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A dynamical systems analysis of assisted and unassisted anterior and posterior hand-held load carriage

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Load carriage is recognised as a primary occupational factor leading to slip and fall injuries, and therefore assessing balance maintenance during such tasks is critical in assessing injury risk. Ten males completed 55 strides under five carriage conditions: (1) unassisted anterior, (2) unassisted posterior, (3) assisted anterior, (4) assisted posterior and (5) unloaded gait (UG). Kinematic data were recorded from markers affixed to landmarks on the right side of each participant, in order to calculate segment angles for the foot, shank, thigh and pelvis. Continuous relative phase (CRP) variability was calculated for each segment pair and local dynamic stability was calculated for each segment in all three movement planes. In general, irrespective of the assistive device or movement plane, anterior load carriage was most stable (lower CRP variability and maximum finite-time Lyapunov exponents). Moreover, load carriage was less dynamically stable than UG, displaying the importance of objectively investigating safe load carriage practices.

Practitioner Summary: Dynamical systems analyses were used to comprehensively evaluate the stability of various hand-held load carriage methods. In general, anterior load carriage was significantly more stable than posterior load carriage, Mover’s assistive device had small but beneficial effects on stability, and load carriage was less stable than UG.

Keywords: load carriage; assistive device; dynamical systems analysis; continuous relative phase variability; Lyapunov exponents

1. Introduction

Slips, trips and falls during gait are a major concern in workplaces due to lost time and the associated financial costs (Yoon and Lockhart 2006). In Ontario, the combined falls category (i.e. fall on the same level, fall to a lower level, fall unspecified) accounted for 18.4% of all allowed lost time claims between 2003 and 2012 (WSIB 2013). Gait is a complex movement involving the coordination of multiple joints and body segments (Stergiou, Scholten, et al. 2001), and it is made more challenging with the addition of external load to the trunk or in the hands. As a result, load-carrying tasks are accepted as a major risk factor for occupational musculoskeletal and slip, trip and fall injuries (Liu and Lockhart 2013; Muslim 2013; Yeoh, Lockhart, and Wu 2013).

One industry that involves continuous load carriage and extensive amounts of physical labour is the moving industry. Professional movers repetitively transfer homeowners’ belongings between homes and the moving truck. The vast majority of belongings are packaged in boxes where box sizes and shapes are uncontrolled, box weights are unpredictable and few boxes have handles (Kudryk 2008). Biomechanical research has placed an emphasis on load carriage, but the literature has predominantly focused on load carriage when the load is coupled to the trunk, often in the form of a vest or backpack (e.g. Birrell and Haslam 2009; Fiolkowski et al. 2006; Liu and Lockhart 2013; Motmans, Tomlow, and Vissers 2006). However, use of backpacks or vests in the moving industry is not feasible because of cost, logistics of moving items and work rate concerns (Kudryk 2008).

Professional movers tend to adopt two very distinct hand-held carrying techniques: in front of the trunk (anterior carriage; AC) and behind the trunk, using a flexed trunk posture to support some of the load’s weight (posterior carriage; PC) (Figure 1) (Kudryk 2008). This PC posture has also been observed in other types of industry and developing countries where backpack use is not feasible (Muslim 2013; Muslim and Nussbaum 2014). Previously, movers reported that the PC technique felt less demanding on the lower back, which was supported by an electromyography comparison in which the PC technique significantly reduced muscular activity in the thoracic and lumbar erector spinae (Kudryk 2008). To further reduce the effort of the upper extremities during AC and PC, an on-body ergonomic assistive device named Mover’s...
assistive device (MAD) was developed (Kudryk 2008). Reductions in muscular activity of the wrist flexors as well as perceived efforts led the authors to deem the MAD an effective ergonomic aid.

Prior to recommending either the AC or PC technique or the use of the MAD to stakeholders, it is important to assess their effects on gait stability. By ‘stability’, we refer to the ability to accommodate perturbations and prevent falling when walking on smooth or rough pavement, steps and stairs (Liu and Lockhart 2013). One method of assessing gait stability is to assess intersegment coordination variability (Seay, van Emmerik, and Hamill 2011). Intersegment coordination (quantified as the continuous relative phase (CRP) angle) can provide a comprehensive description of a movement as it describes the interactions of segments that move a person through complex tasks such as load carriage (LaFiandra et al. 2003; Lu, Yen, and Chen 2008; Stergiou 2004; van Emmerik, Miller, and Hamill 2014). The variability (mean standard deviation (SD)) of this coordination across multiple gait cycles provides information regarding gait stability, as dynamical systems theory suggests that transitions between states of stable coordination patterns are preceded by increased variability (Scholz 1990).

External assistive devices such as the MAD may be beneficial for stability in load carriage. Our recent work quantified transverse plane coordination and coordination variability between the pelvis and trunk and between the trunk and box as a means of minimising upper body torque during AC and PC with and without MAD (Smallman, Graham, and Stevenson 2013). MAD use resulted in decreased perceived discomfort and more in-phase coordination between the trunk and pelvis (no changes in intersegment coordination variability). Furthermore, AC resulted in significantly decreased transverse plane coordination variability between the load being carried and the trunk (Smallman, Graham, and Stevenson 2013). However, it is critical to further assess 3D lower limb coordination variability as a proxy for the ability to maintain stable gait during load carriage.

Figure 1. (A) Example of the AAC technique and the marker triad positions. (B) Example of the APC left hand position. (C) Example of the APC right hand position. Hand positioning remained the same in both the unassisted and assisted trials.
A second dynamical systems method for calculating gait stability is to determine the local dynamic stability of segment movements using maximum finite-time Lyapunov exponents (Brujin et al. 2009b). Although there has been a great deal of work assessing dynamic gait stability and the potential for falls under various scenarios and in different populations (e.g. Granata and Lockhart 2008; Hak et al. 2013; Su and Dingwell 2007), few studies have used these methods to assess the local dynamic stability of gait during load carriage. Recently, the combined effects of cognitive and physical loads on the local dynamic stability of backpack load carriage have been assessed (Qu 2013), as was the effect of increasing load on carriage while wearing a weighted double-pack vest (Liu and Lockhart 2013). In both cases, stability was decreased with increasing load. However, the loads were coupled to the trunk, which is not the case in hand-held load carriage, and thus these results may not generalise to local dynamic stability during hand-held load carriage. This was shown by Muslim (2013), who found that medio-lateral trunk local dynamic stability was improved with increasing PC load, where a 50% body mass load was more stable than either 20% or 35%.

The goal of this study was to expand on the aforementioned research by applying two dynamical systems analysis techniques to comprehensively evaluate changes in lower limb stability when carrying loads in the hands anteriorly and posteriorly, with and without MAD. A secondary goal was to compare stability during load carriage to stability during unloaded gait (UG). Based on previous research, we hypothesised the following:

1. Coordination variability would be higher and dynamic stability lower during the PC task.
2. Coordination variability and stability would be similar with and without use of the MAD.
3. All load carriage tasks would be more variable and less stable than UG.

2. Methods
2.1 Participants
Ten healthy male participants with no reported bodily pain who had not participated in the previous study (Smallman, Graham, and Stevenson 2013) were recruited and participated in this study. Before study onset, participants read and signed an information and consent letter approved by Queen’s University Health Sciences Research Ethics Board. Participant mean (SD) age, height and mass were 23.4 years (± 2.4), 1.79 m (± 0.35) and 80.73 kg (± 1.67), respectively.

2.2 Experimental procedures
After providing informed consent, each participant’s preferred treadmill walking speed (PTWS) was determined using published methods (Dingwell and Marin 2006; Smallman, Graham, and Stevenson 2013). The mean PTWS in this study was 1.23 ± 0.13 m/s. An orientation session was then conducted whereby participants walked on the treadmill at their PTWS for 1 min under each of the four load carriage conditions while carrying a box weighing 5 kg. This protocol allowed for additional familiarisation with the load carriage techniques on a treadmill to reduce the influence of any immediate unfamiliarity/motor-learning effects, and to verify that all of the markers were in view of the motion capture cameras.

Data collection took place on a Sports Art Fitness 6300 treadmill (Sports Art Fitness, Woodinville, WA, USA). Kinematic data were collected using a six-camera Vicon 512 Motion Capture System (Oxford Metrics Group, Oxford, UK) at a rate of 120 Hz. Four rigid-body triads of retro-reflective markers (25 mm diameter) were affixed to the superior-lateral side of the right foot, lateral side of the right shank and thigh, and the posterior aspect of the pelvis. A single marker was affixed to the right heel for stride calculations (Figure 1(A)). For anatomical reference purposes, single retro-reflective markers were fixed to the 1st and 5th metatarsals, medial and lateral malleoli, medial and lateral femoral epicondyles, right and left anterior superior iliac spines and right and left posterior superior iliac spines (Davis et al. 1991).

Before beginning the experimental conditions, an anatomical reference trial was collected to define the relationship between the triad coordinate systems and the segmental anatomical coordinate systems. Participants stood upright on the treadmill with their feet shoulder-width apart and their toes pointed directly in front of them (Miller et al. 2008). In order to calculate the functional hip joint centre, participants performed the star movement followed by the arc movement in a fluid motion with their right leg (Camomilla et al. 2006). All single markers except for the heel marker were then removed, leaving only the four triads of markers and the heel marker remaining for the experimental conditions.

In randomised order, participants completed trials under five different experimental conditions: (1) unassisted anterior carriage (UAC), (2) unassisted posterior carriage (UPC), (3) assisted anterior carriage (AAC), (4) assisted posterior carriage (APC) and (5) UG. Experimental trials consisted of participants walking with the treadmill set at their PTWS for a total of 55 right-foot strides under each condition. During the load carriage trials, participants carried a standardised load equivalent to 20% of their body mass to account for differences in body size, in a standard 2 cubic foot cardboard moving box (46 × 38 × 32 cm) (Kudryk 2008). The method of carriage was standardised across all participants to control for any biases due to a change in carriage technique (Kudryk 2008; Smallman, Graham, and Stevenson 2013) (Figure 1(B),(C)). Researchers fitted
participants with the MAD as per Kudryk (2008) before each assisted condition to ensure consistency across all participants and trials. Two minutes of rest was provided between each trial to minimise any effects of muscular fatigue (Smallman, Graham, and Stevenson 2013).

2.3 Data processing

Custom Matlab® software (The MathWorks, Natick, MA, USA) was used for all data analyses. A shape-preserving cubic spline function was applied in Matlab to smooth over data points when markers went out of view. Next, all data were dual-pass filtered using a second order low-pass Butterworth filter with cut-off frequencies ranging between 7.1 and 8.5 Hz as determined using residual analysis (Winter 2009). The functional hip joint centre was located using a least squares method (Camomilla et al. 2006).

Marker coordinate systems were defined for the foot, shank, thigh and pelvis. Anatomical joint centres of the foot, ankle and knee were defined as per Davis et al. (1991). Local coordinate systems were defined for the foot, shank, thigh and pelvis by using an anatomically based coordinate system (Deluzio and Astephen 2007; Graham, Sadler, and Stevenson 2012) and were represented in a treadmill-based global coordinate system so that segment motions represented movement in the sagittal, transverse and frontal planes. Segment angles were extracted using a standard 3D Euler rotation sequence equivalent to flexion–extension, abduction–adduction, internal–external rotation (Wu and Cavanagh 1995), and segment angular velocities were calculated using the first-order central differences method (Winter 2009). The functional hip joint centre was located using a least squares method (Camomilla et al. 2006).

Phase portrait plots were then created by plotting the normalised angular positions (horizontal axis) against the normalised angular velocities (vertical axis) (Peters et al. 2003). Phase angles (PAs, \( \varphi \)) were calculated for each time point of the gait cycle using the four-quadrant inverse tangent function in Matlab (atan2), and defined as the angle from the right horizontal axis (Seay, van Emmerik, and Hamill 2011). CRP angles were calculated between the pelvis and thigh, between the thigh and shank and between the shank and foot in all planes by subtracting the absolute PA of the distal segment (dist) from the absolute PA of the proximal segment (prox) (Equation (3)).

\[
\text{CRP}_i = |\varphi_{i, \text{prox}} - \varphi_{i, \text{dist}}|.
\]

To eliminate discontinuities in the final CRP angle, any value greater than 180° was subtracted from 360°, resulting in a CRP angle in the range of 0–180° (Seay, van Emmerik, and Hamill 2011). CRP was defined by taking the SD of the CRP measurements at each percentage of the stride cycle and then averaging across all SD values in the stride cycle (\( p \)) (Equation (4)), for each condition and participant (Miller et al. 2008; Stergiou, Jensen, et al. 2001; van Emmerik, Miller, and Hamill 2014).

\[
\nu \text{CRP} = \frac{\sum_{i=1}^{p} \text{SD}_i}{p}.
\]
state-spaces were created from the segment angles at each point in time using the method of delays (Abarbanel et al. 1993).

\[ Y(t) = [x(t), x(t + T_d), x(t + 2T_d), \ldots x(t + (n - 1)T_d)] \]

(5)

where \( Y(t) \) is the \( n \)-dimensional state-space, \( x(t) \) is the original time series data, \( n \) is the number of reconstruction dimensions and \( T_d \) is a constant time delay. A time delay of 10\% of the mean cycle length (12 samples) was used to ensure that all data were processed similarly (Bruijn et al. 2009b; Graham, Sadler, and Stevenson 2012), and a constant embedding dimension of six was chosen since this is more than enough to fully capture the gait dynamics (e.g. Bruijn et al. 2009b; Qu 2013).

Diverging Euclidean distances between nearest neighbours in state space were then calculated as a function of time (\( d_j \)), and then log transformed and averaged over all pairs of initially nearest neighbours to get one divergence curve (\( y \)) per segment per movement condition (Rosenstein, Collins, and De Luca 1993).

\[ y(i) = \frac{1}{\Delta t} \ln d_j(i) > . \]

(6)

The maximum finite-time Lyapunov exponent (\( \lambda_{\text{max}} \)) was then calculated as the slope of the linear part of this average divergence curve (Rosenstein, Collins, and De Luca 1993), which was chosen as 0–0.5 cycles (Bruijn et al. 2009a). A larger exponent is indicative of faster divergence, or less stable behaviour (Rosenstein, Collins, and De Luca 1993).
2.4 Statistical methods

All quantitative statistical analyses were performed with SPSS 21 for Windows (IBM Corporation, Armonk, NY, USA) statistical software. First, separate two-way repeated-measures ANOVAs were applied to determine the within-subject effects of load carriage type (anterior vs. posterior) and assistive device (unassisted vs. assisted) on each dependent variable. The dependent variables were the $v_{CRP}$ for each segment pairing (pelvis–thigh, thigh–shank and shank–foot) in all movement directions, and the local dynamic stability ($l_{max}$) for each segment (pelvis, thigh, shank and foot) in all movement directions. The critical $p$-values were adjusted using the false discovery rate technique, by taking into account the number of ANOVAs run in each case (9 segment pairs for $v_{CRP}$ and 12 segments for $l_{max}$) (Benjamini and Hochberg 1995). The adjusted critical $p$-value for the $v_{CRP}$ data was 0.033, whereas the adjusted critical $p$-value for $l_{max}$ data was 0.029.

Second, separate one-way repeated-measures ANOVAs were applied to determine the overall effect of carriage type (UAC, UPC, AAC, APC and UG) on each dependent variable. Then, post-hoc pairwise comparisons with Bonferroni corrections were used to make mean comparisons between each load carriage condition relative to UG. The critical $p$-value for these analyses was 0.05.

3. Results

A summary of the mean and SD of $v_{CRP}$ and the results of the two-way repeated-measures ANOVAs can be found in Table 1. The AC technique resulted in significantly lower $v_{CRP}$ between all three segment pairings in the sagittal plane, between the pelvis and thigh in the transverse plane, and between the thigh and shank and between the pelvis and thigh in the frontal plane than the PC technique ($p < 0.033$). No significant effects of device or interactions between device and technique were found for $v_{CRP}$ for any of the segment pairings in any movement plane.

A summary of the mean and SD of maximum finite-time Lyapunov exponents and the results of the two-way repeated-measures ANOVA can be found in Table 2. The use of the MAD significantly improved the dynamic stability (decreased $\lambda_{max}$) of the shank in the frontal plane ($p < 0.029$; $-3.55$ and $-4.96$ percentage change for AC and PC techniques, respectively). The AC technique significantly improved stability of the thigh in the sagittal plane, stability of the shank in...
the transverse plane and stability of the pelvis in the frontal plane when compared with the PC technique (\( p \), 0.029). There were significant interactions between device and technique for the pelvis in the sagittal plane, the foot in the transverse plane and the foot in the frontal plane. In these cases, the MAD resulted in slightly increased exponents during AC (7.01%, 7.67% and 1.24% respectively) and slightly decreased exponents during PC (–3.65%, –3.56% and –6.60%, respectively).

Figure 4 displays the vCRP for all five gait conditions. In the sagittal plane, there was a significant main effect of condition on all segment pairings, and significant pairwise differences between UG and both UPC and APC (\( p \), 0.05). In the transverse plane there was a significant main effect of condition on the pelvis–thigh pairing, and significant pairwise differences between UG and both UPC and APC for the thigh–shank and pelvis–thigh (\( p \), 0.05). In the frontal plane, there was a significant main effect of condition on the thigh–shank and pelvis–thigh pairings, significant pairwise differences between UG and APC for the thigh–shank, and significant pairwise comparisons between UG and both UPC and APC for the pelvis–thigh (\( p \), 0.05). No pairwise differences existed between UG and AAC or UAC.

Table 1. Summary mean (SDs) of the vCRP for the shank–foot (S–F), thigh–shank (T–S) and the pelvis–thigh (P–T) across all participants and conditions.

<table>
<thead>
<tr>
<th>Segment pairing</th>
<th>Unassisted</th>
<th>Assisted</th>
<th>ANOVA results p-values</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Anterior</td>
<td>Posterior</td>
<td>Anterior</td>
</tr>
<tr>
<td>Sagittal plane</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>S–F</td>
<td>4.90 (0.79)</td>
<td>5.91 (0.88)</td>
<td>4.81 (1.08)</td>
</tr>
<tr>
<td>T–S</td>
<td>7.38 (1.47)</td>
<td>8.79 (1.69)</td>
<td>7.29 (1.47)</td>
</tr>
<tr>
<td>P–T</td>
<td>20.56 (7.17)</td>
<td>35.61 (9.81)</td>
<td>19.84 (6.17)</td>
</tr>
<tr>
<td>Transverse plane</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>S–F</td>
<td>26.01 (6.08)</td>
<td>29.35 (9.52)</td>
<td>25.31 (4.17)</td>
</tr>
<tr>
<td>T–S</td>
<td>30.59 (6.82)</td>
<td>36.63 (5.38)</td>
<td>32.62 (4.39)</td>
</tr>
<tr>
<td>P–T</td>
<td>32.43 (5.30)</td>
<td>43.76 (4.04)</td>
<td>34.05 (6.15)</td>
</tr>
<tr>
<td>Frontal plane</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>S–F</td>
<td>33.81 (6.45)</td>
<td>38.87 (5.70)</td>
<td>34.65 (5.64)</td>
</tr>
<tr>
<td>T–S</td>
<td>24.58 (4.32)</td>
<td>35.78 (5.78)</td>
<td>25.39 (4.27)</td>
</tr>
<tr>
<td>P–T</td>
<td>21.09 (7.10)</td>
<td>41.95 (7.23)</td>
<td>20.93 (4.39)</td>
</tr>
</tbody>
</table>

Note: ANOVA results are summarised and the bolded values indicate a significant difference at the adjusted \( p \), 0.033.

Table 2. Summary mean (SDs) of the maximum finite-time Lyapunov exponents for the foot, shank, thigh and pelvis across all participants and conditions.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Unassisted</th>
<th>Assisted</th>
<th>ANOVA results p-values</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Anterior</td>
<td>Posterior</td>
<td>Anterior</td>
</tr>
<tr>
<td>Sagittal plane</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Foot</td>
<td>2.216 (0.214)</td>
<td>2.435 (0.292)</td>
<td>2.372 (0.198)</td>
</tr>
<tr>
<td>Shank</td>
<td>2.356 (0.187)</td>
<td>2.519 (0.235)</td>
<td>2.417 (0.189)</td>
</tr>
<tr>
<td>Thigh</td>
<td>2.074 (0.247)</td>
<td>2.346 (0.255)</td>
<td>2.200 (0.268)</td>
</tr>
<tr>
<td>Pelvis</td>
<td>1.963 (0.301)</td>
<td>2.269 (0.347)</td>
<td>2.083 (0.221)</td>
</tr>
<tr>
<td>Transverse plane</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Foot</td>
<td>1.791 (0.200)</td>
<td>1.929 (0.268)</td>
<td>1.913 (0.128)</td>
</tr>
<tr>
<td>Shank</td>
<td>1.839 (0.218)</td>
<td>1.985 (0.212)</td>
<td>1.856 (0.186)</td>
</tr>
<tr>
<td>Thigh</td>
<td>1.764 (0.231)</td>
<td>1.774 (0.159)</td>
<td>1.726 (0.203)</td>
</tr>
<tr>
<td>Pelvis</td>
<td>1.689 (0.332)</td>
<td>2.042 (0.231)</td>
<td>1.791 (0.292)</td>
</tr>
<tr>
<td>Frontal plane</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Foot</td>
<td>1.939 (0.133)</td>
<td>2.069 (0.130)</td>
<td>1.963 (0.154)</td>
</tr>
<tr>
<td>Shank</td>
<td>1.958 (0.215)</td>
<td>2.044 (0.171)</td>
<td>1.887 (0.232)</td>
</tr>
<tr>
<td>Thigh</td>
<td>1.805 (0.169)</td>
<td>1.907 (0.228)</td>
<td>1.887 (0.158)</td>
</tr>
<tr>
<td>Pelvis</td>
<td>1.679 (0.209)</td>
<td>1.975 (0.215)</td>
<td>1.756 (0.171)</td>
</tr>
</tbody>
</table>

Note: ANOVA results are summarised and the bolded values indicate a significant difference at the adjusted \( p \), 0.029.
Figure 5 displays the local dynamic stability ($\lambda_{\text{max}}$) values for all five gait conditions. In all cases but one (UAC for the pelvis in the transverse plane), mean $\lambda_{\text{max}}$ values were lower during UG than during any load carriage condition. In the sagittal plane, there was a significant main effect of condition on all segments except for the shank, significant pairwise differences between UG and all conditions except UAC for the foot and thigh, and significant pairwise differences between UG and both PC techniques for the shank and pelvis ($p < 0.05$). In the transverse plane, there was a significant main effect of condition on the foot and shank, significant pairwise differences between UG and all conditions except UAC for the foot, significant pairwise differences between UG and both PC techniques for the shank and significant pairwise differences between UG and UPC for the thigh and pelvis ($p < 0.05$). In the frontal plane there was a significant main effect of condition on all segments ($p < 0.05$). In terms of pairwise comparisons, UG was significantly more stable than all other conditions for the foot, more stable than all conditions except AAC for the shank, more stable than all conditions except APC for the thigh and more stable than all conditions except UAC for the pelvis ($p < 0.05$).

4. Discussion
The goal of this study was to expand on previous research by applying two dynamical systems analysis techniques to comprehensively evaluate changes in gait stability when carrying loads in the hands anteriorly and posteriorly, with and without the MAD. A secondary goal was to compare stability during load carriage to stability during UG. It was hypothesised that (1) coordination variability would be higher and dynamic stability lower during the PC task, (2)
coordination variability and stability would be similar with and without use of the MAD and (3) all load carriage tasks would have higher coordination variability and be less dynamically stable than UG.

In agreement with our first hypothesis, while performing the AC technique, there was a significant decrease in coordination variability between all three segment pairings in the sagittal plane, between the pelvis and thigh pairing in the transverse plane, and between the pelvis and thigh and between thigh and shank pairings in the frontal plane, when compared with the PC technique. This result corroborates our previous research where PC resulted in significantly greater transverse plane coordination variability between the load being carried and the trunk (Smallman, Graham, and Stevenson 2013). The present local dynamic stability results were also in agreement with our first hypothesis, where the AC technique provided significantly increased stability of the thigh in the sagittal plane, the shank in the transverse plane and the pelvis in the frontal plane compared with the PC technique. As a whole, it appears that AC is more stable than PC, at least in our population of novice materials handlers.

One possible adaptive reason for the decrease in coordination variability during AC is that there is a decrease in visibility of the ground due to load in the hands. Kinematics (step width, toe clearance, etc.) are often altered as a safety mechanism when visibility is limited during obstacle crossing and load carriage (Perry et al. 2010). Since PC allows for greater visibility of the ground while carrying the load (Kudryk 2008), it is possible that gait stability is increased during AC as an inherent safety precaution. It may not be as necessary to have as precise control over the lower extremities during the PC technique because of the improved visibility. Another possible reason is the lack of familiarity of the participants with the PC technique, which is discussed further below.

Figure 5. Maximum finite-time Lyapunov exponents for all segments in all directions of motion during the five carriage conditions: UAC, UPC, AAC, APC and UG. † represents a significant main effect of carriage type. * represents a significant difference between a given load carriage condition and UG.
Concerning use of the MAD, our findings were mostly in agreement with our hypothesis that there would be little effects of the device on either stability metric. Similar to Smallman, Graham, and Stevenson (2013), the MAD did not affect vCRP for any segment pairing and did not differentially affect AC or PC. With respect to local dynamic stability, the MAD induced a small but significant increase in stability of the shank in the frontal plane (3.55–4.96%) and resulted in several significant interactions between carriage type and device, where the MAD slightly improved stability during PC (3.56–6.60%), but slightly hindered stability during AC (1.24–7.67%). Thus, when combining the present results with the fact that the MAD (1) decreases upper limb muscular effort when carrying boxes (Kudryk 2008), (2) improves perceived comfort (Kudryk 2008; Smallman, Graham, and Stevenson 2013) and (3) results in more in-phase coordination between the trunk and pelvis (Smallman, Graham, and Stevenson 2013), it appears to be an effective, acceptable and safe ergonomic aid for workers to use in the moving and other load carriage industries.

When looking at the combination of the vCRP and the Lyapunov exponent results as a whole, assuming that lower vCRP is indicative of greater dynamic stability, UG is more stable than load carriage. These findings agree with previous research (e.g. Liu and Lockhart 2013; Qu 2013), further displaying the importance of studying gait stability during load carriage as a major slip, trip and fall risk factor, since greater dynamic instability can predict an increased risk of falling (Bruijn et al. 2012; Su and Dingwell 2007). Load carriage gait necessitates the continuous conversion between potential and kinetic energy, and the increase in energy expenditure required by the increased load in the hands may be a major contributing factor to the reduction in stability during load carriage (Qu 2013). Here we have taken a major first step in exploring which load carriage techniques most closely replicate UG, which may be very important when recommending safe load carriage practices in industry. At this preliminary point, it appears that AC appears to be the safest from a stability point of view, although further study is needed to confirm this.

A primary limitation in our study was the use of a treadmill in comparison with overground walking. Since gait speed can affect walking stability (Bruijn et al. 2009b), a treadmill was used to maintain a consistent treadmill speed across all load carriage conditions. However, stability can be altered between treadmill and overground walking (Dingwell et al. 2001). Thus, while the findings from this study make an initial comparison of stability across technique and device conditions, results may not be entirely transferable to movers while working in the field. Another limitation of our study was the use of all novice movers, due to the lack of availability of experienced volunteers. The use of novice movers may have affected movement patterns across techniques since they were unfamiliar with the PC technique prior to the study. Experienced movers prefer to use the PC technique (Kudryk 2008), and previous research comparing novice and expert handlers has shown that expert handlers tend to adopt safer handling techniques (Authier, Lortie, and Gagnon 1996). Therefore, while this study provides an initial comparison between techniques, future studies should incorporate overground walking and experienced movers, in order to make these findings completely transferable to the field. A final limitation was the relatively short duration of load carriage under each condition (55 cycles), which was chosen to avoid fatigue across the various conditions.

To the best of the authors’ knowledge, this investigation is the first to simultaneously quantify differences in intersegment coordination variability and dynamic stability during anterior and posterior hand-held load carriage. Findings indicate that the AC technique is more stable than the PC technique, the MAD has small but beneficial effects on overall stability and UG appears to be more stable than all methods of load carriage. Future research should continue to explore safe load carriage practices both in the laboratory and in the field.

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References


