Effect of stable and unstable load carriage on walking gait variability, dynamic stability and muscle activity of older adults.

Abstract

Load carriage perturbs the neuromuscular system, which can be impaired due to ageing. The ability to counteract perturbations is an indicator of neuromuscular function but if the response is insufficient the risk of falls will increase. However, it is unknown how load carriage affects older adults. Fourteen older adults (65±6 years) attended a single visit during which they performed 4 minutes of walking in 3 conditions, unloaded, stable backpack load and unstable backpack load. During each walking trial, 3-dimensional kinematics of the lower limb and trunk movements and electromyographic activity of 6 lower limb muscles were recorded. The local dynamic stability (local divergence exponents), joint angle variability and spatio-temporal variability were determined along with muscle activation magnitudes. Medio-lateral dynamic stability was lower ($p=0.018$) and step width ($p=0.019$) and step width variability ($p=0.015$) were greater in unstable load walking and step width variability was greater in stable load walking ($p=0.009$) compared to unloaded walking. However, there was no effect on joint angle variability. Unstable load carriage increased activity of the Rectus Femoris ($p=0.001$) and Soleus ($p=0.043$) and stable load carriage increased Rectus Femoris activity ($p=0.006$). These results suggest that loaded walking alters the gait of older adults and that unstable load carriage reduces dynamic stability compared to unloaded walking. This can potentially increase the risk of falls, but also offers the potential to use unstable loads as part of fall prevention programmes.

Keywords

Older adults; walking; load carriage; dynamic stability; variability

Introduction

Falls are one of the leading causes of injury and hospital admission (Ambrose et al., 2013), with most falls in older adults occur during walking or other dynamic tasks (Pizzigalli et al., 2011). Age related changes in gait are the result of a number of factors including loss of muscle strength, neuromuscular function (Dingwell et al., 2017; Kang and Dingwell, 2008a) and range of motion (Kang and Dingwell, 2008a, 2008b), fear of falling (Maki, 1997) and reduced certainty when selecting kinematic gait patterns (Kurz and Stergiou, 2003). Studies have linked the loss of stability and an increase in variability of gait, particularly in the medio-lateral direction, to the risk of falling in older adults (Maki, 1997) and retrospectively differentiated fallers and non-fallers (Toebes et al., 2012). Stability during gait can be affected by walking speed (Callisaya et al., 2012), fatigue (Thomas et al., 2013), perturbations (Oliveira et al., 2012) and load carriage (Kim et al., 2014; Kubinski and Higginson, 2012; McGowan et al., 2009).

Ageing results in a decline in neuromuscular function including motor neuron death, decreased corticospinal excitability, impaired somatosensory function and deterioration of the neuromuscular junction (Gonzalez-Freire et al., 2014; Manini et al., 2013; Shaffer and Harrison, 2007). This contributes to a decrease in the ratio of muscle strength to mass (Delmonico et al., 2009; Fragala et al., 2015) and neuromuscular noise is increased (Dingwell et al., 2017; Roos and Dingwell, 2010) which can lead to errors or inaccuracies in the desired movements. Additional load carriage alters the ratio of muscle strength to the mass that must be moved and controlled requiring greater activation of anti-gravity and propulsive muscles and the postural control system to prevent a loss of stability (Arellano et al., 2009). Greater levels of muscle activation result in greater neuromuscular noise in older adults (Singh et al., 2012), therefore loaded walking may increase neuromuscular noise when walking. Arguably, stability is therefore affected more in older adults compared with young adults when walking with additional loads. .

During loaded walking, young adults show an increased spatio-temporal gait variability, double support time, decreased step length (Dames and Smith, 2015; Demura and Demura, 2010; Huang and Kuo, 2014; Qu and Yeo, 2011) and local dynamic stability in the anterior-posterior (Liu and Lockhart, 2013), medio-lateral, and vertical directions (Liu and Lockhart, 2013; Qu, 2013). Older adults have demonstrated a similar adaptation in spatio-temporal gait variables in loaded conditions with increases in double support time and step width (Kubinski and Higginson, 2012). However, it is unknown whether local dynamic stability is affected by load carriage in older adults.

The ability to counteract perturbations and maintain stability is a good indicator of the health of neuromuscular and motor control functions (Hur et al., 2010; Mersmann et al., 2013; Oliveira et al., 2012). Previous research has mainly focused on load carriage of solid, stable items to induce a perturbation. However, the use of a liquid, unstable load would add an additional challenge as individuals must not only support the additional load and produce sufficient propulsive forces, but also actively control and correct perturbations from the unstable load. An unstable load carried on the trunk may magnify the small natural perturbations that occur during gait which must be controlled to prevent a loss of stability that could eventually lead to a fall. Therefore, unstable load carriage could give a greater insight to the neuromuscular control strategies adopted by older adults when normal gait is perturbed than a stable load alone.

The aim of the present study was to investigate how carriage of stable and unstable loads alters the control of older adults gait using measures of dynamic stability, variability and muscle activation. It was hypothesised that both stable and unstable load carriage would decrease dynamic stability, and increase gait variability and lower limb muscle activation compared to unloaded walking. Furthermore, it was hypothesised that unstable load carriage would have a greater effect on gait dynamic stability, variability and muscle activations compared to stable load carriage.

Methods

Participants

Fourteen older adults (n females: 7, n males: 7, age: 65±6 years, height: 1.70±0.10 m, mass: 74±13 kg) volunteered to participate in this study. Participants were excluded if they suffered from neurological conditions such as stroke, Parkinson's disease or dementia. Exclusion criteria also included visual impairment or lower limb conditions that prevented unaided walking. The study received ethical approval from the University research ethics committee. All participants gave written informed consent, were aware of the nature of the study and their right to withdraw at any time.

Procedures

All participants attended a single laboratory visit during which they performed 4 minutes of treadmill walking at their unloaded self-selected walking speed (mean speed: 1.2±0.12 m/s) under 3 conditions, unloaded, with a stable load, and an unstable load. Prior to commencing measurements participants were familiarised with the treadmill walking. Participants walked for 5 minutes on a motorised treadmill to warm up and determine their self-selected comfortable walking speed, which was achieved by participants manually adjusting the treadmill speed until they reached the speed they deemed to be their normal comfortable walking speed. As walking speed has been demonstrated to alter dynamic stability (England and Granata, 2007) and muscle activations (Schmitz et al., 2009) each participants unloaded self-selected speed was used for each load condition to control for effects caused by differences in walking speed.

Both the stable and unstable loads were carried using a backpack with a chest strap and were equivalent to 15% of the participants' body mass (BM), to the nearest 0.1 kg. In each condition 3 water-tight containers, with a volume of 3.6 litres each, were placed inside the backpack (Figure 1). For the stable load, steel weights in denominations of 0.1, 0.5 and 1 kg, were secured inside to prevent movement, and were evenly distributed between the 3 containers. To form the unstable load a volume of water equivalent to a mass of 7.5% of the participants BM was distributed evenly between the 3 containers and steel weights were then added to make up the total mass of the backpack to 15% of the participants BM.

[Figure 1 here]

Participants were fitted with reusable bipolar electrodes with a 2 cm inter-electrode distance (SX230-1000, Biometrics Ltd, UK) to measure the electromyographic (EMG) activity of 6 muscles of the left leg, including the Rectus Femoris (RF), Vastus Medialis (VM), Biceps Femoris (BF), Tibialis Anterior (TA), Gastrocnemius Medialis (GM), and Soleus (SOL) and a reference electrode placed over the left radial head. Specific electrode placements are outlined in Table 1. Prior to the placement of electrodes, the skin was prepared by shaving the area and cleaning with an alcohol wipe. The reusable electrodes were attached to an 8-channel amplifier (range: ±4mV, gain: 1000, impedance: 1MΩ - K800, Biometrics Ltd, UK) before being A/D converted (CA-1000, National Instruments Corp., UK).

[Table 1 here]

Participants were also fitted with retro-reflective markers (diameter: 15 mm) for the measurement of three-dimensional (3D) kinematics of the lower limb, and movements of the trunk. Marker movements in 3D space were recorded using an 8 camera MAC-Eagle motion analysis system (Motion Analysis Corp., USA). Markers were placed on locations based on the modified Helen Hayes marker set (Kadaba et al., 1990) and included a single marker on the sacrum, and markers placed bilaterally over the anterior superior iliac spine, and unilaterally on the left thigh, medial knee epicondyle, lateral knee epicondyle, shank, medial ankle malleolus, lateral ankle malleolus, heel, and base of the great toe (heel and toe markers were placed on the outside of the shoe). In addition, a cluster of 3 markers were placed at the top of the sternum to measure the motions of the trunk (Bruijn et al., 2009a; Qu, 2013). Before commencing measurements, marker positions were recorded with participants stood in the anatomical position to provide reference angles for the hip, knee and ankle joints.

All EMG and 3D kinematic measurements were synchronised and collected for 3 minutes at sampling frequencies of 1000 Hz and 50 Hz respectively using Cortex software (Motion Analysis Corp., USA), from minutes 2-4 of each trial. The first minute of each trial was used to allow participants to adjust the treadmill walking before commencing measurements. The order in which each condition was presented was counterbalanced and randomised across participants to prevent any order effects. Two minutes of rest were provided between each condition.

Gait Variability

The 3D positions of each marker on the lower body were filtered using a dual-pass 2nd order Butterworth filter with a cut-off frequency of 10 Hz before joint angles were calculated. Three-dimensional joint angles of the hip, knee and ankle joints were calculated using the Cardan flexion-abduction-internal rotation sequence of rotations. Sagittal, frontal and transverse plane joint rotations were calculated with respect to the angle of each joint whilst standing in the anatomical position. All joint kinematics were calculated using Cortex software (Motion Analysis Corp., USA). The minimum vertical position of the marker attached to the heel was used to identify heel-strike gait events (Hreljac and Marshall, 2000; Zeni et al., 2008). The heel-strike events were used to separate individual gait cycles, defined as the period from one heel-strike to the next ipsilateral heel-strike.

The spatio-temporal variables calculated included the stride time (ST) and step width (SW). The ST was calculated as the time from one heel-strike to the next ipsilateral heel-strike and SW was calculated as the medio-lateral distance between the positions of the heel marker at heel-strike to that of the next contra-lateral heel-strike. The mean (ST_{MEAN} and SW_{MEAN}) and standard deviation (ST_{SD} and SW_{SD}) were calculated for ST and SW. To quantify the kinematic variability of the hip, knee and ankle in the sagittal, frontal and transverse planes during walking, data for each individual gait cycle were interpolated to 101 data points (0-100%). The standard deviation was then calculated across all gait cycles at each normalised time point. The mean of the standard deviation values (MeanSD) calculated for each normalised time point was then used to represent the kinematic variability for each joint in each plane.

Dynamic Stability

Dynamic stability was calculated as the local divergence exponent (LDE) from the trunk markers in the anterio-posterior (LDE_{AP}), medio-lateral (LDE_{ML}) and vertical (LDE_{VT}) directions using the Rosenstein algorithm (Rosenstein et al., 1993). For the calculation of the LDE, the average position of the 3 markers attached to the sternum for each frame in the anterior-posterior, medio-lateral and vertical directions was used. The application of this method to gait has been described in detail previously (e.g. Bruijn et al., 2009; Dingwell et al., 2001).

Briefly, as accurate calculation of the LDE requires stationary data the first difference of consecutive samples of each averaged trajectory was calculated. To achieve statistical precision, 150 consecutive strides were analysed (Bruijn et al., 2009a). The first differenced signal for each direction over the period of 150 strides was interpolated to 15000 data points. A state space for each direction was constructed using a time delay of 10 samples and embedding dimension of 5 (e.g. Bruijn et al., 2009b; England and Granata, 2007; Liu and Lockhart, 2013). The nearest neighbour (points separated by the smallest Euclidean distance) for each data point in state space was determined and the Euclidean distance of these points was followed for the length of the series creating as many distance-time series as time points in state space. The divergence curve was calculated as the log of the average of all distance-time series and the LDE was calculated as the slope of the linear fit applied to the period equivalent to the average time for 1 step in each condition. The LDE was calculated for the period of 0.5 strides as each step presents an opportunity to correct a perturbation.

Muscle Activations

Processing of all EMG signals was performed using custom programmes written in Matlab software (Mathworks Inc., USA). Raw EMG signals were bandpass filtered using a dual-pass 2nd order Butterworth filter with a 20-450 Hz cut-off frequency before subtracting the signal mean to correct baseline offsets. The bandpass filtered signal was full-wave rectified and low-pass filtered to produce a linear envelope using a dual-pass 2nd order Butterworth filter with a 10 Hz cut-off frequency. The linear envelope was then normalised as a percentage of peak activation of the muscle recorded during unloaded self-selected speed walking. The normalised signals were then separated into individual gait cycles based on the heel-strike events determined by the heel marker and were interpolated to 1001 data points. The EMG activity was then averaged across all gait cycles before the mean EMG activity (EMG_{MEAN}) of the average gait cycle was calculated.

Statistics

All data were tested for normality using the Shapiro-Wilk test and were normally distributed. When data violated the assumption of sphericity a Greenhouse-Geisser correction was used. To determine the effects of load conditions (unloaded, stable and unstable) on gait variability (ST_{MEAN}, ST_{SD}, SW_{MEAN} and SW_{SD}, hip, knee and ankle MeanSD), dynamic stability (LDE_{AP}, LDE_{ML} and LDE_{VT}) and muscle activations (EMG_{MEAN} of all muscles) repeated measures ANOVAs were performed. When significant main effects were present post hoc pairwise comparisons with a Bonferonni correction were performed. The α-level of significance was set at $p<0.05$ for all comparisons. Partial eta squared (η_p^2) was used as an estimate of effect size, values of 0.01, 0.06 and 0.14 were interpreted as small, medium and large effects respectively (Cohen, 1969; Richardson, 2011). All statistical analyses were performed using SPSS software (v22, IBM UK Ltd., UK).

Results

Gait Variability

An effect of load condition was present for SW_{MEAN} ($F(2,26)=5.68$, $p=0.009$, $\eta_p^2=0.30$) and SW_{SD} ($F(2,26)=8.53$, $p=0.001$, $\eta_p^2=0.40$). Unstable load walking induced a significantly higher SW_{MEAN} ($p=0.019$) and SW_{SD} ($p=0.015$) compared with unloaded walking. In addition, stable load walking induced a significantly higher SW_{SD} compared with unloaded walking ($p=0.009$). There were no differences between stable and unstable loaded walking. There were no effects for ST_{MEAN} or ST_{SD} (η_p^2: 0.05 and 0.10 respectively). There were also no effects of load condition on the MeanSD of the hip (sagittal: $\eta_p^2=0.12$, frontal: $\eta_p^2=0.06$ and transverse: $\eta_p^2=0.10$), knee (sagittal: $\eta_p^2=0.10$, frontal: $\eta_p^2=0.01$ and transverse: $\eta_p^2=0.06$) and ankle (sagittal: $\eta_p^2=0.05$, frontal: $\eta_p^2=0.02$ and transverse: $\eta_p^2=0.09$) joints in any rotation plane (Table 2).

[Table 2 here]

Dynamic Stability

An effect of load condition was present for LDE_{ML} ($F(2,26)=7.02$, $p=0.004$, $\eta_p^2=0.35$) with a significantly higher LDE_{ML} for unstable load walking compared with unloaded walking ($p=0.018$), however, stable load walking was not different to either condition (Figure 2). There were no effects for LDE_{AP} and LDE_{VT} (η_p^2: 0.11 and 0.15 respectively).

[Figure 2 here]

Muscle Activation

An effect of load condition was present for EMG_{MEAN} of RF ($F(2,26)=8.96$, $p=0.001$, $\eta_p^2=0.41$) and SOL ($F(1.43,15.89)=5.851$, $p=0.023$, $\eta_p^2=0.310$), both muscles activation were higher for unstable load walking compared with unloaded walking (RF: $p=0.001$ and SOL: $p=0.043$) and RF also increased ($p=0.006$) between unloaded and stable load walking (Figure 3).

There were no effects of load condition for VM, GM or BF (η_p^2: 0.15, 0.16 and 0.13 respectively).

[Figure 3 here]

Discussion

The main findings of this study were that the ML dynamic stability of older adults was reduced when carrying unstable loads compared to unloaded walking. Step width variability was also increased in both loaded conditions compared to unloaded walking and step width was increased when carrying an unstable load compared to unloaded walking. However, joint angle variability was not altered by load carriage. Furthermore, it was found that RF and SOL muscle activation was increased in loaded walking conditions. Combined, these results show that load carriage effects the gait of older adults and that unstable loads have effects on dynamical stability compared to unloaded walking that are not present for stable loads, however this study did not find differences between stable and unstable load carriage.

The present study is the first to demonstrate the effect of unstable load carriage on the dynamic stability of older adults. The increased LDE_{ML} when carrying an unstable load, in the present study, is in agreement with findings in young adults when carrying heavier stable loads than those used in the current study (Liu and Lockhart, 2013; Qu, 2013). In addition to accommodating the added inertia, the unstable load required older adults to attenuate movements of the load, which magnified the natural kinematic perturbations that occur during walking (Dingwell and Marin, 2006). However, a reduction in ML dynamic stability was not present in the stable condition, in contrast with previous findings (Liu and Lockhart, 2013; Qu, 2013). A likely explanation is the relatively lower loads used in the present study for older adults, compared to the young population carrying greater loads. It is suggested that the added perturbation caused by unstable loads was responsible for the decline in stability rather than the added inertia of a load equivalent to 15% BM.

The increased SW_{MEAN}, SW_{SD} and LDE_{ML} with unstable loads compared to unloaded walking found in the present study suggest that the control of ML stability is reduced, but not the control of AP stability. A possible explanation is that humans are mechanically less stable in the ML than the AP direction when walking (Bauby and Kuo, 2000; Rankin et al., 2014; Schrager et al., 2009). It has been demonstrated that in the AP direction an individual is able to rely on passive dynamic properties with limited need for neural feedback control for stability during walking, however, in the ML direction active control is necessary (Bauby and Kuo, 2000; Rankin et al., 2014). An alternative explanation is that the orientation of the unstable load configuration, with the long axis oriented in the ML direction, will result in greater movements of the load in the ML compared to AP and VT directions. The load configuration used will therefore provide greater perturbation in the ML direction than the AP or VT directions.

In loaded conditions, greater muscle output is required, as indicated by the greater RF and SOL muscle activation in the present study. It has been demonstrated that the role of the SOL and RF during gait is different compared to GM and VM, with the SOL contributing more to resisting gravity and forward propulsion than GM (Cronin et al., 2013). It is therefore reasonable to assume that the SOL would contribute more than GM to resist the added load. The role of the RF as a biarticular muscle is to transfer mechanical energy from the hip to knee (Annaswamy et al., 1999), which could lead to a different response in loaded conditions to that of VM. It is also possible to assume that a larger sample size would result in a significant alteration in VM, GM and BF activation given the medium-large effect sizes present (η_p^2: 0.15, 0.16 and 0.13 respectively).

Despite the changes to SW in both loaded conditions and LDE_{ML} when carrying an unstable load there was no change in the MeanSD of any joint or plane of motion. The effect of load carriage on joint kinematic variability has not been demonstrated previously, however, it has been demonstrated that load carriage of 30% BM did not have an effect on sagittal plane joint local dynamic stability (Arellano et al., 2009) and range of motion (Browning et al., 2007; Holt et al., 2003) during walking in young adults. The findings suggest that joint level variability may be more rigidly controlled when walking on a treadmill than trunk stability or step width (Arellano et al., 2009).

Older adult fallers have lower dynamic stability, i.e. larger LDE values, and greater gait variability in the ML direction than age matched non-fallers (Maki, 1997; Toebes et al., 2012). Walking with an unstable load could recreate conditions of increased fall risk in healthy older adults that are found in those with a higher risk of falling, but can be performed in a controlled environment. Consequently, there could be positive effects of training with unstable loads. Future research should therefore focus on the safety and effect of unstable load walking as part of an intervention to reduce falls in healthy older adults.

There were some limitations of the current study. The use of a treadmill limits the external validity of the findings and may also impact upon the natural variability and dynamics of walking as speed is consistent, as is the support surface and position on the treadmill (Kang and Dingwell, 2008b). However, use of a treadmill provides the possibility to analyse a large number of continuous strides that would not be possible during overground walking. The analysis of continuous gait is important for measures of kinematic variability and dynamic stability (Bruijn et al., 2009a; Dingwell and Marin, 2006) and so was accepted for the advantages gained in understanding the dynamics of continuous gait. Another possible limitation is that the speed was the same for each condition. Whilst using the same speed provides consistency between conditions, in reality individuals decrease their walking speed under loaded conditions (Salem et al., 2001).

In conclusion, the findings of this study suggest that in healthy, active older adults load carriage of 15% BM increases step width variability and activation of antigravity and propulsive muscles in the lower limb. In addition, unstable loads decrease ML dynamic stability compared to unloaded walking, a change that is not present when carrying stable loads. However, neither loaded condition altered the variability of hip, knee and ankle kinematics.

Conflicts of Interest

None.

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Tables

Table 1. Electrode placements for the 6 lower limb muscles studied.

Muscle	Electrode position
Rectus Femoris	50% along the line from the anterior superior iliac spine to the superior border of the patella
Vastus Medialis	80% along the line between the anterior superior iliac spine and the joint space in front of the anterior border of the medial ligament
Biceps Femoris	50% along the line between the ischial tuberosity and the lateral epicondyle of the tibia
Tibialis Anterior	33% along the line between the tip of the fibula and the tip of the medial malleolus
Gastrocnemius Medialis	Most prominent bulge of the muscle
Soleus	66% along the line between the medial epicondyle of the femur and the medial malleolus

Table 2. Mean ± standard deviation (SD) values for all spatio-temporal and joint angle gait variability (MeanSD) variables under each load condition.

		Unloaded	Stable	Unstable
Step Width (mm)	Mean	73±34	88±24	97±20*
	SD	22±6	27±5*	31±6*
Stride Time (s)	Mean	1.07±0.09	1.07±0.06	1.08±0.08
	SD	0.04±0.05	0.02±0.01	0.02±0.01
Hip MeanSD (°)	Sagittal	2.7±1.9	4.0±2.3	3.0±2.7
	Frontal	1.8±1.5	1.6±0.6	2.0±1.8
	Transverse	3.4±2.9	4.1±2.8	4.5±5.9
Knee MeanSD (°)	Sagittal	3.2±1.7	4.5±3.3	4.3±3.7
	Frontal	1.8±2.1	1.5±1.5	3.8±2.1
	Transverse	2.3±2.2	4.0±3.2	6.1±4.6
Ankle MeanSD (°)	Sagittal	2.0±0.9	2.7±1.7	6.7±4.4
	Frontal	2.0±2.2	1.9±1.8	3.1±3.1
	Transverse	2.3±1.9	1.9±1.1	4.8±3.3

* indicates that the value is significantly greater than the unloaded condition

Figure Captions

Figure 1. Illustration of the position of containers inside the backpack. Each container held either steel weights for the stable condition or steel weights and water for the unstable condition, distributed evenly between the 3 containers.

Figure 2. Mean ± standard deviation values for local divergence exponent (LDE) values in the anterio-posterior (LDE_{AP}), medio-lateral (LDE_{ML}) and vertical (LDE_{VT}) directions under each load condition.

Figure 3. Mean ± standard deviation values for the average muscle activity (EMG_{MEAN}) of all tested muscles and the coactivation index (CI) of all tested muscle pairs under each load condition.

* indicates value is significantly greater than the unloaded condition