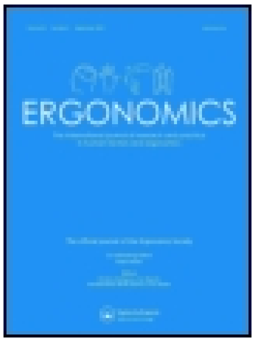

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A biomechanical analysis of anterior load carriage

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Front load carriage is a common occupational task in some industries (e.g. agriculture, construction), but, as compared to lifting tasks, relatively little research has been conducted on the biomechanical loading during these activities. The focus of this study was to explore the low back biomechanics during these activities and, specifically, to examine the effects of load height and walking speed on trunk muscle activity and trunk posture. Eleven male participants participated in two separate front load-carriage experiments. The first experiment called for carrying a barbell (with weight corresponding to 20% of elbow flexion strength) at three heights (knuckle height, elbow height and shoulder height) at a constant horizontal distance from the spine. The second experiment called for participants to carry a bucket of potatoes weighing 14 kg at the same three heights, but with no further restrictions in technique. In both experiments, the participants performed this task while either standing still or walking at a self-selected speed. As they performed these tasks, the activity levels of the right-side muscle of the rectus abdominis, external oblique, biceps brachii, anterior deltoid and three levels (T9, T12 and L3) of the erector spinae were sampled. Mid-sagittal plane trunk posture was also quantified using three magnetic field-based motion sensors at T9, T12 and L3. The results showed a significant effect of both walking speed and load height on trunk posture and trunk muscle activity levels in both the barbell and bucket experiments. In the barbell experiment, the walking trials generated 43% more trunk muscle activity than the standing trials. Trials at shoulder height produced 11% more muscle activity than trials at elbow height in the T9 erector spinae muscles and 71% more muscle activity in the anterior deltoid. In the bucket experiment, trunk muscle activity responded in a similar fashion, but the key result here was the quantification of the natural hyperextension posture of the spine used to balance the bucket of potatoes. These results provide insight into muscle activation patterns

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in dynamic settings, especially (load) carrying biomechanics, and have implications in industrial settings that require workers to carry loads in front of their bodies.

Keywords: Lumbar; Electromyography; Motion analysis; Agriculture

1. Introduction

Physically demanding manual material handling work tasks, such as lifting and carrying have been linked in epidemiological studies to low back problems (e.g. Kuiper *et al.* 1999, Vuuren *et al.* 2005). While considerable research has focused on the biomechanical challenges of lifting tasks (e.g. Waters *et al.* 1993, van Dieën *et al.* 1999), much less research has focused on the biomechanical challenges of anterior load carriage in occupational settings (Magnusson *et al.* 1987). 'Anterior load carriage' is specified here to distinguish it from posterior load carriage (i.e. backpack biomechanics), as a great deal of research has been conducted in this area (e.g. Knapik *et al.* 1996, Goh *et al.* 1998, Stuempfle *et al.* 2004).

Much of the previous work focused on the ergonomics of anterior load carriage has made use of the psychophysical methodology. Lu and Aghazadeh (1994) present a review of a number of psychophysical studies that considered carrying activities and the reviewed studies examined a variety of factors including carrying mode and height, walking speed, time and distance. Of particular relevance to the current study were data generated by Snook (1978), which considered two different heights at which a load was carried (approximately elbow height and approximately knuckle height). Their results showed that psychophysically determined maximum acceptable weights of carry were, on average, 26% (range 14–42%) higher at knuckle height than at elbow height. These results were supported by subsequent work by Snook and Ciriello (1991).

From a biomechanical perspective, much of the current literature on load carriage is focused in two areas: backpack biomechanics; and static modelling of anterior load carriage. The research on posteriorly mounted loads has shown that speed, load weight and height have an effect on energy consumption, spinal loading and coactivation of trunk muscles (e.g. Knapik *et al.* 1996, Goh *et al.* 1998, Holt *et al.* 2003, Stuempfle *et al.* 2004). Stuempfle *et al.* (2004) considered the effect of load position in a backpack on physiological and perceptual variables. Participants in this study carried an internal frame backpack where the load was positioned at a high (~T1–6), central (~T7–12), or low (~L1–5) region of the spine. Results showed that energy costs and perceived exertion decreased with increasing load height. Considering this type of posterior loading task from a more biomechanical perspective, Bobet and Norman (1984) evaluated the electromyographic (EMG) activity of the trapezius and erector spinae muscles when a 19.5 kg load was placed on the back at the C1–C7 region and at the T1–T6 region. Under static conditions, these authors found similar activation levels for both the high and low load placements. Under dynamic walking conditions, however, placing the load in the C1–C7 region created significantly higher levels of muscle activity for both the trapezius and erector spinae muscles as compared to those captured when the load was carried at the T1–T6 region, indicating an interesting interaction between load height and walking speed, two variables that were considered in the present study.

The literature detailing the biomechanics of anteriorly located loads is less expansive and has often been focused on comparing anterior loading with other load carriage locations (e.g. Cook and Neumann 1987) or has been considered under static weight-holding conditions with the goal of assessing spinal stability. Granata and Orishimo (2001) examined the stability of the spine in a static situation where participants stood and held a weighted barbell at a constant distance in front of their body at five prescribed heights ranging from 0–80 cm above the sacrum. They found that EMG activity increased in the trunk flexors as the height of the load was increased and they hypothesized that this demonstrated the neuromuscular response to changes in spinal stability. While this research provided an important detailed assessment of static spinal stability, most occupational tasks involving anterior load carriage are dynamic and muscle activation profiles during dynamic load carriage would be helpful to understand the risks posed.

The objective of the present study was to evaluate the kinetic (muscle activity) and kinematic (trunk posture) responses of human participants during a weight-carrying task with particular emphasis on the effects of walking speed and load height.

2. Methodology

The methods outlined in this paper are a combination of two experimental protocols conducted in the same experimental session, in which participants were asked to carry a load while walking (or standing) on a treadmill. The first protocol employed a barbell that was precisely positioned relative to the spine of the participant to control the moment of the load as they carried it. The second protocol asked participants to carry a bucket full of white potatoes using a self-selected posture limited only by the grip height requirement for that trial. Trials from the two protocols were completely inter-mixed during the experimental sessions.

2.1. *Participants*

A total of 11 healthy male participants from the university population (with no history of low back pain) participated in this experiment. The participant group had a mean age of 25.4 (SD 4.3) years, mean stature of 179.5 (SD 5.2) cm and mean whole body mass of 72.2 (SD 5.3) kg. Of the 11 participants, ten were right-handed and one was left-handed. Participants provided written informed consent prior to participation and this protocol was approved by the university's Institutional Review Board.

2.2. *Apparatus*

2.2.1. Experimental task apparatus. In these experiments, participants stood or walked on a treadmill (RTM-400; Biodex, New York, USA), while carrying either a loaded barbell (figure 1a) or a standard agricultural harvesting bucket filled with 14 kg of potatoes (figure 1b). The bucket containing the potatoes was a standard model that is used by workers for carrying produce in agricultural fields. The weight of the barbell corresponded to 20% of each participant's elbow flexion capacity as established during a maximum voluntary elbow flexion exertion measured on a dynamometer (see section 2.5 for details). The barbell was 0.7 m long (with the load centrally located between the hands) and was equipped with a measuring stick to allow the researchers to ensure that a

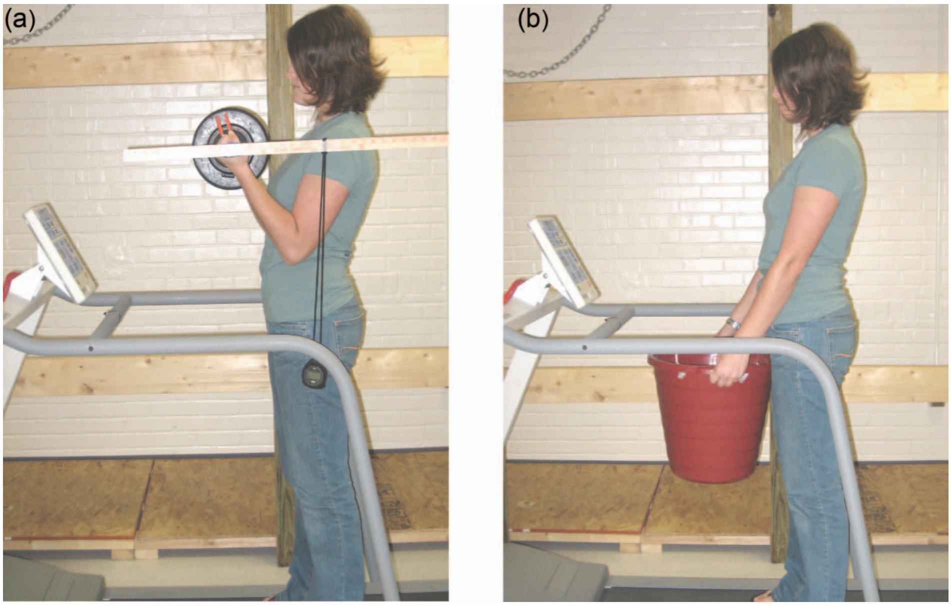

Figure 1. (a) Barbell task with weight held at shoulder height; (b) bucket task with weight held at knuckle height.

constant moment arm of the barbell was maintained about the spine. Because of this level of control, the barbell experiment represents a more controlled study of the response of the low back. The bucket experiment, on the other hand, did not provide the same level of moment arm control and was therefore directed toward analysing the response from a more human performance perspective.

2.2.2. Data collection apparatus. Muscle activation levels and sagittal plane trunk posture were the primary measures considered in this study. Seven pairs of Ag-AgCl bipolar surface electrodes (In-vitro Metric, Healdsburg, CA, USA) were used to monitor the EMG activity of the seven muscles of interest on the right-hand side of the body (load carrying being assumed to be a relatively symmetric task.) These data were carried via shielded cable to the main amplifiers (Model 406; Data Design, Inc., New York, USA) that filtered the EMG data (with a notch filter of 60 Hz and low pass filter of 1000 Hz). Electrode pairs were placed over the right rectus abdominis, the right external oblique, the right biceps brachii and the right anterior deltoid and at three locations of the erector spinae (T9, T12 and L3 levels).

In addition to the muscle activation data collected through EMG, sagittal plane trunk kinematics were also captured. Three motion sensors from a magnetic field-based motion tracking system (Ascension Technology Corporation, Burlington, VT, USA) were used to track the movement of the lower spine during the load carriage trials (system specifications: static resolution position 0.08 cm; static accuracy orientation 0.5° root mean square (RMS); static resolution orientation 0.1° RMS.) This tracking system recorded the time-dependent x, y and z coordinates and the roll, pitch and yaw of the three sensors at a rate of 85 Hz. The motion sensors for the motion tracking system were placed over the spine at the T9, T12 and L3 levels.

Normalized (to maximum) electromyography was employed in this study, so it was necessary to capture the maximum EMG activity levels for each of the muscles of interest. To capture the maximum voluntary contraction (MVC) for the biceps brachii and anterior deltoid, an isokinetic dynamometer (KINCOM #125E; Chattecx Corp., Chattanooga, TN, USA) was used. A lumbar dynamometer (Marras and Mirka 1989) was used to apply static resistance during the lumbar extensor MVC (erector spinae at T9, T12 and L3) with the torso in 10° of flexion. In order to capture MVCs in the abdominal muscles, a strap was placed over the chest to apply manual, static resistance to the participants as they attempted trunk flexion.

2.3. *Independent variables*

The independent variables in both experiments were walking speed and load grip height. Speed had two levels: standing still; or walking at a comfortable pace. Height had three levels: knuckle (arms fully extended downward); elbow; and shoulder heights.

2.4. *Dependent variables*

The dependent variables in the study included the normalized (to maximum) EMG data for the seven muscles sampled (rectus abdominis, external oblique, biceps brachii, anterior deltoid, erector spinae at T9 level, erector spinae at T12 level and erector spinae at L3 level) and the average sagittal angle of the lower spine relative to the normal standing position that was measured by the motion tracking system. A positive angle indicates a hyperextension (i.e. backward lean) from the upright neutral posture.

2.5. *Experimental procedures*

Upon entering the laboratory, each participant was given a 5-min warm-up period consisting of light stretching and walking on the treadmill to determine the 'normal walking pace' that would be used for the walking trials of the study. The mean (across participants) walking speed was 2.5 (SD 0.35) miles per hour (mph) and the range in walking speeds was 1.9–3.3 mph. Surface electrodes were applied to the skin over the muscles of interest using standard preparation techniques (Marras 1990). Participants were then asked to perform a series of MVCs against static resistance provided by a dynamometer for each muscle group tested. During the biceps flexion MVC, the dynamometer recorded the maximum force that was exerted by the biceps with the elbow at 90° and the shoulder in neutral posture (elbow height, weight-carrying posture) and this maximum value was used to establish the 20% load to be used during the barbell trials. The two MVC trunk extensions were performed at 10° of flexion using a lumbar dynamometer (Marras and Mirka 1989). Each maximum exertion lasted for 3 s and there was a 1-min break between maximum effort exertions. After the MVCs were completed, the motion sensors were attached to the skin over the midline of the spine at the T9, T12 and L3 vertebral levels using double-sided tape.

Upon completion of these preliminary activities, participants performed the series of weight-holding tasks with the barbell and the bucket. In these trials, the barbell or bucket was held at knuckle, elbow or shoulder height, while standing still or walking at the participant-selected 'normal walking pace'. Each combination of load height (knuckle, elbow, shoulder) and walking speed (standing, walking) was performed by the participant three times for a total of 36 trials (18 barbell and 18 bucket). The data collection period lasted for 6 s for both static and dynamic trials. During the walking trials, the participants began walking at their normal walking pace and then the load was handed to them. When they reached steady state with this load, the 6-s data collection period began. As soon as the 6-s collection period was over, the experimenter took the weight from the participant. Participants walked no more than a total of 15 s per trial. A break of 30 s was given between experimental trials.

2.6. *Data Processing*

2.6.1. Electromyographic data. All of the unprocessed EMG data were high-pass filtered at 15 Hz and low-pass filtered at 1000 Hz and notch filtered at 60 Hz (ambient electrical noise) and 85.33 Hz (noise generated by the motion monitoring system) and their harmonics. The signals were rectified and averaged across the 6-s data collection period. These task EMG data were then normalized relative to the MVC EMG data.

2.6.2. Motion sensor data. The kinematic data captured during the static and dynamic load carriage trials were processed using the Motion Monitor™ (Innovative Sports Training, Inc., Chicago, IL, USA) software and the sagittal angle value from each sensor was used in the analysis. These data were averaged over the 6-s data collection period. The maximum and minimum of this sagittal angle were also captured from each trial to provide an estimate of the range of motion.

2.7. *Statistical analysis*

ANOVA and multiple ANOVA (MANOVA) were used to statistically analyse the effect of speed, height and speed × height on the dependent variables. Before conducting this analysis, the assumptions of the ANOVA were first tested and verified using the graphical approach advocated by Montgomery (2005). In several cases, the assumption of the homogeneity of variances was violated requiring a transformation (in several cases this required a natural log transformation for the back extensor muscles and required a reciprocal transformation for the abdominal muscles). These transformations were successful in addressing this violation. In all MANOVA and ANOVA analyses, a *p*-value < 0.05 was used as the criterion for significant effect. In only those cases where the MANOVA analysis showed a significant effect were the univariate analyses attempted. If a significant interaction was found in the ANOVA, simple effects analysis was performed to further evaluate any significant main effects that were present. A Tukey-Kramer *post hoc* analysis was performed when any significant effect was found by the ANOVA.

3. Results

3.1. *Barbell experiment*

3.1.1. Electromyographic results. The EMG results of the barbell trials show that muscle activity was higher in the walking trials than in the standing trials (table 1 and figure 2). The increases in activity in the walking trials were 33%, 49% and 47% in the T9, T12 and L3 erector spinae muscles, respectively. Furthermore, muscle activity in the rectus abdominis muscle increased by 51% and the increase was 65% for the external oblique. Load height had a significant effect on muscle activity for the T9 erector

Table 1. Results of the multiple ANOVA (MANOVA)/ANOVA of the electromyographic data from the barbell trials.

	MANOVA	T9	T12	L3	RA	EO	BI	AD
Speed	F = 59.1 $p < 0.001$	F = 175.6 $p < 0.001$	F = 97.1 $p < 0.001$	F = 157.4 $p < 0.001$	F = 50.5 $p < 0.001$	F = 117.1 $p < 0.001$	F = 35.2 $p < 0.001$	F = 7.6 $p = 0.007$
Height	F = 20.9 $p < 0.001$	F = 7.5 $p < 0.001$	F = 1.4 $p = 0.25$	F = 1.2 $p = 0.29$	F = 4.4 $p = 0.01$	F = 2.2 $p = 0.11$	F = 5.5 $p = 0.005$	F = 42.6 $p < 0.001$
Speed*Height	F = 2.6 $p = 0.002$	F = 0.8 $p = 0.44$	F = 0.9 $p = 0.42$	F = 0.0 $p = 0.29$	F = 1.4 $p = 0.26$	F = 10.3 $p < 0.001$	F = 1.6 $p = 0.19$	F = 0.8 $p = 0.46$

RA = rectus abdominus; EO = external oblique; BB = biceps brachii; AD = anterior deltoid.

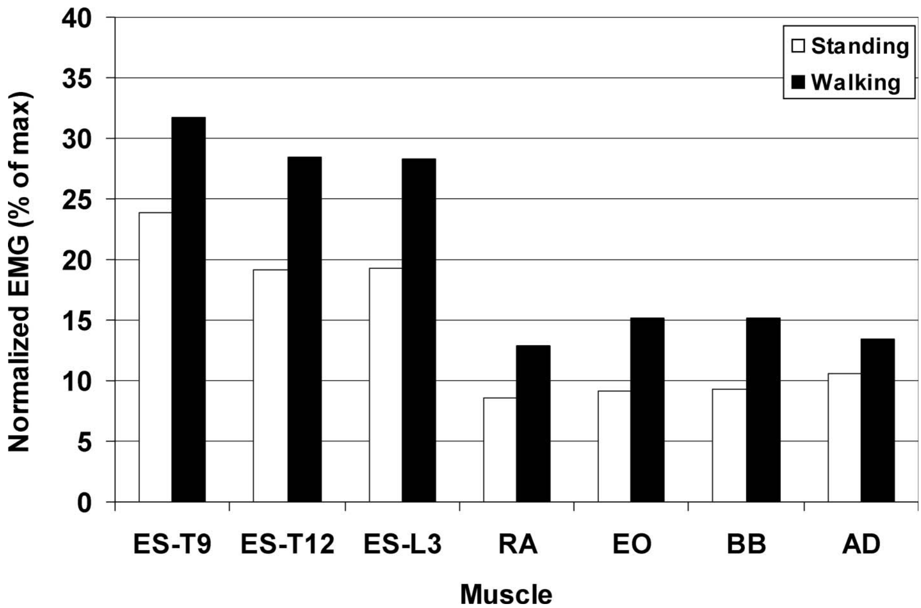

Figure 2. Effect of speed on muscle activity during the barbell experiment. Only statistically significant ($p < 0.05$) effects are shown. EMG = electromyography; ES = erector spinae; RA = rectus abdominus; EO = external oblique; BB = biceps brachii; AD = anterior deltoid.

spinae muscles, rectus abdominis muscle, biceps muscle and anterior deltoid muscle (figure 3).

3.1.2. Kinematic results. The mean sagittal angles showed a significant decrease when going from standing to walking for the T9 and T12 levels (table 2 and figure 4), indicating a greater backward lean during the standing trials. While the walking trials produced a lower average sagittal angle, the dynamic nature of walking trials was also explored and it was found that the average range of motion in the trunk (all three levels) during the walking tasks was about 1.5° greater than the range of motion during the standing tasks ($p < 0.05$). The results also showed a significant effect of load height on the sagittal angles

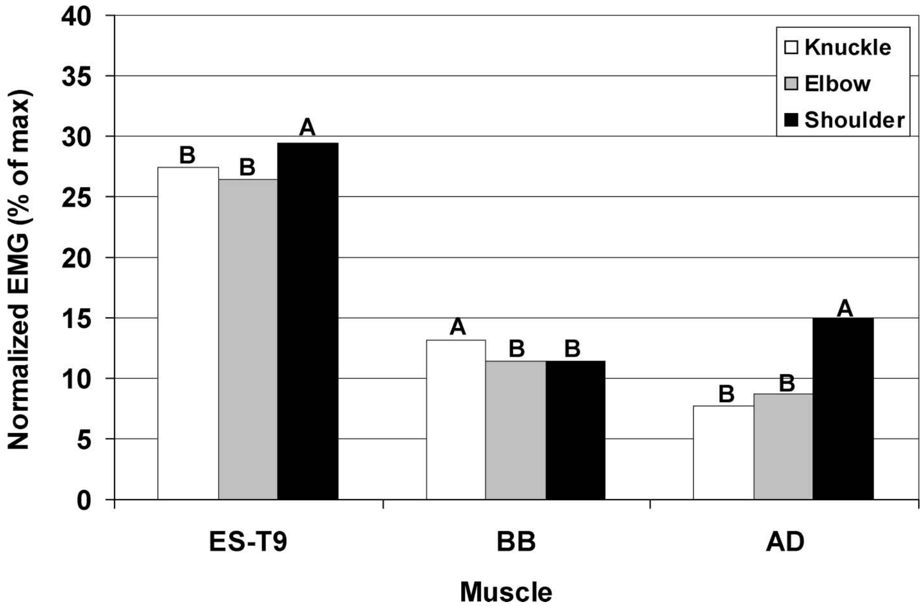

Figure 3. Effect of height on muscle activity during the barbell trials. The letters above each column are from the Tukey-Kramer *post hoc* test. For a given muscle, different letters on the vertical bars indicate that the activation levels were significantly different. EMG = electromyography; ES = erector spinae; BB = biceps brachii; AD = anterior deltoid.

Table 2. Results of the multiple ANOVA (MANOVA)/ANOVA of the average sagittal angle of the lower spine relative to the normal standing position from the barbell trials.

	MANOVA	T9	T12	L3
Speed	F = 16.9 $p < 0.001$	F = 32.2 $p < 0.001$	F = 22.8 $p < 0.001$	F = 0.6 $p = 0.42$
Height	F = 8.2 $p < 0.001$	F = 25.0 $p < 0.001$	F = 10.6 $p < 0.001$	F = 1.5 $p = 0.22$
Speed*Height	F = 1.7 $p = 0.1324$	–	–	–

at the T9 and T12 levels (left side of figure 5). Shoulder height load carriage resulted in larger posture deviations (from upright neutral) than carriage at the two lower heights.

3.2. *Bucket experiment*

3.2.1. Electromyographic results. The muscle activity levels were significantly greater in the walking trials as compared to the standing trials (table 3 and figure 6) with the trends showing a similar response to that shown in the barbell trials. There was a 132% increase in muscle activity for the rectus abdominis from standing to walking trials. The increases

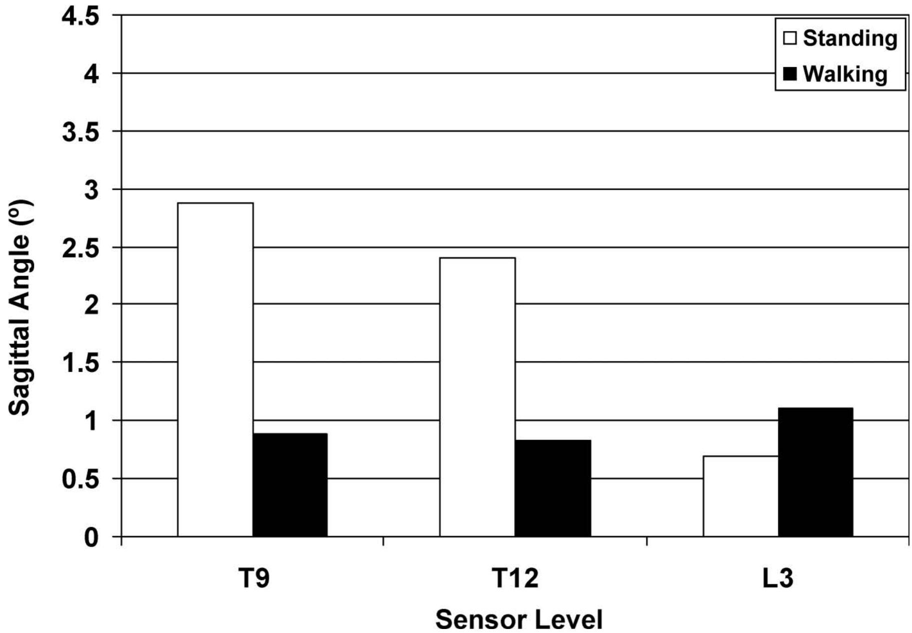

Figure 4. Effect of speed on sagittal angle in the barbell experiment. All effects are shown; L3 effect was not significant, but T9 and T12 were significant at $p < 0.05$.

in erector spinae muscle activity were 35%, 36% and 42% at the T9, T12 and L3 levels respectively. In contrast to the barbell results, the only significant changes in muscle activity with height occurred in the anterior deltoid muscles. The results also showed two interesting significant speed × height interaction effects. For the rectus abdominis, walking generated over a two-fold increase in the activity of the rectus abdominis muscles for knuckle and elbow heights, with a much smaller effect at shoulder height (figure 7). The response of the biceps brachii was much different, with standing while holding the bucket at knuckle height generating much lower responses than any of the other conditions (figure 8).

3.2.2. Kinematic results. Load carriage height significantly affected the mean posture at each of the three sampled spinal levels (table 4 and the right hand side of figure 5). As with the barbell trials, load at shoulder height resulted in a significantly greater average sagittal angle than loads at the two lower heights and the range of motion differences were consistent with those shown in the barbell trials. These effects were slightly more pronounced in the bucket trials reflecting the increased flexibility in the postures allowed in the bucket experiment.

4. Discussion

Most of the archival literature on the biomechanical responses during load carrying is focused on posterior load carriage and is limited to understanding the stresses during carrying work activities. The present study was designed to provide some quantitative data to explore the differences in static vs. dynamic anterior load carrying tasks and to explore the effects of one carrying task design parameter: load height.

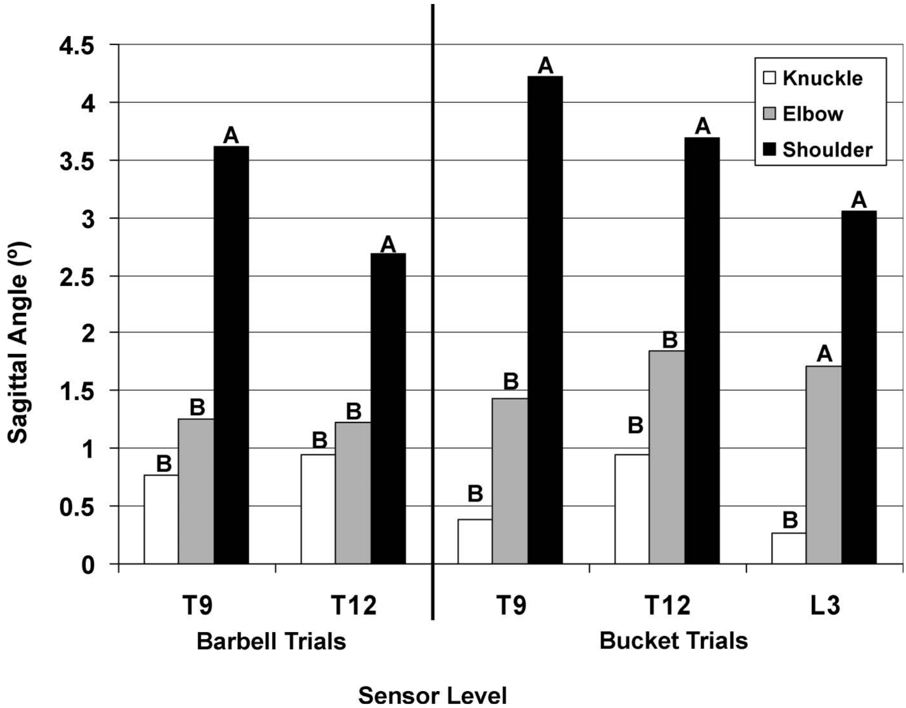

Figure 5. Effect of height on sagittal angle in both the barbell and bucket experiments. The letters above each column are from the Tukey-Kramer *post hoc* test. For a given sensor level, different letters on the vertical bars indicate that the sagittal angles were significantly different.

Table 3. Results of the multiple ANOVA (MANOVA)/ANOVA of the electromyographic data from the bucket trials.

	MANOVA	T9	T12	L3	RA	EO	BI	AD
Speed	$F = 60.7$ $p < 0.001$	$F = 62.8$ $p < 0.001$	$F = 190.0$ $p < 0.001$	$F = 173.5$ $p < 0.001$	$F = 150.8$ $p < 0.001$	$F = 216.3$ **	$F = 14.0$ **	$F = 13.9$ $p < 0.001$
Height	$F = 17.6$ $p < 0.001$	$F = 6.0$ $p = 0.003$	$F = 2.1$ $p = 0.12$	$F = 0.3$ $p = 0.75$	$F = 0.7$ $p = 0.49$	$F = 12.5$ **	$F = 124.3$ **	$F = 124.4$ $p < 0.001$
Speed*Height	$F = 4.2$ $p < 0.001$	$F = 3.3$ $p = 0.04$	$F = 2.3$ $p = 0.11$	$F = 2.3$ $p = 0.06$	$F = 8.6$ $p < 0.001$	$F = 8.2$ $p < .001$	$F = 19.3$ $p < 0.001$	$F = 0.6$ $p = 0.57$

**Simple effects analysis of the interaction of Speed*Height indicates that this was not a significant main effect.
RA = rectus abdominus; EO = external oblique; BB = biceps brachii; AD = anterior deltoid.

The results of this study showed that there are significant differences in the muscle coactivation patterns and trunk posture when changing from load holding (static) to load carrying (dynamic) in terms of both the trunk posture and the muscle activity levels. This was true in both the barbell and bucket experiments. In both of the experiments, there were significant increases in the EMG activity of the back extensors and abdominal

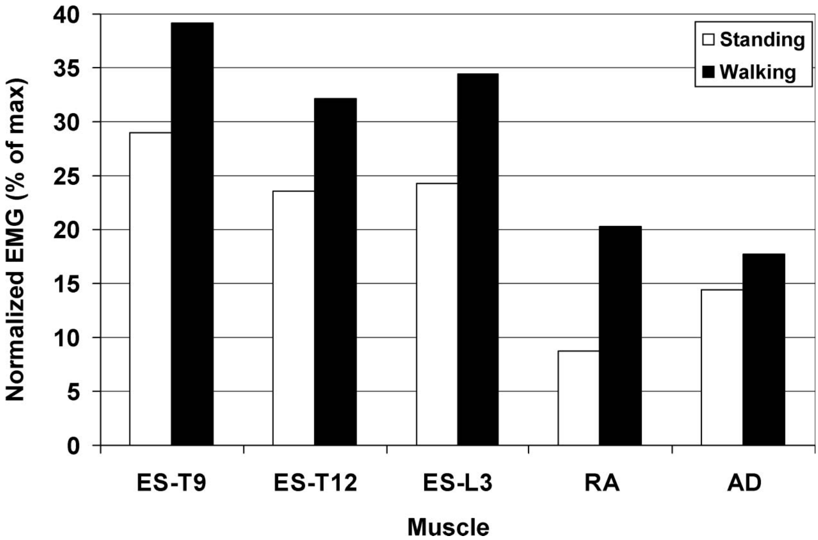

Figure 6. Effect of speed on muscle activity during the bucket experiment. Only statistically significant ($p < 0.05$) effects are shown. EMG = electromyography; ES = erector spinae; RA = rectus abdominus; AD = anterior deltoid.

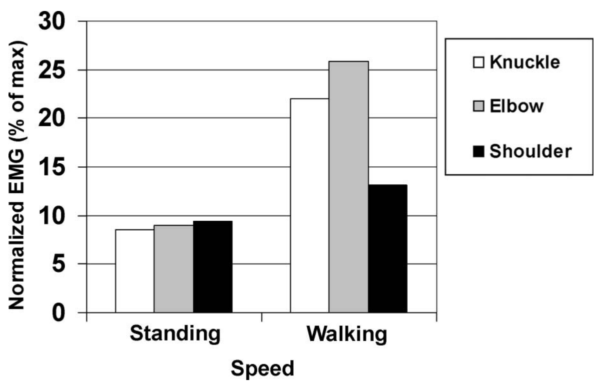

Figure 7. Significant interaction of speed and height on the muscle activity of the rectus abdominis during the bucket experiment. EMG = electromyography.

flexors from the standing to the walking trials. This is most likely due to the fact that during a dynamic task the muscles must constantly compensate for the movement of the body and make corrections to provide dynamic stability to the system. Going from the static condition to the dynamic condition also affected the trunk posture assumed. It has been shown during gait that the trunk exhibits a forward leaning posture when posteriorly loaded (Goh *et al.* 1998) as compared to the more upright posture assumed during quiet standing. The results of the barbell study have shown that this response is still in effect – it is simply shifted posteriorly to shift the centre of mass of the new 'loaded' system to a more balanced location. Interestingly, a similar finding was not found in the bucket study. This can be at least partially explained by the fact that there was greater

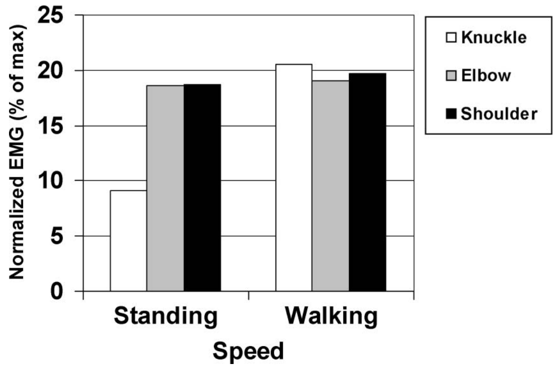

Figure 8. Significant interaction of speed and height on the muscle activity of the biceps brachii during the bucket experiment. EMG = electromyography.

Table 4. Results of the multiple ANOVA (MANOVA)/ANOVA of the average sagittal angle of the lower spine relative to the normal standing position from the bucket trials.

	MANOVA	T9	T12	L3
Speed	$F = 2.7$ $p = 0.048$	$F = 2.8$ $p = 0.10$	$F = 3.6$ $p = 0.06$	$F = 0.3$ $p = 0.61$
Height	$F = 9.8$ $p < 0.001$	$F = 30.1$ $p < 0.001$	$F = 16.8$ $p < 0.001$	$F = 9.1$ $p < 0.001$
Speed*Height	$F = 1.1$ $p = 0.39$	–	–	–

flexibility afforded to the participants in their approach to carrying the bucket, which then generated significant levels of both inter- and intra-subject variability that limited the ability to find statistically significant effects. As noted above, one of the goals of the current study was to expand the understanding of the biomechanical response as described in the static conditions of Granata and Orishimo (2001) to consider these dynamic effects. While it was not possible to replicate all the heights of load holding considered in the previous study (due to safety concerns, i.e. holding weights high while walking) those that were replicated indicate a considerable increase in muscle activation with the walking motion.

The other goal of this study was to begin the exploration of 'design parameters' for a load-carrying task. The parameter that was explored was load height. Compared to speed, load height had a much less universal effect on trunk muscle activation levels, but it had a greater influence on change in sagittal angle of the spine. The height of load had a significant effect in the muscle activity of the erector spinae at the T9 level, anterior deltoid and the biceps brachii in the barbell study. In the bucket study, height only significantly affected muscle activity in the anterior deltoid and the erector spinae at the T9 level. The biomechanical model used by Granata and Orishimo (2001) predicted an increase in the abdominal and paraspinal muscles with an increase in external load height. Results of the present study did not show an increase in the lower erector spinae muscle activity (T12 and L3) with increasing load height, as found by this previous study. This difference could be due to a change in muscle coactivation patterns as a result of the dynamic nature of the study. Further, the greatest increases in the muscle activation levels found by these earlier authors were observed when the weight was held above shoulder level, a level not considered in the present study.

The interesting interactions between speed and height (figure 4) were unexpected. The results showed that the muscle activity in the abdominal muscles (rectus abdominis and external oblique) exhibited differing responses to speed at the different load height levels. The one relatively consistent response was that the trunk flexors (antagonist muscles) had a greater increase in activation in going from standing to walking when the weight was held at knuckle height than at the other heights. In the standing condition, all of these muscles were activated at about 10% of maximum. In the walking conditions, activity in both the rectus abdominis and the external oblique muscles increased for all holding heights, indicating an increase in the stabilizing role that these muscles were playing. In the bucket experiment, the greater increase in the abdominal activation at knuckle height is thought to be driven by the participants holding the weight further away from the legs to provide clearance for the thighs during gait, causing a general increase in the trunk muscle coactivation to maintain stability. A more in-depth interpretation of this response is complicated somewhat by the concomitant changes in trunk posture that occurred at different load heights. Further exploration of this response to the dynamics of gait is warranted.

This study had several limitations that need to be considered. The first limitation was that the study's simulation of dynamic load carriage was conducted on a treadmill instead of walking over ground. This approach limits some of the translational inertial characteristics of the load that might act to change the muscle coactivation patterns and the trunk posture. While there was a load held in the hands (gravitational), this load was not generating the realistic inertial forces that would be created as a person naturally walks with a load. Second, carrying tasks are often performed for much longer than the 6 s duration used in this study and therefore fatigue effects may play a more central role in some of the responses in more realistic occupational settings. In dynamic load carriage tasks, especially anterior ones, fatigue may become a very important consideration in determining the optimal load position.

This study considered the relatively under-explored area of anterior load carriage. Workers in the agriculture industry perform a significant amount of this type of exertion, and understanding the effects of load height and its interaction with walking speed may help ergonomists develop appropriate ergonomic interventions for the prevention of low back injury and fatigue. The development of such interventions is an ongoing area of research in the authors' laboratory and these results have helped to focus their intervention efforts.

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