskeletal overload injury (Lee *et al.* 1991, Kumar 1994, de Looze *et al.* 1995). Studies have revealed that task design influences spinal compression during pushing exertions. Hoozemans *et al.* (2004) observed that spinal compression during a cart-pushing task is influenced by cart weight and handle height. The handle serves as an interface between the worker and the material and can therefore influence apparent load and stability experienced by the worker. Handle or interface design can influence torso muscle co-contraction and spinal load (Davis *et al.* 1998). If the interface allows vertical, horizontal or rotational movement during a pushing exertion then the torso musculature must act to stabilize the exertion in addition to generating the push force. Rancourt and Hogan (2001) observe that interface design can limit the stability of the task and even prohibit successful control of the exertion. For example, it is intuitive that an attempt to push against a wagon handle that has been designed for pulling loads is more difficult than pushing against

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a rigid interface. The goal of the current study was to examine the biomechanics of this phenomenon in order to explain the role of interface stability with regard to torso muscle recruitment and spinal load. This provides insight into interface design for MMH pushing tasks.

Muscle recruitment during MMH tasks influences spinal load and stability. Lee *et al.* (1991) computed spinal compression using a single equivalent muscle model and noted that assuming 'only one muscle is active at one time to stabilize the torso may not be appropriate for the estimation of muscle forces in the lower back'. Trunk muscle co-contraction significantly influences spinal load, accounting for as much as 45% of spinal compression during lifting tasks (Granata and Marras 1995b, Hughes *et al.* 1995, Thelen *et al.* 1995). Recent studies demonstrate that co-contraction during flexion exertions is greater than during equivalent trunk extension efforts and that this co-contraction contributes markedly to spinal load during pushing tasks (Granata *et al.* 2005). Therefore, electromyographic (EMG)-assisted models have been applied to incorporate co-contraction when estimating spinal load (Marras and Granata 1997, Hoozemans *et al.* 2004). Evidence suggests that co-contraction recruitment increases with task instability (Granata and Orishimo 2001). Thus, interface dynamics of the handle in a pushing task may influence co-contraction. Specifically, increased activity of the internal oblique (IO) and external oblique (EO) muscles may be recruited in order to maintain stability against a highly dynamic interface. Recognizing that co-contraction contributes to spinal load, it is expected that interface stability during pushing tasks will influence spinal compression.

It is hypothesized that an interface that is unstable under pushing loads may cause elevated levels of antagonistic co-contraction, thereby resulting in increased spinal compression. The hypothesis was tested by measuring muscle recruitment and estimating spinal load using an EMG-assisted model. Pushing tasks were performed by trunk flexion loading against an unstable handle with 2 degrees of freedom (df) for comparison with data collected while pushing against a rigidly fixed handle.

2. Methods

2.1. Details of subjects

Seven females and seven males with no history of low-back pain volunteered to participate. Mean age was 21.3 (SD 2.7) years and average weight and stature were 68.0 (SD 21.2) kg and 170.9 (SD 10.2) cm respectively. Before participation, all subjects signed an informed consent form approved by the Virginia Tech human subjects review board.

2.2. Experimental procedures

EMG, trunk posture, ground reaction force and push force were recorded while subjects matched a target flexion force displayed on a real-time display. Subjects generated voluntary isometric trunk flexion against a rigid interface and similar flexion exertions against an unstable interface. A 6-df load cell (AMTI, Watertown, MA, USA) was fixed to the wall to record the contact forces between a T-bar interface and the wall (as shown in figure 1). The base of a T-bar was attached to the load cell. This interface extended horizontally toward the subject such that the head of the T-bar was parallel to the shoulders. The height of the load cell and T-bar interface was adjusted such that the elevation of this system was centred 8 cm inferior to the clavicle. The head of the T-bar interface created a surface 8 cm in elevation and 30 cm in breadth such that the subjects could generate a push force by pressing their chest against the interface beam.

Subjects were strapped into a height-adjustable pelvic support and generated a push force by isometric trunk flexion against the interface beam. The protocol was designed to push with the

chest rather than the arms, to separate the variability associated with shoulder and arm dynamics from the recruitment effects of the trunk.

The rigid and unstable interfaces (shown in figure 2) were identical to each other with the exception that the unstable interface allowed trunk lateral and torsional movement, whereas the rigid interface did not. Specifically, when pushing against the rigid interface the subject was fixed in alignment with the wall mounted load cell, thereby prohibiting movement of the torso during the exertions. Conversely, the unstable interface permitted free rotation about a vertical axis (ϕ_1) at the base of the T-bar where it connected to the load cell as well as free rotation (ϕ_2) at the intersection between the stem and head of the T-bar interface. Thus, when pushing against this unstable interface the torso was free to move laterally and to rotate in parallel with the vertical axis of the spine. The interface designs were not intended to represent a specific workplace handle but rather to demonstrate the role of interface instability with regard to muscle recruitment and spinal load. Nonetheless, the experimental paradigm may be generalizable to handle designs that permit movement in the coronal and transverse planes when performing a pushing task. The rigid and unstable interfaces had the same mass and length, 3.6 kg and 89 cm respectively.

A single trial consisted of a subject performing an isometric trunk flexion exertion against the interface for 5 s. The subjects observed a real time display of the applied horizontal push force and were required to match this to a prescribed target force. Three trials were performed at each of three force levels (20%, 35% and 50% of the maximum isometric voluntary exertion (MVE) level). These conditions were tested with the rigid and unstable interfaces in separate trials. Maximum horizontal push forces were recorded as the larger of two steady-state maximal flexion exertions against the rigid interface. Maximum push forces were also recorded while using the unstable interface for comparison with MVE data. The range of force levels of 20% to MVE were chosen to investigate muscle recruitment throughout a range of exertion levels. A minimum of 60 s of rest was provided between trials to avoid effects of fatigue.

Subjects participated in a practice session consisting of ten successful trials at pushing forces of 20% and 35% of MVE using both the rigid and the unstable interfaces. Pilot data qualitatively indicated that these practice trials were sufficient for the subjects to achieve repeatable muscle recruitment patterns.

Trunk postures were recorded using surface mounted electromagnetic tracking sensors (Ascension Technology Corp., Burlington, VT, USA). Sensors were placed over the subject's spinous processes at T10 and S1 and over the manubrium. It should be noted that the upper surface of the T-bar interface was approximately 4 cm below the clavicle and manubrium, thereby avoiding contact between the manubrium sensor and the interface. Although the isometric postures were nominally upright, the flexion, twisting and lateral angles were recorded during each trial for inputs into the EMG-assisted spinal compression model used in the biomechanical assessment. Data were also recorded from the wall mounted 6-df load cell and a ground reaction force platform (Bertec, Columbus, OH, USA).

EMG signals were collected from bipolar surface electrodes (Delsys, Boston, MA, USA) on the left and right rectus abdominus (RA), lumbar paraspinal (LP), EO and IO muscle groups. Electrodes were placed for the RA 3 cm lateral and 2 cm superior to the umbilicus, for the EO 10 cm lateral to the umbilicus with an orientation of 45° to vertical, for the posterior IO 8 cm lateral to the midline within the lumbar triangle at a 45° orientation and for the erector spinae 4 cm lateral to the L3 spinous process. All EMG data were band-pass filtered in hardware between 20 and 500 Hz, sampled at 1000 Hz and rectified. The EMG signals, kinematic and kinetic data were filtered using a 6 Hz, low-pass, seventh-order Butterworth filter in post-processing software (Matlab, Natick, MA, USA). EMG signals from each muscle were

normalized to the mean steady-state values of EMG recorded during maximum voluntary isometric flexion, extension, right-lateral twisting and left-lateral twisting. Mean normalized and integrated EMG amplitudes were determined from the middle 3 s of each 5 s trial.

2.3. Biomechanical assessment

Spinal compression was estimated from a previously published EMG-assisted model of spinal load (Granata and Marras 1995a, Marras and Granata 1997). Model input included the measured and processed EMG, ground reaction force data and trunk kinematic data. Eight muscle equivalents were represented in the model, including the right and left RA, EO, IO and LP muscle groups. Anthropometric origin, insertion and cross-sectional area of each muscle were established from published anatomical data (Jorgensen *et al.* 2001, 2003). The magnitude of force in each muscle was determined from the product of normalized muscle activation, α_i , muscle cross-sectional area, $Area_i$, and muscle Gain (equation 1). Modulating factors were included to represent physiologic effects of force-length, $f(Len_i)$ and force-velocity, $f(Vel_i)$, as described in previous reports of the spine model (Granata and Marras 1995a, Marras and Granata 1997).

$$F_i = Gain \times \alpha_i \times Area_i \times f(Len_i) \times f(Vel_i) \quad i = 1 \dots 8 \quad (1)$$

$$0 < \alpha_i < 1 \quad (2)$$

Since the exertions represented an isometric upright posture, the force-length and force-velocity values were set to unity. Muscle activation, α_i , for each muscle i was set equal to the mean normalized EMG recorded during the experimental trials, i.e. bounded between 0 and 1 (equation 2). Gain represented force capacity per unit cross-sectional area of the muscles and the gains were calibrated for each subject from equations and measurements of trunk moment.

$$M_{ext} = \sum_{i=1}^n F_i \{r_i \times \hat{F}_i\} \quad n = 8 \quad (3)$$

Trunk moment generated by the sum of muscle forces must be equal and opposite to the measured external trunk moment (equation 1). Measured external trunk moments, M_{Ext} , were calculated about the L5/S1 joint using inverse dynamic analyses from measured kinematic data and ground reaction forces (Granata *et al.* 1995). The moment generated by each muscle was described as the product of the muscle force magnitude, F_i , and its moment generating capacity, $\{r_i \times \hat{F}_i\}$. The moment generating capacity of each muscle determined from the vector product between the unit vector direction of muscle force, $\hat{F}_i$, and the muscle origin, r_i . It should be noted that simultaneous solution of equations 1 and 3 with input data of equation 2 provides an estimate of muscle force, F_i , in each muscle, $i = 1 \dots 8$, as well as the Gain coefficient. To be physiologically valid the predicted Gain values representing the force capacity per unit cross-sectional area must be between 30 and 100 N cm⁻² (McGill and Norman 1987, Reid, and Costigan 1987). The vector sum of muscle forces and external loads was used to estimate spinal load during the exertions.

2.4. Biomechanical prediction

The hypothesis investigated in this study was that increased muscle recruitment is necessary to provide compensatory stabilization when pushing against the unstable interface, whereas less co-contraction is necessary when pushing against a stable rigid interface. To demonstrate that this trend is predictable a separate theoretical analysis was conducted by modifying the model described above. The set of muscle activations, α_i , necessary to minimize the sum of squared muscle stresses during the pushing exertions was estimated. The solution set of muscle

activations must satisfy three constraints. First, the predicted set of muscle activations, α_i , must be bounded between zero and 1 to represent normalized EMG activity (equation 2). Second, the set of muscle forces must achieve equilibrium with the external loads (equation 3). Third, the predicted set of muscle forces and associated stiffness must satisfy stability. Stability is achieved when the eigenvalues of the Hessian matrix of potential energy, V , are greater than zero (equation 4) as described in previous studies (Granata and Wilson 2001), eigenvalues

$$\text{eigenvalues} \left\{ \frac{\delta^2}{\delta \theta_i \delta \theta_j} / (V_{trunk} + V_{interface}) \right\} > 0 \quad (4)$$

V_{Trunk} is the potential energy of the musculoskeletal system and θ are the possible trunk rotation angles in the sagittal, lateral and transverse planes. V_{Trunk} has been described elsewhere (Cholewicki and McGill 1996) and includes components from active muscle forces, trunk mass and applied loads:

$$V_{Trunk} = \sum_i F_i (l_i - l_{i0}) + 1 / 2 K_i (l_i - l_{i0})^2 \quad (5)$$

where l_i, l_{i0} and K_i are the instantaneous muscle length, equilibrium length and active stiffness of muscle i respectively. Active stiffness was considered proportional to muscle force, $K_i = q F_i l_i^{-1}$ where q is a non-dimensional proportionality constant $q = 20$ as described by Bergmark (1989). Muscle lengths were determined from the posture dependent origin and insertion geometry from the model. Trunk posture was assumed to be upright and sagittally symmetric. Potential energy attributed to the loads applied through the interface, $V_{Interface}$, was modelled as a 2-df mechanical linkage as shown in figure 2 with potential energy expressed as:

$$V_{interface} = -F_{push} L \cos \phi_1 + 1 / 2 K_1 \phi_1^2 + \{ M_Z \phi_2 + 1 / 2 K_2 \phi_2^2 \} \quad (6)$$

where F_{push} is the horizontal force applied by the subjects, L is the distance from the load cell to the subject, M_Z is the moment about the long axis of the spine and ϕ_1, ϕ_2 are the rotation angles of each df of the interface. Stiffness elements K_1, K_2 , were provided to represent resistance to rotation about the interface axes of ϕ_1 and ϕ_2 . The unstable interface was characterized by setting these stiffness values to zero while the rigid interface was represented with extremely large rotational stiffness, $K_1 K_2$ 2000 Nm/degree.

This theoretical model was used to predict trends in muscle activation, α_i , when using the rigid and unstable interfaces. Input included the measured trunk moments and posture (i.e. upright); the model outputted optimal muscle activations. The model successfully converged to optimal solutions satisfying the three mechanical constraints (equations 2, 3, 4). It should be noted that it is unlikely that the predicted α_i would exactly match the measured EMG due to the simplicity of the model. Nonetheless, results are sufficient to demonstrate the biomechanical necessity of modified co-contraction with interface design. Theoretical trends in spinal load were also computed from the vector sum of external loads, gravitational loads and muscle forces predicted by the optimal muscle activities associated with the rigid and unstable interfaces.

2.5. Statistical analysis

Statistical analyses were performed to determine the effects of interface stability and exertion level on measured muscle EMG activity and spinal compression estimated from the EMG-assisted model. A two factor repeated measures ANOVA was performed to test the effects of within-subject independent variables of interface stability (rigid or unstable) and exertion level (push force of 20%, 35% or 50% MVE). Dependent variables included mean EMG from each of the four recorded muscle groups and spinal compression and were tested in separate analyses. Analyses were performed using commercial software (Statistica 4.5, Statsoft, Inc., Tulsa, OK, USA) with a significance level of $\alpha=0.05$.

3. Results

Results demonstrated that there was no significant difference between left and right muscles within each muscle group. Therefore, reported muscle activity is the average of the left and right muscle of each of the four muscle groups.

Theoretical analyses predicted greater activity in the IO and EO muscles when using the unstable interface compared to recruitment when using the rigid interface. Predicted muscle activities are provided in figure 3. Both the rigid and unstable interface conditions resisted motion in the sagittal plane so very little difference was observed in the paraspinal and RA muscles. Recognizing that spinal load increases with co-contraction (Granata and Marras 1995b), it was not surprising that the theoretical analyses also predicted greater spinal load when pushing against the unstable handle, as illustrated in figure 4.

During the experimental tasks the subjects generated trunk flexion exertions equivalent to 20%, 35% and 50% of their MVE. The applied horizontal forces increased significantly ($p < 0.05$) with exertion effort. Mean (standard deviation) horizontal force during the 20% MVE exertions was 55 (30.5) N and increased to 96 (53.3) N at the 35% MVE level and 137 (76.2) N in the 50% MVE trials. The large variation within each exertion level is attributed to the range of size and strength of subjects. Although all target forces were assigned based upon MVE exertions recorded when using the rigid interface, maximum exertions were also recorded using the unstable interface for comparison. Mean maximum horizontal push force when using the unstable interface, 248 (136) N, was significantly less ($p < 0.02$) than maximum horizontal force recorded when using the stable interface, 295 (133) N.

Measured EMG activity agreed with trends predicted by the theoretical analyses. Normalized EMG activities in the EO and IO were significantly greater ($p < 0.05$) when using the unstable interface compared to the rigid interface. Summary statistics are provided in table 1 and illustrated in figure 5. Mean EMG activity in the EO was 14.5% (8.1%) of maximum when using the rigid interface and 15.1% (8.1%) when using the unstable interface. Average EMG in the IO was 25.3% (15.5%) of maximum when using the rigid interface and 33.3% (19.41%) when using the unstable interface. Muscle activities in the RA and LP did not differ significantly between interface conditions. Muscle activity increased significantly ($p < 0.01$) with flexion exertion effort in the RA, IO and EO muscles. RA increased from 17.4% (14.2%) at the lowest exertion level to 44.3% (31.2%) at the highest exertion level. Similarly EMG from EO and IO increased from 13.3% (5.3%) and 23.6% (18.9%) respectively at the 20% MVE exertion level to 17.1% (6.6%) and 37.5% (20.3%) at the 50% exertion. Activity in the LP muscle failed to achieve statistical significance with main effect of flexion exertion effort.

Spinal compression estimated from the EMG-assisted model was significantly greater ($p < 0.05$) during tasks performed using the unstable interface compared to the stable interface (table 1). During flexion exertions against the stable interface mean spinal compression was 1420 (596) N and this increased by 31.7% to a value of 1870 (608) N during exertions with the unstable interface. Spinal compression also increased significantly ($p < 0.01$) with exertion level. Mean spinal load was 1273 (448) N during the 20% MVE flexion exertions and 2017 (611) N during the 50% MVE exertions. Mean value of Gain calculated by the EMG-assisted model was 68.4 (23.2) N cm^{-1} (equation 1) and thereby fell within the physiologically acceptable range of 30-100 N cm^{-1} .

4. Discussion

The goal of the present study was to examine muscle recruitment and spinal compression with regard to interface stability during isometric pushing tasks. A theoretical model was implemented to demonstrate that interface design must influence co-contraction recruitment

in the torso musculature and associated spinal load. Measured muscle activity agreed with the predicted trends. Muscle recruitment patterns and biomechanical loads on the spine during MMH tasks are affected by external forces and vector directions (Schultz *et al.* 1982). Likewise, the stability of the external load influences the stability of the spine; for example, it is more difficult to maintain spinal stability when external loads impose destabilizing effects (Van Dieën *et al.* 2003). One of the mechanisms employed to control spinal stability is recruitment of torso muscle co-contraction (Bergmark 1989, Granata and Marras 2000). Trunk stiffness increases with muscle activity and antagonistic co-contraction during trunk extension exertions (Cholewicki *et al.* 1999, Gardner-Morse and Stokes 2001, Lee *et al.* 2006) with similar effects expected during lateral and twisting exertions. The unstable interface was designed to allow free motion in the lateral and twisting axes of torso movement during the pushing exertion. The IO and EO muscle groups are the primary contributors to lateral and twisting movement of the trunk (Marras and Granata 1995, 1996). Therefore, to avoid movement error in those directions the model predicted increased activity in the IO and EO muscles when simulating pushing exertions against the unstable interface as shown in figure 3. However, in the sagittal plane of movement both the rigid and the unstable interfaces restricted flexion-extension movement. Therefore, the theoretical analyses predicted no remarkable difference in RA or LP muscle activity between the two interface designs. Measured data agreed with the trends of increased activity in the IO and EO muscle groups associated with the unstable interface. However, measured values of muscle activity were greater than theoretically predicted activities when using the rigid interface and smaller than the predicted activities when using the unstable interface. However, measured data demonstrated a more severe influence of interface stability than predicted by the model with regard to the RA and LP muscles. For example, mean antagonistic activity in the LP muscles predicted by the theoretical analysis was only 4.6% of maximum but measured LP EMG was 14.5% MVE. Despite some differences, both the theoretical analyses and the EMG measurements demonstrated increased muscle recruitment with interface instability, particularly in the IO and EO muscle groups.

Results demonstrated greater measured EMG in the EO and IO muscle groups when pushing against the unstable interface vs. trials recorded with the rigid interface. RA and abdominal EO are considered to be the primary muscles for performing flexion exertions (McGill 1995). Measurements showed a significant increase in activity in these muscles with flexion exertion effort (table 1). Activity in the IO muscles was recorded from the posterior aspect of the torso. Despite the expected contribution to trunk extension the IO increased significantly with flexion exertion. Similarly, notable LP EMG activity was observed in each of the flexion trials despite the fact that the LP muscles provide no mechanical advantage for the generation of a trunk flexion movement. Recent evidence suggests that LP and IO muscle recruitment during flexion exertions may be necessary to maintain the stability of the spine against compressive buckling loads imposed by the forces from RA and EO muscles (Granata *et al.* 2005). This is supported by pair-wise comparison results that demonstrate that EMG measured from the LP muscles was greater during the 50% MVE flexion exertions than during the 20% flexion efforts. However, a more complex model is necessary to predict this antagonistic activity during flexion exertions; that is, the model must include a multi-segment spine to accurately predict LP activity during flexion exertions (Granata *et al.* 2005).

The IO and EO muscle groups demonstrated significant main effects for interface condition in addition to their contribution to control of flexion moment. Van Dieën (2003) observed increased antagonistic co-contraction when lifting an unstable load compared to lifting a similarly heavy but more stable load. In fact, studies demonstrate that co-contraction is modified in association with stability requirements even when trunk moment is held constant across stability conditions (Granata and Orishimo 2001). Likewise, in the present study co-contraction was increased when the applied external load was unstable, even though the applied push force was similar in both interface conditions. Stability of the interface can be interpreted

by considering the potential energy of the interface with regard to the applied load. Clearly the potential energy generated by the push force with regard to the load cell is at a maximum in the illustrated configuration of figure 2 (equation 6). This suggests a tendency for the system to collapse laterally in order to reduce the potential energy. To overcome this tendency, the muscles controlling lateral movement must increase activity to stiffen the trunk-plus-interface system against movement errors. Recognizing that the destabilizing behaviour was proportional to the applied push force, the maximum horizontal force when using the unstable interface was significantly less than the MVE force with the rigid interface. Similar decrements in push force have been observed in stabilization tasks associated with hand tool design (Roy 1999). These results indicate that flexion exertion effort and interface design independently contribute to muscle recruitment patterns.

Interface design influenced compressive load on the spine. Research demonstrates that co-contraction contributes to spinal load (Granata and Marras 1995b, Hughes *et al.* 1995). Trunk flexion-extension moments during pushing tasks are typically less than during lifting exertions, so spinal compression during pushing tasks is often assumed to be less than during lifting (Schibye *et al.* 2001). However, co-contraction during trunk flexion exertions is greater than during extension (Granata *et al.* 2005). Therefore, when estimating spinal load, it is necessary to account for co-contraction of the torso muscles. An EMG-assisted analysis was implemented to estimate spinal compression from measured muscle activity and external forces. Spinal compression when using the unstable interface was 31% greater than when using the rigid interface despite the fact that similar horizontal push forces were applied in the two different conditions. Nonetheless, the estimated compression loads were typically less than 3400 N during the pushing exertions. It should be noted, however, that this estimate of spinal load was limited to steady-state isometric behaviour, whereas co-contraction and spinal load may increase with task dynamics (Marras and Sommerich 1991, Granata and Marras 1995b). Furthermore, both interfaces restricted flexion-extension movement, thereby augmenting sagittal stability. In an industrial task such as pushing a cart, the load may contribute less sagittal stability than measured in the present study and potentially require greater co-recruitment of muscles. Thus, it is believed that the influence of interface design (stability) on spinal load observed here may give conservative estimates for effect size. Nonetheless, the results demonstrate that the handle design or interface for a pushing task influences the biomechanical risks associated with spinal compression.

Although these results provide insight into handle design for MMH tasks, four limitations should be considered. First, the experimental interface used in the present study was not intended to represent any specific industrial situation. Instead, the two interfaces were purposely created to demonstrate possible differences between a rigid and unstable interface in isometric pushing exertions. Results supported the hypothesis that the interface stability in a pushing task can influence risk associated with musculoskeletal load. Specific handle designs must be examined to assess risk effects in the workplace. Second, the experimental protocol attempted to isolate trunk biomechanics by having subjects perform flexion exertions with minimal use of the shoulders and arms. Conversely, in the industrial workplace, pushing tasks place physical demands on the shoulders and arms in addition to biomechanical loads on the torso and spine. Evidence suggests that musculoskeletal compliance in the upper limbs may influence trunk stability, co-contraction and spinal load (Rancourt and Hogan 2001). Further studies are necessary to understand the contribution of shoulder and arm mechanics to musculoskeletal risk during MMH pushing tasks. The surface contact area between the interface and the subject was large, 240 cm², but it is possible that discomfort may have limited the maximum voluntary force when pushing against the interface. However, the surface contact areas of the rigid and unstable interfaces were identical and presentation order of the rigid and unstable interfaces was randomized. Therefore, it is highly unlikely that discomfort contributed to the differences observed between the two different interface conditions. Third, although the

EMG-assisted analysis was capable of accounting for co-contraction, these techniques are based on trunk extension tasks and have never been validated for trunk flexion exertions. Davis and Mirka (2000) demonstrated that the abdominal oblique muscle groups may require multiple vector representations to appropriately represent the lines of force. Likewise, McGill (1996) concluded that the mechanisms of the flexor musculature may be more complex than is typically represented in EMG-assisted models. Ongoing studies are developing improved EMG-assisted estimates of spinal load that will be applicable for flexion exertions. Finally, some have suggested that risk of injury from shear forces on the spine may play a larger role during pushing exertions than during lifting tasks. Further research is necessary to improve the biofidelity of the spinal models in order to investigate the risk of shear overload injury during pushing tasks.

5. Conclusion

The results demonstrate that handle or interface stability may influence spinal load and associated risk of musculoskeletal injury during MMH tasks that include pushing exertions. Pushing and pulling exertions account for half of industrial MMH tasks (Baril-Gingras and Lortie 1995). The design of these workplace tasks can significantly influence muscle recruitment and spinal compression (Hoozemans *et al.* 2004). Modifying the workplace to replace lifting tasks with pushing exertions is often recommended, based upon estimates of spinal load. However, it is necessary for estimates of spinal load to account for muscle co-contraction, which can be significantly greater during trunk flexion exertions than during trunk extension exertions (Granata *et al.* 2005). A biomechanical model predicted that task and handle stability can influence torso muscle co-contraction and associated spinal load. Empirical measurements confirmed that trunk muscle activity in the IO and EO muscle groups was greater when pushing against an unstable interface than when pushing against a rigid interface. This co-contraction resulted in increased spinal load when using the unstable interface. Therefore, it is recommended that handles used for the workplace pushing tasks should be designed for greatest possible stability.

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References

- BARIL-GINGRAS G, LORTIE M. The handling of objects other than boxes: univariate analysis of handling techniques in a large transport company. *Ergonomics* 1995;38:905–925.
- BERGMARK A. Stability of the lumbar spine: a study in mechanical engineering. *Acta Orthopaedica Scandinavica*, Suppl 1989;230:1–54.
- CHOLEWICKI J, JULURU K, MCGILL SM. Intra-abdominal pressure mechanism for stabilizing the lumbar spine. *Journal of Biomechanics* 1999;32:13–17. [PubMed: 10050947]
- CHOLEWICKI J, MCGILL SM. Mechanical stability on the in vivo lumbar spine: implications for injury and chronic low back pain. *Clinical Biomechanics* 1996;11:1–15. [PubMed: 11415593]
- DAMKOT DK, POPE MH, LORD J, FRYMOYER JW. The relationship between work history, work environment and low-back pain in men. *Spine* 1984;9:395–399. [PubMed: 6236564]
- DAVIS KG, MARRAS WS, WATERS TR. Reduction of spinal loading through the use of handles. *Ergonomics* 1998;41:1155–1168. [PubMed: 9715674]
- DAVIS JR, MIRKA GA. Transverse-contour modeling of trunk muscle-distributed forces and spinal loads during lifting and twisting. *Spine* 2000;25:180–189. [PubMed: 10685481]
- DE LOOZE MP, STASSEN ARA, MARKSLAG AMT, BORST MJ, WOONING MM, TOUSSAINT HM. Mechanical loading of the low back in three methods of refuse collecting. *Ergonomics* 1995;38:1993–2006. [PubMed: 7588581]

- GARDNER-MORSE MG, STOKES IAF. Trunk stiffness increases with steady-state effort. *Journal of Biomechanics* 2001;34:457–463. [PubMed: 11266668]
- GRANATA KP, LEE PE, FRANKLIN TC. Co-contraction recruitment and spinal load during isometric trunk flexion and extension in upright postures. *Clinical Biomechanics* 2005;20:1029–1037. [PubMed: 16154249]
- GRANATA KP, MARRAS WS. An EMG assisted model of biomechanical trunk loading during free-dynamic lifting. *Journal of Biomechanics* 1995a;28:1309–1317. [PubMed: 8522544]
- GRANATA KP, MARRAS WS. The influence of trunk muscle coactivity upon dynamic spinal loads. *Spine* 1995b;20:913–919. [PubMed: 7644956]
- GRANATA KP, MARRAS WS. Cost-benefit of muscle co-contraction in protecting against spinal instability. *Spine* 2000;25:1398–1404. [PubMed: 10828922]
- GRANATA KP, MARRAS WS, FATHALLAH FA. A method for measuring external trunk loads during dynamic lifting exertions. *Journal of Biomechanics* 1995;29:1219–1222. [PubMed: 8872281]
- GRANATA KP, ORISHIMO K. Response of trunk muscle coactivation to changes in spinal stability. *Journal of Biomechanics* 2001;34:1117–1123. [PubMed: 11506782]
- GRANATA KP, WILSON SE. Trunk posture and spinal stability. *Journal of Biomechanics* 2001;16:650–659.
- HOOZEMANS M, KUIJER PP, KINGMA I, VAN DIEËN J, DE VRIES W, VAN DER WL, VEEGER DJ, VAN DER BA, FRINGS-DRESEN M. Mechanical loading of the low back and shoulders during pushing and pulling activities. *Ergonomics* 2004;47:1–18. [PubMed: 14660215]
- HOOZEMANS MJ, VAN DER BEEK AJ, FRINGS-DRESEN MH, VAN DIJK FJ, VAN DER WOUDE LH. Pushing and pulling in relation to musculoskeletal disorders: a review of risk factors. *Ergonomics* 1998;41:757–781. [PubMed: 9629062]
- HUGHES RE, BEAN JC, CHAFFIN DB. Evaluating the effect of co-contraction in optimization models. *Journal of Biomechanics* 1995;28:875–878. [PubMed: 7657686]
- JORGENSEN MJ, MARRAS WS, GRANATA KP, WIAND JW. MRI-derived moment-arms of the female and male spinal loading muscles. *Clinical Biomechanics* 2001;16:182–193. [PubMed: 11240052]
- JORGENSEN MJ, MARRAS WS, GUPTA P. Cross-sectional area of the lumbar back muscles as a function of torso flexion. *Clinical Biomechanics* 2003;18:280–286. [PubMed: 12689777]
- KUMAR S. The back compressive forces during maximal push-pull activities in the sagittal plane. *Journal of Human Ergology* 1994;23:133–150. [PubMed: 7730598]
- LEE KS, CHAFFIN DB, HERRIN GA, WAIKAR AM. Effect of handle height on lower-back loading in cart pushing and pulling. *Applied Ergonomics* 1991;22:117–123. [PubMed: 15676806]
- LEE P, ROGERS E, GRANATA KP. Active trunk stiffness increases with co-contraction. *Journal of Electromyography and Kinesiology* 2006;16:51–57. [PubMed: 16099678]
- MARRAS WS, GRANATA KP. A biomechanical assessment and model of axial twisting in the thoracolumbar spine. *Spine* 1995;20:1440–1451. [PubMed: 8623063]
- MARRAS WS, GRANATA KP. Spine loading during trunk lateral bending motions. *Journal of Biomechanics* 1996;30:697–703. [PubMed: 9239549]
- MARRAS WS, GRANATA KP. The development of an EMG-assisted model to assess spine loading during whole-body free-dynamic lifting. *Journal of Electromyography and Kinesiology* 1997;7:259–268. [PubMed: 11369269]
- MARRAS WS, SOMMERICH CM. A three-dimensional motion model of loads on the lumbar spine: II Model validation. *Human Factors* 1991;33:139–149. [PubMed: 1860701]
- MCGILL SM. The mechanics of torso flexion: situps and standing dynamic flexion manoeuvres. *Clinical Biomechanics* 1995;10:184–192. [PubMed: 11415551]
- MCGILL SM. A revised anatomical model of the abdominal musculature for torso flexion efforts. *Journal of Biomechanics* 1996;29:973–977. [PubMed: 8809629]
- MCGILL SM, NORMAN RW. Effects of an anatomically detailed erector spinae model on L4-S1 disc compression and shear. *Journal of Biomechanics* 1987;20:591–600. [PubMed: 3611135]
- RANCOURT D, HOGAN N. Stability in force-production tasks. *Journal of Motor Behavior* 2001;33:193–204. [PubMed: 11404214]

- REID JG, COSTIGAN PA. Trunk muscle balance and muscular force. *Spine* 1987;12:783–786. [PubMed: 2961091]
- ROY, PO. Analysis of contact task stabilization strategies by the upper limb. 1999. M.S. Thesis. University Laval
- SCHIBYE B, SOGAARD K, MARTINSEN D, KLAUSEN K. Mechanical load on the low back and shoulders during pushing and pulling of two-wheeled waste containers compared with lifting and carrying of bags and bins. *Clinical Biomechanics* 2001;16:549–559. [PubMed: 11470296]
- SCHULTZ AB, ANDERSSON GBJ, ORTENGREN R, HADERSPECK K, NACHEMSON A, GOTEGORG S. Loads on the lumbar spine. *Journal of Bone and Joint Surgery* 1982;64:713–720. [PubMed: 7085696]
- THELEN DG, ASHTON-MILLER JA, SCHULTZ AB. Co-contraction of lumbar muscles during the development of time-varying triaxial moments. *Journal of Orthopaedic Research* 1995;13:390–398. [PubMed: 7602401]
- VAN DIEËN JH, KINGMA I, VAN DER BUG JC. Evidence for a role of antagonistic cocontraction in controlling trunk stiffness during lifting. *Journal of Biomechanics* 2003;36:1829–1836. [PubMed: 14614936]

Figure 1.
Experimental arrangement.

Figure 2.

Mechanical schematic of the T-bar interface viewed from above, including the horizontal push force generated by the subject. Potential energy in the system can be expressed as $V_{\text{Interface}} = \{-F_{\text{Push}} L \cos \phi_1 + 1/2 K_1 \phi_1^2\} + \{M_Z \phi_2 + 1/2 K_2 \phi_2^2\}$ where F_{Push} is push force, L is the distance from the load cell to the subject, M_Z is the moment applied about ϕ_2 , and K_1, K_2 represent elastic resistance to rotation about the interface axes of ϕ_1 and ϕ_2 .

Figure 3.

Theoretical prediction of muscle activity in the internal oblique (black lines) and external oblique (grey lines) muscles for the stable (filled symbols) and unstable (open symbols) interfaces. MVE = maximum isometric voluntary exertion; IO = internal oblique; EO = external oblique.

Figure 4.
Theoretical predictions of spinal compression with the stable and unstable interfaces.

Figure 5.

Measured electromyographic (EMG) activity with the rigid interface and the unstable interface (with values averaged across exertion levels). *denotes statistically significant differences at $p < 0.05$. MVE = maximum isometric voluntary exertion; RA = rectus abdominus; IO = internal oblique; LP = lumbar paraspinal; EO = external oblique.

Table 1

Experimental results. Mean (standard deviation) of applied push force, normalized electromyographic (EMG) activity and spinal compression recorded during the isometric flexion exertions.

	Exertion level		
	20% MVE	35% MVE	50% MVE
Rigid interface			
EMG RA (%) ^a	16.7 (13.9)	27.3 (14.8)	42.1 (34.5)
EMG IO (%) ^{,ab}	19.4 (10.8)	25.0 (16.5)	31.6 (16.7)
EMG LP (%)	12.6 (4.2)	12.3 (8.4)	19.9 (10.3)
EMG EO (%) ^{,ab}	12.9 (5.8)	14.4 (6.3)	16.3 (7.4)
Spinal compression (N) ^{,ab}	1016 (445)	1468 (497)	1736 (625)
Unstable interface			
EMG RA (%) ^a	18.1 (15.8)	28.2 (18.8)	46.5 (28.5)
EMG IO (%) ^{,ab}	23.9 (19.7)	26.2 (17.5)	49.9 (17.5)
EMG LP (%)	11.1 (5.0)	12.8 (11.5)	18.2 (12.1)
EMG EO (%) ^{,ab}	13.3 (6.0)	14.8 (6.7)	17.3 (7.7)
Spinal compression (N) ^{,ab}	1530 (457)	1954 (528)	2126 (605)

MVE = maximum isometric voluntary exertion; RA = rectus abdominus; IO = internal oblique; LP = lumbar paraspinal; EO = external oblique.

^a Significant main effect for exertion level.

^b Significant main effect for interface stability.