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Postural control during lifting

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Abstract

Any voluntary motion of the body causes an internal perturbation of balance. Load transfer during manual material handling may increase these perturbations. This study investigates effects of stance condition on postural control during lifting. Nineteen healthy subjects repeatedly lifted and lowered a load between a desk and a shelf. The base of support was varied between parallel and step stance. Ground reaction force and segmental kinematics were measured. Load transfer during lifting perturbed balance. In parallel stance postural response consisted of axial movements in the sagittal plane. Such strategy was accompanied by increased posterior shear forces after lift-off. Lifting in step stance provided extended support in anterior/posterior direction. The postural control mechanisms in the sagittal plane are less complex as compared to parallel stance. However, lifting in step stance was asymmetrical and thus accompanied by distinct lateral transfer of the body. Lateral shear forces were larger as compared to parallel stance. Both lifting techniques exhibit positive and negative aspects. We cannot recommend either one as being better in terms of postural control.

Keywords: Perturbation; Balance; Manual material handling; Center of mass (COM); Center of pressure (COP)

1. Introduction

Ergonomic lifting advice for safe material handling of loads commonly includes the use of a wide stance and performance of the lift with bent knees and a straight back to decrease loads on the spine. Such guidelines are based on knowledge from biomechanical studies that have investigated mechanical aspects of spinal loading (Chaffin, 1987; McGill et al., 1996; Fathallah et al., 1998; Chaffin et al., 1999) and focused to reduce the risk of low back injury within the lumbar disks. Surprisingly, only few studies have investigated postural control strategies during lifting or performing a material handling task (Toussaint et al., 1997; Toussaint et al., 1998; Oddsson et al., 1999). Specific lifting guidelines are lacking, that take into consideration the postural control mechanisms necessary for safe material handling.

Due to the multi-link structure of the various segments of the human body, any voluntary movement will impose a perturbation of equilibrium. These perturbations increase when the movement is performed with an added load such as lifting an object. For example, front-loading of the body will cause a shift of the system center of mass (COM) forward and anticipatory postural adjustments will be applied that counteract the upcoming perturbation (Brown and Frank, 1987). These involuntary "automatic" movements are smoothly incorporated into our movement repertoire to ensure accurate and harmonious motion (Massion and Gahery, 1979; Oddsson, 1990; Timmann et al., 1994; Ioffe et al., 1996). These postural synergies are triggered prior to the onset of voluntary movements and appear to be flexible and task specific (Dietz et al., 2000; Bouisset and Zattara, 1987; Diener et al., 1983). Thereby, a certain voluntary movement may be associated with different automatic postural adjustments depending on the context of the task (Crenna et al., 1987; Pedotti et al., 1989; Shiratori and Latash, 2000; Cordo and Nashner, 1982; Hodges et al., 2000). When voluntary movements are performed under unstable situations these postural adjustments become more complex, and involve more muscles, higher activation levels (Oddsson, 1989, 1990; Aruin et al., 1998), and/or different activation strategies such as

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co-contraction (Humphrey and Reed, 1983; Oddsson, 1990; Nielsen and Kagamihara, 1992; Accornero et al., 1997; Cresswell et al., 1994).

This study investigates effects of stance condition on postural adjustments during lifting. We hypothesized that a larger base of support in step stance would reduce complexity of postural control. In addition we developed a model with postural adjustments absent. Comparison between the passive reaction forces of the model and data recorded from subjects might provide better insight into the postural control strategies.

2. Methods

2.1. Subjects

Nineteen healthy subjects (5 female/14 male) participated in the study. Their age ranged from 20 to 45 years (mean 27a/S.D. 6a); body weight from 51 to 93 kg (mean 71 kg/S.D. 11 kg); and body height from 164 to 191 cm (mean 179 cm/S.D. 8 cm). Before the experiment, all subjects gave written informed consent, which was authorized by the Institutional Review Board.

2.2. Lifting task

Subjects stood comfortably on a force plate in front of a frame with two shelves which were positioned within reaching limit. The lower shelf was 70 cm (desk height) above ground and 40 cm anterior to the subjects' ankles. The upper one was 140 cm (shelf height) above ground and 70 cm anterior to their ankles, respectively.

Subjects lifted and lowered a box 10 times during 1 min. The lifting cycles were of equal periods, paced by a metronome. The pacing indicated to the subject when to lift the load from the shelf or table, respectively. The box was a cube of 25 cm in dimension with two handles placed symmetrically 15 cm above the bottom. The box weighted 4.6 kg, which corresponds to a lifting index of 1 according to 1994 NIOSH-Recommended Weight Limits (Waters et al., 1994; Chaffin et al., 1999) and is suggested to ensure a low risk lifting (Waters et al., 1993).

The lifting tasks were performed with two different stance conditions. In one task subjects stood comfortably with their feet in parallel stance. In the other task they kept their feet in a step stance position with the left foot being placed 20 cm posterior to the right one. In both tasks the feet were apart in pelvis width and the sagittal foot axis was externally rotated between 0 and 10° (Fig. 1a and b). Before the experiment subjects practiced the rhythmic of lifting.

2.3. Measurements

We recorded motion by a position and orientation measurement system, based on switched magnetic fields (Motion Star with extended range transmitter, Ascension Technology Corporation, Burlington, Vermont USA). Accuracy testing with a grid of predefined positions revealed short range accuracy of 4 mm (motion <10 cm), in agreement to technical reports (Handbook of Ascension Tech.). Wide range accuracy testing exhibited larger distortions of 2 cm in our setup (motion >10 cm). Position data was sampled at a frequency of 86 Hz and low pass filtered (Butterworth, 2nd order, $f_{cutoff} = 12$ Hz).

A total of 8 receivers were attached to the trunk, pelvis, legs and arms using double sided tape interfaces and hard foam pads. The trunk marker was placed over the processus spinosus of the 7th cervical vertebrae. The pelvis marker was placed at the base of the sacrum. Two receivers were placed at the upper arms, midway between acromio-clavicular joint and lateral humeral epicondyle. Another two were placed at the lower arms in the proximal third between the processus olecrani and the processus styloideus of the ulna. And two receivers were attached to the thighs, midway between the greater trochanter and the lateral femoral condyle. All cables of the receivers were bundled together on the backs of the subjects. Recordings during quiet upright stance served as reference for the calculation of the relative displacement of positional data.

A force plate (AMTI, Newton, MA, USA) recorded three dimensional ground reaction forces (GRF) and moments. 3D-GRF was sampled at a rate of 100 Hz and low pass filtered (Butterworth, 4th order, $f_{cutoff} = 15$ Hz). COP excursion was calculated in the anterior/posterior and lateral direction from force and moment data.

A contact switch at the bottom of the box was used to identify four different phases during the lifting cycle: (1) transition-up, (2) rest-on-shelf, (3) transition-down, (4) rest-on-desk. Each phase was paced for a duration of 1.5 s. The signals recorded during the intervals of contact between box and desk or shelf were sampled separately by all measurement systems and used as trigger for synchronization of kinetic and kinematic data.

2.4. Model

The biomechanical model consisted of 8 segments (Fig. 1a and b). The segmental parameters of legs, trunk, upper arms, lower arms and load were calculated by body mass proportions (Winter, 1990). Each segment was calculated as collapsed mass at the approximated segmental COM according to geometry and body density distribution. The total body COM was calculated as the weighted sum of the segmental COMs.

The model was used in two different ways. It was either driven by the measured positional data of the receivers to calculated COM displacement for trials of subjects or it was used to calculate COP and COM for simulated lifting in both stance conditions. The simulation task was performed with isolated arm motion, without postural responses of trunk and pelvis segments. In the simulation the motion of the box was simplified to a combined vertical and horizontal displacement following the sinusoidal pattern (Figs. 1 and 2a):

$d = t - \sin(t)$ (Figs. 1 and 2a).

The duration of vertical and horizontal displacement was set to 1 s each with 0.5 s time lag. The simulated COM displacement was calculated with influence of horizontal transfer of load and arms for each subject (Fig. 2d, stat.). In addition load release profiles during rest-on-desk and rest-on-shelf were approximated to a sinusoidal pattern as revealed by vertical components of GRF (Fig. 5a and b; Fig. 2d, rel.). This simulated COM displacements were used to approximate vertical and horizontal GRF in a direct forward solution (Winter, 1990). The simulated COP displacement included dynamic components of vertical and horizontal acceleration of load and arm segments (Fig. 2c and d, d.hor.–d.ver.). As there was no motion of trunk, pelvis or legs, the simulated COM and COP displacements are identical in both stance conditions under assumption that simulated COP displacement does not travel outside a range of ± 10 cm in all directions. The rotational dynamics, such as angular accelerations, are not included in the model.

2.5. Data processing

We analyzed the data of the 4–8th lifting cycles. Within these cycles the pace has stabilized and motion was rhythmic. Kinetic and kinematic raw data were 100-point time normalized for each transition and rest phase, separately. Then the 400-point data was ensemble averaged over all subjects. The variability was assessed by calculation of the coefficient of variation (cv) for time series data (Winter, 1984).

Overall COP displacement in the anterior and lateral direction was calculated for the two transition phases and stance conditions, including simulation. Furthermore, within the initial 300 ms of the transition phases the root mean square (RMS) of horizontal GRF components in the posterior and lateral direction were assessed. The correlation between trunk and pelvis position was used to quantify the activation between the two body segments activities. Comparison of step versus parallel stance condition were performed by two sided t-tests (significant level $p < 0.05$).

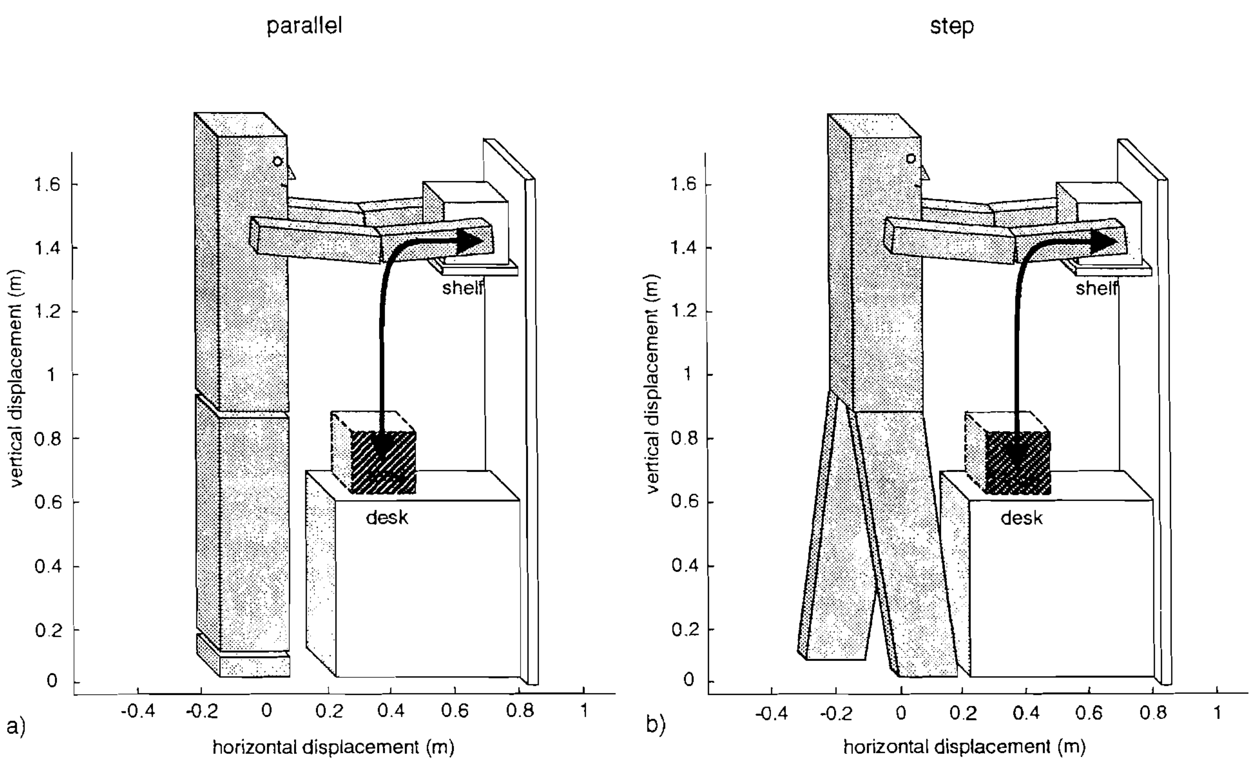

Fig. 1. Biomechanical models with 8 segments for simulation of a lifting cycle (a) in parallel stance and (b) in step stance. Load trajectory is approximated in combined vertical and anterior pattern following a 't-sin(t)' pattern (black arrows).

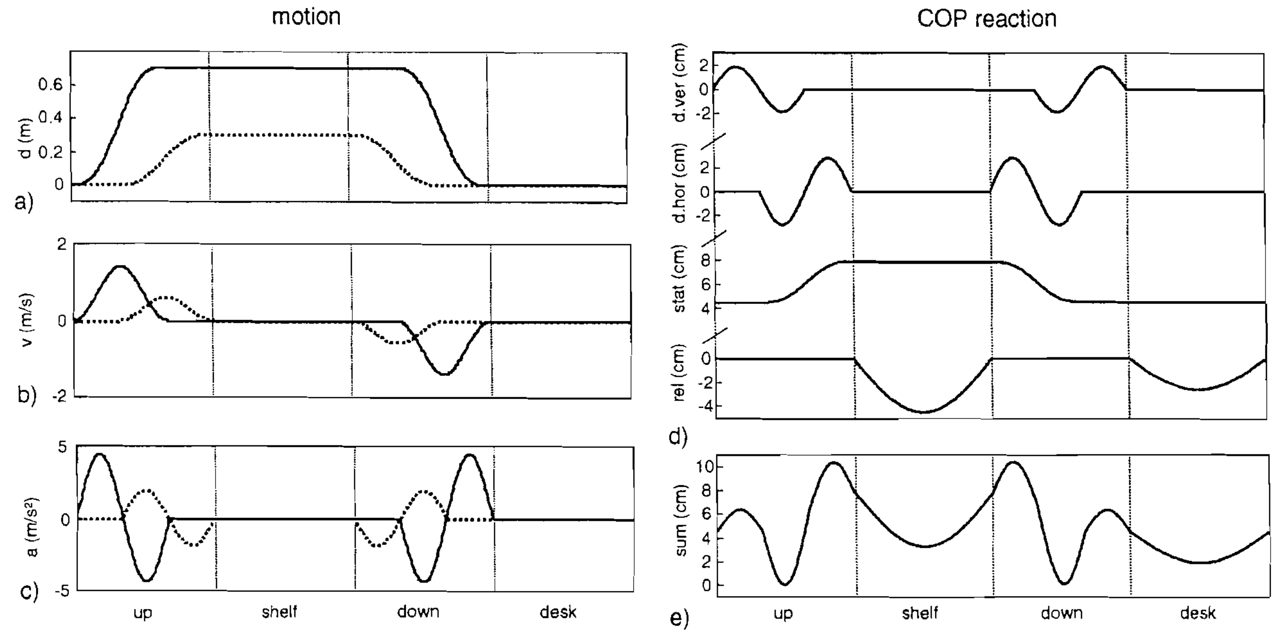

Fig. 2. Patterns of simulated load lifting: components of horizontal (solid) and vertical (dotted) (a) displacement, (b) velocity and (c) acceleration during a lifting cycle; (d) components of calculated displacement of Center of Pressure related to vertical acceleration (*d*.ver), horizontal acceleration (*d*.hor), quasistatic horizontal displacement (stat), and release of load during rest phases (rel); sum of all components (e) exhibits a four-phasic pattern.

3. Results

Mean duration of the lifting cycles were equivalent to the lifting pace of 6 s (Table 1). However, the four phases within each cycle differed significantly from each other. The transition phases where significantly longer than the rest phases. The sum of each transition phase and the following rest phase was equivalent to the pacing duration of 3 s.

3.1. Kinematic parameters

Fig. 3 shows the displacement pattern vs. time of both the pelvis and trunk in the sagittal and frontal planes (right = positive). Inspection of the graphs revealed that subjects used different strategies for the performance of the lifting tasks that were dependent on the base of support. In simulation, no motion of trunk or pelvis was present in both stance conditions (Fig. 3a and d).

Intra- and inter-subject variability of the kinematic parameters were generally low ($cv_{intra} < 0.14$; $cv_{inter} < 0.40$), except for lateral kinematics, when subjects performed the lifts in parallel stance ($cv_{intra,inter} > 1$).

The sagittal trunk displacement during the lifting task was highly congruent for the two stance conditions (Fig. 3b and c) mainly reflecting the primary aspect of the task, i.e. moving the box between the desk and shelf. Range of motion was comparable in both stance situations during transition phases (Table 2). In the frontal plane, when subjects were in parallel stance, minimal lateral movements of the trunk occurred during lifting and lowering the load (Fig. 3f). In step stance, however, a distinct four-phasic pattern of the trunk motion was observed (Fig. 3e).

Both resting phases in both stance conditions showed a tri-phasic trunk displacement pattern in the sagittal plane (Fig. 3b and c). In the frontal plane almost no lateral displacement of the trunk occurred in parallel stance (Fig. 3f). In step stance, however, a distinct bi-phasic pattern occurred (Fig. 3e).

Range of oscillatory pelvic motion in anterior/posterior direction was significantly larger in parallel stance as compared to step stance in both transition phases (Fig. 3b and c; Table 2). In the frontal plane lateral motion of the pelvis was, absent in the parallel stance condition. In step stance, however, a distinct pattern of the pelvis motion, phase-locked to trunk motion occurred (Fig. 3e).

A bi-phasic displacement pattern in the sagittal plane was observed during the resting phases. In the frontal plane no lateral displacement of the pelvis occurred in parallel stance. In step stance a bi-phasic pattern was observed (Fig. 3e and f).

3.2. Kinetic parameters

COP and COM patterns of the lifting tasks are given in Fig. 4. A simulation of the lifting and lowering task revealed a specific three-phasic pattern of COP

Table 1
Mean duration (S.D.) of the different phases within each lifting cycle given for the parallel and step stance condition separately

	Transition-up (s)	Rest-on-shelf (s)	Transition-down (s)	Rest-on-desk (s)
Parallel stance	1.58 (0.15)	1.42 (0.16)*	1.64 (0.19)	1.36 (0.18)*
Step stance	1.60 (0.18)	1.41 (0.19)*	1.59 (0.14)	1.41 (0.17)*
Transition + rest	3.01 (0.07)		3.00 (0.12)	
Cycle duration	6.01 (0.08)			

*Statistically significant difference between duration of transition phase and following rest phase ($p<0.05$).

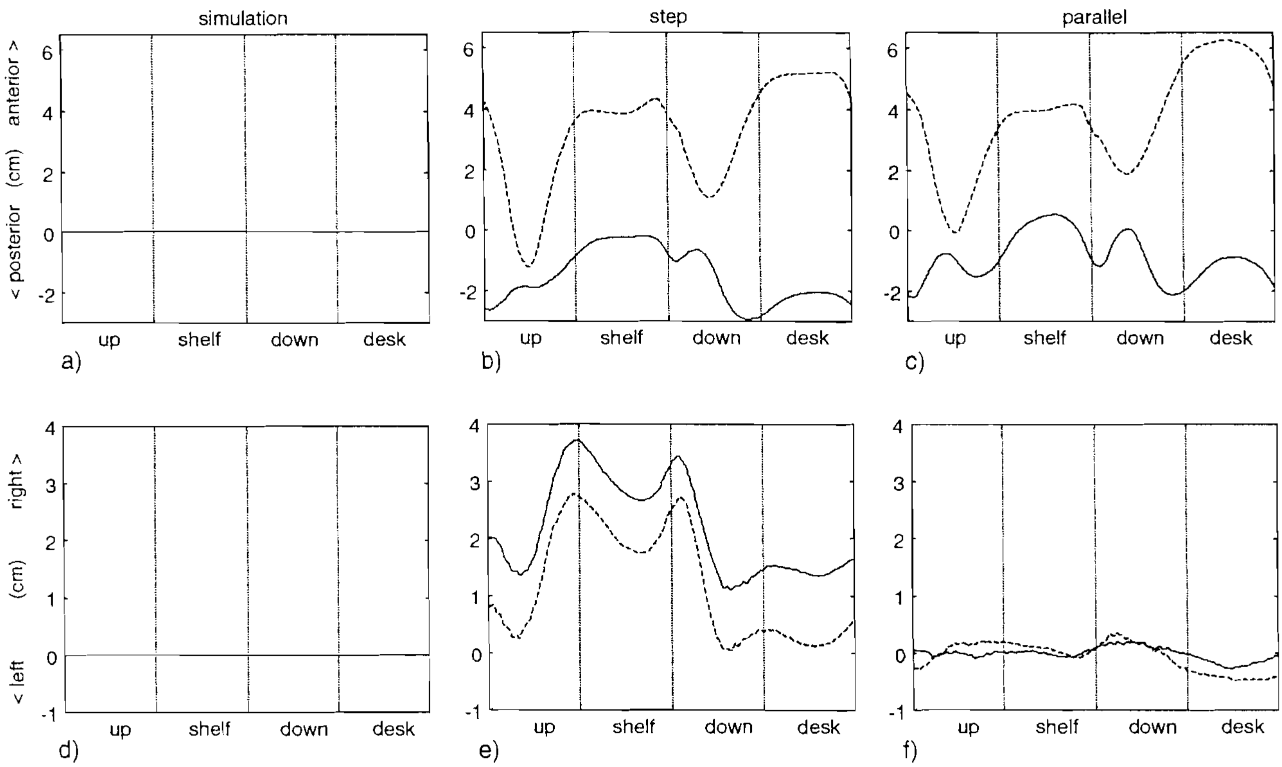

Fig. 3. Mean displacement of pelvis (solid) and trunk (dashed) in the anterior/posterior and lateral direction: (a, d) simulation of lifting and lifting in (b, e) step stance and (c, f) parallel stance condition.

Table 2
Amount of oscillatory motion during transition phases for trunk and pelvis

	Trunk		Pelvis	
Phase	Transition-up (cm)	Transition-down (cm)	Transition-up (cm)	Transition-down (cm)
Parallel stance	3.6 (3.0)	2.1 (3.5)	0.8 (0.8)	1.2 (1.1)
Step stance	5.0 (3.7)	3.2 (4.0)	0.1 (0.5)	0.3 (0.9)
Comparison	n.s.	n.s.	$p=0.002$	$p=0.02$

displacement in both stance conditions (Fig. 4a). During the rest phases the COP displacement revealed a bi-phasic pattern. The overall COP displacement was larger during rest-on-shelf ($\text{range}_{\text{shelf}}=4.5$ cm) than during rest-on-desk phases ($\text{range}_{\text{desk}}=2.6$ cm). The model did not provide COP changes in the frontal plane in both stance conditions (Fig. 4d).

Subjects' data revealed that the COP pattern is equivalent to the one, predicted by the model, (Fig. 4b and c). During the transition phases, the overall COP displacement was significantly smaller in parallel stance conditions as compared to step stance condition or as predicted by the model (Table 3, $p<0.01$). In the frontal plane, COP remained almost unchanged when subjects lifted and lowered the box in the parallel stance condition (Fig. 4f). In the step stance condition a four-phasic lateral COP displacement pattern occurred, when subjects lifted and lowered the load (Fig. 4e).

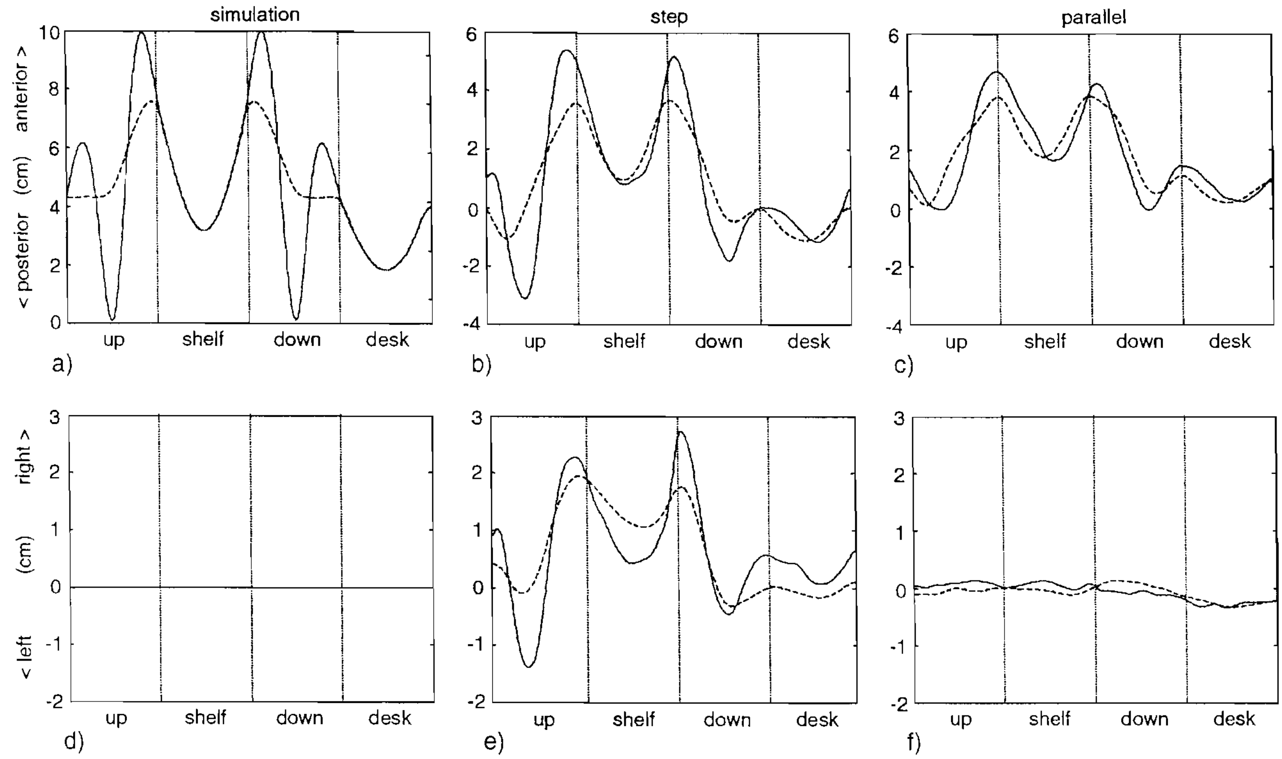

Fig. 4. Mean displacement of COP (solid) and COM (dashed) in the anterior/posterior and lateral direction: (a, d) simulation of lifting and lifting in (b, e) step stance and (c, f) parallel stance condition.

Table 3
Mean (S.D.) displacement of COP/COM (cm) in anterior/posterior direction. (Note that both parameters are reduced in parallel stance during transition phases as compared to simulation but not in step stance condition)

	Stance	Transition-up (cm)	Rest-on-shelf (cm)	Transition-down (cm)	Rest-on-desk (cm)
COP	Parallel	6.6 (1.5)*	3.9 (1.9)	6.0 (2.3)*	2.2 (1.4)
	Step	9.8 (4.2)	5.3 (2.1)	9.8 (3.6)	2.5 (1.1)
COM	Parallel	3.8 (1.4)*	2.9 (0.5)	3.8 (1.5)*	1.3 (0.6)
	Step	5.3 (2.0)	3.2 (0.4)	5.1 (2.2)	1.7 (0.9)

*Denotes statistically significant difference to simulation ($p<0.05$).

During the resting phases a similar COP displacement in the sagittal plane was found during both stance conditions as predicted by the model (Table 3). Data showed a bi-phasic COP displacement pattern in the anterior–posterior direction (Fig. 4b and c). In the frontal plane, step stance was accompanied by a bi-phasic COP displacement, whereas parallel stance showed no motion (Fig. 4e and f).

As predicted by the model, COM showed a specific pattern of displacement that was in phase with the COP displacement in both stance conditions (Fig. 4a). The amplitude of COM displacement was smaller than the COP displacement in all phases ($p<0.01$, Table 3).

As in COP investigation, subjects data revealed that the COM pattern was equivalent to that predicted by the model (Fig. 4e and f). The over all COM displacement was significantly smaller in parallel stance conditions as compared to step stance condition or as predicted by the model (Table 3). Step stance was accompanied by larger lateral COM displacement when compared to parallel stance or simulation (Fig. 4d–f).

Lifting and lowering of the box in parallel stance was accompanied by an early negative peak in the sagittal plane shear force which occurred in the first 10–20% (134 ± 130 ms after lift-off) of the lifting and lowering phases of the task (Fig. 5c). This force was closely associated with the initial backward displacement of the pelvis in the lift-off phases of the task (Fig. 3c). In step stance, this initial posterior directed peak in shear force was almost absent (transition-up) or markedly decreased (transition-down) as compared to the parallel stance condition (Fig. 5d, Table 4). In the remainder of both

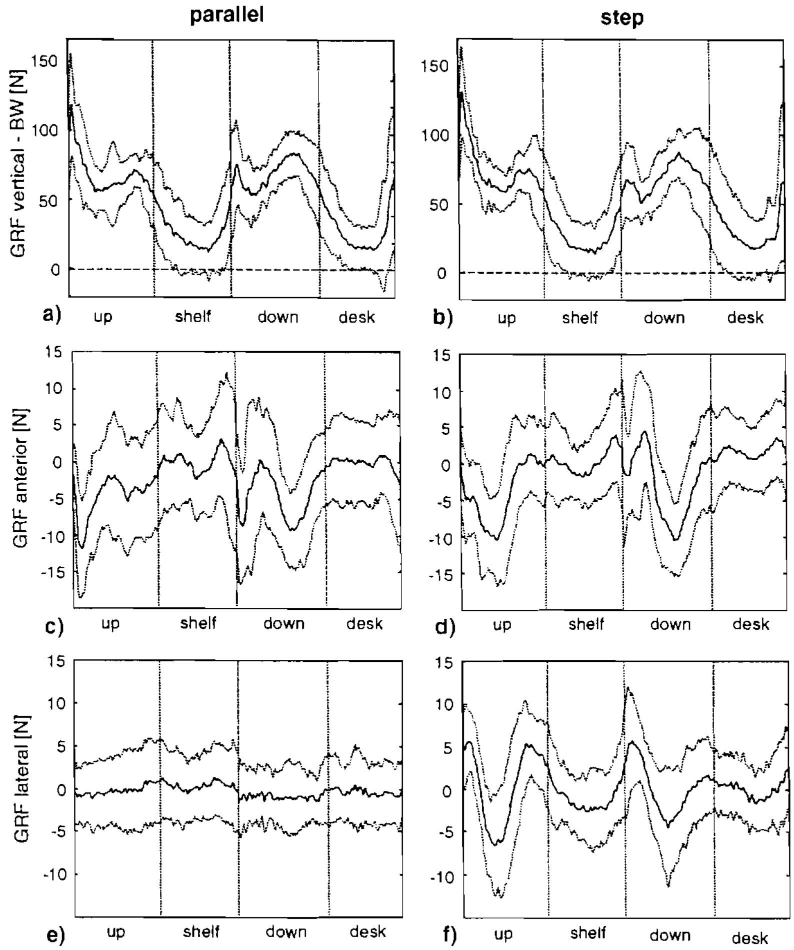

Fig. 5. Mean vertical, sagittal and lateral Ground Reaction Force (±S.D. dotted) of lifting in (a, c, e) parallel stance and (b, d, f) step stance.

Table 4
Mean (S.D.) of Root Mean Square in horizontal GRF (N) during the first 300 ms of each transition phase. (Note the increase of anterior/posterior shear force in parallel stance in transition up and the general increase of lateral shear force in step stance condition)

Phase	Anterior/posterior		Lateral	
	Transition-up (N)	Transition-down (N)	Transition-up (N)	Transition-down (N)
Parallel stance	10.3 (4.9)	10.4 (4.4)	2.8 (0.7)	2.9 (0.9)
Step stance	6.7 (2.6)	8.4 (2.9)	5.3 (2.0)	7.3 (2.4)
Difference	$p = 0.007$	n.s.	$p = 0.001$	$p = 0.000$

transition phases, changes in GRF were significantly larger in the step stance condition than in the parallel stance condition ($P<0.01$) indicative of an increased shear support (Fig. 5c and d). Lateral shear forces were minimal in parallel stance condition (Fig. 5e). However, lifting in step stance was accompanied by significantly increased lateral shear forces as compared to parallel stance (Table 4). It showed a pronounced four-phasic pattern t, closely associated with the acceleration of trunk and pelvis displacement, respectively (Fig. 5f).

During the resting phases, sagittal shear forces showed smaller values than during transition phases (Fig. 5c and d). There were no differences in magnitude of GRF between the two stance conditions. In the frontal plane, there were no lateral GRF in the parallel stance condition. Resting in step stance was accompanied by a bi-phasic lateral displacement pattern.

3.3. Relationship between the kinematic and kinetic parameters

When subjects lifted or lowered the load in parallel stance, the spatial coordination of their trunk, pelvis and the resultant COP/COM displacement were out of phase. The trunk moved in the same direction as the shift of the COP/COM. The pelvis movement, however,

Table 5
Mean correlation coefficient between trunk and pelvis displacement during the different phases of lifting cycle. A positive correlation coefficient is indicative of trunk and pelvis motion into the same direction. A negative correlation coefficient is indicative of trunk and pelvis motion into opposing directions. Values close to '0' represent no correlation and close to '1' or '−1' excellent correlation and anti-correlation, respectively

Direction	Stance	Lifting phases			
		Transition-up	Rest-on-shelf	Transition-down	Rest-on-desk
Anterior	Parallel	−0.76*	+0.79*	−0.94*	+0.78*
	Step	−0.04	+0.28	−0.46	+0.84*
Lateral	Parallel	n.a.	n.a.	n.a.	n.a.
	Step	+0.97*	+0.94*	+0.97*	+0.93*

*Denotes statistically significant correlation between trunk and pelvis ($p<0.05$).
n.a. indicates not applicable correlation calculation, due to too low range of motion.

was in the opposite direction to the trunk and COP/COM (Table 5, negative correlation). No lateral displacement of the trunk and pelvis, or of the COP/COM were observed during lifting in the parallel stance condition.

When subjects performed the tasks in step stance, the trunk and pelvis moved in phase, especially in the frontal plane (Table 5, positive correlation). In the initial phase of the lift, the pelvis sagittal motion briefly opposed that of the COP shift and the trunk displacement in the sagittal plane. In the further course of the load transfer, the direction of the pelvis and trunk displacement was in phase with the COP/COM shift. Thus, the coordination of hip, trunk displacement and COP/COM observed for the step stance condition appeared to fit the inverted pendulum model better, than did the parallel stance condition.

4. Discussion

Evidence based on kinematic and kinetic parameters has been presented which indicates that subjects lifting a load with different base of support use different postural control strategies. In parallel stance, postural response consisted of axial movements in the sagittal plane. Such strategy was accompanied by increased posterior shear forces after lift-off. In step stance postural response included lateral transfer of the body. Lateral shear forces were increased as compared to parallel stance.

4.1. Model

A stiff 'Inverted Pendulum Model' for predicting the postural restoration following postural perturbations has been proposed (Johansson et al., 1988). It is applicable in quasistatic tasks, such as quiet erect stance. The COP displacement determines the COM acceleration (Guersen et al., 1976). Accordingly, the predicted COP changes cover voluntarily controlled aspects that are related to ankle torque in the sagittal plane (Winter et al., 1998). It is valid only if external perturbation is as small as the COM remains within the base of support (Rietdyk et al., 1999). Other literature defines a preferred zone with change of support absent (Popovic et al., 2000). If the COM sways beyond this zone the 'Inverted Pendulum Model' would predict a fall, as there is no force to restore balance and stepping is not part of the model (Otten, 1999; Maki and McIlroy, 1997; McIlroy and Maki, 1993). Hence, stiffness cannot be assumed and multi segmental motion, allowing hip movements and accompanying postural reactions would be required to explain compensation responses (Horak and Nashner, 1986). The choice of the 'Inverted Pendulum Model' in the present experiment was justified as the amount of perturbations caused by the load and lifting movements was not sufficient to shift the COM outside the preferred zone in the base of support. A limitation to obtained results may be that measured motion of critical data (pelvis) is in the order of long range measurement accuracy. As relative displacement data is used exclusively, the better short-range accuracy is applicable. The simplified model (no angular acceleration) may lead to underestimated range of simulated COP displacement. The low dynamics in the simulated task limits the influence of simplification.

4.2. Dependence of postural control on the base of support

The simulation of the lifting task using the inverted pendulum model did not take into account postural control efforts. Thus, the COM displacement was a consequence of the lifting task and the COP displacement directly reflects the destabilizing effect of the task. Furthermore, in consensus with previous studies (Nashner and McCollum, 1985), the GRF was assumed to point into the same direction as did the COM, in order to achieve static balance during the simulation of the lift. Postural adjustments during task execution were described as mechanisms to keep or to restore the situation where the GRF pointed to the COM (Horak and Nashner, 1986; Pedotti et al., 1989). Voluntary movements were possible only by an active torque generation around the ankle, that would correspond to the voluntary activation of the ventral and plantar muscles around the ankle, an "ankle strategy". The ankle torque produced had to be of sufficient magnitude to accelerate the COM during lifting or lowering the box in order to balance compensation when hip and trunk move in synchrony. The acceleration of the COM is directly related to the difference between COP and COM in quiet standing (Winter et al., 1998). This is not by active or passive control, but is due to pure mechanical coupling during stiff posture (Morasso and Schieppati, 1999).

Lifting in parallel stance was associated with an acceleration of the COM, which could not be compensated by purely activating the ventral or plantar ankle muscle groups alone. Thus, subjects used an axial postural compensation strategy, where they pushed their pelvis forward and simultaneously moved their trunks backward in order to maintain equilibrium.

This actual behavior with the simultaneous activation of both an ankle and opposing hip strategy did not fit the inverted pendulum model. Instead, subjects initially shifted their pelvis backward in order to generate backward momentum that could be used to initiate the lift-off of the load. This strategy was associated with an early posterior peak in the GRF (Fig. 5c). These findings suggest that a lift performed with a short base of support requires a more complex whole body co-activation pattern in order to maintain equilibrium than the one observed during step stance. Lifting in parallel stance provided a solid lateral stability as indicated by the absence of lateral displacement of the body (Fig. 3f).

When subjects lifted with an increased base of support in step stance, hip and trunk motion were out of phase with the COP/COM shifts only in the early phase of the lift. During lift-off, subjects appeared to behave more like a "crane" by utilizing the increased leverage provided in the sagittal plane by the step stance while showing a postural compensation pattern similar to that when lifting in parallel stance. The inverted pendulum model appeared to fit the behavior during step stance better than during parallel stance, at least in the anterior–posterior direction. These findings suggest that in the anterior–posterior direction the postural control strategies required to maintain balance during lift-off were less complex than in the parallel stance condition. In the further course of the lift performed in step stance, subjects appeared to phase-lock the pelvis and trunk and simultaneously shift them in the horizontal plane, across the base of support to accomplish the lifting task. In this phase of the lift, subjects took advantage of the increased mechanical support, which simplified postural control since trunk and pelvis were coordinated as one unit. In this phase of the lift postural compensation may be sufficiently achieved by muscles around the ankle. However, lateral movement increased in the step stance condition due to an active lateral shift of the COM between the two feet (Fig. 3e). This aspect of the behavior is comparable to the load transfer in the double support phase of gait (Chao et al., 1983; McMahon, 1986). The lateral control of the COM was associated with increased lateral shear forces that may make subjects more susceptible to external lateral perturbations, especially when the load is centered over one leg.

4.3. Implications

Both lifting techniques exhibit positive and negative aspects. Lifting in step stance reduces complexity of postural control in the anterior/posterior direction. But, the task is asymmetric and decreases lateral stability when compared with lifting in parallel stance. We cannot recommend either one as being better in terms of postural control.

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