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Low-Back Biomechanics and Static Stability During Isometric Pushing

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Pushing and pulling tasks are increasingly prevalent in industrial workplaces. Few studies have investigated low-back biomechanical risk factors associated with pushing, and we are aware of none that has quantified spinal stability during pushing exertions. Data recorded from 11 healthy participants performing isometric pushing exertions demonstrated that trunk posture, vector force direction of the applied load, and trunk moment were influenced ($p < .01$) by exertion level, elevation of the handle for the pushing task, and foot position. A biomechanical model was used to analyze the posture and hand force data gathered from the pushing exertions. Model results indicate that pushing exertions provide significantly ($p < .01$) less stability than lifting when antagonistic cocontraction is ignored. However, stability can be augmented by recruitment of muscle cocontraction. Results suggest that cocontraction may be recruited to compensate for the fact that equilibrium mechanics provide little intrinsic trunk stiffness and stability during pushing exertions. If one maintains stability by means of cocontraction, additional spinal load is thereby created, increasing the risk of overload injury. Thus it is important to consider muscle cocontraction when evaluating the biomechanics of pushing exertions. Potential applications of this research include improved assessment of biomechanical risk factors for the design of industrial pushing tasks.

INTRODUCTION

Biomechanical analyses of pushing and pulling tasks are necessary to improve the current understanding of spinal load, spinal stability, and the associated risk of musculoskeletal injury. Lifting has been historically cited as a significant risk factor for occupationally related low-back disorders (Andersson, 1981; Marras et al., 1995). Industry has responded to this risk by modifying the workplace in order to decrease lifting and carrying tasks, often replacing them with pushing and pulling exertions. However, there is epidemiologic risk associated with pushing and pulling tasks as well. Of all industrial back injuries in the United States, Canada, and the U.K., 20% have been attributed to push or pull activities (Damkot, Pope, Lord, & Frymoyer, 1984; Hoozemans, van der Beek, Frings-Dresen, van Dijk, & van der Woude, 1998). It is expected that this injury rate will increase in response to the trend toward a growing number of push-related tasks in the workplace. Despite the fact that 50% of industrial manual materials handling includes pushing and pulling tasks (Baril-Gingras & Lortie, 1995), the biomechanics of pushing exertions remains poorly understood (Schibye, Sogaard, Martinsen, & Klausen, 2001; van der Beek, Hoozemans, Frings-Dresen, & Burdorf, 1999). Specifically, we are aware of no previously published analyses attempting to quantify spinal stability during pushing exertions.

Biomechanical risk factors for low-back disorders include trunk moment and external force, given their relationship with spinal load and stability (Chaffin & Page, 1994; Granata & Marras, 1996; National Institute for Occupational Safety and Health, 1981). Few studies have quantified

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trunk moment during pushing exertions (deLooze, van Greuningen, Rebel, Kingma, & Kuijer, 2000; Kumar, 1994). Research conducted in industrial settings has reported that workers lean against the objects to be pushed, resulting in vertical as well as horizontal forces (deLooze et al., 1995; van der Beek, Kluver, Frings-Dresen, & Hoozemans, 2000). The vector direction of the external force tends to pass close to the lumbosacral junction of the spine during these pushing tasks, thereby minimizing trunk moments (deLooze et al., 2000). Because moment is thought to be less during pushing than during lifting, preliminary estimates suggest spinal load is also less during pushing than during lifting (deLooze et al., 1995; Schibye et al., 2001). However, those analyses neglected the influence of trunk muscle cocontraction (Lee, Chaffin, Waikar, & Chung, 1989). Trunk muscle cocontraction is known to dramatically increase spinal load and may contribute to risk of low-back injury during pushing tasks. Estimates of stability suggest high levels of cocontraction must be recruited during pushing exertions. Therefore spinal stability during pushing exertions should be investigated.

To maintain spinal stability, the bending stiffness of the spinal column must increase in proportion to the applied compressive load (Meakin, Hukins, & Aspden, 1996). In the absence of muscular support, the bending stiffness of the osteoligamentous spine is small (Stokes, Gardner-Morse, Churchill, & Laible, 2002), and the spinal column is unstable under combined external and anatomic loads that exceed 88 N (Crisco & Panjabi M.M., 1992; Crisco, Panjabi, Yamamoto, & Oxland, 1992). Fortunately, stiffness of active skeletal muscles increases with force (Kearney & Hunter, 1990; Morgan, 1977) such that recruitment of the paraspinal muscles can augment the bending stiffness of the trunk and spine (Cholewicki, Jurulu, Radebold, Panjabi, & McGill, 1999; Gardner-Morse & Stokes, 2001; Kettler, Hartwig, Schultheis, Claes, & Wilke, 2002). Paraspinal muscle activation increases with lifting effort (Chaffin, 1969; Schultz & Anderson, 1981). Hence, during lifting exertions, the muscle activity recruited to achieve equilibrium also contributes to spinal stability.

Conversely, during pushing exertions very little paraspinal muscle activation is necessary to achieve equilibrium (deLooze et al., 2000; Kumar, 1994). This may suggest that equilibrium conditions of pushing may be less stable than equilibrium conditions of lifting. Additional stability can be achieved through neuromotor recruitment of muscle stiffness by means of cocontraction (Bergmark, 1989; Cholewicki, Panjabi, & Khachatryan, 1997; Gardner-Morse, Stokes, & Laible, 1995; Granata & Wilson, 2001). When equilibrium conditions become increasingly unstable, then greater demand is placed on the neuromuscular controller to recruit antagonistic cocontraction. The purpose of the current study was to quantify external load vectors and trunk moment during pushing to estimate the equilibrium level of stability at a variety of effort levels and handle heights. Results illustrate the need to consider muscle cocontraction when considering pushing exertion.

METHODS

Model

Stability associated with equilibrium levels of exertion were estimated from a two-dimensional biomechanical model (see Appendix). A simple sagittal-plane inverted-pendulum representation of the spine (Figure 1) shows that potential energy with respect to the base of the spine is related to the external force, F_{Ext}, the weight of the trunk, Mg, and muscle controlled trunk stiffness, k,

$$V = Mg\, d_{\text{cm}} \cos\theta_{\text{cm}} + F_{\text{Ext}} \{d_{\text{cm}} \cos(\theta_{\text{cm}} - \phi) + d_{\text{F}} \cos(\theta_{\text{F}} - \phi)\} + \tfrac{1}{2}k(\theta_{\text{cm}} - \theta_0)^2, \quad (1)$$

in which d_{cm} represents length of the vector from L5-S1 junction to the trunk center of mass (CM) and d_{F} is the vector length from CM to the applied force at the push handle. Angles θ_{cm}, θ_{F}, and ϕ are the angles of the vector d_{cm}, d_{F}, and the angle of the force vector with respect to vertical. The neutral angle θ_0 represents the equilibrium of the vector d_{cm} (Cholewicki & McGill, 1996). Trigonometric terms result from the scalar product between the forces, F_{Ext}, Mg, and position vectors, d_{F} and d_{cm}. Equilibrium is determined from the negative first derivative with respect to trunk angle, θ_{cm} (Appendix),

$$M_{\text{LS}} = Mg\, d_{\text{cm}} \sin\theta_{\text{cm}} + F_{\text{Ext}}\, d_{\text{cm}}\sin(\theta_{\text{cm}} - \phi), \quad (2)$$

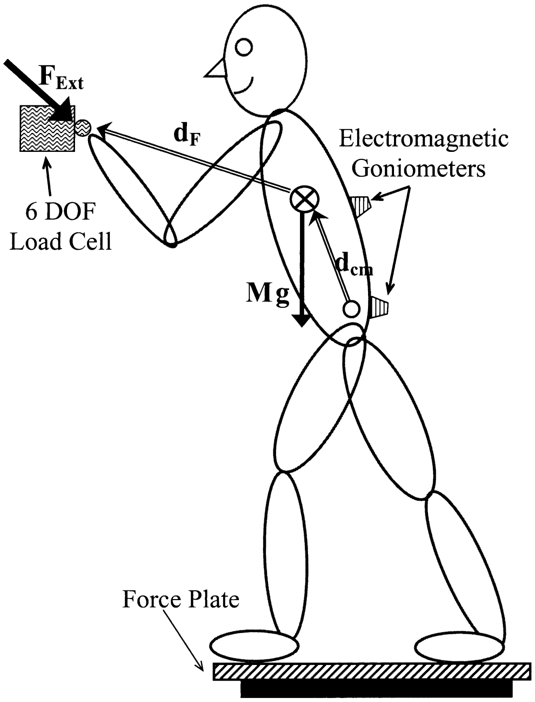


Figure 1. Schematic of isometric push experiment and stability model.

in which M_{LS} represents the internal trunk moment attributed to muscle activation about the lumbosacral junction. Because the system is at an equilibrium posture, $\theta_{cm} = \theta_0$, the term associated with trunk stiffness is zero and therefore is not included in Equation 2. Spinal load can be estimated from the vector sum of muscle-generated forces and external forces, F_{Ext}. Muscle-generated forces can be estimated from the equilibrium trunk moment, M_{LS}. This underestimates spinal load because muscle cocontraction and antagonistic recruitment patterns are ignored (Granata & Marras, 1995b; Hughes, Bean, & Chaffin, 1995). Future efforts must include muscle recruitment patterns in calculations of spinal load during pushing (Marras & Granata, 1997). To maintain static stability, the second derivative of potential energy must be greater than zero (Crisco & Panjabi, 1992; Granata & Orishimo, 2001). Hence trunk rotational stiffness must be greater than critical stiffness, k_{Cr}, which satisfies the stability equality

$$k > k_{Cr} = Mg\, d_{cm} \cos\theta_{cm} + F_{Ext}\, d_{cm} \cos(\theta_{cm} - \phi). \quad (3)$$

Bergmark represented musculoskeletal stiffness as a linear relationship with muscle force (Bergmark, 1989). It has been demonstrated that joint rotational stiffness is proportional to joint moment (Granata, Wilson, & Padua, 2002; Weiss, Hunter, & Kearney, 1988), with similar trends observed in the trunk (Gardner-Morse & Stokes, 2001; Cholewicki, Simons, & Radebold, 2000). Therefore, for small angle disturbances with respect to the equilibrium posture, θ_{cm}, trunk stiffness can be represented as

$$k = q\, M_{LS}, \quad (4)$$

in which q is the stiffness gradient. Some authors have reported the relation in Equation 4 scaled by muscle length (Cholewicki & McGill, 1996; Gardner-Morse et al., 1995), but this does not change the static behavior of our model. By combining Equations 2 through 4, the minimum stiffness gradient necessary for stability can be determined (i.e., a critical gradient q_{Cr}). Small values of q_{Cr} represent improved stability potential. Consequently, stability of equilibrium conditions will be defined as the inverse of q_{Cr},

$$q_{Cr}^{-1} = \frac{Mg\, d_{cm} \sin\theta_{cm} + F_{Ext} d_{cm} \sin(\theta_{cm} - \phi)}{Mg\, d_{cm} \cos\theta_{cm} + F_{Ext} d_{cm} \cos(\theta_{cm} - \phi)}. \quad (5)$$

Note that this metric describes the stability related to equilibrium levels of trunk moment without consideration of cocontraction. Additional stability can be achieved through neuromotor recruitment of muscle stiffness by means of cocontraction (Bergmark, 1989; Cholewicki et al., 1997; Gardner-Morse et al., 1995; Granata & Wilson, 2001). Hence, as the external stability, q_{Cr}^{-1}, decreases, it may be necessary to increase stability recruited through cocontraction. During pushing exertions gravitational and applied moments are often in opposite directions, so the numerator can approach zero, creating conditions of low external stability (Figure 2). Although published literature suggests that compressive load may be reduced during pushing exertions, the simple model described in Equation 5 indicates that the external stability during pushing exertions may be lower than that during equivalent lifting because trunk moment approaches zero. The current study was designed to quantify this external stability and the influence of pushing task design on these biomechanical factors.

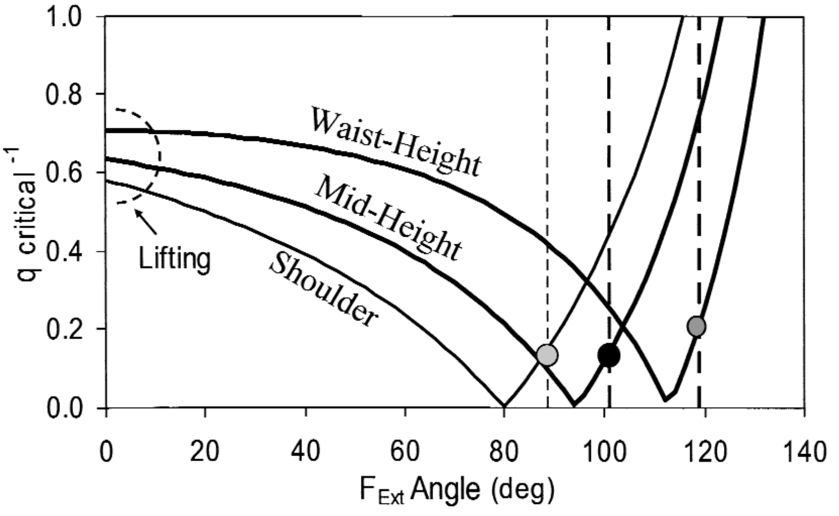


Figure 2. Critical stability, q_{Cr}^{-1}, estimated by the model for an applied total force 15% BW as a function of force direction. Model parameters were determined from empirically measured data during isometric pushing exertions. Force angle 0° is a lifting force and 90° is a horizontal push. The vertical (dashed) lines represent the average measured angles of applied forces from the experiment. Note the stability of the pushing solutions (circles) is much less than that of lifting.

Factors that influence external stability must be compensated for by cocontraction and associated changes in spinal load.

Experimental Protocol

Eleven healthy participants (7 men, 4 women) 20 to 23 years of age volunteered to participate in this study of pushing biomechanics. Mean (standard deviation) participant height and mass were 175.5 (10.6) cm and 72.3 (9.4) kg, respectively. Mean shoulder and anterior-superior iliac spine (ASIS) height were 145.8 (7.1) cm and 108.4 (7.1) cm, respectively. Average ASIS height was 74.3% of the shoulder height. All participants signed informed consent forms approved by the Human Investigations Committee of the University of Virginia.

Participants pushed on a stationary bar adjusted to three separate elevations: shoulder height, ASIS (waist) height, and at a height midway between the shoulder and waist height (midheight). The push apparatus consisted of a 3-cm diameter aluminum bar handle attached to a frame via a six-degree-of-freedom (6-DOF) load cell (Bertec, Columbus, OH). Participants stood on a force plate (Bertec, Columbus, OH) with equivalent surface area of 60 × 60 cm and a leading edge 42 cm horizontal distance from the bar (Figure 1). Participants wore their own athletic shoes; coefficient of friction between a typical athletic shoe and the force platform was approximately $\mu = 0.70$. They were asked to push for 10 s using a comfortable posture at three separate effort levels: maximum voluntary effort (MVE), 30% of their body weight (30% BW), and 15% of their body weight (15% BW). Participants controlled the push force by observing a display of the measured horizontal load applied to the bar handle. Two different stances were examined: feet parallel (equidistant from the plane of force application) and feet staggered (one foot forward of the other).

Before the trials at any one handle height, participants were given practice trials and asked to find the most comfortable foot position for the designated bar height and stance. After their preferred foot position was established, the locations of their feet were measured and marked on the platform so that participants could return to the same foot position should their feet move between trials. The participants were required to maintain the same foot position for all exertions for a given bar height and stance. Presentation order of bar height and stance were counterbalanced between subjects, whereas the order of effort was randomized within subjects.

Forces were recorded from both the 6-DOF load cell located in the push bar and a ground reaction force plate on which the participants stood. Trunk position was recorded from electromagnetic sensors (Ascension Technology, Burlington, VT) secured to the skin by double-sided tape over the spinous processes of S1 and T10. Prior to testing, the sensors were calibrated relative to the center of the force plate.

Analyses

Vertical and horizontal forces, torso angle, and moment on the trunk at L5/S1 were determined from the recorded data and foot position from static measurements. The kinetic and kinematic values used for analyses represented averages of a 5-s sample window with the smallest variability in horizontal force from the 10-s data collection. Trunk moments were computed by methods described by Granata, Marras, and Fathallah (1995) and included the measured forces and gravitational loads on the body segments determined using anthropometric estimates of segment masses and center of masses from Winter (1990) and Erdmann (1997). Separate repeated measures analyses of variance (ANOVAs) were performed with independent variables of handle height, exertion level, and stance to assess the applied horizontal force, force angle, torso angle, moment at L5/S1, and stability. Analyses were performed using commercial statistical software (Statistca 5.1, Statsoft, Tulsa, OK) with a significance level of $\alpha < .01$ for all tests.

RESULTS

Pushing kinematics were recorded during the isometric tasks and revealed trends with handle height and push-force exertion level. Participants selected their foot placement, and then the horizontal distance from the handle to the feet was recorded. However, distance was not significantly affected by handle height (Table 1). Recall that participants were required to maintain the same foot placement for all exertion levels at each handle height. Foot placement distance

TABLE 1: Repeated Measures ANOVA Summary

	Variable	*df*	*F*	*p*
	Kinematic Variables			
Distance (handle–foot)				
Parallel stance	Bar height	2	0.20	*ns*
Staggered stance (front/rear)	Bar height	2	0.66/0.02	*ns/ns*
Trunk angle (sagittal)	Stance	1	0.04	*ns*
	Bar height	2	24.08	<.001
	Effort	2	42.47	<.001
	Kinetic Variables			
Max. horizontal force	Stance	1	1.34	*ns*
	Bar height	2	12.02	<.001
Force vector angle	Stance	1	2.09	*ns*
	Bar height	2	210.28	<.001
	Effort	2	48.55	<.001
	Height × Effort	4	33.74	<.001
	Biomechanical Variables			
Trunk moment (L5/S1)	Stance	1	20.79	<.001
	Bar height	2	19.04	<.001
	Effort	2	30.09	<.001
	Stance × Effort	2	10.02	<.001
	Height × Effort	4	4.79	<.003
Stability	Stance	1	47.35	<.001
	Bar height	2	17.15	<.001
	Effort	2	7.94	<.003
	Height × Effort	4	8.37	<.001

Note. Only statistically significant interactions are listed. Foot placement (i.e., stance) was held constant across effort levels by experimental design. Results suggest trunk moment and stability are influenced by the design of the pushing task.

during parallel stance was almost exactly midway between the front and rear feet recorded during staggered stance trials. Torso angle relative to vertical reflects both geometric and force demands of the task. Participants always leaned forward, at least 15°, while pushing. For a prescribed level of effort, participants always leaned significantly more forward for the lowest (waist) handle elevation. Participants also leaned significantly more forward during the maximal effort trials than during submaximal exertions. Average torso angle was more than 35° from the vertical for the maximum effort trials at the waist handle elevation. Trunk flexion angle was not significantly different in parallel stance versus staggered stance.

Kinetics of the pushing task were recorded directly from the handle and were significantly influenced by task design parameters. Participants were able to match the prescribed horizontal force. For the 15% BW trials the average horizontal force was 14% of the participant's body weight, and for the 30% BW trials it was 27%. During MVE trials, mean horizontal force was approximately 50% of body weight and was significantly greater in the waist height conditions than when the bar was at shoulder level (Table 1). Mean horizontal force of 396 N during the waist-height trials was more than 33% greater than the maximum horizontal force achieved when pushing at shoulder height. In most trials participants applied a substantial vertical force to the handle in addition to the desired horizontal force. There was a consistent and significant trend of greater upward force for greater exertion and higher handle height (Figure 3). At maximal exertion participants averaged almost 250 N of upward force on the bar. Thus the reactive force on a participant's hands was downward and posterior.

Force direction was compared with trunk flexion angle to compute external force components along the axial direction (i.e., parallel to the trunk contributing to spinal compression) and tangential directions (i.e., contributing to anterior-posterior shear load). Results revealed that the axial or external compressive load could be larger than the applied horizontal force. This axial force averaged 112% of the horizontal force in MVE trials, 55% of the horizontal force in the 30% BW trials, and was almost absent for the 15% BW trials. Force direction was not significantly different in parallel stance versus staggered stance.

Biomechanical variables, including trunk moment and stability estimates (Equation 5), were computed from the measured data. Sagittal plane trunk moments computed about L5/S1 were typically small (i.e., mean external flexion moment of 72 Nm) but were significantly influenced by exertion level, handle height, and stance (Table 1 and Figure 4). There were also significant

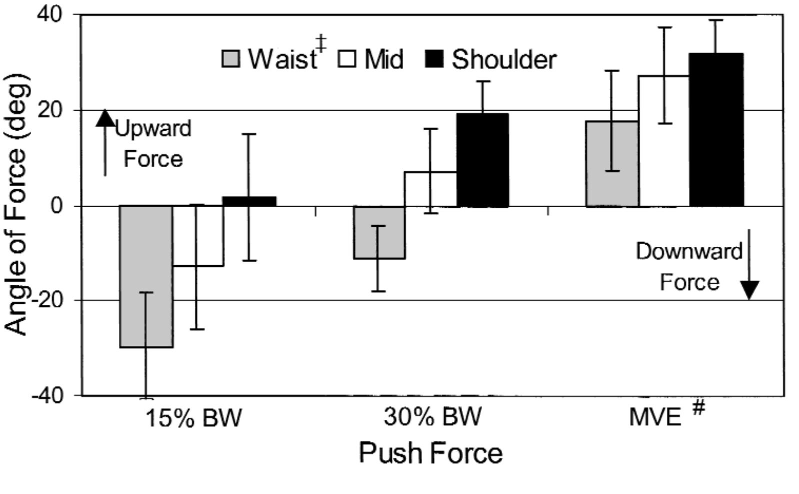


Figure 3. Angle of the applied force relative to horizontal was more upward for greater exertion and a higher bar. Positive angles indicate combined push and upward force, whereas negative angles indicate push and downward force. #Angle of push force was significantly greater at MVE exertion levels. ‡Angle of push force was significantly influenced by elevation of the push bar.

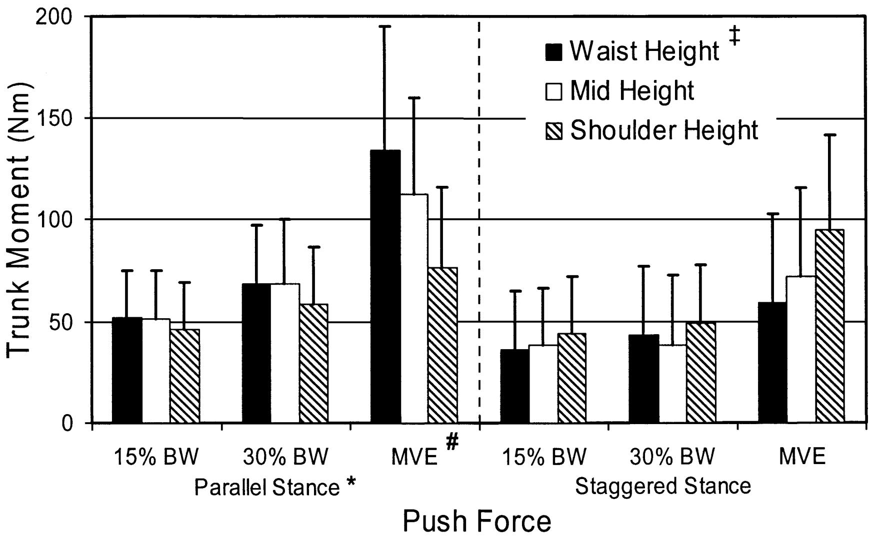
stance-by-exertion and height-by-exertion interactions. The parallel stance resulted in larger sagittal trunk moments. This was especially pronounced for the maximum effort trials, in which the moments with the parallel stance were 44% larger than the staggered stance values. The moments during waist elevation pushing exertions were significantly larger than the midheight or shoulder-height elevations for the MVE trials. This is particularly notable in the parallel stance condition. The opposite trend was noted for staggered stance MVE conditions, but this effect failed to reach statistical significance.

The inverse stiffness gradient reflects the stability of the trunk and was significantly influenced by bar height, effort, and stance (Table 1 and Figure 5). The parallel stance had higher values of q^{-1} (i.e., was more stable) than did the staggered stance, .26 versus .18. This reflects the smaller moments on the lumbar spine with the staggered stance but is limited to sagittal plane analyses. The ratios were also inversely proportional to the bar height. Pushing at shoulder height was the least stable, q_{Cr}^{-1} = .17, followed by midheight, q_{Cr}^{-1} = .21, and waist-height exertions, q_{Cr}^{-1} = .26.

To illustrate stability during the pushing tasks, average measured angles of the applied force at the three bar heights are indicated in Figure 2. Intersections with the modeled critical inverse stiffness values demonstrate that the postures and forces applied by the participants resulted in relatively low stability situations. Note also that the stability was much less than would be the case for lifting predicted by the model – that is, lifting represented by F_{Ext} angle equal to 0° (dotted hemicircle at the left of Figure 2).

DISCUSSION

As industrial manual materials handling progresses toward workplace designs that involve less lifting and more pushing and pulling, it is necessary to understand spinal biomechanics and musculoskeletal risk factors associated with these tasks. It has been estimated that 50% of industrial manual materials handling includes pushing and pulling tasks (Baril-Gingras & Lortie, 1995). These account for 20% of all industrial back injuries (Damkot et al., 1984; Hoozemans et al., 1998). Clearly the vertical load is reduced during pushing exertions when compared with lifting tasks. This may contribute to reduced trunk moment and spinal load (deLooze et al., 1995; Schibye et al., 2001). However, we



\# trunk moment was significantly greater at MVE exertion levels

‡ trunk moment was significantly influenced by elevation of the push-bar

* trunk moment was significantly influenced by foot placement, i.e. stance

Figure 4. Sagittal trunk moments were larger for the parallel stance, maximal effort, and waist-height pushing exertions.

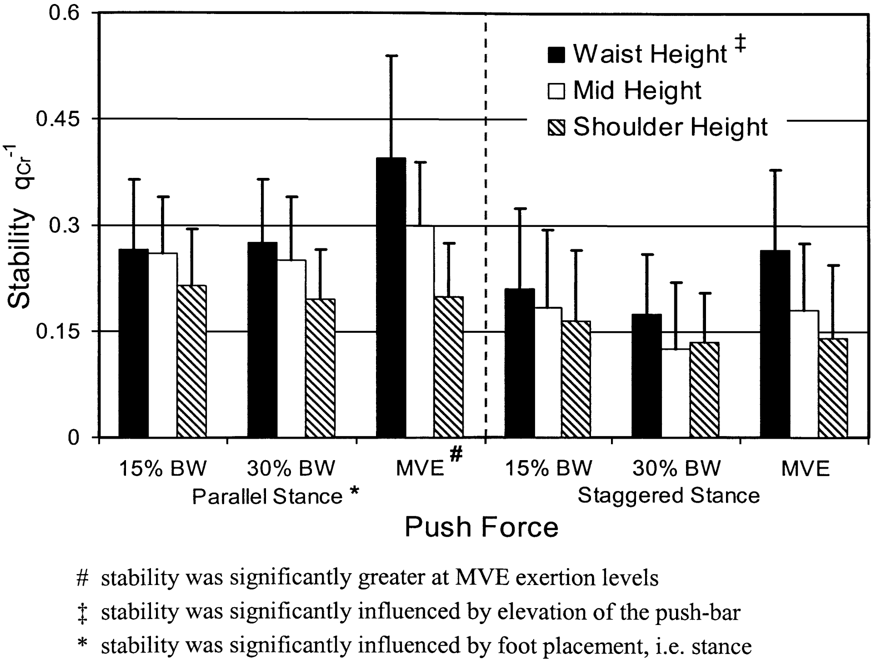


Figure 5. Stability (inverse of stiffness gradient) was significantly influenced by exertion level and push height for parallel and staggered stances.

are aware of no analyses that have attempted to quantify spinal stability and associated risk of musculoskeletal injury during pushing exertions. The goal of this study was to quantify the biomechanics of pushing exertions and provide estimates of spinal stability during these exertions.

Measured horizontal forces and kinematic results agree with levels and trends observed in controlled studies and industrial environments. Although results are limited to isometric exertions, the measured horizontal MVE forces (~300 N) were similar to values recorded during dynamic pushing tasks. Van der Beek et al. (2000) measured peak push forces of more than 450 N generated by experienced postal workers performing typical work tasks. Horizontal forces as high as 225 N were reported by Schibye et al. (2001) during the initiation of pushing (i.e., acceleration phase of pushing two-wheeled waste containers). Recording peak acceleration forces while participants operated an industrial hoist system, Woldstad and Chaffin (1994) found push forces exceeding 250 N. Thus industrial workplace tasks may require peak horizontal pushing forces that approach the maximal isometric capacity.

Maximal horizontal forces were significantly influenced by handle elevation. The largest horizontal MVE forces were generated at the lowest (waist) handle height, agreeing with the trends reported by Chaffin, Andres, and Garg (1983). However, their middle height of 109 cm was closer to our minimum height, which averaged 108.5 cm. At this bar elevation, our participants generated mean horizontal force of 392 N, 14% larger than the values reported by Chaffin et al. (1983). Kumar (1995) also found maximum strength near waist height (100 cm) with tests above and below this height. Ayoub (1974) reported maximum force generation for pushes at 70% of shoulder height, which was only slightly above our waist height.

Vertical force components were large for many trials. This force component is related in part to the participant's posture at each handle height and exertion level. Participants chose to modify trunk angle not by adjusting foot position but rather by modulating elbow flexion. Elbow flexion and associated trunk flexion increased with level of exertion. The net effect was to align the axis of the spine with the external force vector, thereby reducing external trunk moment and potentially reducing spinal compression and spinal stability.

There are three obvious reasons to apply an upward force component. First, it increases the normal ground reaction load and associated friction, thereby allowing the participant to minimize the potential for slipping (Chaffin, 1987). Because the horizontal push force may be limited by the floor friction, the mechanics of the pushing exertions, including factors of vertical force, trunk moment, and stability, may be influenced by the coefficient of friction. Second, the vertical force reduces the moment required at the low back and, potentially, reduces the required moment and strength in the shoulders. Third, if one wishes to apply leg extension strength to the push force, then clearly a vertical lift component is necessary and accompanied by an axial alignment.

Moreover, the posture creates an engineering truss structure wherein gravitational body weight is applied to the handle with considerable forward push force and posterior vertical ground reaction force but minimal muscular effort during low-exertion trials. However, if the handle were free to move in the horizontal plane, then dynamics would notably influence this truss effect and associated muscle recruitment (Nussbaum, Chaffin, & Baker, 1999). Therefore the static handle position is a limitation of the experimental protocol. As handle height and exertion level increased, the reaction force vector moved toward the lumbosacral junction, thereby creating large push force without large trunk moments. In similar results reported by deLooze et al. (2000), the reaction force vector rotated downward with increased handle height and level of exertion when participants pushed on a stationary bar while walking on a treadmill. Although this moment-reducing strategy creates a more efficient force transfer from the legs to the handle, it may also challenge spinal stability as suggested by the low moment versus force ratios.

External trunk moment was generally in the extension direction, requiring abdominal muscles to generate the appropriate internal flexion moments. Lifting exertions are always associated with paraspinal activity to generate internal trunk extension moment. Thus sagittal plane pushing and lifting exertions are unique from the perspective of muscle recruitment (i.e., which muscles perform the role of prime movers). Chaffin and Park (1973) reported that risk of occupational low-back injury is associated with force requirements of a job relative to the individual's strength. Because trunk flexion strength is typically weaker than extensor strength (Hamill & Knutzen, 1995; Parnianpour, Campello, & Sheikhzadeh, 1991), the lift-strength ratio may be adversely affected in pushing. Trunk moments were strongly affected by handle height for the MVE trials only. These MVE pushing exertions, with their large vertical and axial forces and flexed torso postures, appear to represent a regime of force solutions different from submaximal exertions. For the 30% and 15% BW trials, the low-back moments were small and unaffected by the height of the handle. The magnitude of the moments for the 30% and 15% BW trials (<68 Nm) were similar to the values of deLooze et al. (2000) reported at similar levels of force and handle heights.

Estimates of spinal load during pushing suggest lower compressive loads than those during equivalent lifting tasks. However, risk of spinal compression overload during pushing tasks may be secondary to shear load damage. Only two studies were found that reported spinal shear forces (deLooze et al., 1995; Schibye et al., 2001), but they failed to achieve consensus. Estimated values ranged from 100 to 1400 N, whereas published estimates of compressive load range from 500 N (Schibye et al., 2001) to 5500 N (Kumar, 1994; Lavender, Conrad, Reichelt, Johnson, & Meyer, 2000), with the large difference resulting from exertion level and experimental design (deLooze et al., 1995; Lee, Chaffin, Herrin, & Waikar, 1991; Resnick & Chaffin, 1995).

Assuming muscle moment arms of 5 to 10 cm and mean measured moments from the current data of 36 to 135 Nm, one may be tempted to estimate compressive loads of 360 to 2700 N plus external axial force and trunk weight. Published compression loads during industrial lifting tasks range from 3400 to 13,000 N (Karwowski, Caldwell, & Gaddie, 1994; Kumar, 1996; Norman et al., 1998). However, the estimates of spinal load during pushing ignore the influence of trunk muscle cocontraction, which can dramatically increase the spinal load (Granata & Marras, 1995b; Hughes et al., 1995). Published models (Gardner-Morse & Stokes, 1998) and empirical data (Cholewicki et al., 1997; Granata & Orishimo, 2001) suggest that antagonistic cocontraction is recruited to augment spinal stability. Recognizing that the model described in Equation 5 indicates reduced spinal stability during pushing exertions, one must expect the neurocontroller to recruit high levels of trunk muscle cocontraction to restore stability.

In post hoc pilot measurements, we recorded electromyographic (EMG) activity from 2 participants during pushing exertions and observed extraordinarily high levels of trunk muscle cocontraction – that is, extensor versus flexor EMG activity ranged from 30% to 200%. Biomechanical and muscle recruitment predictions are analogous to observed increases in cocontraction of up to 400% while using lift-assisted devices versus manual lifting exertions (Chaffin, Stump, Nussbaum, & Baker, 1999; Nussbaum et al., 1999). Nussbaum et al. (1999) concluded that "while alleviating the spinal loads, [these devices may have] increased the need for stability or control" (p. 1615). Ongoing analyses will quantify this effect more thoroughly. We agree with Lee et al. (1989), who concluded that "a simple biomechanical model with only one muscle active at a time may not be appropriate for the estimation of muscle forces of the low back" (p. 1562), and Chaffin et al. (1999) had similar observations. EMG-assisted models have been developed to include the influence of trunk muscle recruitment when computing spinal compression (Granata & Marras, 1995a; McGill & Norman, 1986). Unfortunately, these techniques have never been calibrated or validated for trunk flexion exertions such as in pushing. Nevertheless, one can safely assume that spinal load was influenced by handle height and push force as per trends in trunk moment. Future research must investigate the spinal load during pushing tasks while accounting for the notable contribution of trunk muscle cocontraction.

The purpose of recording trunk moments and postures during the pushing exertions was to investigate spinal stability. Estimates of spinal stability suggest that the low moment and potentially high axial load combinations may present a risk of musculoskeletal instability. The model (Equation 5, Figure 2) indicates stability of the equilibrium system may decline to unsafe levels unless the neurocontroller augments trunk stiffness by means of cocontraction. In a survey of 403 industrial jobs, Marras et al. (1993, 1995) reported average workplace lifting parameters, including mean lifted load, lifting horizontal moment arm, and trunk flexion angles recorded in situ. Data reported in that study were entered into Equation 5 for comparison with the push stability, resulting in a mean stability estimate exceeding .50 to represent industrial lifting. Results from our pushing measurements suggest that stability was influenced by handle height and exertion level, ranging from .13 to .40. Most of these values for pushing exertions were much less stable than values estimated from industrial lifting tasks. Model results (Figure 6) demonstrate that external stability must change with the angle of the handle reaction force and trunk angle. Thus, to maintain spinal stability, increased trunk stiffness was clearly required, potentially explaining the observed levels of muscle cocontraction in the 2 pilot participants and as reported by Lee et al. (1989). When recruiting cocontraction, the risk of instability must be balanced against the risk of overload injury.

When considering the results, it is necessary to consider limitations of the analysis of stability. The biomechanical model of stability is a static representation of neuromuscular control and ignores the dynamic feedback associated with reflex and voluntary response dynamics. Reflex is known to contribute to effective muscle stiffness and may therefore contribute to stability by means of active feedback control (Kearney & Stein, 1997; Nichols & Houk, 1976). Empirical measures of neuromuscular response dynamics reveal an abnormal response in patients with low-back pain, although it is unclear whether this behavior was a contributing cause to low-back pain or a subsequent effect of discomfort guarding (Radebold, Cholewicki, Panjabi, & Patel, 2000; Radebold, Cholewicki, Polzhofer, & Green, 2001). However, the time delay associated with paraspinal response ranges from 30 to 200 ms, so the role of the response dynamics for control of spinal stability is unclear (Granata, Slota, Bennett, & Kang, 2004; Lavender et al., 1989). Future efforts should investigate the role of neuromuscular response dynamics.

The model assumed a linear relation between

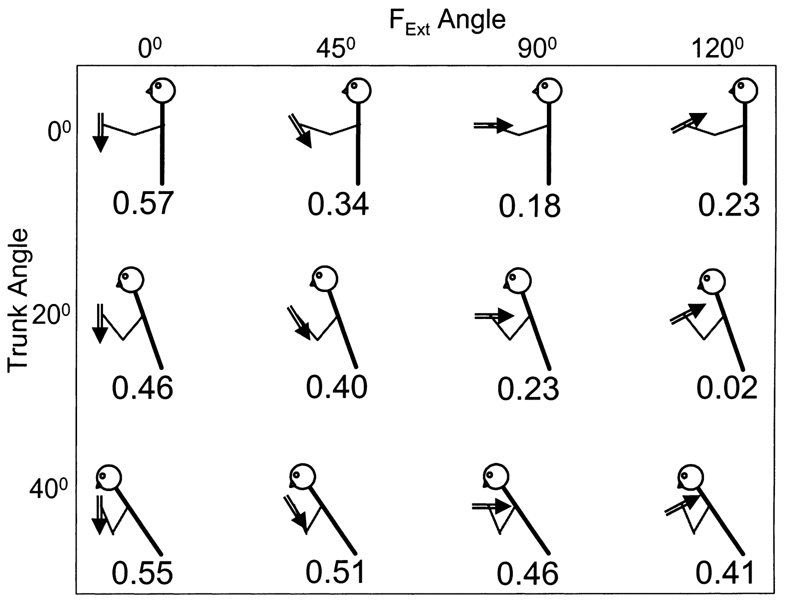


Figure 6. Values under each stick figure represent modeled stability, q_{Cr}^{-1}, for the posture, θ_{cm}, and angle of external force, ϕ, illustrated. For demonstrative purposes, these model results are based on shoulder height exertions with simplifying assumptions wherein the distance from L5-S1 to the handle is fixed. Trends demonstrate reduced external stability as the angle of external force approaches horizontal, thereby suggesting the need to increase recruitment of cocontraction.

trunk stiffness and external trunk moment (Equation 4). Although this linear behavior is well accepted for other joints (Hunter & Kearney, 1982; Wilson, Wood, & Elliott, 1991), it remains to be validated for the trunk. However, the model was limited to small angle perturbations, and it is therefore a reasonable approximation to active stiffness. Although the model accounts for changes in external load related to posture, it has also been assumed that stiffness is not largely influenced by trunk posture. Several models of spinal stability assume muscle stiffness is scaled by its equilibrium length (Cholewicki & McGill, 1996; Gardner-Morse et al., 1995); others treat stiffness as independent of length (Cholewicki et al., 1997; Granata & Wilson, 2001). Although several physiologic measurements of stiffness from muscle activation have shown effects of muscle length (Tai & Tobinson, 1999; Zhang, Nuber, Butler, Bowen, & Rymer, 1998), some researchers have concluded that length dependence is related primarily to passive characteristics at end range of motion (Gottlieb & Agarwal, 1978; Kearney & Hunter, 1990). Further effort is necessary to quantify this effect in the trunk. Although this assumption will not influence the effect of external force direction on external stability, it may influence the stability levels reported for the separate handle height conditions.

Finally, it must be understood that the estimated external stability represents the contribution of equilibrium forces to stability and neglects effects of cocontraction. This was done to demonstrate the relative role of external factors in push versus lifting exertions.

In summary, during horizontal pushing exertions ancillary vertical loads are also applied to the handle, probably for the purpose of improving performance and reducing the risk of slipping. These contribute to low trunk moments during the pushing exertions, which may potentially reduce spinal compression levels, as compared with industrial lifting exertions. However, the spinal load must be related to load tolerance of the spine. Stability represents a controllable tolerance estimate of structural integrity of the spinal column. Results indicate that extraordinary levels of trunk muscle cocontraction should be recruited to compensate for the fact that equilibrium mechanics provide little intrinsic trunk stiffness and stability during pushing exertions.

Thus it is particularly important to consider muscle cocontraction when evaluating the biomechanics of pushing exertion. If one maintains stability by means of cocontraction, then additional spinal load is created, increasing the risk of overload injury. As manual materials handling in industry is rapidly redesigned to replace lifting tasks with pushing and pulling exertions, it is necessary to proactively identify risk factors for low-back pain and design parameters that will aid in the control of musculoskeletal injury.

APPENDIX

Stability analyses were developed from the potential energy of the biomechanical system represented in Figure 1. Position of the CM relative to L5-S1, $\vec{x}_{cm}$, can be expressed as the product of a length vector, $\vec{d}_{cm}$, and a rotation matrix, R_{cm}. Similar expressions represent the position of the point of applied force at the push handle, $\vec{x}_{F_{Ext}}$,

$$\vec{x}_{cm} = R_{cm}\vec{d}_{cm} \tag{A1a}$$

$$\vec{x}_{F_{Ext}} = R_{cm}\vec{d}_{cm} + R_F\vec{d}_F, \tag{A1b}$$

in which the two-dimensional Euler rotation matrix is simply

$$R_i = \begin{bmatrix} \cos\theta_i & -\sin_i \\ \sin\theta_i & \cos_i \end{bmatrix}. \tag{A2}$$

Potential energy of the system with respect to the base of the spine is described as the scalar product of the external loads, $\vec{F}_{Ext}$, $M\vec{g}$ with vector distances to the base of the spine plus and strain energy stored in the muscle tissues. The sum of strain energies in multiple muscles can be estimated in terms of a rotational stiffness, k, and the trunk angle,

$$V = M\vec{g} \cdot \vec{x}_{cm} + \vec{F}_{Ext} \cdot \vec{x}_{F_{Ext}} + ½k(\theta_{cm} - \theta_0)^2, \tag{A3}$$

in which $\vec{g}$ is the two-dimensional gravitational vector with magnitude g and direction $[0,-1]^t$, M is the trunk mass and external force at the push handle, and $\vec{F}_{Ext}$ has magnitude $|F_{Ext}|$ with an arbitrary unit-vector direction $\vec{F}_{Ext} = R_\phi[0,-1]^t$ at an angle ϕ with respect to vertical. By specifying the force angle ϕ, one can readily estimate the effects of push versus lifting forces at the hands. Expanding the equation of potential energy into its trigonometric components results in

$$V = Mg\, d_{cm} \cos\theta_{cm} + F_{Ext}\{d_{cm}\cos(\theta_{cm} - \phi) + d_F\cos(\theta_F - \phi)\} + ½k(\theta_{cm} - \theta_0)^2. \tag{A4}$$

The first derivative with respect to angle represents equilibrium,

$$\begin{pmatrix} \dfrac{\delta}{\delta\theta_{cm}}V \\ \dfrac{\delta}{\delta\theta_F}V \end{pmatrix} = \begin{pmatrix} -Mg\, d_{cm}\sin\theta_{cm} - F_{Ext}d_{cm}\sin(\theta_{cm} - \phi) + k(\theta_{cm} - \phi_0) \\ -F_{Ext}d_F\sin(\theta_F - \phi) \end{pmatrix} \tag{A5}$$

$$= \begin{pmatrix} -M_{LS} + M_{cm} \\ -M_{cm} \end{pmatrix}.$$

External moments about L5-S1 and about the CM are supported by the muscle-generated internal moments, M_{LS} and M_{cm}, respectively. Simultaneous solution of these equations allows one to estimate the static moment, M_{LS} (Equation 2 in the Methods section). It is assumed that the system is at static equilibrium such that $\theta_{cm} = \theta_0$. The second derivative (Hessian matrix) must be greater than zero (positive definite matrix) to ensure static stability.

$$\begin{pmatrix} \dfrac{\delta}{\delta\theta_{cm}}\dfrac{\delta}{\delta\theta_{cm}}V & \dfrac{\delta}{\delta\theta_F}\dfrac{\delta}{\delta\theta_{cm}}V \\ \dfrac{\delta}{\delta\theta_F}\dfrac{\delta}{\delta\theta_{cm}}V & \dfrac{\delta}{\delta\theta_F}\dfrac{\delta}{\delta\theta_F}V \end{pmatrix} = \begin{pmatrix} Mg\, d_{cm}\cos\theta_{cm} - F_{Ext}d_{cm}\cos(\theta_{cm} - \phi) + k & 0 \\ 0 & -F_{Ext}d_F\cos(\theta_F - \phi) \end{pmatrix} \tag{A6}$$

The eigenvalues are on the diagonal of the matrix, and each must be greater than zero. The second eigenvalue (lower diagonal value) represents destabilizing effects of external loads about the CM. It is interesting to note that control of the moment about the CM will influence stability of the spine. Nonetheless, we assumed there is sufficient control at the CM and focus on the upper-diagonal eigenvalue representing the L5-S1 spine.

To be stable, the trunk rotational stiffness, k, must satisfy the stability constraint

$$k > Mg\, d_{cm} \cos\theta_{cm} + F_{Ext}\, d_{cm} \cos(\theta_{cm} - \phi). \quad (A7)$$

If a linear relationship is assumed between trunk stiffness and equilibrium moment, M_{LS} (Equation 4 in the Methods section) for small angular disturbances, $\delta\theta_{cm}$, about equilibrium, θ_{cm}, then the stiffness term can be replaced by M_{LS} multiplied by the stiffness gradient q.

$$\begin{aligned} &q\{Mg\, d_{cm} \sin\theta_{cm} + F_{Ext}\, d_{cm} \sin(\theta_{cm} - \phi)\} \\ &> Mg\, d_{cm} \cos\theta_{cm} + F_{Ext}\, d_{cm} \cos(\theta_{cm} - \phi) \end{aligned} \quad (A8)$$

The minimum value of q that satisfies this relation, q_{Cr}, represents the stiffness gradient necessary to achieve stability at equilibrium conditions. Small values of q_{Cr} represent improved stability from equilibrium conditions. Larger values of q_{Cr} represent less stable equilibrium conditions and require increased muscle cocontraction to maintain stability of the spine. We operationally define stability in terms of q_{Cr}^{-1} (i.e., larger values of q_{Cr}^{-1} represent improved stability). Using empirically measured data to estimate values of M, F_{Ext}, d_{cm}, d_F, θ_{cm}, and θ_F, one can readily determine the value of q_{Cr}^{-1} for a range of external force directions, ϕ (Figure 2).

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