THE DETERMINANTS OF INDIVIDUAL LOAD CARRIAGE ECONOMY

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I confirm that the work submitted is my own, except where work has formed part of jointly authored publications has been included. The contribution of the candidate and the other authors to this work has been explicitly indicated below. I confirm that the appropriate credit has been given within the thesis where reference has been made to the work of others.

Chapter 4

> **Hudson, S.**, Cooke, C. and Lloyd, R., 2017. The reliability of the Extra Load Index as a measure of relative load carriage economy. *Ergonomics*, *60*(9), pp.1250-1254.
>
> I was responsible for the conception and design of the study, conducted all experiments, processed and analysed the data, interpreted the results and wrote and edited the manuscript. Professor Carlton Cooke and Professor Ray Lloyd assisted with interpreting the results and edited and approved the manuscript.

Chapter 5

> **Hudson, S.,** Cooke, C., Davies, S., West, S., Gamieldien, R., Low, C., & Lloyd, R. (2018). A comparison of economy and sagittal plane trunk movements among back-, back/front- and head-loading. *Ergonomics*, *61*(9), 1216-1222.
>
> I completed a secondary analysis of data collected prior to the start of my PhD, which involved processing and analysing the data, interpreting the results and writing the manuscript. Professor Ray Lloyd was responsible for conception and design of the study. Professor Ray Lloyd, Professor Simeon Davies, Dr. Sacha West and Mr. Raeeq Gamieldien conducted the experiments. Professor Ray Lloyd, Professor Carlton Cooke and Dr. Chris Low assisted with interpreting the results. All authors edited and approved the manuscript.

Hudson, S., Cooke, C., Davies, S., West, S., Gamieldien, R., Low, C., & Lloyd, R. (2020). Inter-individual variability in load carriage economy and comparisons between different load conditions. *Applied Ergonomics*, *82*, 102968.

This publication was also part of the secondary analysis of data collected prior to the start of my PhD. The author contributions were the same as stated for the first publication from Chapter 5.

Chapter 7

Hudson, S. Low, C. Cooke, C. Vanwanseele, B. and Lloyd, R., (2020), The effects of step width control on load carriage economy. *Proceedings of the 38th International Conference of Biomechanics in Sports.* Liverpool, UK: International Society of Biomechanics in Sports

I was responsible for the conception and design of the study, conducted all experiments, processed and analysed the data, interpreted the results and wrote and edited the conference paper. Professor Benedicte Vanwanseele facilitated data collection at KU Leuven. Professor Carlton Cooke, Professor Ray Lloyd and Dr. Chris Low assisted with interpreting the results. All authors edited and approved the paper.

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Abstract

Energy saving phenomena have been identified for load carriage on the head, the back, and evenly distributed between the back and front of the torso (back/front-loading), but the mechanics explaining these phenomena are unknown. This research aimed to identify the determinants of individual load carriage economy. Three empirical studies and the development of a theoretical deterministic model (TDM) are presented. Study 1 showed that the Extra Load Index (ELI), a measure of relative load carriage economy, and loaded walking gait kinematics have good test-retest reliability with 7 and 20 kg (e.g. largest coefficients of variation (CV) = 4.17%). Study 2 showed that there is no significant difference in ELI for head-, back- and back/front-loading across a range of load mass (3 – 20 kg) for experienced head-loaders. However, there were significant differences in gait kinematics between methods. For example, forward lean increased from 3 to 20 kg for back- (10.7°) and back/front-loading (2.4°) but decreased for head-loading (-2.2°). Study 2 also supported the existence of considerable inter-individual variation for both ELI (e.g. CV of up to 16%) and load carriage kinematics (e.g. change in forward lean from unloaded walking of +24% to –8% for back-loading with 20 kg). The TDM provides a framework to analyse the biomechanics of load carriage, as in study 3. Study 3 showed that a combination of reduced trunk movement and stride pattern perturbations from unloaded walking are associated with an improved economy for some load conditions (back/front-loading with 20 kg and head-loading with 12 kg), however this finding was not consistent across all load method and masses. In conclusion, a loaded walking gait closer to that of unloaded walking is beneficial for some load carriage conditions and may be an important determinant of load carriage economy. However, this does not explain individual load carriage economy variability.

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List of Abbreviations

Abbreviation	Definition
ANOVA	Analysis of variance
ASIS	Anterior superior iliac spine
B/F	Back and front
BF	Breathing frequency
BM	Body mass
BMI	Body mass index
COM	Centre of mass
CV	Coefficients of variation
Cw	Energy cost of walking per unit distance
ELI	Extra load index
GCS	Global coordinate system
GRF	Ground reaction force
HJC	Hip joint centre
HS	Heel strike
I	Inertia
ICC	Intraclass correlation coefficients
IK	Inverse kinematics
LCS	Local coordinate system
LHJC	Left hip joint centre
LoA	Limits of agreement
m	mass
MLM	Multi-level model
mlO2L	Oxygen consumption when carrying additional load
mlO2U	Oxygen consumption when walking unloaded
PSIS	Posterior superior iliac spine
RER	Respiratory exchange ratio
RHJC	Right hip joint centre
ROM	Range of motion
RPE	Rating of perceived exertion
SD	Standard deviation
SEM	Standard error of measurement

DOF	Degrees of freedom
TEM	Technical error of measurement
TO	Toe off
TV	Tidal volume
VAS	Visual analogue scale
$\dot{V}CO_2$	Rate of carbon dioxide production
V_E	Minute ventilation
$\dot{V}O_2$	Rate of oxygen consumption

Chapter 1. Introduction

The requirement for humans to manually carry external load remains prevalent throughout the modern world. It is an occupational necessity for personnel in the military and emergency services (Knapik et al., 1996). It is a daily activity for school children (Singh and Koh, 2009, Motmans et al., 2006) and individual's living in rural areas of developing countries where transport infrastructure is poor (Lloyd et al., 2010d). It is also a convenient way of transporting external load for recreational outdoor pursuits such as hiking and mountaineering (Lobb, 2004). Consequently, many scientific investigations have attempted to document the physiological, biomechanical and subjective perceptual effects of load carriage. Within these studies, various methods of load carriage have been considered (Soule and Goldman, 1969, Datta and Ramanathan, 1971, Lloyd and Cooke, 2000b), with walking speed (Harman et al., 2001, Hsiang and Chang, 2002), gradient (Lloyd and Cooke, 2000b), load placement (e.g. Obusek et al., 1997) and load mass (Harman et al., 2000) often manipulated.

Research examining the physiological consequences of load carriage has predominantly focused on the associated metabolic energy cost. Carrying external load does not simply change the mass of the carrier; if this was the case then the energetic cost of load carriage would simply increase in direct proportionality with all load carriage methods. Instead, altering the method (e.g. in a backpack, in the hands or on the head) appears to alter the associated economy of energy expenditure (e.g. Datta and Ramanathan, 1971, Lloyd et al., 2010b, Soule and Goldman, 1969). As such, differences in how an external load interacts with the locomotor system, when carried in different locations on the body, is likely to explain differences in economy between load carriage methods. For the purpose of this thesis, load carriage economy is defined as the mass specific rate of oxygen consumption required to support and move a given external load at a given walking velocity, where the oxygen consumption serves as a proxy for the metabolic energy demand.

It is generally accepted that carrying a load further from the body's centre of mass (COM), such as in the hands or on the feet, results in a worse load carriage economy compared to the same load carried closer to the body's COM, such as in a backpack (Soule and Goldman, 1969, Datta and Ramanathan, 1971, Legg and Mahanty, 1985, Abe et al., 2004). The metabolic energy cost required to transport a load close to the COM of the body tends to rise in proportion to the mass of the additional load (Datta and Ramanathan, 1971, Taylor et al., 1980, Huang and Kuo, 2014). Yet, energy saving phenomena have been reported for loads carried on the head, which places the load directly above the body's COM (Maloiy et al., 1986, Charteris et al., 1989), on the back (Abe et al., 2004), and evenly distributed between the back and front of the torso (back/front-loading) (Lloyd and Cooke, 2000b) when carried at walking speeds of ~3 $km \cdot h^{-1}$. Despite attempts to identify the potential mechanisms that may contribute to the energy saving phenomena observed in these methods of loading (Jones et al., 1987, Heglund et al., 1995, Abe et al., 2004, Lloyd and Cooke, 2011), the determinants remain unclear.

Research by Lloyd et al. (2010b), Lloyd et al. (2010c) and Lloyd and Cooke (2011) highlighted a large magnitude of individual variability in load carriage economy when different methods of load carriage are compared. Despite this, all previous literature in this area has focused on mean data, with no mention of variability within or between individuals. Therefore, the determinants of the variability reported by Lloyd et al. (2010b), Lloyd et al. (2010c) and Lloyd and Cooke (2011) are, as yet, unknown. Lloyd and Cooke (2011) also identified a large level of individual variation in walking gait adaptations to load carriage. Understanding how different individuals adapt their walking gait to load carriage and how different gait adaptations can influence economy, could help to elucidate why large individual differences in load carriage economy appear to exist for different methods. To investigate the extent to which load carriage induced walking gait adaptations can influence economy, an approach that considers both loaded and loaded walking seems appropriate. Therefore, measures of relative load carriage economy that account for unloaded walking, such as the Extra Load Index (ELI) (Lloyd et al., 2010a), seem more appropriate to investigate the determinants of individual load carriage economy than measures that only consider the metabolic cost associated with loaded conditions (e.g. gross metabolic rate).

The design of load carriage systems, particularly backpacks, has evolved over time to improve the physiological, biomechanical and perceptual responses to carrying a load (Seay, 2015, Orr, 2010). For example, most modern backpacks now include internal frames, hip belts and improved shoulder straps to reduce compression under the armpits. Furthermore, methods that more evenly distribute load around the trunk, such as the doublepack, appear to be more prevalent, with research identifying potential benefits for this method over more traditional backpacks (Datta and Ramanathan, 1971, Kinoshita, 1985, Lloyd and Cooke, 2000b, Dahl et al., 2016). An improved understanding of the determining factors for individual differences in load carriage economy and walking gait patterns could help to inform the development of future load carriage systems, perhaps through an increased degree of customisation.

1.1. Aim and Objectives

The review of existing literature in Chapter 2 reveals equivocal evidence for the economy associated with head-loading, and a large level of individual variation in load carriage economy with head- and back-loading methods. The aim of this thesis was to identify the key biomechanical factor(s) that determine an individual's load carriage economy with methods that place the load close to, or in alignment with, the centre of mass of the body.

To achieve the aim, the objectives for this research project were:

1. To assess the suitability of the Extra Load Index (ELI) as a measure of relative load carriage economy.
2. To establish the extent of individual variation in load carriage economy and walking gait alterations as a consequence of load carriage, for methods that place load close the centre of mass of the body, or in vertical alignment.
3. To identify potential determinants of individual load carriage economy through the analysis of load carriage economy and alterations in walking gait characteristics caused by carrying external load.

4. To conduct cause and effect trials on the identified determinants of load carriage economy, manipulating the identified key determinants in an attempt to manipulate individual load carriage economy.

To address Objective 1, the test-retest reliability of the ELI was investigated using light (7 kg) and heavy (20 kg) loads, at slow (3 $km \cdot h^{-1}$), fast (6 $km \cdot h^{-1}$) and self-selected walking speeds (Chapter 4).

Objective 2 was addressed in two phases. Firstly, the research in Chapter 4 was designed to investigate the magnitude of day-to-day variations in the ELI and walking gait alterations as a consequence of load carriage. The day-to-day variations could then be factored into the interpretation of the magnitude of individual variation. Secondly, the research in Chapter 5 was designed to assess inter-individual differences in economy and walking gait characteristics in three common methods of load carriage (back-, back/front- and head-loading) with a range of load mass (3 – 20 kg).

In order to address Objective 3, relationships were assessed between load carriage economy and sagittal plane kinematics frequently reported in the load carriage literature (Chapter 5). Based on a lack of strong relationships between ELI and kinematic variables from the work in Chapter 5, a theoretical deterministic model for walking speed was created in Chapter 6, to use as a framework to assess the biomechanical walking gait perturbations to load carriage. The biomechanical factors identified in the deterministic model were measured in the research in Chapter 7 to assess relationships between load carriage economy and factors that might improve economy.

Objective 4 was to show causation for the factors identified as potential candidate variables for improved load carriage economy from the correlation analysis in Chapter 5. The final study (Chapter 7) was designed to allow for manipulation of a variable identified as a potential key determinant of load carriage economy. As no candidate variables for improved economy were identified in Chapter 5, objective 4 was addressed by selecting candidate variables from factors included in the deterministic model in Chapter 6.

1.2. Structure of this thesis

This introductory chapter outlines the rationale, aims and objectives for this research. The next chapter (Chapter 2) is a review of existing literature, which provides a synthesis of what is currently known about load carriage economy and its determinants, and what is unknown or equivocal. The general methods used in multiple experimental studies within this thesis are detailed in Chapter 3. Chapters 4, 5 and 7 are the three experimental studies that were conducted to achieve the aims of this this thesis, while Chapter 6 is a theoretical chapter that details the development of a deterministic model used as an analysis tool in the subsequent experimental study (Chapter 7). The findings of the three experimental studies and the theoretical model developed in Chapter 6 are brought together in a general discussion (Chapter 8), along with a reflection of the aims set out here in the introduction (Chapter 1), conclusions, limitations and directions for future research. Appendices are also included with information to support the main thesis. Each item included in the appendices is cited in the relevant chapters of this thesis.

Chapter 2. Literature Review

The purpose of this review is to evaluate literature on the metabolic economy of load carriage and its potential biomechanical determinants. This informs the subsequent experimental chapters in this thesis.

2.1. Exercise economy

Economy can be defined as the metabolic cost of exercise, measured as the rate of oxygen consumption ($\dot{V}O_2$) per kilogram per minute for a given locomotion speed and gradient (Cooke, 2013). Measurements of exercise economy are based on the calculation of oxygen consumption from expired air under steady state conditions (Poole and Jones, 2011). This can be achieved by using indirect calorimetry to measure pulmonary gas exchange between $\dot{V}O_2$ and carbon dioxide production ($\dot{V}CO_2$) during exercise (Poole and Jones, 2011, Winter et al., 2006, Eston and Reilly, 2013). At a constant sub-maximal exercise intensity, $\dot{V}O_2$ reaches a level that is sufficient to meet the energy demands of the tissues. Other physiological variables such as heart rate, cardiac output and breathing frequency also plateau and a steady-state condition is achieved (Waters and Mulroy, 1999). Measuring the rate of oxygen consumption at this point provides a reflection of the energy expenditure required for an activity. During moderate exercise intensities (VO_2 < 2 litres·min^{-1}) at a constant work rate, the rate of oxygen consumption appears to reach a steady-state value within approximately 2-3 minutes from a resting baseline in healthy young individuals (Poole and Jones, 2011). The deficit between energy demand and oxygen uptake prior to achieving a steady-state value is covered by intramuscular oxygen stores, and depletion of phosphocreatine (Jones et al., 2013). In line with this, studies investigating the economy of load carriage energy expenditure have measured $\dot{V}O_2$ after 3 minutes of exercise at a constant work rate (Lloyd and Cooke, 2000b, Lloyd et al., 2010b, Abe et al., 2004, Huang and Kuo, 2014).

Most of the energy consumed during human locomotion can be attributed to the energy consumed by the muscles to generate force and perform mechanical work (Fedak et al., 1982). Williams and Cavanagh (1987) reported that several mechanical factors relate to economy when running, including postural excursions, COM oscillations, step parameters and ground reaction force-time curves. Thus, running mechanics appear to directly affect the metabolic energy cost. Given the kinematic adaptions that occur in response to carrying an external load (Kinoshita, 1985, Lloyd and Cooke, 2011), it seems reasonable to speculate that mechanical adaptions to load carriage could influence economy when walking.

2.2. Measuring load carriage economy

Load carriage economy has been reported as $\dot{V}O_2$ in absolute terms (Chung et al., 2005), relative to body mass (Lloyd and Cooke, 2000b, Legg and Mahanty, 1985, Quesada et al., 2000) and relative to the combined mass of the body and external load (Balogun et al., 1986). In order to provide a better measure of the energy expenditure attributable to the action of walking, some authors have reported economy as the net energy cost by subtracting the $\dot{V}O_2$ when stood at rest from the $\dot{V}O_2$ when walking (Abe et al., 2004, Bastien et al., 2005, Browning and Kram, 2009). Metabolic rate, often reported as metabolic power normalised to mass (W/kg), has also been used as a measure of load carriage economy, in order to account for substrate utilisation (Huang and Kuo, 2014, Kipp et al., 2018). Measures of metabolic rate calculate the rate of energy production from measured $\dot{V}O_2$ and $\dot{V}CO_2$ by including coefficients for $\dot{V}O_2$ and $\dot{V}CO_2$ based on the assumptions for energy release from carbohydrates and lipids. The Brockway equation (Equation 1; Brockway, 1987) appears to be the most cited measure of metabolic rate used in walking, running and load carriage research (Browning and Kram, 2009, Arellano and Kram, 2011, Huang and Kuo, 2014).

$$\text{Metabolic rate} = 16.58\ \dot{V}O_2 + 4.51\ \dot{V}CO_2 - 5.90\ N \qquad \text{Equation 1}$$

The limitation with these methods when calculating load carriage economy is that they do not account for the energy expenditure of unloaded walking. In an attempt to mitigate this limitation, the logic of calculating net energy cost has been further extended with the development of the ELI (Equation 2) as a measure of relative load carriage economy (Lloyd et al., 2010a).

$$\text{ELI} = \frac{\text{mlO}_{2L} \cdot \text{kg total mass}^{-1} \cdot \text{min}^{-1}}{\text{mlO}_{2U} \cdot \text{kg body mass}^{-1} \cdot \text{min}^{-1}} \qquad \text{Equation 2}$$

Equation 2 mlO_{2L} refers to oxygen consumption for the combined mass of the individual and external load at a given walking speed. mlO_{2U} refers to oxygen consumption for unloaded walking at the same given walking speed.

The ELI, developed from the seminal work of Taylor et al. (1980) (Equation 3), accounts for the rate of oxygen consumption during unloaded walking, providing a better understanding of the oxygen consumption attributable to supporting and moving an external load (Lloyd et al., 2010a).

$$\frac{\dot{V}O_{2L} / \dot{V}O_2}{m_L/m} \qquad \text{Equation 3}$$

Equation 3 $\dot{V}O_{2L}$ refers to oxygen consumption for the combined mass of the individual and external load at a given walking speed. $\dot{V}O_2$ is the oxygen consumption at the same given walking speed without a load. m_L is the combined mass of the individual and the external load. m is the mass of the individual.

An ELI value of 1 indicates that the additional energy expenditure required to carry a load is increased in direct proportion to the mass of the additional load supported by the muscles. An ELI >1 indicates a reduced economy, while an ELI <1 indicates an improved economy (Lloyd et al., 2010a). Lloyd et al. (2010a) conceptualized the energetic cost of load carriage as: the energy cost of unloaded walking at a given speed + the energy cost required to support and move a given external load ± the net change in the energy cost of movement due to changes in the kinematics and kinetics of movement as a result of the interaction between the load mass, speed and load carriage method. The final term in this expression reflects changes in movement economy associated with loaded locomotion. When this final term is 0, the ELI will be 1 (i.e. the cost of carrying the external load has risen in proportion to the mass of the load being supported). In this instance, it is possible that there are no changes in the kinetics or kinematics of movement with load carriage or, more likely, the positive and negative effects cancel out. Any deviation from an ELI value of 1 suggests that the final term is not 0 and, thus, the changes in the kinematics and kinetics of movement have had a net positive or negative effect on load carriage economy. Based on this concept, the ELI appears to be a more appropriate measure of load carriage economy compared to measures that do not account for the energy cost of unloaded walking, particularly for research investigating the determinants of load carriage economy.

Few load carriage studies include a measure of unloaded oxygen consumption or energy expenditure and it could be argued that this is a serious omission from much of the current load carriage literature. Any additional energy expenditure above that required for unloaded walking, when carrying external load, is likely associated with biomechanical changes that are perturbations from an individual's normal gait pattern (Lloyd and Cooke, 2011). Based on ELI values calculated from previously published literature, Lloyd et al. (2010a) demonstrated that the ELI is sensitive enough to differentiate between load placements. They found higher ELI values for load carried on the feet (ELI values ranging from 1.45 – 1.73) and in the hands (ELI values ranging from 1.07 – 1.32) compared to on the back (ELI values ranging from 0.97 – 1.01) and evenly distributed around the trunk (ELI values ranging from 0.96). Lloyd et al. (2010a) also reported that the ELI is independent of the magnitude of external load (with loads of 10 – 30% of body mass), body composition and walking speed by finding no strong correlations between any of these variables and ELI. Consequently, the ELI could represent a useful tool for comparing the relative economy of different load carriage systems. As yet, no studies have assessed the reliability of the ELI. Knowledge of the ELI's reliability is important if the measure is to be used with confidence.

2.3. Load placement and walking economy

There is a substantial amount of literature on the physiological demands associated with load carriage. However, many studies have focused on a single method of load carriage and there are often differences in the walking speed and load mass employed. This makes it difficult to directly compare the findings of these studies and evaluate the effect of different load placements on economy. In order to make such comparisons, ELI values can be calculated for studies that include a measure of unloaded oxygen consumption. Table 1, an adapted version of the table from Lloyd et al. (2010a), shows that loads carried more distally (e.g. in the hands or on the feet) appear to result in a worse relative economy (higher ELI values) (Soule and Goldman, 1969, Kamon and Belding, 1971) compared to loads carried closer the body's centre of mass (COM), which produce a more proportional response (ELI values of approximately 1.00) (Legg and Mahanty, 1985, Lammert and Garby, 1985). This is in agreement with earlier studies comparing the effect of different load placements on economy (Soule and Goldman, 1969, Datta and Ramanathan, 1971, Legg, 1985), all of which concluded that the optimum methods of load carriage place the COM of the external load close to the COM of the body. Of these early studies, Datta and Ramanathan (1971) compared economy in the largest range of load carriage methods (*n* = 7) which included a traditional backpack, doublepack (load split evenly between the front and back of the torso), rice bag (sack held by hands over each shoulder), yoke (load supported by a bamboo pole across shoulders), in the hands, directly on the head and indirectly on the head (load placed on back and supported by a head strap). The study included seven male participants and although load carriage experience was not reported, six of participants were reported to have sedentary jobs whilst the seventh was employed in unskilled manual work. For each mode, 30 kg was carried for 1 km on level ground at a speed of 5 $km \cdot h^{-1}$. The authors reported clear differences in economy between the modes, with the doublepack being the most economical ($\dot{V}O_2$ = 1.01 $l \cdot min^{-1}$) and in the hands being the least ($\dot{V}O_2$ = 1.46 $l \cdot min^{-1}$). The doublepack was associated with significantly better economy than all other methods, except for direct head-loading. The percentage increases in $\dot{V}O_2$ above the doublepack method were 2.8% for direct head-loading, 9.5% for backpack, 14.7% for indirect head-loading, 20.9% for rice bag, 28.8% for yoke and 45% for hands. Unfortunately, Datta and Ramanathan (1971) did not include a measure of unloaded walking economy, and as such, it is not possible to make direct comparisons between their findings and those of the studies included in Table 1.

In agreement with Datta and Ramanathan (1971), Lloyd and Cooke (2000b) found a 6-9% decrease in $\dot{V}O_2$ when carrying 25.6 kg in a commercially available doublepack system (load evenly distributed between the front and back of the trunk) compared to a backpack, on flat and uphill gradients (up to 20%). Legg and Mahanty (1985) compared five modes of carrying load close to the trunk (35% of body mass, mean load mass: 24.9 kg) and found no significant difference between any of the load carriage devices. However, they did report that a doublepack was associated with a 6.4% decrease in $\dot{V}O_2$ compared to the load carried on the back. Based on this evidence, evenly distributing a heavy load (>20 kg) around the trunk appears to be more economical than carrying the load on the back alone. Although more subtle differences between trunk loading methods, such as external versus internal frames, appear to produce minimal differences in economy (Holewijn, 1990, Kirk and Schneider, 1992).

Table 1. Calculated mean ELI values from previously published data for different forms of load carriage (Adapted from Lloyd et al. (2010a); BM = Body Mass). Standard deviations could not be calculated for ELI due to a lack of individual data.

Reference	Loading Methods		Participants	Speed ($km \cdot h^{-1}$)	ELI	Comments
	Position	Mass				
Soule and Goldman (1969)	Feet	12 kg (6 kg on each foot)	10 males	4.0, 4.8 & 5.6	1.57 – 1.86	Increase in ELI with increase in speed. ELI = 1.57, 1.79 & 1.87 for 4.0, 4.8 & 5.6 $km \cdot h^{-1}$, respectively.
Soule and Goldman (1969)	Hands	8 & 14 kg (4 & 7 kg in each hand)	10 males	4.0, 4.8 & 5.6	1.08 – 1.26	ELI increased with load mass increase but not with increase in speed. Lowest ELI with 8 kg was 1.08 at 4.8 $km \cdot h^{-1}$. Lowest ELI with 14 kg was 1.22 at 4.8 $km \cdot h^{-1}$.
Kamon and Belding (1971)	Hands	10, 15 & 20 kg	3 males	4.0 & 5.0	1.07 – 1.32	4.0 & 5.0 $km \cdot h^{-1}$, Increasing ELI with increasing load from 10 - 20kg on 0% gradient.
Francis and Hoobler (1986)	Hands	1.82 & 3.62 kg	5 males, 5 females	4.8 & 5.6	1.02 – 1.05	Light loads – 1.82 and 3.64 kg. No difference in ELI between load mass. Small difference in ELI between walking speeds (1.02 for 5.6 $km \cdot h^{-1}$ and 1.05 for 4.8 $km \cdot h^{-1}$).
Gordon et al. (1983)	Back	20%, 30%, 40% & 50% BM	10 males	4.8	0.97 - 1.01	Small decrease in ELI with increasing load. All walking performed at a 10% gradient.
Legg and Mahanty (1985)	Back	35% BM	5 males	4.5	1.02	ELI value of 1.02 reported for a military backpack with an internal frame and the same backpack without a frame.

Reference	Loading Methods		Participants	Speed ($km{\cdot}h^{-1}$)	ELI	Comments
	Position	Mass				
Legg and Mahanty (1986)	Back	35% BM	5 males	4.5	1.02	35% BM of the average participant represents 24.9 kg. ELI = 1.34 with 30% BM carried in backpack and 5% BM in weighted boots (2.5% on each foot).
Rorke (1990)	Back	20% & 40% BM	10 males	4.8 & 6.1	0.93 - 1.05	20% & 40% BM, 4.8 & 6.1 $km{\cdot}h^{-1}$, increasing ELI with increases in speed and load.
Quesada et al. (2000)	Back	15% & 30% BM	12 males	6.0	1.04 - 1.05	ELI values of 1.04 and 1.05 for 15% and 30% BM, respectively.
Lloyd and Cooke (2000b)	Back	25.6 kg	5 males, 4 females	3.0	1.12 - 1.27	Varying gradients from -27% to 20%. 25.6 kg was 35% of the average body mass.
Lloyd et al. (2010b)	Back	10-70% BM	24 females	SS (mean = 3.08)	0.93 - 1.09	Load mass increased until voluntary secession. Seven participants manged to carry 70% BM. No significant change in ELI between load mass. All participants manged 10-25% BM with ELI range of 0.94 - 0.99.
Hinde et al. (2017)	Back	18.2 kg	7 males, 4 females	4.0	0.97 – 0.99	Loads carried at 0% (ELI = 0.99) and 10% (ELI = 0.97) gradient. Study included temperatures of -10°C - 20°C (only 20°C data is reported here).

Reference	Loading Methods		Participants	Speed (km·h^{-1})	ELI	Comments
	Position	Mass				
Prado-Nóvoa et al. (2019)	Back	5, 10 & 15 kg	27 males, 21 females	4.0	0.93 – 0.99	No difference in economy between males and females. Lowest ELI values occurred with 15 kg for males (0.93) and females (0.95).
Vickery-Howe et al. (2020)	Weighted vest	20% & 40% BM	15 males, 15 females	SS (mean = 4.7)	0.95 – 1.00	No difference in self-selected speed between males and female or load carriage conditions (including unloaded walking). No difference between males and females. 20% body mass represented 14.8 kg and 12.3 kg for males and females, respectively. 40% body mass represented 29.7 kg and 24.6 kg for males and females, respectively.
Legg and Mahanty (1985)	Back/ Front	35% BM	5 males	4.5	0.96	Half of the load carried in a military backpack. The other half carried in a front pack (slightly smaller commercially available pack carried on the chest).
Lloyd and Cooke (2000b)	Back/ Front	25.6 kg	5 males, 4 females	3.0	1.04 - 1.24	Varying gradients from -27% to 20%. 25.6 kg was 35% of the average body mass. ELI increased as the gradient increased up to 20%.
Soule and Goldman (1969)	Head	14 kg	10 males	4.0, 4.8 & 5.6	0.99 - 1.04	14 kg (steel helmet with added lead weights. ELI = 1.02, 0.99 & 1.04 for 4.0, 4.8 & 5.6 km·h^{-1}, respectively.

Reference	Loading Methods		Participants	Speed ($km \cdot h^{-1}$)	ELI	Comments
	Position	Mass				
Nag and Sen (1979)	Head	60, 80 & 100 kg	4 males	3.2 & 3.7	0.87 - 1.22	Head strap method (forehead strap) Increase in ELI with increase in load and increase in walking speed. At 3.2 $km \cdot h^{-1}$, ELI = 0.87 – 1.06. At 3.7 $km \cdot h^{-1}$, ELI = 0.96 – 1.22.
Lloyd et al. (2010b)	Head	10-70% BM	24 females	SS (mean = 3.08)	0.95 - 1.11	Direct head loading. Load mass increased until voluntary secession. Experienced (*n* = 13) and inexperienced (*n* = 11) head-loaders. Two participants (experienced) manged to carry 70% BM. No significant change in ELI with increase in load. All participants managed 10-15% BM with ELI range of 1.03 - 1.07.

* BM = Body mass; SS = self-selected

Table 1 shows some consistency in the literature for the economy associated with load carried on the trunk, close to the body's COM, with the energetic cost increasing roughly in proportion to the mass of the additional load. This is indicated by ELI values of approximately 1.00 (e.g. a load of 20% body mass would result in a 20% increase in energy expenditure) across a range of walking speeds and load mass when walking at 0% gradient. There is, however, one form of load carriage, head-loading, that has produced inconsistent findings in the literature. Head-loading positions the load either directly over the body's COM (direct head-loading) or on the back supported by a strap around the forehead (indirect head-loading). Both methods are widely used in Africa and Asia. For inexperienced head-loaders, Soule and Goldman (1969) showed a proportional increase in $\dot{V}O_2$ to load carried directly on head-loading (Table 1), and Datta and Ramanathan (1971) found no significant difference in $\dot{V}O_2$ between head-loading (both direct and indirect methods) and back-loading.

In contrast, Maloiy et al. (1986) and Charteris et al. (1989) reported that head-loading can be a very economical method of load carriage for experienced head-loaders. Maloiy et al. (1986) reported that African women (of the Luo and Kikuyu tribes) with head-loading experience are able to carry loads of up to 20% body mass on the head with no additional energy expenditure (assessed via $\dot{V}O_2$) above that required for unloaded walking. Furthermore, Maloiy et al. (1986) showed that these women could carry loads above 20% body mass with a proportional increase in energy expenditure (e.g. a load of 30% of body mass would result in a 10% increase in energy cost). Carrying 20% of body mass with no additional energy expenditure above that required for unloaded walking would imply an ELI value of 0.83, which is somewhat lower than the ELI reported for other methods of load carriage or indeed other head-loading studies (Table 1). The findings by Maloiy et al. (1986) were supported by Charteris et al. (1989), who reported that loads of up to 25% of body mass can be carried directly on the head by African (Xhosa) women with several years of head-loading experience before energy expenditure increased above that required for unloaded walking. This phenomenon has been termed the 'free-ride' hypothesis (Charteris et al., 1989).

2.4. The 'free-ride' hypothesis

Based on the work of Soule and Goldman (1969), Datta and Ramanathan (1971), Maloiy et al. (1986) and Charteris et al. (1989), it would appear that the free-ride hypothesis might be explained by head-loading experience. This is further supported by the work of Maloiy et al. (1986) who, in addition to investigating head-loading economy in African women, also compared back-, head- and combined back and head-loading for three inexperienced head-loaders. They found the same proportional increase in $\dot{V}O_2$ for head-loading (ELI of 1.00 ± 0.04), back-loading (ELI of 1.00 ± 0.04) and combined back and head-loading (ELI of 1.00 ± 0.04). Maloiy et al. (1986) suggested that experienced head-loaders might have some form of mechanical advantage when head-loading and/or might have some anatomical adaptation as a result of carrying load since childhood, which could account for the improved economy. However, research by Das and Saha (1966) on professional Nepalese porters (*n* = 6), who regularly carry load on the head, found no advantage for direct or indirect head-loading compared to back-loading. They reported an increased $\dot{V}O_2$ above that required for back-loading of 7.3% (0% gradient), 7.7% (10% gradient) and 3.3% (20% gradient) for indirect head-loading and 4.5% (0% gradient), 28.5% (10% gradient) and 23.6% (20% gradient) for direct head-loading. Das and Saha (1966) provide no explanation for the poor economy associated direct head-loading. It may be that walking on an incline gradient while balancing a load on top of the head requires more effort to maintain posture compared to back-loading or indirect head-loading.

The free-ride hypothesis is based on limited data, with very small participant numbers of five and six used by Maloiy et al. (1986) and Charteris et al. (1989). Furthermore, Maloiy et al. (1986) allowed participants to carry load in their customary manner, with three women carrying load on top of the head (direct head-loading) while two carried the load on the back supported by a strap around the forehead (indirect head-loading). As such, the energy saving phenomenon reported by Maloiy et al. (1986) appears to be independent of head-loading method. This is unexpected as the kinematics of the two methods appear to be very different, with indirect-head-loading likely to evoke a greater increase in forward lean, which is a factor associated with reduced economy (Lloyd and

Cooke, 2000c). However, no studies have directly compared the kinematics of the two methods. The proposed mechanism for improved load carriage economy are explored in the next section of this literature review (section 2.5).

Lloyd et al. (2010c) attempted to provide a more comprehensive investigation of the ‘free-ride’ hypothesis. They assessed the physiological consequences of head-loading compared to back-loading for twenty-four Xhosa women, thirteen of which had at least 10 years of head-loading experience and eleven had no head-loading experience. The authors reported a large level of individual variation in load carrying economy, regardless of method, with some women more economical for head-loading while others were more economical in back-loading. Interestingly, Lloyd et al. (2010c) also found head-loading economy to be independent of experience with three of the four most economical head loaders being inexperienced. Of the individuals that were more economical at head-loading than back-loading (9 of the 24 participants), five were experienced head-loaders and four were inexperienced. The difference in findings between Lloyd et al. (2010c) and those of Maloiy et al. (1986) and Charteris et al. (1989) might be explained by differences in sample size. Lloyd et al. (2010c) showed that it is possible to select a subset of women who achieved remarkable levels of economy, similar to those reported in earlier studies (Maloiy et al., 1986, Charteris et al., 1989). This shows that the ‘free-ride’ is not a generalisable finding when testing a larger sample of women and is not explained by head-loading experience.

An energy saving phenomenon similar to the ‘free-ride’ has also been reported with light loads (~10 - 15% body mass) carried on the back at slow walking speeds of 2.4 $km \cdot h^{-1}$ to 3.6 $km \cdot h^{-1}$ (Abe et al., 2004). Furthermore, Lloyd and Cooke (2000b) demonstrated that carrying a heavy load (25.6 kg) evenly distributed between the anterior and posterior trunk appears to be more economical than carrying the same heavy load on the back alone. Both of these studies indicated that freedom of movement in the trunk could be an important factor for improved load carriage economy, with light loads carried on the back allowing for a similar level of trunk angle excursion to unloaded walking, while combined back and front loading appears to allow for increased trunk angle excursion compared to back loading with a heavy load (Lloyd and Cooke, 2000b).

2.5. Proposed mechanisms for improved load carriage economy

Despite evidence that the 'free-ride' is not a generalisable finding (Lloyd et al., 2010b, Lloyd et al., 2010c), previous attempts to explain the phenomenon have resulted in a number of proposed mechanisms for improved load carriage economy, particularly when head-loading (Maloiy et al., 1986, Jones et al., 1987, Heglund et al., 1995). Considering these proposed mechanisms might be useful in attempting to elucidate the key factors associated with an individual's load carriage economy and could help explain the individual variation reported by Lloyd et al. (2010b), (2010c).

Maloiy et al. (1986) suggested that experienced head-loaders might be able to carry a load on the head more steadily, without it moving or accelerating and decelerating as much as the body, allowing it to be carried without additional energy cost. However, there is no data to show whether African women can move loads more steadily. In any case, reducing the movement of the load would not guarantee an energetic advantage if dependent on additional muscle activity to alter walking gait mechanics, in order to reduce load movement. Maloiy et al. (1986) also suggested that carrying loads on the head from early childhood might lead to anatomical adaptations, allowing individuals to support loads < 20% body mass using non-metabolizing structural elements. However, Alexander (1986) argued that the spine would have to be soft enough to compress by 0.25 metres in order to make the body compliant enough to prevent vertical movement of the load moving while walking, which does not seem feasible.

Jones et al. (1987) proposed that body composition might be responsible for African women being more economical head loaders. They reported that leaner women from a sample of eight Mandinka women (with body fat ranging between 16-34%) appeared to exhibit the 'free ride' phenomenon identified by Maloiy et al. (1986) for direct head-loading, while women with a higher percentage of body fat produced a more proportional response. Jones et al. (1987) concluded that Mandinka women (from Keneba, Gambia) can carry up to 40% of their fat free mass as either body fat, an external load or as a combination of both, before needing to increase their energy expenditure above that associated with unloaded walking. However, the findings of Lloyd et al. (2010c) contradict this conclusion as the eleven most economical head loaders in their study (average ELI values below 0.9) were women with a body mass index that showed them to be slightly overweight (BMI = 26.0 ± 4.1 kg/m^2). Therefore, given the small number of participants used by Jones et al. (1987) (n = 8), the findings might have been a consequence of one or two individuals in the lean group being very economical with this method. Unfortunately, only mean data were reported. Lloyd et al. (2010c) reported that three out of the four most economical head-loaders in their study were women with no previous head-loading experience, which suggests that anatomical adaptations to head-loading are also unlikely to explain head-loading economy.

Heglund et al. (1995) speculated that some of the energy required to accelerate and decelerate the body when walking could be conserved from an improved exchange of energy between potential and kinetic forms, which would reduce the required mechanical work during each step. This exchange is similar to that of an inverted pendulum (Cavagna et al., 1977), with the body's COM vaulting up and over the support leg during the gait cycle. In a perfect system, as the body rises and falls through each step, the energy transfer from kinetic energy to potential energy and then back to kinetic energy would be complete (100%). However, in humans, the energy transfer has been estimated to be up to 65% during the stride cycle (Cavagna et al., 2002, Cavagna et al., 1977) and, as such, muscle activity is required and energy is expended to propel the body forward. Using the inverted pendulum theory, Heglund et al. (1995) proposed that African women might have a more complete energy transfer between potential and kinetic energy when head loading, allowing them to do less mechanical work. However, no evidence exists for how this improved energy transfer occurs when a load is placed on the head, and therefore how mechanical efficiency can be improved. Furthermore, Heglund et al. (1995) only analysed the kinetic and potential energies of the body's COM, which is a method of walking gait analysis that has been criticised by Winter

(1979) because it does not account for the energy exchanges that occur in the reciprocal movements of the limbs.

Abe et al. (2004) proposed that an energy saving phenomenon when walking at slow speeds (2.4 - 3.6 $km \cdot h^{-1}$) with a light loads of 9 and 12 kg (~10 - 15% body mass) carried on the back could be caused by increased rotative torque about the lower limb (rotative torque = radius of rotation between the COM of the body and load x load mass). This theory suggests that light loads do not constrain posture as much as heavy loads, allowing for increased flexion/extension in the trunk. Having an additional load on the back without constrained trunk movement could contribute to an increased momentum of the torso in the sagittal plane, which could increase forward momentum through the gait cycle. Although no empirical evidence exists to support this, theoretically, an increase in momentum would reduce the propulsive force that the muscles need to generate for a given walking speed, reducing the metabolic energy cost. Therefore, an increased freedom of movement in the trunk with a load carried on the torso might be a factor in understanding the determinants of load carriage economy with methods that load the trunk. However, it does not explain how some individuals can be more economical when head-loading, a method that appears to constrain posture in an upright position in order to balance a load on the head.

The individual variation in load carriage economy identified by Lloyd et al. (2010c) and Lloyd and Cooke (2011) indicates that the focus of load carriage research may benefit from focusing on mechanisms to explain the variation in economy between methods to focusing on why some individuals are more economical with certain methods of load carriage than others. Research on unloaded walking has shown that alterations in an individual's natural walking gait can influence energy expenditure (Donelan et al., 2001, Högberg, 1952, Heinert et al., 1988). It seems reasonable to hypothesis that differences in gait alterations between individuals when carrying an external load might provide an explanation for the individual variation in load carriage economy previously reported. Even acute perturbations to an individual's gait associated with a particular load carriage system could be an important factor when investigating the mechanisms that determine individual load carriage economy. The subsequent sections of this literature review will consider the biomechanical adaptations associated with load carriage systems that place the load close the body's COM to try and elucidate the factors that might determine individual load carriage economy.

2.6. Postural adjustments to load carriage

Joint angle kinematics have been frequently reported in load carriage research designed to examine postural alterations with load carriage. Movements of the trunk, hip, knee and ankle appear to be the most frequently investigated parameters, probably because this is where most movement occurs during the human walking gait. The perturbations caused by load to each of these joint angles from unloaded walking, and how perturbations might influence economy, will be considered in this section.

2.6.1. Trunk movement

As with physiological research on load carriage, biomechanical research has focused on the back-loading method, with few studies investigating the biomechanics of other trunk loading methods or head-loading. Many studies have found that back-loading increases forward lean in a load dependent manner (Kinoshita, 1985, Martin and Nelson, 1986, Goh et al., 1998, Harman et al., 2000, Lloyd and Cooke, 2000b, Attwells et al., 2006) (Table 2). For example, Attwells et al. (2006) reported an increase in mean forward lean during the stance phase of 17.8° for a 50 kg load, of which ~ 42 kg was carried on the back (with 8 kg carried in the form of a rifle and helmet), compared to an 8 kg load consisting of a helmet and a military rifle carried in the arms.

Backpacks shift the COM of the combined body and backpack system (combined system) in the posterior direction and forward lean appears to occur in an attempt to counter the posterior shift and improve postural stability (Kinoshita, 1985, Martin and Nelson, 1986, Goh et al., 1998, Harman et al., 2000). From a mechanical perspective, forward lean may help to propel the body forward into the next step (Kinoshita, 1985) and be necessary in reducing the risk of falling by keeping the COM of the combined system over the base of support (Harman et al., 2001). It may also help to keep the COM lower, which is likely to increase stability, particularly if the individual is walking over uneven terrain (Harman et al., 2001).

There is little data on the forward lean associated with load carriage methods that distribute the load around the trunk. Kinoshita (1985) reported that both back-loading and back/front-loading are associated with increased forward lean but that the forward lean associated with back-loading (11°) was greater than back/front-loading (4°) when carrying 40% of body mass. Lloyd and Cooke (2011) also found a greater increase in forward lean for back-loading (22°) compared to back/front-loading (9°) when carrying 25.6 kg. The findings of Kinoshita (1985) and Lloyd and Cooke (2011) seem logical given that evenly distributing a load between the anterior and posterior of the trunk will not shift the combined system COM from the body's unloaded position as much compared to back- loading. Further, Lloyd and Cooke (2011) also showed that the reduced forward lean for back/front-loading compared to back-loading alone occurs across the stance phase, with at least 9° less forward lean for back/front-loading at heel-strike, mid-support and toe-off gait events.

Table 2. A summary of forward lean values reported in load carriage literature.

Reference	Loading Methods		Participants	Speed ($km \cdot h^{-1}$)	Forward Lean	Trunk ROM	Comments
	Position	Mass					
Martin and Nelson (1986)	Back	0, 9.5, 17.7, 30, 36.8 kg	11 males, 11 females	6.4	-1.2 – 6.6°	-	Increase in forward lean with 30 kg and 36.8 kg. No significant difference between males and females.
Goh et al. (1998)	Back	0, 15% and 30% BM	10 males (infantry soldiers)	4.40 - 4.72	-8.38 – 4.26°	4.72 – 5.51°	Increase in forward lean with increased mass. No significant difference ROM. -8.3±1.4° reported for unloaded walking.
Attwells et al. (2006)	Back	8, 16, 40 and 50 kg (including rifle and helmet)	20 male soldiers	SS for each condition (mean = 5.4)	-4.8 - 13°	-	Rifle carried in the arms and helmet in all conditions (8 kg). Increase in forward lean with increase in mass. No difference in ROM (values not reported).
Wood and Orloff (2007)	Back	15% BM	13 females	4.68	10°	-	No change in forward lean during 30 minutes of walking.
Singh and Koh (2009)	Upper back; Lower back	0, 10%, 15%, 20% BM	17 boys	SS (mean data not reported)	2.23 – 11.75°	-	Increase in forward lean with increase in mass. No significant difference in forward lean between upper and lower back configurations.
Kinoshita (1985)	Back; B/F	0%, 20%, 40% BM	10 males (infantry soldiers)	4.5	Back = 7-11º; B/F = 4°	4°	Back-loading = 7° with 20% BM, 20º with 40% BM. B/F = 4° with both 20% and 40% BM. No difference in ROM between methods.
Lloyd and Cooke (2011)	Back; B/F	0, 25.6 kg	9 males	3	9 - 22°	-	9° forward lean in the B/F condition and 22° in the Back condition.

* SS = self-selected; ROM = Range of motion; B/F = back and front combined loading.

* Negative values represent trunk extension; positive values represent trunk flexion.

To date, Lloyd and Cooke (2011) are the only authors to have investigated the relationship between forward lean and load carriage economy. They reported a strong negative relationship (r = -0.867) between ELI and increased forward lean from heel-strike to mid-support when using a doublepack (back/front-loading) that was not present when carrying the same load with a backpack (r = 0.454). Economical load carriage systems that evenly distribute load around the trunk, such as the doublepack, are associated with more upright postures (reduced forward lean) compared to back loading (Kinoshita, 1985, Lloyd and Cooke, 2011). Lloyd and Cooke (2011) found that a doublepack allowed for a slightly greater freedom of movement in the sagittal plane of the trunk from heel-strike to mid-support compared to back loading. They tentatively suggested that the increased change in forward lean and associated increase in momentum through the stance phase might act as an energy saving mechanism. Furthermore, Lloyd and Cooke (2011) found that the differences in trunk angle at heel-strike, mid-support and toe-off events between unloaded walking and back-loading was strongly related to a worse economy (r = 0.643, r = 0.670 and r = 0.794 for heel-strike, mid-support and toe-off, respectively).

Trunk movements associated with head-loading have not been reported in the literature. It would seem logical to suggest that direct head-loading requires an upright posture (minimal forward lean) and minimal trunk ROM in order to balance the load on the head. As such, it is unlikely that any mechanical benefits from increased forward lean or trunk ROM would explain individual load carriage economy when head-loading. Further research is required to establish the relationship between freedom of movement of the trunk in the sagittal plane and load carriage economy.

The postural adjustments of the trunk with load carriage are likely to be associated with a change in muscle activity. Collecting the activity patterns of muscle during load carriage can be challenging, particularly using surface electromyography, due to interactions between the external load and the electrodes often interfering with the signal. Nevertheless, the muscle activity associated with some load carriage conditions have been reported. Carlsöö

(1964) found that back loading reduces sacrospinalis and erector spinae activity, while increasing the activity of the rectus abdominus. Carlsöö (1964) also noted that the activity of the erector spinae increased as forward lean increased. Motmans et al. (2006) investigated the muscle activity associated with a number of load carriage methods (unloaded, shoulder bag, backpack, front pack and doublepack) and found the doublepack to be closest to that of unloaded walking in terms of muscle activation, with similar muscle activity in both the rectus abdominus and erector spinae. In agreement with Carlsöö (1964) they found that carrying additional loads in a backpack significantly reduced erector spinae activity and concomitantly increased the activity of the rectus abdominus. To date, no peer-reviewed study has investigated muscular activation during head loading. For direct head loading, the upright posture that is likely required to balance the load on the head might be expected to be associated with greater erector spinae activity and reduced rectus abdominus activity compared to back loading. Due to the relatively low absolute level of activity in the postural muscles, and the trade-off in muscular activity associated with forward lean between the anterior and posterior muscle of the torso, it is unlikely that changes in muscle activation associated with forward lean explain differences in economy between different load carriage methods and individuals. However, an interaction between forward lean and other joint positions might be important in determining load carriage economy and is worthy of future study.

2.6.2. Hip, knee and ankle kinematics

Along with trunk angle, many studies focusing on the biomechanics of load carriage have also reported hip, knee and ankle joint angles (e.g. Kinoshita, 1985, Harman et al., 2000, Attwells et al., 2006, Majumdar et al., 2010). Furthermore, most studies have reported angular displacements in the sagittal plane for these joints because these changes are the most pronounced displacements in the walking gait (Kinoshita, 1985, Attwells et al., 2006, Majumdar et al., 2010). Table 3, Table 4 and Table 5 show a summary of some of the load carriage literature that has included measures of hip, knee and ankle angle in the sagittal plane, respectively.

Sagittal plane hip angle appears to decrease (larger hip flexion) at heel-strike and increase at toe-off (larger hip extension) as the mass of a load increases when carried on the back, causing an overall increase in hip angle range of motion (also known as hip angle excursion) across the stance phase (Harman et al., 2000, Majumdar et al., 2010, Wang et al., 2013). This is supported by a meta-analysis from Liew et al. (2016) which demonstrated that back-loading tends to increase hip range of motion compared to unloaded walking. This might be expected given the increased trunk forward lean associated with loads carried solely on the back. Evenly distributing the load around the torso (via a weighted vest) has been associated with peak flexion angles of 30° - 34° for loads of 10 - 40% of body mass (Silder et al., 2013, Wills et al., 2019, Vickery-Howe et al., 2020). In comparison, back-loading has been associated with larger hip flexion angles than those reported in the weighted vest studies, with Wang et al. (2013) showing a hip flexion angle of 45° at heel-strike with 32 kg. However, no studies have directly compared hip angle with back/front-loading compared to a back-loading. Hip angles have not been reported in literature for head-loading, but it might be expected that head-loading results in a greater level of hip extension compared to back- and back/front-loading to create a more upright posture to balance a load directly on the head. However, further research is warranted to assess hip movements with head-loading, particularly as sagittal plane motions of the trunk might contribute to load carriage economy in back- (Abe et al., 2004) and back/front-loading (Lloyd and Cooke, 2011).

Load carriage methods that position the load on the trunk have been shown to consistently increase knee flexion during the stance phase compared to unloaded walking (Table 4) (Harman et al., 2000, Silder et al., 2013, Vickery-Howe et al., 2020, Wang et al., 2013). Vickery-Howe et al. (2020) reported a significant increase in knee flexion of 3.5° with load from 0% - 40% of body mass carried using a weighted vest, which evenly distributed load around the torso. This supports the earlier work of Silder et al. (2013) who reported a 4° increase in knee flexion with load from 0% - 30% body mass carried with a weighted vest. For back-loading, Wang et al. (2013) found an increase in peak knee flexion of 6° from 0 – 32 kg when unfatigued, which increased with fatigue. Although no studies have directly compared knee flexion with different load carriage methods

Birrell and Haslam (2009) found that sagittal plane knee kinematics were not altered until heavier loads of 24 and 32 kg were carried, although the authors did not provide data. With heavier loads carried on the trunk, Birrell and Haslam (2009) suggested that knee angle ROM decreased as the mass of the load increased. A similar decrease in knee ROM for back-loading has also been reported by Harman et al. (2000). No previous research has assessed the knee angle movements associated with head-loading. It seems reasonable to speculate that head-loading would be associated with similar knee flexion angles to those reported with a weighted vest, as both Vickery-Howe et al. (2020) and Silder et al. (2013) reported an upright posture with this method, which would also be expected for direct head-loading in order to balance load on the head. Dynamic walking gait simulations have predicted that unloaded walking with increased knee flexion requires increased muscle activity (Steele et al., 2010), which is likely to result in an increase in metabolic cost (Waters and Mulroy, 1999). Furthermore, Ortega and Farley (2005) found that simultaneously increasing, hip, knee and ankle flexion during unloaded walking, to flatten the COM trajectory, doubled the metabolic cost compared to normal walking. While it's likely that an increased knee flexion could result in a greater metabolic cost, no experimental data exists to support the influence of isolated knee joint flexion on the metabolic cost of walking.

Load carriage with a backpack appears to increase ankle dorsiflexion at heel-strike and through the stance phase, and increase in plantarflexion during the initial part of the swing phase (Attwells et al., 2006, Majumdar et al., 2010). This leads to an increased ankle ROM compared to unloaded walking. Majumdar et al. (2010) suggested that an increase in dorsiflexion of the ankle during the stance phase might help absorb an increase in the impact forces that occur with increased load and facilitate increased knee flexion to further help absorb impact forces. More evenly distributing load around the torso (via weighted vests) has been shown to not alter peak ankle plantarflexion or dorsiflexion from unloaded walking when carrying up to 40% of body mass (Silder et al., 2013, Vickery-Howe et al., 2020) (Table 5). The role of the ankle joint in unloaded walking economy has been the subject of previous investigations, with impaired ankle movement at push-off appearing to require a greater level of energy expenditure to walk at a given speed (Doets et al., 2009, Van Engelen et al., 2010). Huang et al. (2015) restricted ankle plantarflexion during unloaded walking using a modified ankle-foot orthosis. This reduced the amount of work performed by the ankle during the push-off phase, leading to greater mechanical work by the knee and ankle joints during mid-stance and a greater overall metabolic cost. As such, if load carriage reduces ankle plantarflexion during push-off, it's possible that this could worsen an individual's load carriage economy.

Although changes to the hip, knee and ankle angles with load carriage have been studied for back-loading, the relationship between these changes and the metabolic energy requirements for carrying additional load have not been reported. A major contributor to the metabolic cost of walking is the mechanical work performed by the muscles to propel the body forward from one step to another (Donelan et al., 2002a). Huang and Kuo (2014) reported the estimated work performed at the hip, knee and ankle joints during loaded walking, which might provide an insight into the role these joints have in the energy expenditure associated with load carriage. The authors used inverse dynamics to estimate joint work with loads of up to 40% body mass carried in a backpack. They reported a large increase in positive work per stride for loaded walking compared to walking unloaded, which they attributed to the ankle joint at push-off and the knee joint after impact. The largest increase in positive work was attributed to the ankle during push-off. Huang and Kuo (2014) found that positive work and metabolic cost increased linearly with increased load and concluded that most of the increased metabolic cost with load carriage was explained by increased positive mechanical work, particularly at the ankle and knee. While this provides a useful indication of the distribution of work among the lower limb joints, the use of inverse dynamics to measure actual muscle work is imperfect and does not necessarily indicate the actual metabolic cost of exercise because it does not account for passive work performed through the passive stretching and shortening of tendons (Dean and Kuo, 2011).

Table 3. A summary of sagittal plane hip angles reported in the literature

Reference	Loading Methods		Participants	Speed (km·h^{-1})	Flexion (º)	Extension (º)	ROM (º)
	Position	Mass					
Vickery-Howe et al. (2020)	Weighted vest	0%, 20% & 40% BM	15 males, 15 females	SS (mean = 4.7)	Female peak angle (SD) = 33 (7), 34 (7), 33(9) for 0%, 20% and 40%, respectively. Male peak angle (SD) = 33 (8), 33 (7), 34(5) for 0%, 20% and 40%, respectively.	Female peak angle (SD) = -9 (5), -10(7), -11(7) for 0%, 20% and 40%, respectively. Male peak angle (SD) = -7(6), -8(6), -8(6) for 0%, 20% and 40%, respectively.	Female ROM calculated from peak values = 42, 44, 44 for 0%, 20% and 40%, respectively. Male ROM calculated from peak values = 42, 44, 44 for 0%, 20% and 40%, respectively.
Wills et al. (2019)	Weighted vest	23 kg	13 males	5.5	Peak angle (SD) = 34(7) Heel-strike = 32(6)	Peak angle (SD) = -16(7)	Calculated from peak values = 50
Silder et al. (2013)	Weighted vest	0%, 10%, 20% and 30% BM	17 males, 12 females	SS (mean = 4.64 ± 0.28)	Peak angle (SD) = 29(5), 30(6), 30(5), 32(5) for 0%, 10%, 20% and 30%, respectively.	Peak angle (SD) = -16(7), -16(7), -17(6), -16(6) for 0%, 10%, 20% and 30%, respectively.	Calculated from peak values = 45, 46, 47, 48 for 0%, 10%, 20% and 30%, respectively.
Wang et al. (2013)	Back (MOLLE)	0, 32 kg	18 males	6	At heel-strike = 32(4) and 45(5) for 0 kg and 32 kg walking, respectively.	-	-
Harman et al. (2000)	Back (ALICE)	6, 20, 33, 47 kg	16 males	4, 4.8, 5.4	114(7), 140(7), 137(7), 133(6) for 6 kg, 20 kg, 33 kg and 47 kg.	1946), 192(6), 191(6), 188(6) for 6 kg, 20 kg, 33 kg and 47 kg, respectively.	ROM (SD) = 50(5), 52(5), 54(5), 55(5) for 6 kg, 20 kg, 33 kg and 47 kg, respectively.

*MOLLE = Modular Lightweight Load-Carrying Equipment; ALICE = All-Purpose Lightweight Individual-Carrying Equipment; SS = self-selected
*Data from Harman et al. (2000) did not use anatomical zero. Negative values = extension; positive values = flexion.

Table 4. A summary of sagittal plane knee angle reported in literature.

Reference	Loading Methods		Participants	Speed ($km \cdot h^{-1}$)	Flexion (°)	Extension (°)	ROM (°)
	Position	Mass					
Vickery-Howe et al. (2020)	Weighted vest	0%, 20% & 40% BM	15 males, 15 females	SS (mean = 4.7)	Females peak angle (SD) = 23(12), 25(12), 29(14) for 0%, 20% and 40%, respectively. Males peak angle (SD) = 21(6), 21(6), 22(6) for 0%, 20% and 40%, respectively.	-	-
Wills et al. (2019)	Weighted vest	23 kg	13 males	5.5	Peak angle (SD) = 72(6) Heel-strike = 8(3)	Peak angle (SD) = -2(3)	-
Silder et al. (2013)	Weighted vest	0%, 10%, 20% and 30% body mass	17 males, 12 females	SS (mean = 4.64 ± 0.28)	Peak during stance (SD) = 22(5), 23(6), 24(6), 26(6) for BW, 10%, 20% and 30%, respectively. Peak during swing (SD) = 70(5), 70(5), 71(4), 71(6) for BW, 10%, 20% and 30%, respectively.	-	-
Wang et al. (2013)	Back (MOLLE)	0, 32 kg	18 males	6	Heel-strike = -3(3) and 4(3) for 0 kg and 32 kg, respectively. Peak knee flexion (SD) at stance 19(3) and 25(5) for 0 kg and 32 kg, respectively.	-	-

Reference	Loading method		Participants	Speed (km·h^{-1})	Flexion (°)	Extension (°)	ROM (°)
	Position	Mass					
Harman et al. (2000)	Back (ALICE)	6, 20, 33, 47 kg	16 males	4, 4.8, 5.4	Peak knee flexion (SD) = 111(3), 112(4), 113(5), 112(5) for 6 kg, 20 kg, 33 kg and 47 kg, respectively.	Peak knee extension (SD) = 178(5), 178(5), 179(6) and 179(6) for 6 kg, 20 kg, 33 kg and 47 kg, respectively.	Knee ROM (SD) = 67(6), 66(5), 66(6) and 65(6) for 6 kg, 20 kg, 33 kg and 47kg, respectively.
Attwells et al. (2006)	Back	7.95, 15.95, 39.95, 50.05 kg	20 male soldiers	SS	-	-	Approximately 21, 23, 26 and 26 for 7.95, 15.95, 39.95, 50.05 kg, respectively (actual values not reported by Attwells et al. (2006)).

*MOLLE = Modular Lightweight Load-Carrying Equipment; ALICE = All-Purpose Lightweight Individual-Carrying Equipment. SS = self-selected
*Data from Harman et al. (2000) did not use anatomical zero. Negative values = extension; positive values = flexion

Table 5. A summary of sagittal plane ankle angles reported in literature.

Reference	Loading Methods		Participants	Speed ($km{\cdot}h^{-1}$)	Dorsiflexion (°)	Plantarflexion (°)	ROM (°)
	Position	Mass					
Vickery-Howe et al. (2020)	Weighted vest	0%, 20% & 40% BM	15 males, 15 females	SS (mean = 4.7)	Females peak angle (SD) = 13(5), 14(5), 13(5) for 0%, 20% and 40%, respectively.	Females peak angle (SD) = -21(6), -21(8), -22(7) for 0%, 20% and 40%, respectively.	Female ROM calculated from peak values = 34, 35, 35 for 0%, 20% and 40%, respectively.
					Males peak angle (SD) = 11(4), 11(3), 11(4) for 0%, 20% and 40%, respectively.	Males peak angle (SD) = -22(6), -22(5), -23(6) for 0%, 20% and 40%, respectively.	Male ROM calculated from peak values = 33, 33, 34 for 0%, 20% and 40%, respectively.
Wills et al. (2019)	Weighted vest	23 kg	13 males	5.5	Peak dorsiflexion (SD) = 8(3) Heel-strike = 1(4)	Peak angle (SD) = -22(7)	Calculated from peak values = 30
Silder et al. (2013)	Weighted vest	10%, 20% and 30% body mass	17 males, 12 females	SS (mean = 4.64 ± 0.28)	Peak dorsiflexion (SD) = 13(4), 13(4), 13(4) and 14(4) for BW, 10%, 20% and 30%, respectively.	Peak plantarflexion (SD) = -12(5), -13(6), -12(5), -12(5) for BW, 10%, 20% and 30%, respectively.	Calculated from peak values = 25, 26, 25, 26 for 0%, 10%, 20% and 30%, respectively.
Wang et al. (2013)	Back (MOLLE)	0, 32 kg	18 males	6	Dorsiflexion (SD) at heel-strike = 8(2) and 7(3) for unloaded and loaded walking, respectively.	-	-
Harman et al. (2000)	Back (ALICE)	6, 20, 33, 47 kg	16 males	4, 4.8, 5.4	Peak dorsiflexion (SD) = 108(3), 108(5), 108(5), 108(5) for 6 kg, 20 kg, 33 kg and 47 kg, respectively.	Peak plantarflexion (SD) = 137(5), 137(6), 138(6) and 139(6) for 6kg, 20 kg, 33 kg and 47 kg, respectively.	Ankle ROM (SD) = 29(4), 29(3), 30(3) and 30(3) for 6 kg, 20 kg, 33 kg and 47 kg, respectively

*MOLLE = Modular Lightweight Load-Carrying Equipment; ALICE = All-Purpose Lightweight Individual-Carrying Equipment. SS = self-selected

* Negative values = extension; positive values = flexion

Much of the load carriage literature has focused on joint angle kinematics in the sagittal plane, and as such there is a general lack of data for how load carriage influences kinematics in the frontal and transverse planes. This is presumably because most movement occurs in the sagittal plane for walking and running gaits (Whittle, 2014). Trunk loading methods that constrain posture in the sagittal plane, such as back-loading with heavy load mass, are likely to have a similar effect on movements in the transverse plane. Indeed, a few studies have reported a decrease in pelvic rotation from unloaded walking with the back-loading method (LaFiandra et al., 2002, Sharpe et al., 2008, Birrell and Haslam, 2009). LaFiandra et al. (2002) found that back-loading with 40% body mass at 2.16 - 5.76 $km \cdot h^{-1}$ resulted in a decrease in transverse plane pelvic and trunk rotation, but an increase in upper body torque of 225% compared to unloaded walking. It's possible that an increase in freedom of movement of the trunk in the transverse plane could benefit load carriage economy, by allowing for a walking gait pattern that is closer to that of unloaded walking. However, increasing rotational movements in the upper body with heavy loads could also increase upper body torque, which increases the potential for low back injury and would require an increase in muscular effort to counteract the torque (LaFiandra et al., 2002). As such, it's possible that an increase in upper body rotation with heavy loads could have a negative impact on load carriage economy. The use of a hip belt when back-loading transfers ~30% of the vertical load from the shoulders to the hips during level walking (LaFiandra and Harman, 2004) and allows for greater amplitudes of pelvic and trunk rotation with heavy load (40% body mass) compared to not using a hip belt (Sharpe et al., 2008). This increase in trunk and pelvic rotation with a hip belt could be due to the reduced vertical load on the shoulders, lessening the upper body torque associated with back-loading.

Hip abduction appears to increase from unloaded walking for back-loading, with greater abduction as the mass of the load carried increases (Birrell and Haslam, 2009). An increase in hip abduction with an increase in the load carried could be a consequence of increased step width, widening the base of support to increase stability. Donelan et al. (2001) showed that young healthy individuals preferred an energetically optimal step width of 0.13 ± 0.03 *L,* where *L* is step width expressed as a fraction of leg length, compared to wider and shorter steps widths which require a greater metabolic cost. Forced perturbations to widen step width increase the energy cost of unloaded walking by increasing the mechanical work required to redirect the centre of mass from step-to-step (Donelan et al., 2001, Donelan et al., 2002a). Narrow step widths (through forced perturbations), when step widths are narrower than the width of the foot, appear to increase the mechanical work required laterally to move the swing leg to avoid the stance leg which increases the energy cost of unloaded walking (Shorter et al., 2017). As such, alterations in step width with load carriage could lead to alterations in economy, particularly if load carriage causes an individual to take much wider or narrower steps than their preferred unloaded walking step width. Previous research on the effect of load carriage on step width has found no difference in step width as a percentage of stature with weighted vests between 10-30% body mass (Silder et al., 2013). However, no studies have assessed the association between step width and economy in head-loading or back-loading, which might require an increase in lateral stabilisation compared to other methods that evenly distribute load around the torso.

2.7. Spatiotemporal walking gait adjustments to load carriage

Spatiotemporal parameters of the human walking gait describe the timing and positional characteristics. They include factors such as step length and cadence, as well as timings of specific phases of the gait cycle (Bowker and Messenger, 1988). The perturbations caused by load to these spatiotemporal variables from unloaded walking, and how perturbations might influence economy, will be considered in this section

2.7.1. Stride length and cadence

The most economical stride length/stride frequency combination when walking unloaded is suggested to be similar to the one freely chosen by an individual (Högberg, 1952, Cotes and Meade, 1960, Knuttgen, 1961, Cavanagh and Williams, 1982). Furthermore, when walking at a given speed, individuals appear to choose a stride frequency that minimises energy expenditure (Cotes and Meade, 1960, Zarrugh et al., 1974). Literature on unloaded walking and running has indicated a U-shaped relationship between the stride length/frequency combination and energy expenditure, for a given speed (Högberg, 1952, Cotes and Meade, 1960, Cavanagh and Williams, 1982). Therefore, smaller deviations from the optimal stride length/stride frequency combination appear to result in a smaller effect on $\dot{V}O_2$ compared to larger deviations.

Table 6 provides a summary of studies that have investigated the effect of load carriage on stride length. Most studies have found little change in stride length between load carriage conditions (e.g. Goh et al., 1998, Harman et al., 2000, Wood and Orloff, 2007, Silder et al., 2013, Huang and Kuo, 2014). The relatively few that have reported a change in stride length with load carriage appear to have found a slight shortening, compared to unloaded walking at 0% and uphill gradients (Martin and Nelson, 1986, Harman et al., 2000, LaFiandra et al., 2003b). For a given walking speed, this would mean that as the load mass increases, stride length decreases with a concomitant increase in stride frequency. These studies have tended to find a shortening of stride length under heavy load carriage and/or fast walking speed conditions. Studies comparing different methods of trunk loading have found little difference in the stride length – cadence combination (Kinoshita, 1985, Lloyd and Cooke, 2011). There appears to be differences in loaded stride length between males and females (Martin and Nelson, 1986), which as might be expected, appears to not be the case when stride length is normalised to stature (Silder et al., 2013). Like much of the load carriage literature, nearly all studies examining stride length/stride frequency have only reported mean data. Lloyd and Cooke (2011) reported a high level of individual variability for change in stride length (+12% to -6%) from unloaded to loaded (with 25.6 kg) when walking on the flat. Thus, considering individual differences in stride length alterations from load carriage might be useful when attempting to determine individual differences in economy. While small perturbations are unlikely to affect the energy cost of load carriage with any particular method (Högberg, 1952, Knuttgen, 1961, Heinert et al., 1988), larger alterations might be detrimental. When Maloiy et al. (1986) first reported the free ride phenomenon, they found no difference in stride frequency between head-loading and unloaded walking. It is possible that the most economical method of load carriage for an individual is one that causes no change in stride parameters from unloaded walking.

Table 6. A summary of stride length values reported in load carriage literature.

Reference	Loading Methods		Participants	Speed (km·h^{-1})	Stride Length (m)	Comments
	Position	Mass				
Martin and Nelson (1986)	Back	0, 9, 17, 29, 36kg	11 males, 11 females	6.4	M = 0.88 – 0.90 F = 0.82 – 0.86	Significant decrease in stride length as load mass increased. Women had significantly shorter stride lengths (increased cadence).
Goh et al. (1998)	Back	0, 15% and 30% body mass	10 males (infantry soldiers)	SS (range = 4.40 - 4.72)	1.27 – 1.43	No difference in stride length between loads.
Harman et al. (2000)	Back	6, 20, 33, 47kg	16	4, 4.8, 5.4	1.57 – 1.60	(ALICE) US military backpack. No difference in stride length between load mass from 6 - 33 kg. Significant reduction with 47 kg.
LaFiandra et al. (2003b)	Back	0 and 40% BM	5 males, 7 females	2.16, 2.88, 3.60, 4.32, 5.04, 5.76	L = 0.85 – 1.37 U = 0.89 – 1.42	Increased walking speed significantly increased stride length. Significant decrease from unloaded (U) to loaded (L).
Wood and Orloff (2007)	Back	15% body mass	13 females	4.68	1.40 -1.41	No change in stride length over a 30-minute period of walking.

Reference	Loading Methods		Participants	Speed ($km{\cdot}h^{-1}$)	Stride Length (m)	Comments
	Position	Mass				
Singh and Koh (2009)	Back	10%, 15%, 20% body mass	17 boys (9 ± 1.58 years)	SS (absolute velocity not reported)	-	Absolute stride length values not reported. No significant differences between in normalised stride length between loads.
Huang and Kuo (2014)	Back	0, 10.6, 15.1, 19.6, 24.2 kg	6 males, 2 females	4.5	-	Absolute stride length values not reported. Normalised step length did not change with an increase in load mass.
Kinoshita (1985)	Back; B/F	0%, 20%, 40% BM	10 males (infantry soldiers)	4.5	1.46 – 1.48	No difference in stride length between methods or mass.
Lloyd and Cooke (2011)	Back, B/F	0, 25.6 kg	5 females, 4 males	3	-	No difference in stride length between unloaded, backpack and doublepack when walking on the flat.
Silder et al. (2013)	Weighted vest	0, 10%, 20% and 30% body mass	17 males, 12 females	SS (mean = 4.64 ± 0.28)	-	Absolute stride length values not reported. No differences in normalised stride length between males and females or loads.

*U = unloaded; L = loaded; M = males; F = females; BM = body mass; SS = self-selected; B/F = Back and Front combined.

2.7.2. Stance time

Load carriage appears to result in longer walking gait stance times compared to unloaded walking (Kinoshita, 1985, Kram et al., 1987, Lloyd and Cooke, 2000a, Birrell et al., 2007, Birrell and Haslam, 2010, Silder et al., 2013). A longer stance time would suggest that the normal function of the lever system of the foot is impeded during the push-off phase of walking. A longer stance time could be accounted for by the need to apply greater forces with increased load if there is no change in the rate of force application. It has also been suggested that a longer stance time is necessary when an extra load is being carried due to the need for extra stability (Schiffman et al., 2006), particularly if the load carriage system causes the COM to be shifted further away from the body's normal unloaded position (Birrell and Haslam, 2010). Kinoshita (1985) reported no change in stance time between a backpack and a doublepack with 20% or 40% body mass. However, on closer examination of the mean data, there was a shorter stance time with the doublepack compared to the backpack when carrying both 20% body mass (0.737 s backpack versus 0.730 s doublepack) and 40% body mass (0.745 s backpack versus 0.737 s doublepack). A shorter stance time when carrying load in a doublepack compared to a backpack is supported by the work of Lloyd and Cooke (2000a) who also found slightly shorter stance times with a commercially available doublepack compared to a traditional backpack. Stability is suggested to play an important role in stance time (Schiffman et al., 2006), with increased stability likely to reduce stance time. This might be explained the difference in findings between the conditions, with the even loading of the doublepack between the front and back of the torso likely to increase stability and reduce stance time compared to uneven loading with a backpack.

Additionally, the doublepack has been associated with a more upright posture compared to the backpack. This reduces the time it takes for the COM to pass over the foot during each stride, reducing stance time (Birrell et al., 2007). Birrell and Haslam (2010) observed shorter stance times for a backpack with webbing that distributed some of the load around the trunk compared to a standard backpack when a light load (8 kg) was carried. However, in contrast to previous research, the authors reported that stance time was significantly shorter when carrying heavier loads (24 kg and 32 kg) in the standard backpack compared to loads more evenly distributed around the trunk. They suggested that an increased perceived discomfort with the heavier load in the backpack compared to other conditions might have been responsible for the findings by causing the participants discomfort to alter their walking gait, overriding the biomechanical effects such as changes in the COM position or forward lean. To date, only Lloyd et al. (2011) have reported stance time as a parameter when investigating the effects of head-loading on the walking gait. They reported that stance time when head-loading did not change from unloaded walking while stance time when back-loading was significantly longer than unloaded walking (an increase of 0.031 seconds). This could be a consequence of the upright posture required to balance the load directly on top of the head while walking, which has been shown to decrease stance time compared to load carriage methods that are associated with greater forward lean.

2.8. Load carriage kinetics

Force platforms provide a means of recording and measuring the three components (vertical, antero-posterior, and medio-lateral) of foot-floor reaction force during human locomotion (Bowker and Messenger, 1988). Carrying an additional load increases the vertical and antero-posterior force produced during the walking gait (Kinoshita, 1985, Lloyd and Cooke, 2000a, Harman et al., 2000, Birrell and Haslam, 2008, Tilbury-Davis and Hooper, 1999).

Previous studies comparing the anteroposterior forces between loading conditions have tended to find no difference when there are only subtle differences in design between load carriage systems (e.g. different types of backpack) (Harman et al., 1999, LaFiandra et al., 2003a, Birrell et al., 2007). However, differences have been reported when there are substantial differences between load carriage methods (e.g. different load placements) (Kinoshita, 1985, Birrell and Haslam, 2010, Lloyd and Cooke, 2000a). Birrell and Haslam (2010), Kinoshita (1985) and Harman et al. (2000) all identified a smaller maximum braking force (in the antero-posterior direction) with load evenly distributed around the torso compared to the same load positioned on the back. Kinoshita

(1985) reported that peak braking force increased from unloaded walking by 45% for back-loading compared to 39% for back/front-loading when carrying 40% body mass. Birrell and Haslam (2010) found a 10% increase in braking force for back-loading compared to back/front-loading with a 32 kg load. A larger braking force for back-loading is likely to be a consequence of increased forward lean that has also been associated with backpack load carriage compared to doublepack (load evenly distributed around the torso) loading. In contrast, Lloyd and Cooke (2000a) identified no difference in maximum braking force between backpack and doublepack conditions but did find a larger maximum propulsive force with a backpack. The authors suggested that the larger propulsive force with a backpack could have resulted from a decrease in trunk movement through the stride cycle compared to a greater freedom of movement in the trunk with a doublepack. With this theory, more movement in the trunk could lead to an increase in momentum of the upper body, which could reduce the propulsive force requirements (Lloyd and Cooke, 2000a). Yet, there is conflicting evidence for reduced propulsive force with load evenly distributed around the torso compared to a backpack, with Kinoshita (1985) finding no difference in propulsive forces associated with either a backpack or a doublepack. Theoretically, a reduced propulsive force (for a given speed) could be beneficial to load carriage economy, as it would suggest that less energy is required to propel the body forward.

Birrell and Haslam (2008) suggested that load carriage methods restricting arm movement could increase both maximum braking and propulsive force compared to load carriage methods that did not impede the arms. The authors suggested that the mechanism for this could be a reduced involvement of the arms to drive the body forward (Birrell and Haslam, 2008). This concept could have implications for head loading, where the arms are usually required to be in a fixed position in order to support the load. Lloyd et al. (2011) found no difference in maximum braking and propulsive force between head loading and back loading. It might be that the greater trunk range of motion associated with more upright postures are neutralised by the need to restrict trunk motion in order to balance a load on the head (Lloyd et al., 2011).

Regarding vertical force, there is consistent evidence showing that the magnitude of peak vertical forces are approximately equal to the added load, even when different load carriage methods are concerned (Kinoshita, 1985, Tilbury-Davis and Hooper, 1999, Birrell et al., 2007, Birrell and Haslam, 2010, Lloyd and Cooke, 2000a, Lloyd et al., 2011). This implies that the increased peak vertical force with load carriage is predominantly due to the static effect of the load rather than changes in the acceleration of the system (combined body and load). The vertical force minimum (vertical force at mid-stance), however, does appear to be sensitive to differences in load carriage method. Kinoshita (1985), Lloyd and Cooke (2000a) and Birrell and Haslam (2008) all identified a significantly greater minimum vertical force associated with a back/front-loading compared to back-loading. This could be caused by the more upright posture associated with a doublepack resulting in a more vertical application of force. Given the position of the COM when head loading, this method might also be expected to increase the minimum vertical force compared to the same load carried on the back. In line with this, Lloyd et al. (2011) reported a slightly smaller force minimum for back-loading compared to head-loading. Birrell and Haslam (2010) reported a significant reduction in peak vertical force at the toe-off gait event (2^{nd} vertical peak) with load placed on the back compared to the same load more evenly distributed around the trunk. However, this is not a consistent finding in the literature with Kinoshita (1985) and Lloyd and Cooke (2000a) both finding no difference in the 2^{nd} peak vertical force between back- and back/front-loading. Furthermore, LaFiandra et al. (2003a) also reported no difference in the 2^{nd} peak vertical force between three different backpack designs.

Few studies have reported the medio-lateral forces associated with load carriage. This could be due to the large variability that has been reported (Lloyd et al., 2011), making changes in this force component when carrying load difficult to interpret. The available evidence suggests that mediolateral force is more sensitive to changes in speed than changes in load (Harman et al., 2000, Harman et al., 2001). Birrell et al. (2007) reported significant increases in medio-lateral impulse for load carried in front of the body (rifle carriage), which could imply less stability with this method of load carriage. The authors suggested that an increase in medio-lateral impulse occurs if the load being carried shifts the body's COM further away from its usual position when unloaded. Indeed, reducing the amount the COM is displaced has been shown to increase static stability when supporting a load (Schiffman et al., 2006).

To date, only one study has explored relationships between ground reaction force variables and load carriage economy (Lloyd and Cooke, 2011). Comparing a backpack to a doublepack system, Lloyd and Cooke (2011) reported that a better load carriage economy with a doublepack was associated with a smaller lateral impact peak force and a smaller maximum braking force. Additionally, a smaller difference in maximum braking force between loaded (with 25.6 kg) and unloaded walking with a doublepack is strongly related to improved economy (r = 0.797) (Lloyd and Cooke, 2011). A smaller difference in ground reaction forces from unloaded walking relating to better economy supports the concept that an individual's normal walking gait represents the most economical for that individual (Martin and Morgan, 1992). Individual variation in load carriage economy might, in part, be a consequence of some individuals being able to walk more naturally with certain methods of load carriage than others.

Lloyd and Cooke (2000a) identified large intra- and inter- individual variations in the ground reaction forces for load carriage with a backpack and doublepack. However, they did not quantify this variation as part of their study. Individual variability in the responses of kinetic variables to load carriage have not been reported elsewhere. Given the level of individual variation in load carriage economy and kinematic variables reported by Lloyd et al. (2010c) and Lloyd and Cooke (2011), respectively, it might be expected that a large level of individual variation exists for ground reaction forces during load carriage.

2.9. Subjective perceptions of load carriage

Analysing subjective responses could assist in differentiating between different methods of load carriage when physiological and biomechanical differences are indistinguishable (Legg et al., 1997). Furthermore, subjective perceptions are an important determinant for individuals when selecting a method of load carriage

(Legg et al., 1997). A variety of methods have been employed to assess perceptual responses to load carriage, these include whole body Rating of Perceived Exertion (RPE) (Legg, 1985, Legg and Mahanty, 1985), Differentiated RPE (Kirk and Schneider, 1992), Visual Analogue Scales (VAS) (Ling et al., 2004), Category Rating Scales (CRS) (Mackie et al., 2003) and interviews (Birrell and Hooper, 2007).

As might be expected, RPE scores appear higher when carrying a load compared to unloaded walking and to significantly increase as the mass of a load increases (Gordon et al., 1983). The nature of whole body RPE makes it an overall measure of exertion, which may mask local effects at certain body positions. Thus, more sensitive methods might be required to gain a greater understanding of subjective perceptions associated with subtle differences between load carriage methods. Using differentiated RPE, Kirk and Schneider (1992) showed that despite a constant metabolic cost, the RPE increased throughout an hour of exercise for both the shoulders and legs when carrying a backpack. Visual Analogue Scales allow individuals to rate perceived pain/discomfort at different areas of the body during load carriage (Lloyd et al., 2010d). Mackie et al. (2003) suggested that subjective scores given by participants might reflect psychological characteristics, such as a willingness to use extremes on a scale. However, Lloyd et al. (2010d) argued that raw values provided by the VAS are suitable if a repeated measures design is used as the comparison is effectively intra-participant and the researchers acknowledge that values will mirror an individual's pain sensitivity and psychological characteristics. When the difference between load carriage conditions are minimal (e.g. mass distribution with a load carriage device), more sensitive methods of perceived comfort than VAS might be required. Legg and Mahanty (1985) found that VAS failed to distinguish between a backpack with or without a frame. However, short questionnaires administered immediately after carrying each load indicated that the backpack with the external frame was the easiest load carriage system to put on and take off.

While most research has focused on the perceived exertion associated with back-loading, Lloyd et al. (2010d) compared both head-loading and back-loading. They found a general preference for back loading compared to head loading, in experienced head-loaders, primarily due to the greater feeling of stability. Moreover, using VAS, Lloyd et al. (2010d) provided evidence that head loading invoked a level of neck pain that outweighed the greater discomfort in most other areas of the body associated with back loading. Measures of perceived exertion, particularly VAS scales, might be useful when trying to explain differences between individual gait adaptations to load carriage if there are differences in pain/discomfort. Differences in walking gait adaptations due to pain/discomfort differences could impact individual load carriage economy, although this has not been reported previously in the literature.

2.10. Summary of the literature review

The physiological and biomechanical consequences of load carriage have been widely studied. Despite this, the economy associated with different methods of load carriage remains equivocal. Carrying a load closer to the COM of the body has generally been associated with a better economy compared to loads carried more distally (i.e. in the hands or on the feet). However, literature on the metabolic energy cost associated with methods that place the load close to the body's COM is more equivocal, particularly for head-loading. The early work of Das and Saha (1966), Soule and Goldman (1969) and Datta and Ramanathan (1971) found that energy expenditure for head-loading increased in proportion to the mass of the load. In contrast, Maloiy et al. (1986) and Charteris et al. (1989) showed an energy saving phenomena for head-loading for individuals with experience of using this method. A limitation of these studies is the small sample sizes used ($n \leq 6$); in a larger sample ($n = 24$; 13 experienced and 11 inexperienced head-loaders) Lloyd et al. (2010c) found considerable individual variation in load carriage economy, from which a subset of participants demonstrated the energy saving phenomenon reported by Maloiy et al. (1986) and Charteris et al. (1989). Interestingly, Lloyd et al. (2010c) also found head-loading economy to be independent of head-loading experience, with 38.5% of the experienced head-loaders group being more economical in head-loading than back-loading and 36.4% of an inexperienced group showing the same tendency. Previous work has attempted to identify mechanisms that may contribute to improved head-loading economy (e.g. Jones et al., 1987, Heglund et al., 1995), yet the determinants remain unclear. Much of the research on head-loading has focused on the physiological response and there is very little research examining the associated biomechanics, with those that have focusing on stride frequency (Maloiy et al., 1986, Charteris et al., 1989) and ground reaction forces (Lloyd et al., 2011). A better understanding of the kinematics and kinetics associated with head-loading might help elucidate factors that determine economy with this method.

Energy saving phenomena have also been reported with light loads (~10-15% body mass) carried on the back (Abe et al., 2004) and heavy loads (~ 25 kg) evenly distributed between the front and back of the torso (Lloyd and Cooke, 2000b). It's possible that the energy saving phenomenon reported for these trunk loading methods are associated freedom of movement of the trunk. Abe et al. (2004) suggested that light loads carried on the back might contribute to forward momentum during the gait cycle because they do not constrain posture as much as heavy loads (i.e. lighter loads might allow for a similar level of trunk flexion/extension to unloaded walking). A similar mechanism might be responsible for the improved economy reported for back/front-loading compared to back-loading with heavy loads (Lloyd and Cooke, 2000b), with evenly distributing a load between the front and the back of the torso possibly allowing for a greater freedom of movement in the trunk compared to the same load carried on the back alone. A greater momentum associated with a greater freedom of movement of the trunk might contribute to a lower peak propulsive force, which was reported for back/front-loading compared to back-loading by (Lloyd and Cooke, 2000a). As such, understanding the role of trunk movements in determining load carriage economy warrants further research, particularly for methods that place the load on the torso.

A considerable degree of individual variation has been identified for load carriage economy in head- and back-loading (Lloyd et al., 2010c), and for the change in stride length from unloaded walking for back- and back/front-loading (Lloyd and Cooke, 2011). Few studies have reported individual variation, possibly due to small sample sizes. Individual variation in load carriage economy and the kinematics and kinetics associated with load carriage warrants further attention. It's possible that different factors might combine in individuals to influence economy rather than a single set of factors for each method. If this is the case, further research is needed to establish the nature of the factors and how they interact in individuals. This could help to establish why some individuals appear to be more economical with certain methods of load carriage compared to others.

Returning to the objectives for this research project set out in the introduction chapter, this review of literature has highlighted that the ELI appears to be the most suitable measure of load carriage economy for making comparisons between different individuals and loading methods because it accounts for the metabolic energy cost attributable to unloaded walking. The ELI has been shown to be a valid measure of relative load carriage economy (Lloyd et al., 2010a), but it's reliability is unknown. As such, assessing the suitability of the ELI for the research in this thesis, by investigating its test-retest reliability, was the first objective. The second objective of this research project was to establish the extent of individual variation in load carriage economy and walking gait alterations as a consequence of load carriage, for methods that place load close the centre of mass of the body, or in vertical alignment. This objective was based on the large individual variation in load carriage economy reported by Lloyd et al. (2010c) and the small sample sizes (n < 10) used by studies that have reported energy saving phenomena for load carried on the head (Maloiy et al., 1986, Charteris et al., 1989), back (Abe et al., 2004) and in a doublepack (Lloyd and Cooke, 2000b). This review of literature identified several proposed mechanisms for improved load carriage economy (e.g. Heglund et al., 1995), yet there is a lack of empirical evidence for the relationships between load carriage economy and the walking gait adaptations to load carriage, particularly for methods that place load on the head. As such, the third objective was to identify potential determinants of individual load carriage economy through the analysis of load carriage economy and alterations in walking gait characteristics caused by carrying external load. The final objective of the research in this thesis was to conduct cause and effect trials on the identified determinants of load carriage economy, manipulating the identified key determinants in an attempt to manipulate individual load carriage economy.

Chapter 3. General methods and methodological considerations

3.1. Introduction

This chapter details the methodological considerations for the studies in Chapters 4, 5 and 7, and describes the methods that are consistent across those studies. The individual experimental protocols are detailed in the methods section of each experimental chapter. Data collection for Chapter 5 was conducted prior to this PhD by Professor Ray Lloyd and colleagues from the Cape Peninsula University of Technology (Professor Simeon Davies, Dr Sacha West and Raeeq Gamieldien). As such, the methods of data collection for Chapter 5 are briefly described but the focus is on the secondary analysis of the data, conducted as part of this PhD research.

A summary of the sample size, participant sex and the loading carriage conditions across all studies is provided in Table 7. A justification for the chosen participant population and loading conditions described in each experimental chapter is provided in section 3.3 and 3.6, respectively.

Table 7. A summary of participants and the load carriage conditions for each experimental Chapter.

	Chapter 4	Chapter 5	Chapter 7
Participants	12 males, 5 females	18 females	10 males, 5 females
Load carriage method	Back	Back, Back/Front, Head	Back, Back/Front, Head
Load carriage mass	7 & 20 kg	3, 6, 9, 12, 15 & 20 kg	3, 12 & 20 kg
Head-loading experience	-	Yes (≥ 5 years)	No
Back-loading experience	Yes (≥ 5 years)	Yes	Yes (≥ 5 years)
Back/Front-loading experience	-	No	No

3.2. Ethical approval

All experimental research received ethical approval. Experiments in Chapter 4 were approved by the departmental ethics committee at Leeds Trinity University (Appendix A) and experiments in Chapter 7 were approved by the school ethics committee at Leeds Trinity University (Appendix B). Experiments in Chapter 7 were also approved by the institutional ethics committee at KU Leuven (Appendix C). The nature and purpose of each investigation was clearly outlined in participant information sheets (Appendix D and E) and explained verbally to all volunteers. Each participant completed a health screening questionnaire (Appendix F) and were verbally informed of any potential risks or discomforts before being accepted into the studies. Participants were given the opportunity to ask questions and then gave consent when they were satisfied with the study requirements. Participants were excluded if they were suffering from any musculoskeletal pain/discomfort or had a history of neck or back injury, as this could have been further aggravated by load carriage, and impacted walking gait alterations with load carriage. All participants were asked to provide written and verbal consent (Appendix G and H), while retaining the right to withdraw from the studies without having to provide an explanation. Consent to scientific illustration was obtained when video was recorded in Chapters 4 and 7 (Appendix I).

3.3. Participants

All participants were aged between 18-50 years, apparently healthy, free from any known injury or illness, and had no history of back or neck pain. Participants physical characteristics are detailed in each experimental Chapter. Male and female volunteers were recruited for the research in Chapters 4 and 7. It was considered unlikely that sex differences would exist for load carriage economy because males and females have been reported to have similar waking gait adaptations to load carriage (Silder et al., 2013, Krupenevich et al., 2015). This has since been supported with evidence from Prado-Nóvoa et al. (2019), who found no difference in load carriage economy between males and females for the same relative load, and Godhe et al. (2020), who provided evidence that the dominant factor in the $\dot{V}O_2$ required to carry heavy load (≥ 20 kg) is body mass, not sex differences.

The inclusion criteria for participant recruitment for the research described in Chapter 5 was Xhosa women with a minimum of 5 years of head-loading experience and accustomed to carrying 20 kg on the head. A secondary analysis of the data collected enabled the assessment of the physiological, biomechanical and subjective perception responses to load carriage for African women with several years of head-loading experience, a population for which there is some evidence of a reduced energy expenditure for head-loading (Maloiy et al., 1986) and large individual variation for head- and back-loading economy (Lloyd et al., 2010b).

Load carriage experience was assessed through a questionnaire (Appendix I). The participants in all of the studies reported in this thesis had experience of back-loading and no experience of back/front-loading. The participants in the research in Chapters 4 and 7 had no experience of head-loading. The decision to investigate head-loading economy with inexperienced head-loaders in Chapter 7 was based on evidence from Lloyd et al. (2010c) that showed a large amount of individual variation in head-loading economy for both experienced (ELI value range = ~0.8 – 1.4) and inexperienced (ELI value range = ~0.9 – 1.4) head-loaders. Furthermore, Lloyd et al. (2010c) reported that 36.4% of participants with no head-loading experience exhibited better head-loading economy than back-loading, whilst 38.5% of experienced head-loaders exhibited the same tendency. It was also more feasible to recruit inexperienced head-loaders for Chapter 7, as the study took place in Belgium, where the head-loading method of load carriage is uncommon.

3.4. Preliminary measures

In every experiment, stature was measured using a portable stadiometer (Seca 217, Seca Ltd, UK) on the first visit to the laboratory. Body mass (barefoot) was measured at the beginning of every trial so that dependant variables, such as

$\dot{V}O_2$, could be normalised to body mass or total mass (body mass plus the mass of the load carriage device). Body mass was recoded using calibrated digital scales in the studies described in Chapter 4 (Seca 813, Seca Ltd, UK) and Chapter 7 (Seca 875, Seca Ltd, UK).

3.5. Experimental design

All investigations used a repeated measures experimental design. Unloaded walking was assessed in every trial which allowed for analysis of loaded walking relative to unloaded walking. The independent and dependant variables selected for each experiment are described in the experimental chapters. For all trials, periods of walking lasted 4 minutes, in order to achieve a steady state of oxygen consumption measured in the final minute of each stage. Poole and Richardson (1997) demonstrated that, for a constant work rate at moderate intensity (an intensity where there is an equilibrium between blood lactate production and clearance), healthy individuals achieve a steady state of $\dot{V}O_2$ within 3 minutes. Furthermore, four-minute walking periods have previously been used to assess load carriage economy for walking (Lloyd and Cooke, 2000b, Lloyd et al., 2010b). Where trial conditions were randomised, a Latin square design was used to ensure a balanced order. Participants were always asked to maintain a similar diet and refrain from moderate-vigorous exercise and alcohol consumption in the 24 hours prior to each test.

3.6. Load carriage conditions

3.6.1. Load carriage devices

The research reported in this thesis is focused on the economy associated with load carriage methods that position the load close to, or in vertical alignment with, the centre of mass of the body. Specifically, back-loading, combined back and front-loading, and head-loading were chosen for the research in this thesis because evidence of a reduced metabolic energy expenditure has been reported for each of these methods (Abe et al., 2004, Lloyd and Cooke, 2000b, Maloiy et al., 1986, Charteris et al., 1989). Abe et al. (2004) provided evidence of a reduced metabolic energy expenditure for back-loading using a backpack device, however this device was not described in detail by the authors (the make and model of the backpack was not provided). Back-loading in this thesis used commercially available backpacks with hip belts for support. The backpack devices differed between studies but all shared common design features including an internal frame, hip belt and adjustable padded shoulder straps. The research in Chapter 4 used two backpack devices, one to carry 7 kg (Prototype Featherlight Freedom - without front balance pockets, AARN, New Zealand) and another to carry 20 kg (Jura 35, Karrimor, UK). Two separate backpacks were used to avoid adjusting the mass of the packs between load mass conditions (load mass conditions are described in the next section, 3.6.2), which could have altered the position of the load within the device between the test/retest conditions.

The backpack device used for the research in Chapter 5, conducted prior to this PhD, was a Karrimor device (Karrimor, UK). The backpack device used in Chapter 7 (Aeon pack, Lowe Alpine, USA) was smaller (25 litres) than the backpack devices used in Chapter 4 (35 litres and 50 litres), to allow anatomical markers to be positioned on the posterior superior iliac spine for three-dimensional motion capture. The technical specifications of the backpacks used in Chapter 4 and Chapter 7 can be seen in Appendix K.

Lloyd and Cooke (2000b) provided evidence of a reduced metabolic energy expenditure for carrying heavy load (25.6 kg) with a device that evenly distributed load between the back and front of the torso (back/front-loading) compared to back-loading alone. The back/front-loading device they used was an AARN balance pack (AARN design ltd, New Zealand) which is a back-loading system with front balance pockets that attach to the shoulder straps and hip belt. This type of load carriage design allows for a load to be evenly distributed between the anterior and posterior of the torso. The study described in Chapter 5 used an AARN balance pack (Prototype Featherlite Freedom, AARN design ltd, New Zealand) similar to the one used by Lloyd and Cooke (2000b). For the study described in Chapter 7, the back/front-loading device was made up from the Aeon backpack, used for the back-loading condition, with balance pockets from the prototype AARN balance pack, attached to the shoulder straps and hip belt of the backpack. This allowed anatomical markers to be positioned on the posterior superior iliac spine for three-dimensional motion capture. For the back- and back/front-loading methods, shoulder straps and hip belts were tightened to participant comfort.

For head-loading, neither Maloiy et al. (1986) or Charteris et al. (1989) describe the device they used for direct head loading in any detail. Other research on the direct head-loading method has used a plastic bucket (Lloyd et al., 2011) or a crate (Lloyd et al., 2010b, Lloyd et al., 2010c) to carry load on the head. The studies described in Chapter 5 and 7 used a 20-litre plastic bucket with a piece of cloth used as a cushion between the head and the bucket.

3.6.2. Load mass

Load carriage research has used absolute loads (e.g. Harman et al., 2000) and relative loads representing a percentage of body mass (e.g. Lloyd et al., 2010b). Absolute load was preferred to relative load in all experiments in this thesis because absolute load was deemed more ecologically valid. Individuals are unlikely to pack load carriage systems to a percentage of their body mass, particularly personnel in the military services and individuals living in developing countries. For these populations, the mass of the load is dependent on the task and not the individual’s body size. For example, in some developing countries, lack of safe water access results in domestic water carrying that typically consists of 20-25 litres being transported per trip, using the head-loading method (Geere et al., 2010).

The study in Chapter 4 included loads of 7 kg and 20 kg, to represent a light and heavy load, respectively. Loads of 10% of body mass (Lloyd et al., 2010b), 6 kg (Harman et al., 2000) and 8 kg (Birrell and Haslam, 2010) have been previously used in load carriage research to represent a light load. As such, the mass of the light load for this research is in line with published literature (7 kg is 10% of body mass for a 70 kg participant). The study was originally designed with 35 kg for the heavy load (50% of body mass for a 70 kg individual), however, it became clear in pilot testing that some individuals could not complete the protocol with a load mass above 20 kg. Previous research, particularly those on military personnel, have used loads in excess 40 kg (Harman et al., 2000), however, 20 kg loads have been used to represent a heavy load in the literature (e.g. Lloyd et al., 2011, Birrell and Haslam, 2009) and deemed appropriate due to the untrained nature of some participants in terms of load carriage. The mass of the loads in Chapters 4 and 5 were made up of the load carriage system plus sandbags, measured to the nearest 50 g using digital scales (Seca 813, Seca Ltd, UK). For the research in Chapter 4, the sandbags were placed in plastic food storage containers to evenly distribute the load within the load carriage device. Three food containers containing sandbags, each 3.6 litres in size, were stacked vertically inside each backpack device.

The experiments described in Chapter 5 used loads of 3, 6, 9, 12, 15 and 20 kg, which enabled the investigation of load carriage physiology, biomechanics and subjective perceptions across a range of light and heavy loads. The study in Chapter 7 included loads of 3, 12 and 20 kg. These loads were chosen because they represented a very light (3 kg), medium (12 kg) and heavy (20 kg) load from the range of loads investigated in Chapter 5. The mass of the loads in Chapter 7 were made up of the load carriage system plus rubber weights (between 1 and 5 kg) and metal weights from a Monark cycle ergometer (Monark Exercise AB, Sweden) (between 0.1 and 0.5 kg), to the nearest 100 g using digital scales (Seca 875, Seca Ltd, UK). Plastic food storage containers (3 litres in size) were used to evenly distribute the load in the back-loading conditions (backpack and back-loading component of the back/front method). Weights were placed directly in the bucket for the head-loading method and directly in the front balance pockets for the back/front-loading method. The study in Chapter 7 used a portable gas analysis system (Oxycon Mobile, Jaeger, Germany), because it was the only system available at the location of the experiments. The portable system was carried on the anterior of the trunk in all load carriage conditions and had a total mass of 1kg (including housing vest). For the Back/Front-loading method, the portable gas analysis system sat in the centre of the chest, with the front balance pockets positioned on the shoulder straps either side. Figure 1 shows the configuration of the front balance pockets and portable gas analysis system in the Front/Back-loading condition in Chapter 7.

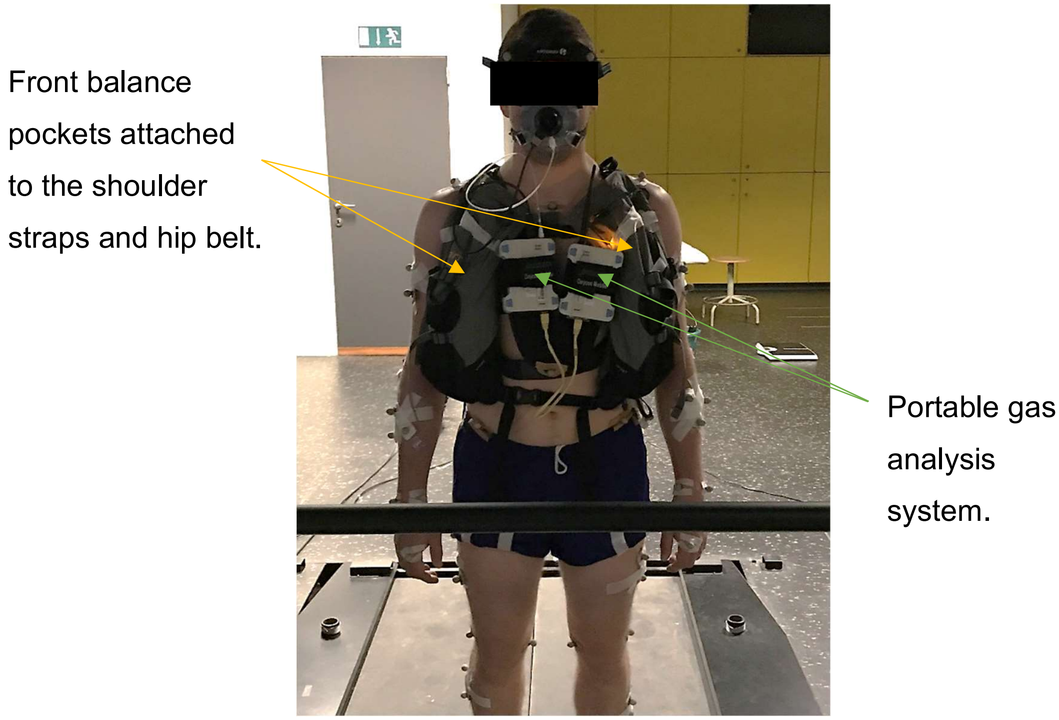


Figure 1. Balance pocket configuration for Back/Front-loading in Chapter 7.

3.7. Walking speed

Changes to walking gait mechanics during load carriage have been investigated at controlled speeds (e.g. Huang and Kuo, 2014, Harman et al., 2001) and self-selected speeds (e.g. Attwells et al., 2006, Majumdar et al., 2010). Although allowing participants to adjust their self-selected speed in response to the load would be more ecologically valid, particularly when investigating recreational load carriage, doing so it makes it difficult to decouple the effects of load and walking speed. As such, controlled walking speeds were used for the research described in the thesis. The walking trials described in Chapters 5 and 7 were conducted at 3 $km \cdot h^{-1}$. This speed was selected based on the work of Maloiy et al. (1986), Charteris et al. (1989), Abe et al. (2004) and Lloyd and Cooke (2000b) who provided evidence for improved load carriage economy at speeds of ~3 $km \cdot h^{-1}$. Furthermore, this energy saving phenomenon has not been reported at faster walker speeds (Abe et al., 2004). The walking trials described in Chapter 4 were conducted at a range of walking speeds (3 $km \cdot h^{-1}$, 6 $km \cdot h^{-1}$ and a self-selected pace) to provide a robust assessment of the reliability of the ELI. All walking trials were conducted on motorised treadmills (Chapter 4: Mercury, HP Cosmos, Germany; Chapter 5: Genesis, South Africa; Chapter 7: Forcelink, Motekforce, Netherlands). Walking on a treadmill was preferred to over-ground because it allowed for speed to be tightly controlled over each period of walking, which made it easier to achieve steady-state conditions for the measurement of load carriage economy. Furthermore, any variability in gait across strides was then independent of variations in average walking speed and terrain.

3.7.1. Treadmill speed verification

In Chapter 4, treadmill speed was verified at 3 $km \cdot h^{-1}$, 4 $km \cdot h^{-1}$, 5 $km \cdot h^{-1}$ and 6 $km \cdot h^{-1}$ prior to the start of the investigation. The procedure for this involved measuring the length of the treadmill belt (3.33 metres), measuring the total distance travelled by the belt in 20 revolutions (66.6 metres) and the time taken for the belt to complete 20 revolutions at the four speeds. Speed was the calculated using the known formula:

$$\text{Speed} = \text{Distance}/\text{Time} \qquad \text{Equation 4}$$

This process was repeated 3 times at each speed with and without an 84 kg male carrying a heavy (20 kg) rucksack (combined mass = 104 kg) (trial 1). This process was then repeated one week later (trial 2), and the results are presented in Table 8. Raw data from the treadmill verification process for the study in Chapter 4 is presented in Appendix K.

In Chapter 7, treadmill speed was verified at 3 $km \cdot h^{-1}$ because this was the walking speed used in that experiment (Table 9). The procedure for measuring treadmill speed was identical to the procedure described above for the research in Chapter 4, except that the protocol was not repeated after seven days because the focus of the investigation in Chapter 7 was not test-retest reliability. The treadmill used for the experiment described in Chapter 7 was a dual-belt treadmill, and the speed of both belts was assessed. An 86.2 kg male participated in the treadmill speed verification. The participant walked on the treadmill unloaded and carrying a 20 kg rucksack (total combined mass = 106.2 kg).

Table 8. Mean ± SD treadmill speeds for the test-retest verification and reliability in experimental Chapter 4.

	Unloaded treadmill			Loaded treadmill (104 kg)		
	Trial 1	Trial 2		Trial 1	Trial 2	
Displayed speed ($km{\cdot}h^{-1}$)	Actual speed ($km{\cdot}h^{-1}$)	Actual speed ($km{\cdot}h^{-1}$)	Mean difference from displayed speed ($km{\cdot}h^{-1}$)	Actual speed ($km{\cdot}h^{-1}$)	Actual speed ($km{\cdot}h^{-1}$)	Mean difference from displayed speed ($km{\cdot}h^{-1}$)
3.00	3.04 ± 0.00	3.03 ± 0.00	0.03	2.95 ± 0.01	2.96 ± 0.00	-0.05
4.00	4.08 ± 0.00	4.05 ± 0.04	0.06	4.01 ± 0.01	4.01 ± 0.01	0.01
5.00	5.09 ± 0.00	5.09 ± 0.00	0.07	5.06 ± 0.01	5.04 ± 0.01	0.04
6.00	6.07± 0.01	6.08 ± 0.01	0.07	5.99 ± 0.00	6.03 ± 0.01	0.01

* Trial 1 = Initial verification test; Trial 2 = Repeat verification test

Table 9. Mean ± SD Treadmill speeds for verification of treadmill speed in the experiment reported in Chapter 7.

	Displayed Speed ($km{\cdot}h^{-1}$)	Unloaded Treadmill Speed ($km{\cdot}h^{-1}$)	Unloaded Walking Speed ($km{\cdot}h^{-1}$)	Loaded Walking Speed ($km{\cdot}h^{-1}$)
Belt 1	3.00	3.03 ± 0.03	2.99 ± 0.01	2.99 ± 0.00
Belt 2	3.00	3.03 ± 0.01	2.99 ± 0.01	2.99 ± 0.01

* Mass for unloaded walking = 86.2 kg; Mass for loaded walking = 106.2 kg

Chapter 5 involved a secondary analysis of data collected prior to the beginning of this PhD. The verification of treadmill speed for Chapter 5 was completed in an manner similar to those described above, as verified by the principal investigator who is one of the supervisors of this work.

3.8. Familiarisation protocol

Every experiment included a familiarisation period prior to completing the first trial, to reduce any possible effects of trial order and ensure participants were able to complete all load carriage conditions. During the familiarisation, participants were screened for any potential contraindications to exercise, and for the studies described in Chapters 5 and 7, participants were asked to complete questionnaires relating to their load carriage history. Participants were then habituated to the experimental protocol and equipment. A typical habituation period lasted approximately 20 minutes and involved participants waking on the treadmill at the walking speed(s) for the experiment, with and without each of the load carriage systems. The facemask for the online gas analysis system was also fitted, in order for participants to become accustomed to breathing through it.

3.9. Physiological measurements

3.9.1. Collection and analysis of expired air

Expired air variables were measured using online gas analysis systems. Different systems were used in Chapters 4, 5, and 7 because the studies described in each Chapter took place in different laboratories (Chapter 4: Metalyzer 3B, Cortex, Germany; Chapter 5: K4b2, COSMED, Italy; Chapter 7: Oxycon Mobile, Jaeger, Germany). Despite the different manufacturers, the online gas analysis systems measured pulmonary gas exchange in the same way by measuring expired gas volumes continuously using a volume transducer fixed to a facemask and measuring expired oxygen and carbon dioxide concentrations sampled continuously through a sample line attached to the volume transducer. Fifteen minutes prior to the commencement of exercise in each trial, the online gas analysis systems were calibrated in accordance with the manufacturer's guidelines. Briefly, this involved running a calibration reference gas through the systems via a sample line (Metalyzer 3B system = 17.05% O_2, 4.98% CO_2, balance N_2; Oxycon Mobile system = 15.99% O_2, 5% CO_2, balance N_2) and then verifying the calibration gas against ambient air (20.93% O_2 and 0.03% CO_2). A volume calibration was then performed for both devices using a standard 3-litre syringe (Hans Rudolph, Inc., USA). Expired air was collected continuously throughout each period of exercise. On the completion of each test, the data was averaged for 60-second intervals and data from the final minute of exercise was analysed in line with previous load carriage economy studies (Lloyd et al., 2010b, Lloyd et al., 2010a, Hinde et al., 2017). Means and standard deviations were calculated for $\dot{V}O_2$ ($l \cdot min^{-1}$) and any other pulmonary gas exchange variables of interest. Relative load carriage economy was calculated using the ELI as follows:

$$ELI = \frac{mlO_{2L} \cdot kg\ total\ mass^{-1} \cdot min^{-1}}{mlO_{2U} \cdot kg\ body\ mass^{-1} \cdot min^{-1}} \qquad \text{Equation 5}$$

where mlO_{2L} refers to oxygen consumption when carrying an additional load and mlO_{2U} refers to oxygen consumption when walking unloaded. The ELI was preferred to other measures of load carriage economy such as $\dot{V}O_2$ in absolute or relative (to body mass or total mass) terms and net energy cost, which subtracts the $\dot{V}O_2$ required for standing from the $\dot{V}O_2$ required for walking, because the ELI accounts for the energy expenditure required for unloaded walking and provides a single value for economy. The reliability of the ELI is evaluated in Chapter 4 of this thesis.

3.9.2. Heart rate

Heart rate (HR) was measured continuously throughout all trials in each experimental study using a Polar heart rate monitoring system (Polar, Finland). In Chapters 4 and 7, heart rate was measured during the rest period to monitor recovery and ensure participants had returned to their resting HR before beginning the next bout of load carriage.

3.10. Subjective Perceptions

Subjective perceptions of load carriage were recorded and analysed in all experimental studies. Ratings of Perceived Exertion (RPE) and Visual Analogue Scales (VAS) were used to measure perceptions of whole-body exertion and localised pain/discomfort, respectively. Ratings of perceived exertion were measured using a whole body RPE scale (Borg, 1982) and has been used in previous load carriage research (e.g. Holewijn et al., 1992, Lloyd et al., 2010d). Visual Analogue Scales (VAS) were used to measure the participant's perceived degree of pain/discomfort for fifteen areas of the body, following each period of exercise. The VAS used were identical to those created by Lloyd et al. (2010d), consisting of body pictures with clearly shaded areas and a 100mm scale below each image with anchor points of 'No Pain' at one end and 'Pain as bad as it could possibly be' at the other end (Figure 2). The sheets were laminated, and participants were asked to mark each VAS using a fine point washable marker pen, to indicate the pain/discomfort. The marked point on each scale was then measured to the nearest millimetre to identify a pain score for each region of the body (Neck, back of shoulders, front of shoulders, chest, upper back, abdomen, lower back, hips, buttocks, front of thigh, back of thigh, knees, lower leg, ankles and feet).

On the Scale below indicate the amount of pain/discomfort you feel in the area indicated

<u>Front of Thigh</u> ***<u>Back of Thigh</u>***

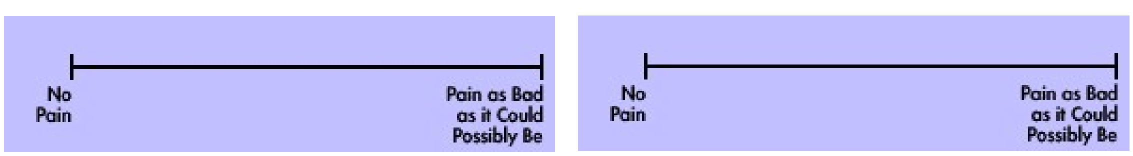


Figure 2. Sample VAS data collection sheet (Quadriceps and Hamstrings).

3.11. Kinematic measurements

Previous research has analysed the kinematics of load carriage using two-dimensional (2D) video-based motion analysis (e.g. Lloyd and Cooke, 2011, Harman et al., 2000) and three-dimensional (3D) motion analysis (e.g. Birrell and Haslam, 2009, Wills et al., 2019). Two-dimensional video-based motion analysis was used to record sagittal plane kinematics for the studies presented in Chapters 4 and 5. The deterministic model developed in Chapter 6 highlighted the need for a 3D analysis of load carriage kinematics for the research in Chapter 7, to gain a better understanding of the potential biomechanical determinants of load carriage economy.

3.11.1. Two-dimensional motion analysis

3.11.1.1. Filming procedures

Sagittal plane kinematics were assessed using a standard digital video camera (Chapter 4: Casio EX-ZR700, Japan; Chapter 5: Panasonic, Japan), set at 50Hz and placed perpendicular to the treadmill. To avoid aliasing error, the sampling frequency used for the studies in Chapters 4 and 5 were roughly ten times greater than the anticipated highest frequency in the signal (Payton and Burden, 2017). In both studies, the distances from the lens of the camera and the treadmill and the lens of the camera and the floor were measured and kept constant for all trials. A calibration instrument (1m x 1m) for the vertical and horizontal axis was placed on the treadmill and recorded prior to each trial for the study in Chapter 4. For the study in Chapter 5, the treadmill was marked on the vertical (0.5m) and horizontal (1m) axis and recorded prior to each trial for calibration. Video footage was collected during the final 60 seconds of each exercise stage, in line with the analysis of expired air data.

3.11.1.2. Marker placement

Superficial joint markers were placed on the shoulder, hip, knee, ankle and toe on the side of the body facing the camera, in order to measure trunk, hip, knee and ankle angles at heel-strike and toe-off. Data from Plagenhoef (1971) were used to identify the exact position of each marker. The shoulder marker was placed 5cm inferior to the acromion process, midway between the anterior and posterior surface. The hip marker was placed 3 cm superior and 1 cm anterior to the greater trochanter. The knee marker was placed at the midpoint of the femoral epicondyles. The ankle marker was placed at the most distal point of the lateral malleolus. The toe marker was placed on the lateral side of the head of the 5th metatarsal.

3.11.1.3. Digitising procedures

Video files collected for the research in Chapters 4 and 5 were uploaded to SIMI motion (SIMI motion 8.5.6, Germany) to be manually digitized by a single observer. Intra-observer reliability of digitising was assessed by repeatedly digitising a single video frame at heel-strike and toe-off 10 times each for one participant. Intra-observer measurement error of digitising was assessed using a calculation for technical error of measurement (TEM) (Goto and Mascie-Taylor, 2007). This process was repeated one week later to assess day-day measurement error.

Absolute TEM = $\sqrt{\frac{\sum D^2}{2n}}$ Equation 6

where D is the difference between the two measurements taken on the independent measurements and n is the number of measurements used. Absolute TEM was then be transformed into relative TEM in order to obtain the error expressed as a percentage corresponding to the total average of the variable analysed. Relative TEM was calculated according to the following equation:

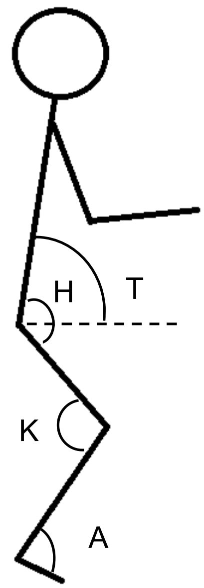
Relative TEM (%) = $\frac{\boldsymbol{TEM}}{\bar{\boldsymbol{x}}}$ x 100 Equation 7

where $\bar{x}$ is the variable average value. A relative TEM of less than 1% was deemed as acceptable (Perini et al., 2005). The results for digitising reliability are provided in each experimental chapter and the raw data is available in Appendix M and N.

Following tests of intra-observer reliability, the calibration files for each trial were digitised to provide a scale for the trial videos. Videos for each trial were then manually digitised, frame by frame, to obtain two-dimensional coordinates for each of the anatomical landmarks specified with markers, for six consecutive strides. The analysis of six stance phases is in line with other load carriage research that have investigated the biomechanics of load carriage using between 3-10 stance phases to achieve a representative sample of the walking gait (Lloyd and Cooke, 2011, Harman et al., 2000, Silder et al., 2013, Birrell and Haslam, 2009, Wills et al., 2019, Chow et al., 2005). Once the reconstruction was complete, joint angles were calculated by the software for each step at two events of the step cycle (heel-strike and toe-off). Step events were visually identified from the video footage. Heel-strike was identified as the frame where the foot appeared to make contact with the treadmill and toe-off was identified as the frame where the foot appeared to no longer be in contact with the treadmill. Arellano et al. (2009) indicated that the walking gait pattern while carrying external load is less stable in the sagittal plane during the stance phase than the swing phase. As such, the focus of the analysis was the stance limb because this appears to be the part of the gait cycle where the body directly experiences the effects of additional load in the sagittal plane. Joint angle excursion was measured as the change in joint angle from heel-strike to toe-off in each step.

3.11.1.4. Joint angle and spatiotemporal measurements

Joint angle kinematics and spatiotemporal gait parameters measured in the research in Chapters 4 and 5 were selected for analysis based on previous literature assessing the sagittal plane biomechanics associated with backpacks and back/front-loading (e.g. Lloyd and Cooke, 2011, Harman et al., 2000, Kinoshita, 1985). Figure 3 illustrates the sagittal plane joint angles used to analyse dynamic posture for the studies described in Chapter 4 and Chapter 5. Angles of the trunk, hip, knee and ankle joints were measured. Trunk forward lean was measured as the angle of the trunk from the horizontal. Therefore, 90º represents a vertical trunk position and angles less than 90º indicate forward lean.



T = Trunk angle: angle between trunk and a horizontal line.

H = Hip angle: the absolute angle between the thigh and the trunk segments.

K = Knee angle: dorsal angle between the shank and thigh segments.

A = Ankle angle: the absolute angle between the foot and the shank segments.

Figure 3. Sagittal plane joint angles used to analyse posture through the stride cycle.

Gait events were visually identified from the video footage. Heel-strike was identified as the frame where the foot appeared to make initial contact with the treadmill and toe-off was identified as the frame before the foot appeared to no longer be in contact with the treadmill. For the research presented in Chapter 4 and Chapter 5, step cadence was defined as the average step time across 12 consecutive steps. Step length was then calculated from the known walking speed and step cadence by dividing the speed of the treadmill by step cadence. Stance time was calculated as the average contact time over the six strides on the right leg. The contact time for each stance phase was measured as the time between initial contact of the right foot to the final frame before the right foot broke contact with the treadmill belt, at the point of take-off. Step time was measured as the duration of time taken from one-foot contact to the consecutive contralateral foot contact. Durations of double stance were measured as the time taken between heel-strike with one foot to the consecutive toe-off with the opposite foot. Single stance durations were measured as the time taken between toe-off with one foot to the subsequent heel-strike with the same foot.

3.11.2. Three-dimensional motion capture

Three-dimensional motion capture was used to record the 3D kinematics of load carriage for the research in Chapter 7. The methodological considerations for recording and analysing 3D kinematics of the whole-body, are outlined in this section. The experimental protocol used to record 3D kinematics is described in the method section of Chapter 7.

3.11.2.1. Coordinate systems

To quantify the three-dimensional motion of each participant, reference systems were fixed to the environment and each body segment. The reference system fixed to the environment is known as the Global Coordinate System (GCS). The GCS represents the three-dimensional space in which motion capture occurs, also known as the capture volume. The reference system fixed to each body segment is known as the Local Coordinate System (LCS). Two methods to track the position and orientation of the segment LCS are the six degrees of freedom (6 DOF) method (also known as segment optimisation) and Inverse Kinematics (IK) (also known as global optimisation) (Robertson et al., 2013). The six DOF method refers to the independent coordinates required to characterise a body, or systems, position (Zatsiorsky, 1998). A rigid body, freely suspended, has a maximum of six degrees of freedom. It can translate along, and rotate about, three independent axis (longitudinal, vertical, frontal). With the six DOF method, each segment requires a minimum of three non-collinear tracking markers in order to define the segments' position and orientation (Bartlett and Payton, 2008). For example, markers can be placed on the proximal and distal ends of a bone to define a segment, and a third non-collinear marker can be used to define the orientation of the vector between the two endpoints. The six DOF approach tracks each segment independently, decoupling the calculation of a segment's orientation from an adjacent segment, without imposing joint constraints (Schmitz et al., 2016). This method assumes that segments are linked implicitly by the motion capture data and that the segments will not dislocate because the participants joints did not come apart when the motion was captured. However, because the six DOF method does not constrain the endpoints of the proximal and distal segment, some segment dislocation can occur, predominantly as a result of soft tissue artefact (Lu and O'connor, 1999, Leardini et al., 2017). Soft tissue artefact is the discrepancy between the movements of the marker attached to the skin moving, relative to the underlying bone (Bartlett and Payton, 2008). It can result from muscle contraction and relaxation, and skin sliding across joints, particularly joints with large rotations (Leardini et al., 2017).

The IK method uses a least squares approach to search, in each data frame, for an optimal pose of the multi-link model that minimises the differences between the measured and model-determined marker coordinates across all body segments (Lu and O'connor, 1999). In contrast to the six DOF method, the IK approach includes joint constraints that restrict the relative motion between segments, minimising the effects of surface tissue artefact and measurement error. However, it is important to determine the appropriateness of applying joint constraints. IK is an extension of the six DOF position and orientation estimation because if a joint is ascribed six degrees of freedom, the outcome is the same as the six DOF method. For analysis of load carriage and the walking gait, the six DOF method was preferred to IK because IK uses a best fit across all the markers, so the head markers can affect the position and orientation estimation of the feet segments, and vice versa.

3.11.2.2. Marker set

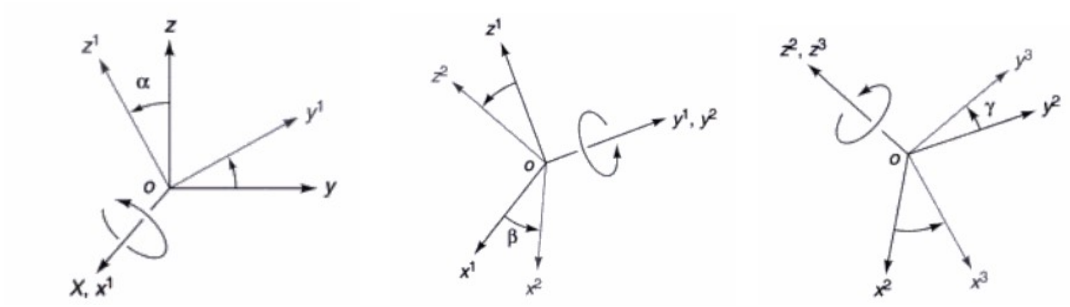
For three-dimensional analysis, a marker set is simply a configuration of markers that are used to establish the LCS for a body segment. The British Association of Sports and Exercise Science (Payton and Burden, 2017) recommended, based on the work of Cappozzo et al. (1995), that marker sets should adhere to the following criteria:

- A minimum of three non-collinear markers are required per rigid segment.
- Movement should be minimised between markers and the underlying bone.
- Markers should be clearly visible to at least two cameras at every instant during the recording.

Standardised marker sets, such as the Plug-in Gait and Helen Hayes model (Davis et al., 1991, Kadaba et al., 1990) have been developed to analyse the walking gait. An advantage of these marker sets is that they have been tested in many laboratories and included in many studies over the past 30 years. A limitation of these marker sets is that they use a relatively small number of markers, particularly on the foot and shank which only have two markers each. As such, while these marker sets may be good for analysing the walking gait, they may not be valid for many other sporting movements. An alternative method is to design a custom marker set. This would allow the limitations of the standardised marker sets to be overcome. A limitation of developing a custom marker set is that additional measures of accuracy and reliability would be required to ensure the marker set is appropriate. As the research in this thesis is based on the walking gait, the Plug-in gait, based on the work of Bell et al. (1990), Davis et al. (1991), Kadaba et al. (1989) and Kadaba et al. (1990), was deemed appropriate for the analysis of upper and lower body movements for the research in Chapter 7. Additional markers were included to improve segment tracking and the identification of joint centres at the knee and ankle.

3.11.2.3. Calculation of joint angles

Two common methods to measure joints angles in three-dimensional space are Cardan/Euler angles (Davis et al., 1991) and helical angles (Woltring, 1991). The most widely used method are Cardan/Euler angles (Robertson et al., 2013). Using a Cardan rotation sequence, the orientation of one LCS with respect to another LCS can be represented by three successive rotations about unique axis (Figure 4). The choice of rotation order can affect the joint angles calculated and the x-y-z rotation sequence, recommended by the International Society of Biomechanics (Wu and Cavanagh, 1995), is the most commonly used in biomechanics research (Robertson et al., 2013). The Cardan rotation sequence x-y-z involves the first rotation about the x-axis, which leads to new orientations of the y- and z-axes (y^1 and z^1). The x-axis stays in the same orientation and becomes x^1. The second rotation about the y^1-axis leads to new positions for the x^1- and z^1-axes (x^2 and z^2). The third rotation about the z^2-axis leads to new orientations for the x^2- and y^2-axes (x^3 and y^3) (Robertson et al., 2013).



a. b. c.

Figure 4. Cardan x-y-z rotation sequence. First (a) the x-axis about the stationary coordinate system (α); then (b) about the new y^1-axis (β); finally (c) about the z^2-axis (γ) (Robertson et al., 2013).

Helical angles are an alternative method of defining the orientation of one LCS to another LCS (Woltring, 1991). This technique is based on the finite helical axis (Woltring et al., 1985) in which a position vector and an orientation vector are defined. Briefly, a finite helical axis is defined from the translation (t) and rotation (θ) about the helical axis (n) from point (P_1) to point (P_2) (Figure 5) (Spoor and Veldpaus, 1980, Woltring et al., 1985).

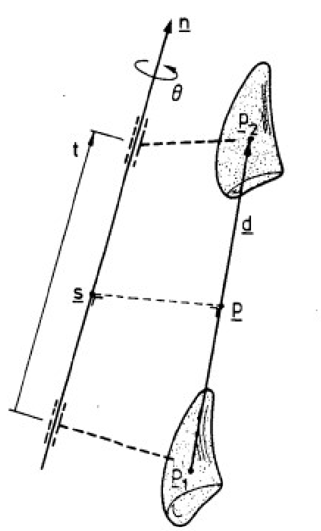


Figure 5. A finite helical axis (Woltring et al., 1985)

An advantage of the finite helical method is that it does not exhibit gimbal lock, which occurs when two axis systems achieve the same position. Robertson et al. (2013) suggested that this is a bigger issue in the upper extremity than the lower extremities and is unlikely to occur for the walking gait, which involves minimal motion of the upper extremities. An advantage of Cardan angles over helical angles is that helical angles are very sensitive to noisy coordinate data, which needs to be substantially smoothed before helical angles can be calculated (Robertson et al., 2013). Cardan angles are also widely used in biomechanics and provide a well-understood representation of joint angles (Robertson et al., 2013), which allows for direct comparisons between studies that have used Cardan angles. As such, Cardan angles with an x-y-z rotation sequence were used for the research in Chapter 7

3.11.2.4. Body segment inertial parameters

For the study in Chapter 7, body segment inertial parameters (BSIP) (mass, centre of mass and moment of inertia) were estimated using data from Zatsiorsky (1990) that were adjusted by De Leva (1996). Most early methods developed to estimate body segment parameters are based on cadaver studies (Dempster, 1955, Clauser et al., 1969, Chandler et al., 1975). A disadvantage of these methods is that the density of tissue of cadavers may differ from living tissue. The decision to use the De Leva (1996) model to estimate BSIP was based on Zatsiorsky (1990) using a gamma ray scanning technique to quantify the density of each segment of live, young males and females (100 males, 15 females; mean ages: 24 and 19 years, respectively), enabling estimations of mass and COM for each segment. The age of the population used by Zatsiorsky (1990) is similar to the participants in the studies in this thesis, which is important because muscle mass and bone density decreases as the body ages, decreasing the density of body segments. However, Zatsiorsky (1990) used bony landmarks as reference points to define the segment lengths with some of the bony landmarks a considerable distance from the actual joint centre (De Leva, 1996). De Leva (1996) adjusted the mean relative COM positions and the radii of gyration from Zatsiorsky (1990) data so that they were referenced to joint centres instead of bony landmarks. Another method of estimating inertial properties of human body segments is mathematical modelling (Hanavan Jr, 1964, Hatze, 1980). This method models segments as rigid bodies represented by geometric shapes and assumes that mass is uniformly distributed within each segment. An advantage of scanning and imaging techniques, such as Zatsiorsky (1990), over mathematical modelling is that they quantify the density of each segment.

3.12. Signal processing

Common smoothing methods for raw digitized data in biomechanics include digital low pass filters (e.g. Butterworth filter), spline curve fitting (e.g. Quintic spline) and frequency domain techniques (e.g. Fourier series truncation) (Winter, 2009). Research on walking gait and load carriage kinematics typically use a Butterworth filter to low-pass filter displacement data with a cut-off frequency of 6 Hz (e.g. Lloyd and Cooke, 2011, Dames and Smith, 2015). As such, a 2nd order Butterworth filter was applied to raw data in the studies described in Chapters 4, 5 and 7. A limitation of low pass filters, such as the Butterworth filter, is that they are inefficient when processing signals with frequencies that vary dramatically over time, such as the dramatic accelerations and decelerations associated with kicking a soccer ball (Nunome et al., 2006). However, the signal frequencies associated with the loaded and unloaded walking do not change dramatically over time as with quicker movements like sprinting or kicking a soccer ball, therefore low pass filtering appears appropriate for load carriage at walking speeds.

For the kinematic data in Chapters 4, 5 and 7, the residual analysis method recommended by Winter (2009) was used to determine the appropriate cut-off frequencies by comparing the difference between raw and filtered signals over a range of cut-off frequencies (1 – 20 Hz). Figure 6 shows a theoretical plot for the residual between filtered and unfiltered signals over a range of cut-off frequencies (f_c). In order to estimate the optimal cut-off frequency from the residual plot, a line is drawn from point *e* (residual at the highest cut-off frequency) that mirrors the gradient of the plot at the higher frequencies (*e* to *d*) and continues until intercepting the vertical axis at point *a*. A second line is then drawn perpendicular to the vertical axis at point *a.* Finally, a vertical line is drawn from point *b*, where the drawn perpendicular line from point *a* on the vertical axis intersects the residual line (point *b*), to the horizontal axis. The f_c^1 frequency represents the chosen optimal cut-off frequency with *bc* representing the signal distortion at this frequency.

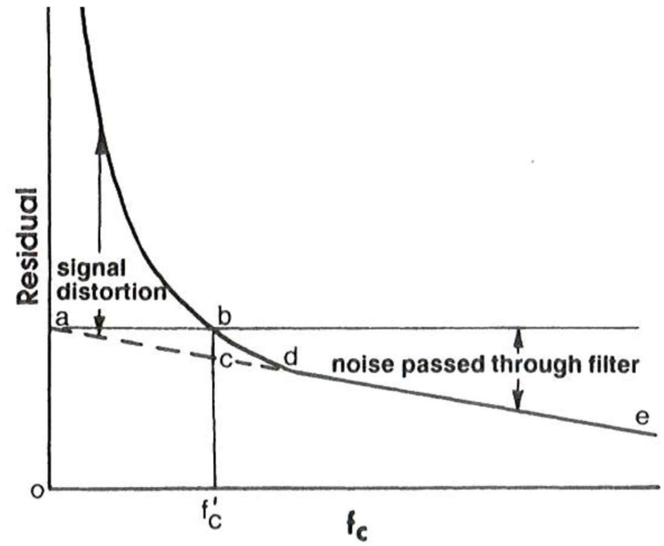


Figure 6. Residual plot between an unfiltered and filtered signal as a function of the filter cut-off frequency. The cut-off frequency is shown on the horizontal axis (fc). The residual is shown on the vertical axis (mm). Image taken from Winter (2009).

A residual analysis was performed on horizontal and vertical displacement data for one joint marker on each segment digitised for studies in Chapters 4 and 5, and for one tracking marker for each of the 15 segments modelled for the study in Chapter 7. A residual analysis calculator provided by the British Association of Sport and Exercises Sciences (BASES) (https://members.bases.org.uk/spage-resource_library-practitioner_and_researcher_resource_centre.html) was used to estimate the optimal cut-off frequencies for each joint angle/body segment. In each experimental chapter, a comparison was made between three separate participants using the same loading method to estimate the appropriate cut-off frequencies for each segment. The results from the residual analysis for each experimental chapter and the raw data are available in Appendix L. The estimated cut-off frequencies were similar (within 2 Hz) between participants for both the horizontal and vertical displacement values from markers on the same segment/joint angle for all experimental Chapters. Based on these data, a cut-off frequency of 6 Hz was determined to be most appropriate for each experimental Chapter and was used for all participants across all conditions. Further, the estimated cut-off frequency of 6 Hz in the present research is in line with the low pass filter frequency of 6 Hz used in previous literature on walking with load carriage (e.g. Dames and Smith, 2015, Lloyd and Cooke, 2011, Vickery-Howe et al., 2020).

Using different cut-off frequencies for force and position data can cause artefacts, particularly for high impact movements (Bisseling and Hof, 2006, Kristianslund et al., 2012). Kristianslund et al. (2012) suggested that force and movement data should be processed with the same filter and at the same cut-off frequency in order to reduce error. As such, the kinetic data presented in Chapter 7 were filtered at the same frequency as the kinematic data in the study in Chapter 7, using a low pass second order Butterworth filter.

3.13. Data analysis

In line with the conclusions of Lloyd et al. (2010a), all physiological, kinematic and kinetic data measured during loaded walking trials were analysed as the change from (Δ) unloaded walking, to accommodate for individual differences in unloaded walking gait. Another method used to distinguish between individual gait patterns is the scaling of gait data to body size by creating dimensionless numbers relating to gait mechanics (Hof, 1996, Pierrynowski and Galea, 2001, Pinzone et al., 2016). This removes variability due to physical characteristics such as leg length and body mass. Although non-dimensional normalisation of gait data is favourable when making comparisons between the unloaded walking gaits of different individuals, it was deemed unnecessary for the research in this thesis because of the repeated measures design employed in each of the studies to focus on how the addition of different external loading conditions can alter individual gait patterns from that of unloaded walking.

3.14. Statistical analysis

The statistical analysis performed on the data from each experiment is detailed in the appropriate chapter. Briefly, all statistical tests were conducted using SPSS version 24 (IBM SPSS Statistics, SPSS inc., Chicago, IL, USA). Descriptive statistics (mean ± SD) were calculated for all outcome measures. All mean data in the subsequent experimental studies were tested for normality of distribution using Shapiro-Wilk as this test has more power to detect differences from normality for samples sizes of less than n = 50 compared to the Kolmogerov-Smirnov test, another common test to assess if data is normally distributed (Field, 2013). Data were also checked for outliers by visually exploring boxplots and histograms. A repeated-measures two-way (method x mass) or three-way (method x mass x body position) analysis of variance (ANOVA) was used to test for differences between several means. If sphericity was violated, then the Greenhouse-Geisser correction was used $\varepsilon < 0.75$, with Huynh-Feldt corrections used for less severe asphericity. Post-hoc tests were employed using a Bonferroni correction for multiple comparisons. The Bonferroni correction was used to control the Type I error rate. The Bonferroni was used over Tukey's test because the Bonferroni has more power when the number of comparisons is small (Field, 2013). Effect sizes are also reported in Chapters 5 and 7 using partial eta squared (η^2), with partial η^2 classified as small (0.010 - 0.059), medium (0.060 - 0.137) and large (>0.138) (Richardson, 2011). For correlations in the studies presented in Chapters 4, 5 and 7, a Pearson's product moment correlation coefficient was used. The correlation coefficients were interpreted using intervals of negligible correlation (0.0-0.09), weak correlation (0.10-0.39), moderate correlation (0.40-0.69), strong correlation (0.70-0.89) and very strong correlation (>0.90) (Schober et al., 2018). In addition, the coefficients of determination (r^2) expressed as a percentage were calculated from the r value. Statistical significance was set at $p < 0.05$ in all experimental chapters. Where $p < 0.10$, the results are reported as being close to statistical significance.

Chapter 4. The reliability of the Extra Load Index, loaded walking kinematics and subjective perceptions

Part of this work has been published in a peer-reviewed journal:
Hudson, S., Cooke, C. and Lloyd, R., 2017. The reliability of the Extra Load Index as a measure of relative load carriage economy. *Ergonomics*, *60*(9), pp.1250-1254.

4.1. Introduction

The ELI appears to be a useful and valid tool to measure relative load carriage economy (Lloyd et al., 2010a) but its reliability has yet to be reported. Knowledge of reliability is important if ELI is to be used with confidence. The reproducibility of the rate of oxygen consumption ($\dot{V}O_2$) during treadmill running has been frequently reported (e.g. Brisswalter and Legros, 1994, Pereira and Freedson, 1997, Pereira et al., 1994), however, few studies have determined the day-to-day variation of walking economy in healthy populations. Furthermore, no studies appear to have assessed the reliability of load carriage economy. Of those that have reported the reproducibility of walking economy in healthy adult populations (Wergel-Kolmert and Wohlfart, 1999, de Mendonça and Pereira, 2008), the day-to-day variation appears to be less reliable compared to running economy, with a coefficients of variation (CV) between ~ 8 - 9% and ~ 1.5 - 5% for walking and running economy, respectively. Furthermore, the reliability of $\dot{V}O_2$ appears to decrease at lower intensities of both running (Pereira *et al.* 1994) and walking (de Mendonça and Pereira, 2008). A number of different exercise intensities have been employed in the load carriage literature with walking speeds ranging from ~ 3 $km{\cdot}h^{-1}$ (Maloiy et al., 1986, Lloyd et al., 2010b) to ~ 6 $km{\cdot}h^{-1}$ (Quesada et al., 2000), and loads ranging from 10% body mass (Abe et al., 2004, Singh and Koh, 2009) to in excess of 50% body mass (Lloyd et al., 2010b). For this reason, knowledge of the reproducibility of load carriage economy across a range of exercise intensities would be beneficial, particularly at lower intensities were the reliability of $\dot{V}O_2$ appears to be lessened.

The reliability of both 2D and 3D kinematic analysis has been assessed for a range of different movements, including the unloaded walking gait (McGinley et al., 2009). In healthy individuals, unloaded walking kinematics appear to have good reliability both within (Wilken et al., 2012) and between (Benedetti et al., 2013) laboratories. However, the reproducibility of walking gait kinematics with load carriage do not appear to have been reported in the literature. In order to investigate the biomechanical factors that could determine load carriage economy, knowledge of the loaded walking gait's reliability could be beneficial to help understand the variable effect of loading. As such, investigating the reliability of load carriage kinematics seems warranted.

The reliability of subjective perceptions of load carriage have also not been reported in the literature. Whole body rating of perceived exertion (RPE) is a method frequently used to assess the subjective perceptions of load carriage (e.g. Goslin and Rorke, 1986, Lloyd et al., 2010d, Simpson et al., 2011). The reliability of whole body RPE has been questioned during progressive treadmill exercise, starting at an intensity of 12.8 $km \cdot h^{-1}$ (Lamb et al., 1999), but it has been suggested to be a reliable measure of perceived exertion during a range of activities including cycling, stepping, walking and jogging (Stamford, 1976). As such, investigating the reliability of loaded walking RPE appears warranted. Balogun et al. (1986) and Lloyd et al. (2010d) reported that whole body RPE was not sensitive enough to differentiate between different load carriage conditions. Visual analogue scales (VAS), to measure pain/discomfort in a number of locations on the body, appear to be a better measure of subjective perceptions in terms of differentiating between different loads and load carriage methods (Lloyd *et al.,* 2010e; Simpson *et al*., 2011). However, the reliability of VAS scales to assess load carriage has not been reported.

The aim of this study was to establish the reliability of the ELI, kinematics and subjective perceptions associated with load carriage, across a range of walking speeds (slow, self-selected and fast) with both light and heavy loads. It was hypothesised that the rate of oxygen consumption ($\dot{V}O_2$) and relative load carriage economy would show good reliability between repeated trials when walking unloaded and when carrying loads of 7 kg and 20 kg at slow, fast and self-selected walking speeds. It was also hypothesised that there would be good reliability between test-retest trials for loaded walking kinematics and subjective perceptions when walking unloaded and carrying loads of 7 kg and 20 kg.

4.2. Methods

4.2.1. Participants

Seventeen apparently healthy volunteers (12 males, 5 females) took part in the study (age 29 ± 10.7 years, mass 77.5 ± 13.9 kg, stature 1.77 ± 0.09 metres). All volunteers had no history of back pain and gave written informed consent to participate. The study received approval from the institutional ethics committee at Leeds Trinity University.

4.2.2. Experimental design

All trials were conducted at Leeds Trinity University. Participants attended the laboratory on six occasions in order to complete test-retest reliability of three different trial conditions. Trial conditions differed in walking speed, with a slow speed (3 $km \cdot h^{-1}$), fast speed (6 $km \cdot h^{-1}$) and a self-selected speed (4.4 ± 0.7 $km \cdot h^{-1}$). Trial conditions were completed in a randomised order, separated by a minimum of 48 hours and repeated identically seven days later. The order in which trial conditions were undertaken was randomised via a Latin square design with participants randomly assigned (by drawing lots) to one of three speeds. Trials involved 4x4 minute periods of walking, each separated by 5 minutes of rest. The initial stage was performed unloaded followed in a randomised order by a second unloaded period and walking with backpacks of 7 kg and 20 kg. The order of loading was identical for each of the initial trial and repeat trials. In an attempt to control for possible circadian variations in walking economy, test-retest trails were performed at approximately the same time of day for each individual. Participants were also asked to maintain a similar diet and refrain from moderate-vigorous exercise and alcohol consumption in the 24 hours' prior to each test.

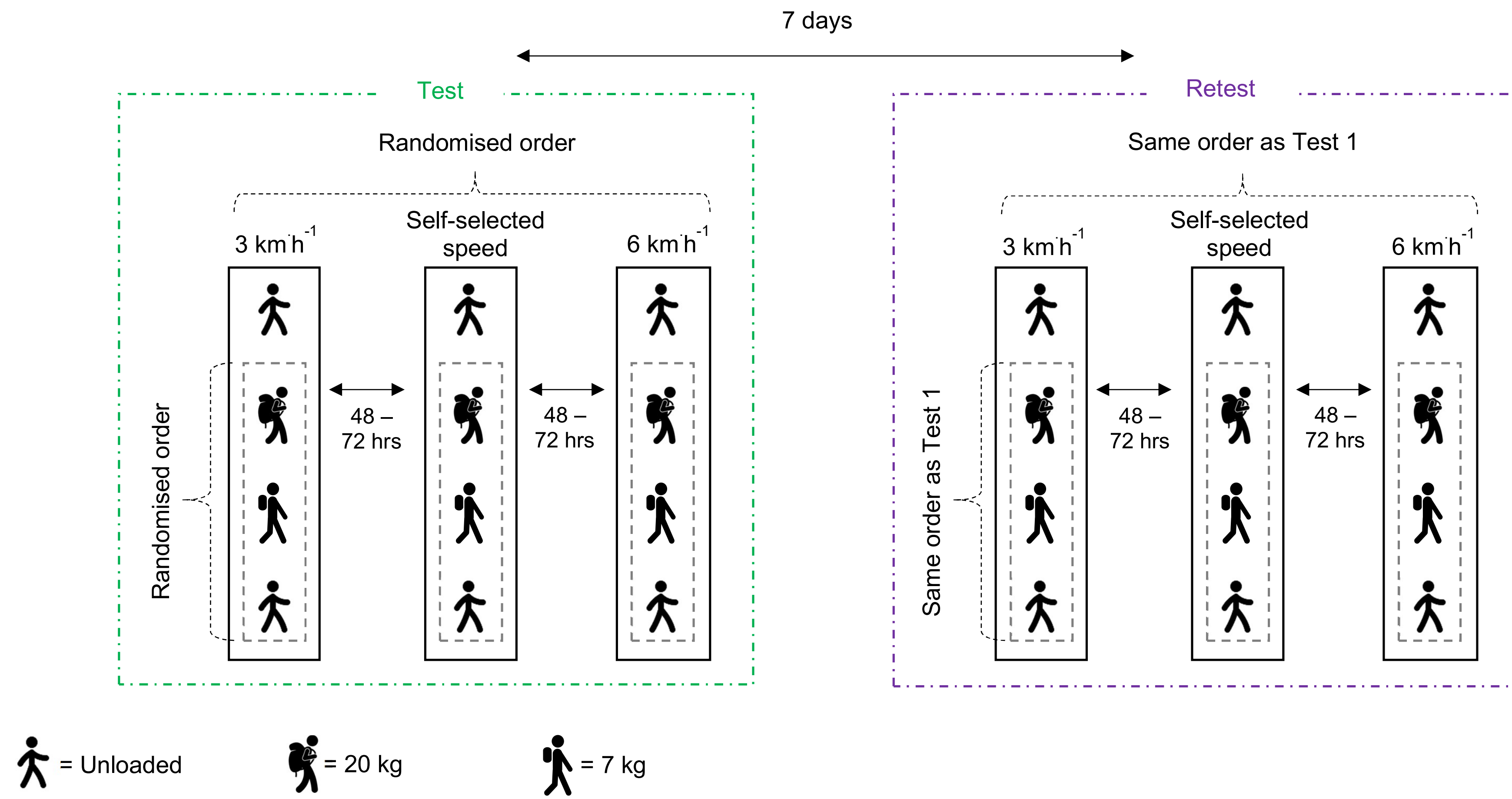


Figure 7. Overview of the experimental design in Chapter 4.

4.2.3. Experimental procedures

4.2.3.1. Treadmill speed verification

The procedures and results from the treadmill speed verification in this study are presented in Chapter 3.

4.2.3.2. Loading methods

For each loading condition, participants were fitted with a traditional back-loading rucksack, with a hip belt for support (7 kg = Featherlight freedom, AARN, New Zealand; 20 kg = Karrimor Jura 35, Karrimor, UK) (Figure 8). The mass of the load was made up of the rucksack itself plus sandbags and water bottles, stored in plastic containers to help evenly distribute the load and improve stability within the rucksack. Participants were asked to wear a t-shirt, shorts and the same footwear during each test, in order to minimise the influence of clothing.

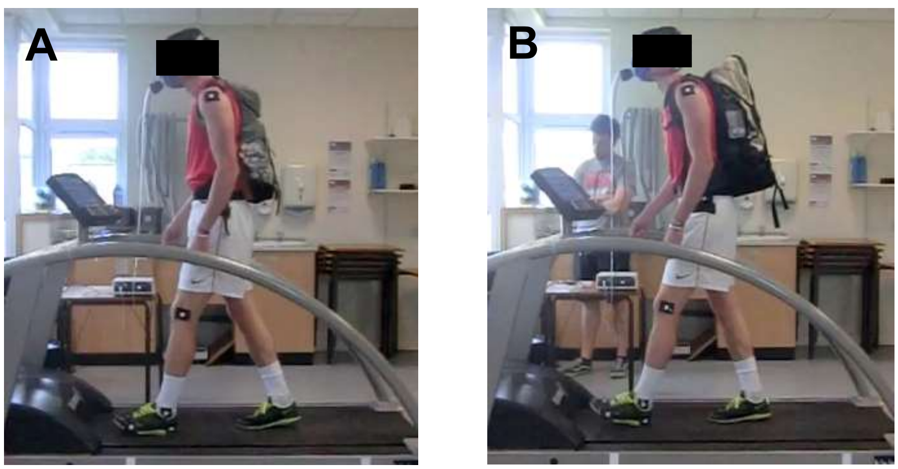


Figure 8. (A) Sagittal plane view of a participant completing the 7 kg condition. (B) Sagittal plane view of a participant completing the 20 kg condition.

4.2.3.3. Initial screening and habituation

The first laboratory visit included an initial screening of participants for any contraindications to exercise. Body mass and stature were measured, followed by a habituation period lasting ~ 20 minutes, which involved walking on the motorised treadmill (Mercury, HP Cosmos, Germany) at each of the walking speed conditions, with and without the 7 kg and 20 kg backpacks. The facemask for the online gas analysis system (Metalyzer 3B, Cortex, Germany) was also fitted, in order for participants to become accustomed to breathing through it. The self-selected walking speed established during the habituation period, recorded as the speed at which participants felt most comfortable while walking unloaded, was used as the self-selected walking speed in subsequent trials.

4.2.3.4. Main trials

Each trial began with the recording of the participant's body mass in order to calculate the ELI for that trial. Resting heart rate (Polar, H7, Finland) and oxygen uptake were then measured for two minutes prior to exercise. Exercise began with participants walking unloaded at 0% gradient for four minutes at a speed determined by the trial condition. After four minutes, there was a five-minute rest period, during which, participants stepped off the treadmill and removed the facemask. Heart rate was monitored during the rest period to ensure that participants returned to the baseline resting level established before exercise began. The final minute of each rest period was used to refit the facemask and rucksack. The procedure of four minutes walking followed by five minutes of rest was then repeated with the light load, heavy load and unloaded walking for a second time, in a randomised order if it was the first trial or in an identical order to the first test if it was a repeat trial.

4.2.3.5. Physiological measures and subjective perceptions

$\dot{V}O_2$ ($l{\cdot}min^{-1}$) was recorded for each period of walking and was used to calculate the ELI for each load carriage condition. RPE was recorded in the final 30 seconds of each period of walking. Pain/discomfort for each area of the body was recorded using VAS during each rest period.

4.2.3.6. Kinematic data

Sagittal plane kinematics were measured for the 3 $km{\cdot}h^{-1}$ trials to assess the test-retest reliability of walking gait perturbations caused by load carriage. The reliability at this walking speed was assessed to inform the future studies in this thesis, which are all performed at 3 $km{\cdot}h^{-1}$, in line with previous research that has demonstrated energy saving phenomena for load carriage (Maloiy et al., 1986, Lloyd and Cooke, 2000b, Abe et al., 2004). Sagittal plane kinematics were measured for six consecutive strides with each trial. Video files were manually digitised by a single observer using SIMI motion (SIMI 8.5.6, Germany). Intra-observer reliability of digitising was assessed using video files from one participant (20 kg loading condition for participant 1). Overall, there was good intra-observer reliability for the digitising of trunk, hip, knee and ankle angles at heel-strike and toe-off. The largest deviation occurred in trunk angle with a relative technical error of measurement of 0.2% at heel-strike and 0.2% at toe-off. The full results from the intra-observer reliability test are presented in Appendix M.

Raw joint angle data were filtered using a 2nd order Butterworth filter set at 6 Hz. Results from a residual analysis conducted on three participants to determine the optimal cut-off frequency of the filter are presented in Appendix L. Step time, double stance time and single stance time were also measured by visually inspecting each video for time periods between heel-strike and toe-off gait events.

4.2.4. Statistical analysis

Paired samples t-tests were used to test for significant differences between test-retest trials for each loading condition for ELI, $\dot{V}O_2$, spatiotemporal variables, joint angle kinematics and subjective perceptions. Bland-Altman plots were generated to assess the systematic bias and 95% limits of agreement limits of agreement (LoA) were measured as the mean of the differences $\pm$ 1.96 SD of the differences for each trial condition (Bland and Altman, 1986). Prior to creating the Bland-Altman plots, heteroscedasticity was formally assessed by plotting the absolute differences between the two trials against the individual means and calculating the correlation coefficient. Coefficient of variation (CV) and standard error of measurement (SEM) were also assessed following the guidelines of Atkinson and Nevill (1998). Test/retest intraclass correlation coefficients (ICC) were measured using a freely available Microsoft Excel spreadsheet (www.sportsci.org/2015/ValidRely.htm) (Hopkins, 2017). CV's <10% were considered as showing good absolute reliability (Atkinson and Nevill, 1998). Atkinson and Nevill (1998) suggested that, while some researchers have interpreted CV's of 10% or below as an indicator of good reliability, this measure should be interpreted with caution because CV reflects the repeated test error of the average individual and not all individuals. As such, CV's were interpreted alongside LoA, SEM and ICC. ICC's were interpreted using the guidelines from Koo and Li (2016) with values of less than 0.5, between 0.5 - 0.75, between 0.75 and 0.90 and greater than 0.90 indicating poor, moderate, good and excellent reliability, respectively.

4.3. Results

4.3.1. Oxygen consumption and relative load carriage economy

ELI values did not differ significantly between test-retest trials in any of the walking speed conditions with either of the additional loads (all conditions, $p > 0.05$). Following confirmation that heteroscedasticity was not present in any of the trial conditions (Figure 9), the systematic bias and 95% LoA were determined and are presented in Table 10 and Figure 10. Table 10 also shows the CV and SEM, which were small in all conditions with the highest CV (4.17%) and SEM (0.04), recorded when walking at 3 $km \cdot h^{-1}$ with 7 kg. ELI values did increase significantly with walking speed ($p = 0.018$).

Table 10. Reliability measures for the ELI at different walking speeds with 7 kg and 20 kg loads.

	3 $km \cdot h^{-1}$		Self-selected speed		6 $km \cdot h^{-1}$	
	7 kg	20 kg	7 kg	20 kg	7 kg	20 kg
Trial 1	0.94	0.95	0.98	0.99	0.97	1.00
Trial 2	0.95	0.95	0.96	0.96	0.98	1.00
Systematic Bias	-0.01	0.00	0.01	0.03	-0.02	0.00
95% LoA (±)	0.11	0.10	0.05	0.09	0.09	0.07
CV (%)	4.17	2.74	1.75	3.42	3.51	2.51
SEM	0.04	0.03	0.02	0.03	0.03	0.03

LoA = limits of agreement; CV = coefficient of variation; SEM = standard error of measurement

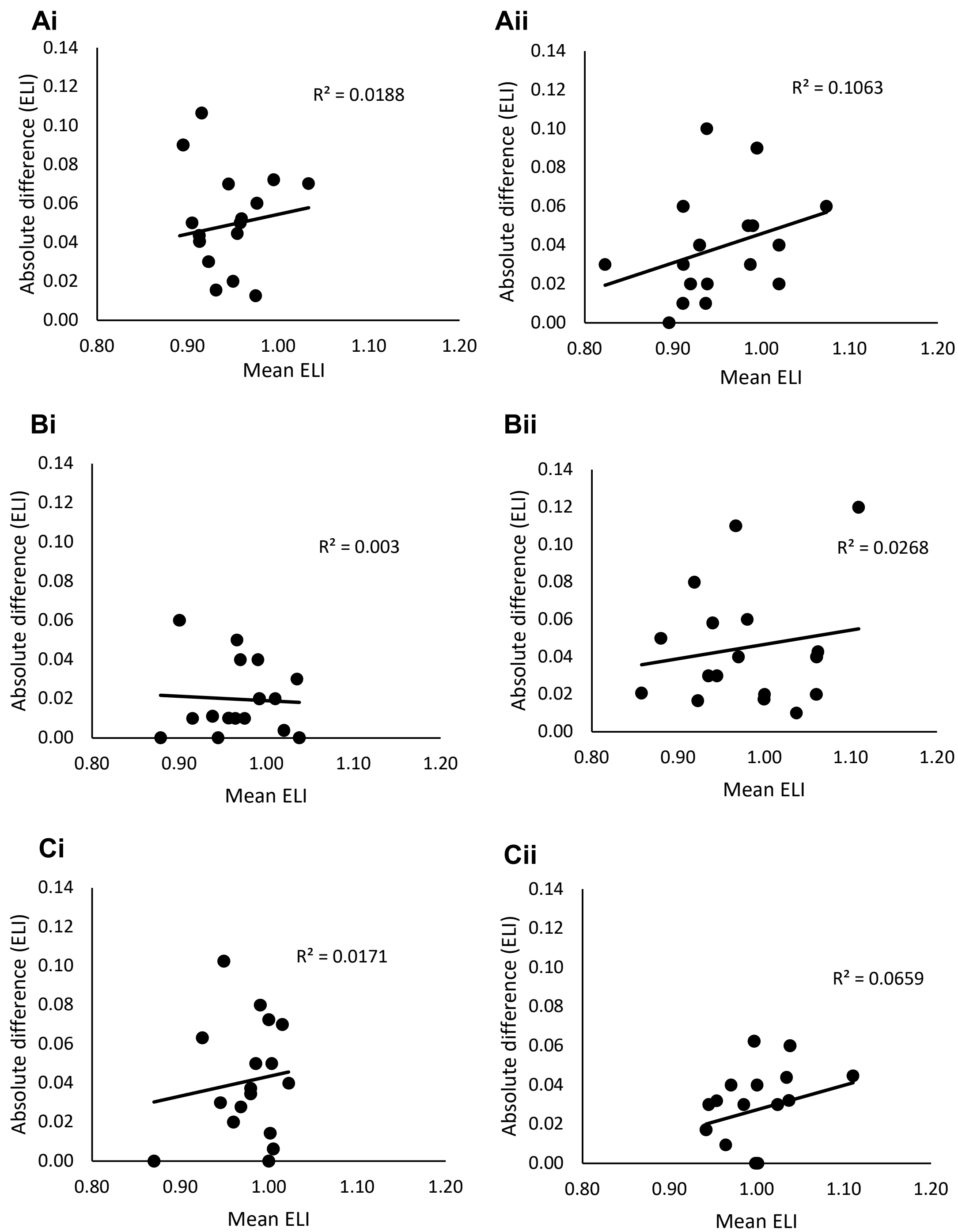


Figure 9. Absolute difference plots between the tests and the individual means for the examination of heteroscedasticity for each of the walking speeds (A = 3 km·h^{-1}; B = self-selected; C = 6 km·h^{-1}) with the light (i) and heavy loads (ii).

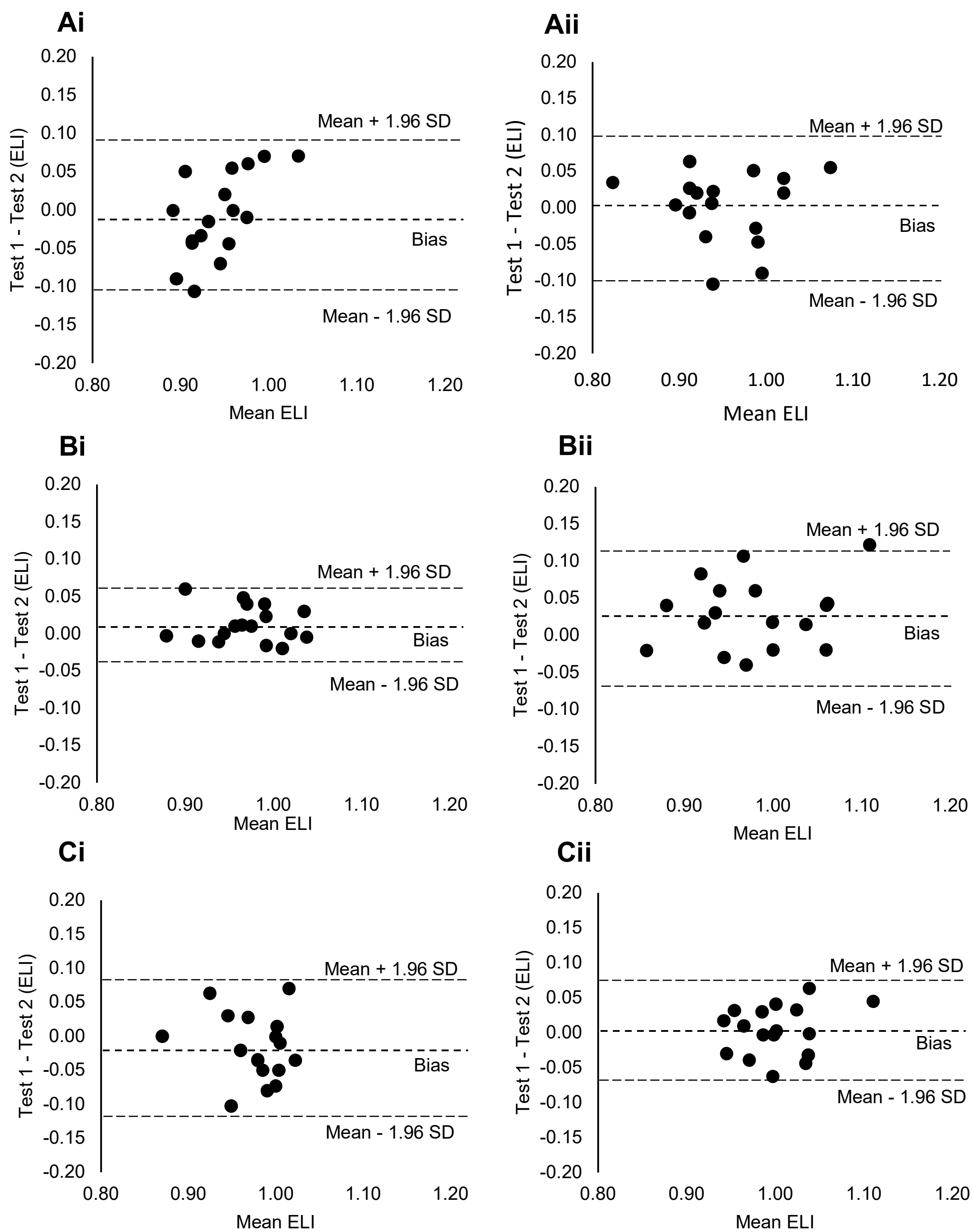


Figure 10. Bland-Altman plot illustrating systematic bias and 95% limits of agreement for each of the walking speeds (A = 3 km·h^{-1}; B = self-selected; C = 6 km·h^{-1}) with the light (i) and heavy loads (ii).

There was no significant difference in $\dot{V}O_2$ between the two unloaded periods of walking performed in each of the trial conditions ($p = 0.235$). The variations in $\dot{V}O_2$ between the unloaded periods of walking in each trial are presented in Table 11. There was no significant difference in $\dot{V}O_2$ between test-retest trials (all conditions, $p > 0.05$). Walking at 6 km·h^{-1} with a load of 20 kg produced the largest LoA and SEM of ± 0.19 l·min^{-1} and 0.06 l·min^{-1}, respectively (Table 12). The largest CV (4.50%) was measured for the self-selected speed when carrying 20 kg. $\dot{V}O_2$ did significantly increase with an increase in walking speed ($p = 0.001$) and when the mass of the load carried increased ($p = 0.001$).

Table 11. Reliability measures for $\dot{V}O_2$ (l·min^{-1}) between repeated bouts of unloaded walking within the same trial.

	3 km·h^{-1}		Self-selected speed		6 km·h^{-1}	
	Trial 1	Trial 2	Trial 1	Trial 2	Trial 1	Trial 2
$\dot{V}O_2$ (l·min^{-1}) unloaded 1	0.69	0.70	0.87	0.89	1.27	1.24
$\dot{V}O_2$ (l·min^{-1}) unloaded 2	0.67	0.68	0.87	0.88	1.25	1.24
Systematic Bias	-0.02	-0.02	0.00	-0.01	-0.01	0.00
95% LoA (±)	0.07	0.07	0.06	0.06	0.06	0.09
CV	3.62	3.68	2.30	2.63	1.86	2.72
SEM	0.02	0.03	0.02	0.02	0.02	0.03

LoA = limits of agreement; CV = coefficient of variation; SEM = standard error of measurement

Table 12. Reliability measures for $\dot{V}O_2$ ($l{\cdot}min^{-1}$) at different walking speeds with 7 kg and 20 kg loads.

	3 $km{\cdot}h^{-1}$				Self-selected speed				6 $km{\cdot}h^{-1}$			
	U1	U2	7 kg	20 kg	U1	U2	7 kg	20 kg	U1	U2	7 kg	20 kg
Trial 1 $\dot{V}O_2$ ($l{\cdot}min^{-1}$)	0.69	0.67	0.71	0.83	0.87	0.87	0.93	1.09	1.27	1.25	1.34	1.59
Trial 2 $\dot{V}O_2$ ($l{\cdot}min^{-1}$)	0.70	0.68	0.72	0.84	0.89	0.88	0.94	1.08	1.24	1.24	1.33	1.56
Systematic Bias	-0.01	-0.01	-0.02	-0.01	-0.02	-0.01	-0.01	0.01	0.03	0.01	0.01	0.03
95% LoA (±)	0.06	0.07	0.05	0.09	0.07	0.09	0.08	0.14	0.16	0.15	0.14	0.19
CV (%)	3.78	4.08	3.59	4.32	3.62	4.05	3.62	4.50	3.80	4.01	3.58	3.64
SEM	0.03	0.03	0.03	0.04	0.03	0.04	0.03	0.05	0.05	0.05	0.05	0.06
ICC	0.96	0.95	0.96	0.94	0.96	0.96	0.97	0.94	0.95	0.95	0.96	0.94

U1 = Unloaded; U2 = Unloaded 2; LoA = limits of agreement; CV = coefficient of variation; SEM = standard error of measurement; ICC = intraclass correlation coefficients

4.3.2. Subjective perceptions

There were no significant differences between the test-retest trials for RPE in any of the loading conditions at 3 $km \cdot h^{-1}$, self-selected speed and 6 $km \cdot h^{-1}$ (all conditions, *p* > 0.05), although the difference between the repeat trials was close to significance with 20 kg at 3 $km \cdot h^{-1}$ ($p = 0.051$) and 6 $km \cdot h^{-1}$ ($p = 0.083$). After confirming that heteroscedasticity was not present in any of the trial conditions, the systematic bias, 95% LoA, SEM and CV's were determined and are presented in Table 13. RPE increased as the load mass increased at all speeds, except for the 6 $km \cdot h^{-1}$ repeat trial with 20 kg, where the mean RPE was the same between the 7 kg and 20 kg conditions. The largest systematic bias (0.88), LoA (± 3.39), SEM (1.22) and CV (13.27%) all occurred with the 20 kg load at 3 $km \cdot h^{-1}$.

There was no pain/discomfort (value of 0) reported for any of the 15 areas of the body during the unloaded walking trials at 3 $km \cdot h^{-1}$ or at the self-selected walking speed. There was also no pain/discomfort walking at 3 $km \cdot h^{-1}$ with 7 kg at the knees, ankles and feet. There were large test-retest differences in pain/discomfort scores at the upper body sites with the 20 kg load when walking at 3 $km \cdot h^{-1}$. The LoA and SEM with 20 kg were large in the neck (LoA ± 12.26 mm; SEM = 4.42 mm), front shoulders (LoA ± 19.00 mm; SEM = 6.86 mm), back shoulders (LoA ± 10.08 mm; SEM = 3.64 mm), upper back (LoA ± 20.26 mm; SEM = 7.31 mm), chest (LoA ± 10.57 mm; SEM =3.81 mm) and lower back (LoA ± 21.25 mm; SEM 7.67 mm).

Table 13. Ratings of perceived exertion with each load and speed combination.

	3 $km \cdot h^{-1}$				Self-selected speed				6 $km \cdot h^{-1}$			
	U1	U2	7 kg	20 kg	U1	U2	7 kg	20 kg	U1	U2	7 kg	20 kg
Trial 1	6	6	7	9	7	7	8	10	9	8	10	13
Trial 2	6	6	8	10	7	7	8	10	8	8	10	12
Bias	0.12	0.12	0.29	0.88	0	0.06	0.12	0.12	-0.29	-0.18	0.18	-0.53
95% LoA (±)	1.18	1.18	1.51	3.39	1.39	1.47	2.29	2.58	1.93	2.52	2.22	2.31
CV (%)	6.74	6.74	7.33	13.27	7.46	7.78	10.54	8.95	8.25	10.93	8.27	6.8
SEM	0.42	0.42	0.55	1.22	0.5	0.53	0.82	0.93	0.7	0.91	0.8	0.83
ICC	0.18	0.18	0.92	0.76	0.49	0.61	0.72	0.86	0.87	0.75	0.88	0.86

U1 = Unloaded; U2 = Unloaded 2; LoA = limits of agreement; CV = coefficient of variation; SEM = standard error of measurement; ICC = intraclass correlation coefficients

4.3.3. Spatiotemporal variables

There were no significant differences between test-retest trials for step time (all conditions, $p > 0.05$), double stance time (all conditions, $p > 0.05$) or single stance time (all conditions, $p > 0.05$) when walking at 3 km·h^{-1}. At 3 km·h^{-1} the 20 kg produced the largest 95% LoA for step time (± 0.04 seconds), double stance time (± 0.02 seconds) and single stance time (± 0.03 seconds). The largest CV (4.55%) was measured for the single stance time with the 20 kg load (Table 14).

Table 14. Step time, double stance time and single stance time reliability at 3 km·h^{-1}.

	3 km·h^{-1}			
	U1	U2	7 kg	20 kg
Step time (seconds)				
Trial 1	0.41	0.4	0.41	0.41
Trial 2	0.4	0.4	0.4	0.4
Systematic Bias	0.01	0	0	0.01
95% LoA (±)	0.03	0.04	0.03	0.04
CV (%)	2.6	3.3	3.03	3.12
SEM	0.01	0.01	0.01	0.01
ICC	0.86	0.67	0.69	0.74
Double stance time (seconds)				
Trial 1	0.17	0.17	0.17	0.18
Trial 2	0.16	0.17	0.17	0.17
Systematic Bias	0	0	0	0
95% LoA (±)	0.02	0.01	0.01	0.02
CV (%)	4.37	2.85	3.15	3.61
SEM	0.01	0	0.01	0.01
ICC	0.70	0.55	0.83	0.81
Single stance time (seconds)				
Trial 1	0.24	0.24	0.24	0.23
Trial 2	0.24	0.23	0.23	0.23
Systematic Bias	0	0.01	0	0
95% LoA (±)	0.02	0.03	0.03	0.03
CV (%)	3.41	4.3	4.13	4.55
SEM	0.01	0.01	0.01	0.01
ICC	0.86	0.52	0.75	0.68

U1 = Unloaded; U2 = Unloaded 2; LoA = limits of agreement; CV = coefficient of variation; SEM = standard error of measurement; ICC = intraclass correlation coefficients

4.3.4. Sagittal plane joint kinematics

The test-retest differences in the measured sagittal plane joint angles for walking at 3 $km \cdot h^{-1}$ are presented in Table 15. There were no significant differences in joint angles between test-retest trials (all conditions, $p > 0.05$). The LoA, CV and SEM were small for all joint kinematics at both heel-strike and toe-off. Considering all load conditions, the largest LoA and SEM were in hip angle at both heel-strike and toe-off. The largest systematic bias occurred for hip angle at heel-strike with 7 kg (1.72º). The largest LoA occurred for hip angle at toe-off with 7 kg (± 10.73º). The largest CV and SEM occurred for trunk angle at heel-strike when unloaded (CV = 3.03%) and hip angle at toe off with 7 kg (SEM = 3.87º), respectively. The ICC values showed moderate to good reliability (ICC = 0.5 - 0.9) for all kinematic variables, except for hip angle at heel-strike in the second unloaded walking, 7 kg and 20 kg conditions which all showed poor reliability (ICC < 0.5).

Table 15. Trunk, hip, knee and ankle angle reliability at heel-strike and toe-off in the 3 km·h^{-1} condition.

	Heel-strike				Toe-off			
	U1	U2	7 kg	20 kg	U1	U2	7 kg	20 kg
Trunk angle (º)								
Trial 1	91.59	91.55	86.81	80.1	89.43	89.57	86.97	80.77
Trial 2	91.74	91.64	86.47	80.76	89.35	89.52	86.55	81.11
Bias	0.15	0.09	-0.34	0.66	-0.08	-0.05	-0.43	0.35
95% LoA (±)	6.81	7.43	6.76	5.61	5.84	6.42	6.29	5.38
CV (%)	2.78	3.03	2.61	2.03	2.32	2.56	2.43	1.96
SEM	2.46	2.68	2.44	2.03	2.11	2.32	2.27	1.94
ICC	0.65	0.55	0.45	0.66	0.75	0.58	0.58	0.65
Hip angle (º)								
Trial 1	163.79	163.97	158.82	150.05	174.48	174.31	172.55	166.75
Trial 2	164.42	163.98	157.1	150.48	173.44	173.47	170.87	167.07
Bias	-0.63	0	1.72	-0.43	1.04	0.84	1.68	-0.32
95% LoA (±)	9.38	10.63	10.04	10.65	8.53	9.73	10.73	10.01
CV (%)	2.06	2.34	2.29	2.56	1.77	2.02	2.25	2.16
SEM	3.38	3.83	3.62	3.84	3.08	3.51	3.87	3.61
ICC	0.58	0.37	0.38	0.36	0.77	0.66	0.60	0.51
Knee angle (º)								
Trial 1	177.1	177.4	176.95	172.48	122.8	122.18	122.98	122.73
Trial 2	178.31	178.34	176.5	172.54	121.66	121.27	121.56	122.26
Bias	-1.21	-0.94	0.45	-0.06	1.14	0.9	1.41	0.46
95% LoA (±)	6.06	6.51	6.85	10.46	4.96	3.24	5.82	7.07
CV (%)	1.23	1.32	1.4	2.19	1.46	0.96	1.72	2.08
SEM	2.19	2.35	2.47	3.77	1.79	1.17	2.1	2.55
ICC	0.77	0.69	0.62	0.50	0.68	0.91	0.70	0.54
Ankle angle (º)								
Trial 1	117.57	118.17	118.44	116.91	124.66	125.39	124.58	125.76
Trial 2	118.65	119.24	119.13	117.59	124.11	125.57	125.26	125.35
Bias	-1.08	-1.07	-0.69	-0.68	0.56	-0.17	-0.69	0.41
95% LoA (±)	6.75	8.25	5.93	5.93	6.82	8.43	6.86	5.59
CV (%)	2.06	2.51	1.8	1.82	1.98	2.42	1.98	1.61
SEM	2.44	2.97	2.14	2.14	2.46	3.04	2.48	2.02
ICC	0.61	0.52	0.61	0.58	0.71	0.67	0.77	0.82

U1 = Unloaded; U2 = Unloaded 2; LoA = limits of agreement; CV = coefficient of variation; SEM = standard error of measurement; ICC = intraclass correlation coefficients; Bias = Systematic Bias

4.3.5. Summary of the results

The reliability of the ELI:

- Neither $\dot{V}O_2$ or ELI differed significantly between test-retest trials in any of the walking speed or load conditions.
- The ELI demonstrated good reliability across the walking speed and load conditions. The largest systematic bias (0.03) occurred with 20 kg at a self-selected walking speed (4.4 ± 0.7 $km \cdot h^{-1}$). The largest LoA (± 0.11) and the highest CV (4.17%) and SEM (0.04) were recorded for the 3 $km \cdot h^{-1}$ speed with 7 kg (Table 10).

The reliability of subjective perceptions:

- Considering RPE, the largest systematic bias (0.88), LoA (± 3.39), SEM (1.22) and CV (13.27%) all occurred with the 20 kg load at 3 $km \cdot h^{-1}$ (Table 13).
- The largest test-retest differences for pain/discomfort scores occurred at the upper body sites with the 20 kg load when walking at 3 $km \cdot h^{-1}$. The front shoulders (LoA ± 19.00; SEM = 6.86), upper back (LoA ± 20.26; SEM = 7.31) and lower back lower back (LoA ± 21.25; SEM 7.67) exhibited the largest test-retest differences.

The reliability of sagittal plane kinematics at 3 $km \cdot h^{-1}$:

- Spatiotemporal variables and sagittal plane joint angles did not differ significantly between test-retest trials in any of the walking speed or load conditions.
- Of the spatiotemporal measures, single stance time when carrying 20 kg was associated with the largest CV (4.55%) and step time when carrying 20 kg was associated with the largest LoA (± 0.04 seconds) (Table 14).
- Considering joint angles, the largest systematic bias (1.72º), LoA (± 10.73º) and SEM (3.87º) occurred for hip angle with the 7 kg load. The largest CV occurred for trunk angle when walking unloaded (CV = 3.03%) (Table 15).

4.4. Discussion

The aim of this study was to establish the reliability of the ELI, kinematics and subjective perceptions associated with load carriage.

This discussion is split into three parts. The first part is focused on the reliability of the ELI as a measure of relative load carriage economy (section 4.4.1). The second part is focused on the reliability of load carriage kinematics (section 4.4.2). The third part is on the reliability of subjective perceptions (section 4.4.3).

4.4.1. The reliability of the ELI

This is the first study to examine the reliability of the ELI as a measure of relative load carriage economy. The ELI demonstrated good reliability at slow, fast and self-selected walking speeds with both a relatively light and heavy load. The systematic bias was small in all conditions, with the largest LoA within ± 0.11, the largest SEM was 0.04 and the highest magnitude of CV was 4.17%. The ELI was found to be most reliable at the self-selected speed with the light load (95% LoA = 0.05; CV = 1.75%; SEM = 0.02). The self-selected speed (4.4 ± 0.7 $km{\cdot}h^{-1}$) was also the only condition where the CV appeared larger when carrying the heavy load than the light load. This is, perhaps, because the speed-load combination of the self-selected speed with a light load was closest to representing the participant's natural walking pattern, and therefore, the between day variation was smallest in this condition. Additionally, the self-selected speed was chosen unloaded, which might have led to greater variability with the heavier load.

The ELI was assessed across a range of walking speeds with both relatively light and heavy loads because a range of speed-load combinations are employed in a variety of applied scenarios. Individuals in the military services are regularly required to carry heavy loads in excess of 30 kg at walking speeds of between 5 - 6 $km{\cdot}h^{-1}$ (Harman et al., 2001), while school children and individuals in rural areas of developing countries often adopt a slower walking pace of around 3 $km{\cdot}h^{-1}$ with both light and heavy loads (Singh and Koh, 2009, Lloyd et al., 2010b). Although previous research, particularly those on military personnel, have used loads in excess 40 kg (Harman et al., 2000), 20 kg was chosen in this study due to the untrained nature of some participants and because similar loads have been frequently used to represent a heavy load in the literature (e.g. Lloyd et al., 2011, Birrell and Haslam, 2009). As much of the literature on unloaded exercise suggests that reliability of energy expenditure increases as the exercise intensity increases and there was no difference in the reliability of ELI across a range of exercise intensities in the present study, we would expect ELI values to demonstrate good reliability with loads in excess of 20 kg.

As expected, given the ELI results, $\dot{V}O_2$ also demonstrated good test-retest reliability with the largest LoA within $\pm$ 0.19 $l{\cdot}min^{-1}$, a highest SEM of 0.06 $l{\cdot}min^{-1}$ and a highest CV of 4.50% (Table 12). Furthermore, there appears to be little difference in test-retest reliability between unloaded and loaded $\dot{V}O_2$. This demonstrates a better level of reliability than previously reported for walking economy at speeds of 4-5 $km{\cdot}h^{-1}$ (Wergel-Kolmert and Wohlfart, 1999, de Mendonça and Pereira, 2008) and is similar to the CV of 4.4% reported when walking intensity is increased by gradients up to 10% (de Mendonça and Pereira, 2008). In the present study, the CV for $\dot{V}O_2$ did not reduce as a result of increasing walking speed or when carrying an external load. Furthermore, the present study showed that the LoA and SEM were lower at 3 $km{\cdot}h^{-1}$ compared to 6 $km{\cdot}h^{-1}$, which is somewhat unexpected, given that previous research has suggested that an increase in exercise intensity increases reliability of $\dot{V}O_2$ (Pereira et al., 1994, de Mendonça and Pereira, 2008). However, the difference in $\dot{V}O_2$ between 3 $km{\cdot}h^{-1}$ and 6 $km{\cdot}h^{-1}$ in LoA and SEM were small and there was no difference in CV between speeds.

Unloaded $\dot{V}O_2$ was measured twice in each trial to assess its reliability between repeated bouts of walking on the same day because of its important role as the denominator in the calculation of the ELI. Based on previous literature, it was predicted that $\dot{V}O_2$ during unloaded walking might be less reliable than $\dot{V}O_2$ during loaded walking, as the exercise intensity is lower. However, there was no difference in $\dot{V}O_2$ between the two unload periods in each trial (Table 11) and as such, $\dot{V}O_2$ from the first unloaded period of each trial was used in the calculation of the ELI.

4.4.2. The reliability of load carriage kinematics

The results of this study show that step parameters exhibit good reliability at a slow walking speed of 3 $km{\cdot}h^{-1}$ with both a light and heavy load. Studies that have shown an economical advantage for a particular method of load carriage, in comparison to others, have done so at slow walking speeds of ~ 3 $km{\cdot}h^{-1}$ (Maloiy et al., 1986, Charteris et al., 1989, Lloyd and Cooke, 2000b, Abe et al., 2004). As such, the reliability of load carriage kinematics in this study were assessed at 3 $km{\cdot}h^{-1}$, which informs all studies in this thesis that focus on load carriage economy at this walking speed. Studies examining the reliability of unloaded gait analysis in healthy adults have shown good reliability for step parameters (Stolze et al., 1998) and both 2D and 3D joint angle kinematics (McGinley et al., 2009, Ross et al., 2015). Single stance time was slightly less reliable between tests, with both the light and heavy load, compared to double stance time and overall step time. This is in line with the findings of Stolze et al. (1998), who found the swing phase time of the gait to be the least reliable of all step parameters in healthy individuals walking unloaded.

This study focused on the sagittal plane trunk, hip, knee and ankle joint angles because this is where most movement occurs in the human walking gait and, as such, are the variables that have been assessed in many reliability studies for the unloaded walking gait (Besier et al., 2003, Growney et al., 1997, Kadaba et al., 1989, Wilken et al., 2012, Tsushima et al., 2003, Ross et al., 2015). Measures of hip joint angle at both heel-strike and toe-off, but particularly heel-strike, appeared to be the least reliable of the kinematic measures. The systematic bias, CV and SEM measures suggest a good level of reliability for hip angle at heel-strike. However, the ICC suggests poor reliability and the LoA are higher for this joint angle compared to the others. The decreased reliability of the hip joint measure could be, in part, a consequence marker reapplication error due to the increased difficulty in locating the hip joint centre compared to that of the knee and ankle. Indeed, in a comparison of gait analysis between seven different laboratories, Benedetti et al. (2013) found hip angle measures to have the highest inter-laboratory differences and suggested that this was due to difficulties in modelling the thigh segment and locating the hip joint centre. A number of studies have assessed the between assessor reliability of walking gait data kinematic because differences in marker placement accuracy can influence results (Besier et al., 2003, McGinley et al., 2009). In this study, the same researcher applied all of the markers, so measures of inter-assessor reliability were not applicable. Ensuring that the hip joint centre is accurately marked when the load carriage device includes a hip belt is important for accurate and reliable measures of hip joint angle. In the present study, the marker for hip joint centre was positioned just below the backpack hip belt (Figure 8). However, for some trials, the arm obstructed the hip joint centre marker, particularly at heel-strike, which might have increased the error of locating the actual hip joint centre during the digitising process.

4.4.1. The reliability of subjective perceptions

Ratings of perceived exertion appeared to be reliable with load carriage and tended to be slightly less reliable with the 20 kg load compared to 7 kg. The least reliable condition for RPE was 20 kg at a walking speed of 3 $km{\cdot}h^{-1}$. This finding is in line with previous research suggesting that the test-retest reliability of RPE using the Borg scale decreases as the intensity of exercise increases (Lamb et al., 1999). RPE increased as the load mas increased. A similar finding was reported by Goslin and Rorke (1986) who showed that RPE increased linearly from 20%-40% carried in a backpack. The use of visual analogue scales to measure pain/discomfort of loaded walking at various body sites appears to lack reliability. This is somewhat expected given the subjective nature of visual analogue scales. Given the lack of reliability in pain/discomfort using visual analogue scales found here, the results of subjective pain/discomfort associated with load carriage in future studies should be interpreted with caution. Despite this, subjective measures could still be a useful tool to identify if large changes in an individual loaded walking biomechanics are a consequence of pain or discomfort to a particular area of the body. The test-retest reliability of pain/discomfort are only reported for 3 $km{\cdot}h^{-1}$, to inform the other studies presented in this research.

4.5. Conclusion

Based on the evidence provided in the chapter, the ELI appears to be a reliable measure of relative load carriage economy that can be easily interpreted by developers and manufacturers as well as scientific researchers. As such, the ELI represents a useful and reliable tool for comparing the relative economy of different load carriage systems. The test-retest reliability of sagittal plane joint angles at the trunk, knee and ankle appear reliable at heel-strike and toe-off gait events during load carriage. Measurements of test-retest 2D hip angle showed less reliability than the other angles and should be interpreted with caution. This could have been caused by the arms obstructing the hip marker, and not necessarily a lack of reliability in the true value for hip angle and heel-strike. Step parameters and whole body RPE during short duration, steady-state load carriage tasks also appear to demonstrate a good level of reliability. VAS scales with load carriage appear less reliable in the upper body than lower body sites with load carriage but this is likely to be dependent on the load carriage method employed.

Chapter 5. A comparison of economy and sagittal plane kinematics among back-, back/front- and head-loading.

Parts of this work has been published in peer reviewed journals:

Hudson, S., Cooke, C., Davies, S., West, S., Gamieldien, R., Low, C., & Lloyd, R. (2018). A comparison of economy and sagittal plane trunk movements among back-, back/front-and head-loading. *Ergonomics*, *61*(9), 1216-1222.

Hudson, S., Cooke, C., Davies, S., West, S., Gamieldien, R., Low, C., & Lloyd, R. (2020). Inter-individual variability in load carriage economy and comparisons between different load conditions. *Applied ergonomics*, *82*, 102968.

5.1. Introduction

The research described in Chapter 4 showed that the ELI has good test-retest reliability at slow (3 $km \cdot h^{-1}$), fast (6 $km \cdot h^{-1}$) and self-selected walking speeds with light (7 kg) and heavy (20 kg) loads. As such, the ELI is used for the research in this study to assess if the relative economy associated with loads carried on the head, back and evenly distributed between the back and front of the torso can be explained by alterations in sagittal plane kinematics from unloaded walking as a consequence of the load carried.

Energy saving phenomena have been reported with loads carried on the head (Maloiy et al., 1986, Charteris et al., 1989), back (Abe et al., 2004) and evenly distributed between the front and back of the torso (Lloyd and Cooke, 2000b). Much work has been done to identify potential mechanisms that may contribute to these energy saving phenomena (e.g. Jones et al., 1987, Heglund et al., 1995, Abe et al., 2004, Lloyd and Cooke, 2011), yet the determinants remain unclear. Abe et al. (2004) and Lloyd and Cooke (2000b) identified potential energy saving mechanisms for back and back/front-loading, respectively. Abe et al. (2004) proposed that back-loading can be very economical with light loads (up to 12kg) at low speeds (2.4 – 3.6 $km \cdot h^{-1}$), due to a mechanism that they characterised as the contribution of rotative torque about the lower limb. Prior to the findings of Abe et al. (2004), Lloyd and Cooke (2000b) had reported back/front-loading to be more economical than back-loading with heavier loads, due to a mechanism that they characterised as the contribution of trunk momentum to the energy required for walking. Although characterised slightly differently, these proposed mechanisms appear similar and suggest that increased sagittal plane trunk movement during load carriage might act as an energy saving mechanism. Indeed it seems plausible that increased trunk movement through the step cycle, when carrying a load at slow speeds, could contribute to forward momentum, thus reducing the amount of force required to propel the body forward with each step (Lloyd and Cooke, 2000a, Lloyd and Cooke, 2011).

Unlike back and back/front-loading, head-loading is likely to require a constrained, upright posture to maintain equilibrium of the load, regardless of the mass. If constraining the trunk increases the energy cost of load carriage, then head-loading, in theory, would be less economical than methods that load the trunk. Yet, research on head-loading economy is equivocal. Some studies have reported that the energy cost of head-loading rises in proportion to the mass of the external load (Soule and Goldman, 1969, Datta and Ramanathan, 1971, Datta et al., 1973, Lloyd et al., 2010b, Lloyd et al., 2010c), while others have reported that head-loading could represent a remarkably economical method for certain individuals, with African women able to carry loads of up to 20% body mass on their head with no additional energy cost above that required for unloaded walking (Maloiy et al., 1986, Charteris et al., 1989). However, these latter studies used small sample sizes ($n \leq 6$) and are not generalisable. More recently, Lloyd et al. (2010c) demonstrated a large level of individual variation in economy for both head- and back-loading, with some individuals being remarkably economical at head-loading, while others were very economical at back-loading. Furthermore, they investigated load carriage economy in both experienced ($n = 13$) and inexperienced ($n = 11$) head-loaders and found that 38.5% of experienced head-loaders had better economy in head-loading than back-loading, while 36.4% of inexperienced head-loaders exhibited the same tendency. This led Lloyd et al. (2010c) to suggest that load carriage economy with head-loading might be independent of previous experience and, therefore, not a result of structural adaptation. As the mechanisms underpinning individual variation in energy cost of load carriage are yet to be established, examining the role of postural adjustments associated with transporting a load seems warranted, particularly given the potential energy saving role of sagittal plane trunk movements that have been suggested for methods that load the trunk and the remarkable levels of economy that have been reported for head-loading in some individuals and small sample studies.

Despite the substantial individual variation in load carriage economy reported by Lloyd et al. (2010c), to date, it has not been reported elsewhere. Individual variation in energy expenditure could help explain the contradictory evidence that exists for load carriage economy with different methods, particularly given the small sample sizes ($n < 10$) used in previous studies (Maloiy et al., 1986, Lloyd and Cooke, 2000b, Abe et al., 2004). Lloyd and Cooke (2011) also reported a high level of individual variation in step parameters when carrying load on the back and back/front, with changes in stride length from those associated with unloaded walking ranging from +12% to -6%. As with load carriage economy, individual variation in loaded walking gait kinematics has not been reported elsewhere. Information on the presence and extent of individual variation in loaded walking gait kinematics could help to elucidate why there appears to be individual variation in load carriage economy. It could also, perhaps, explain why some individuals are more economical with certain methods of load carriage.

There were two main aims for this study. The first aim was to assess the economy and sagittal plane kinematics associated with three methods of load carriage that have all been reported as economical, but all constrain posture differently. The second aim was to assess the amount of inter-individual variation in economy and sagittal plane kinematics associated with each method of load carriage. It was hypothesised that the load carriage method that allowed for the greatest freedom of movement of the trunk, for a given load mass, would be associated with the best associated economy. Head-loading was expected to constrain posture in an upright position, and, as such, be the least economical method. Combined back/front-loading was expected to allow for greater movement for the trunk with heavier loads compared to back-loading (i.e. closer to that of unloaded walking) and therefore, be more economical at heavier loads. It was also hypothesised that that, in a larger sample of participants than reported in much of the published load carriage literature, there would be a considerable amount of inter-individual variation in load carriage economy and load carriage kinematics.

5.2. Methods

A secondary analysis was conducted on data collected prior to this PhD by one of my PhD supervisors, Professor Ray Lloyd, and colleagues from the Cape Peninsula University of Technology (Professor Simeon Davies, Dr Sacha West and Raeeq Gamieldien). This section provides an outline of the methods that were used for data collection, as reported by Professor Ray Lloyd, and a detailed description of the methods used for secondary analysis.

5.2.1. Outline of data collection methods

5.2.1.1. Participants

Eighteen apparently healthy female volunteers with a minimum of 5 years' experience of head load carriage were recruited (age 23 ± 3.8 years, mass 61.1 ± 10.7 kg, stature 1.59 ± 0.08 metres). All participants were accustomed to carrying 20 kg loads on the head (typical load for water carrying; Porter et al., 2013). All volunteers gave written informed consent to participate. A *post hoc* power calculation performed using G*Power© software determined that 95% power was achieved using a sample size of 18, based on an anticipated medium effect size (Richardson, 2011).

5.2.1.2. Experimental design

All trials were conducted at the Human Performance Laboratory at Cape Peninsula University of Technology. Figure 11 provides an overview of the study design. Participants attended the laboratory on four separate occasions in order to complete a habituation session and three different trial conditions. Trial conditions differed in load carriage method, with load carried on directly the head (Head), on the back (Back) and evenly distributed between the front and back (Back/Front). Each participant chose, at random, the loading method for each experimental trial (via the picking of a marked piece of paper from a hat). Trials involved seven, four-minute periods of walking at 3 $km{\cdot}h^{-1}$, with each period separated by two minutes of rest. The initial stage was performed unloaded, followed by loads of 3, 6, 9, 12, 15 and 20 kg. Participants were asked to maintain a similar diet and refrain from moderate-vigorous exercise and alcohol consumption in the 24 hours prior to each test.

Overview of the study:

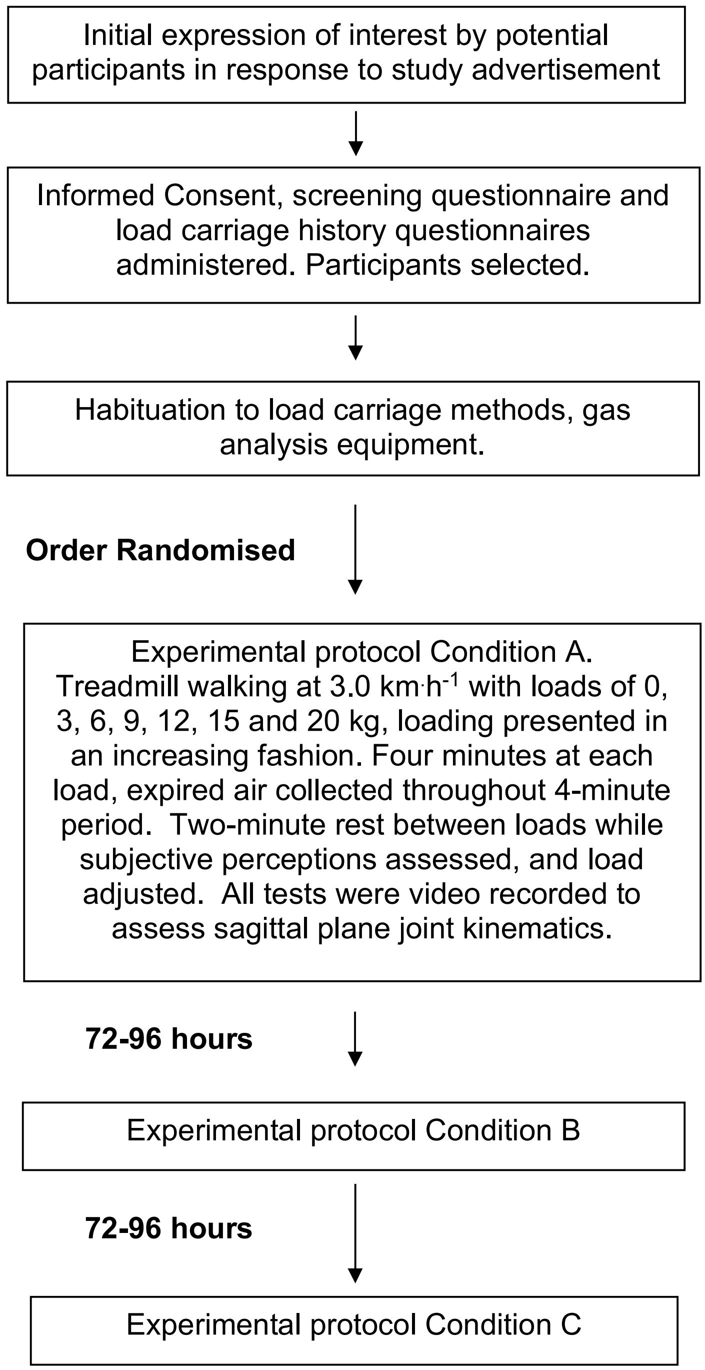


Figure 11. Overview of the experimental study design in Chapter 5

5.2.1.3. Experimental procedures

5.2.1.3.1. Loading methods

A traditional 45 litre rucksack (Karrimor, UK) was used for back-loading, a 20 litre plastic bucket was used for head-loading and a load carriage system with front balance pockets was used for front/back loading (AARN design, New Zealand) (Figure 12). A piece of rolled up material was allowed to provide a cushion between the head and the bucket when head-loading. The mass of the load was made up of the load carriage device itself plus sandbags to the nearest 50 g.

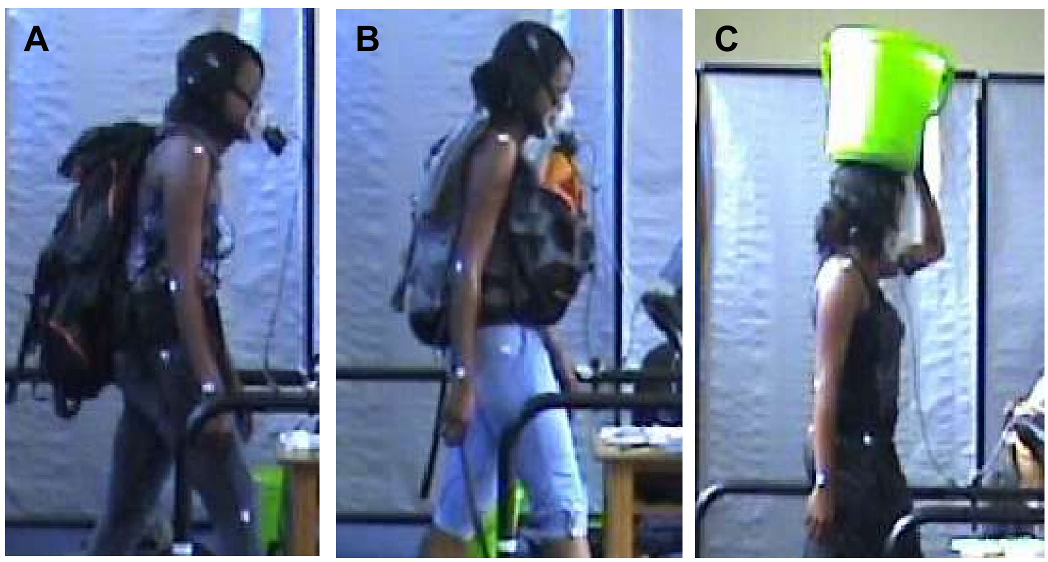


Figure 12. Still images showing the load carriage devices used in each condition. (A) Sagittal plane view of the Back condition. (B) Sagittal plane view of the Back/Front condition. (C) Sagittal plane view of the Head condition.

5.2.1.3.2. Main trials:

Each trial began by measuring the participant's body mass in order to calculate the ELI for that trial. Participants were then fitted with a face mask and a heart rate monitor (Polar, Finland) and asked to walk unloaded on the treadmill at 3 $km{\cdot}h^{-1}$ for four minutes at 0% gradient. After four minutes, there was a two-minute rest period during which the participants were fitted with the appropriate loading device for the trial. The initial load was set at 3 kg. At the end of the rest period, participants recommenced walking at the same speed for a further four minutes. This pattern of work and rest continued with loads of 6, 9, 12, 15 and 20 kg being carried in subsequent stages. Ratings of perceived exertion were recorded in the final 30 seconds of each stage and VAS were completed in between each period of walking to assess subjective perceptions of perceived pain/discomfort.

5.2.2. Secondary analysis methods

5.2.2.1. Expired gas analysis

$\dot{V}O_2$ in the final minute of each walking period was used to calculate the ELI and the energy cost of walking per unit distance (Cw; Equation 8; Abe et al., 2004) for each loading condition

$$Cw = ml / [BM + L] / m \qquad \text{Equation 8}$$

where ml refers to millilitres of $\dot{V}O_2$, BM refers to the body mass of the participant, L is the additional load mass and m is the distance covered in metres.

The gross metabolic rate per kilogram of body mass (W/kg) was also calculated from $\dot{V}O_2$ and $\dot{V}CO_2$ using the Brockway (1987) equation and assuming zero protein metabolism.

5.2.2.2. Kinematic analysis

Video files were manually digitised to analyse sagittal plane kinematics (SIMI 8.5.6, Germany). Six steps from the final minute of each stage were digitized. Intra-observer digitising reliability was completed prior to the full digitisation of trials. Overall, there was a good level of reliability with a highest relative TEM of 0.6% for trunk angle when walking unloaded at heel-strike. The full results of intra-observer digitising reliability conducted prior to digitising can be seen in Appendix L. Once the reconstruction was complete, raw data were filtered using a 2nd order Butterworth filter at 6 Hz. Joint angles were for each step at two events of the step cycle (heel-strike and toe-off).

Trunk, hip, knee and ankle joint angles were calculated by the SIMI motion software (SIMI motion, Germany) at heel-strike, mid-support and toe-off gait events. A single value for each joint angle was also calculated as the average from the three events each step cycle. Joint angle excursions were measured as the change in angle from heel-strike to toe-off in each step. Step length, cadence, step time, double stance time and single stance time were measured by visually inspecting each video for time periods between heel-strike and toe-off gait events.

5.2.2.3. Data and statistical analysis:

Mean, SD and CV were calculated for each dependant variable. Joint angles, joint angle excursions and step parameters were analysed as the change from unloaded to loaded walking. Normal distribution of data was verified using the Shaprio Wilk test and visually exploring boxplots. A one-way ANOVA with repeated measures was used to test for significant main effects of method for all unloaded walking variables. To assess for differences between conditions, a two-way repeated measures ANOVA (load method x load mass) was conducted to establish any significant main effects and interactions. A three-way repeated measures ANOVA was used to assess VAS data (method x mass x body position). Post-hoc tests for significant main effects were conducted using a Bonferroni correction. Pearson's Product Moment Correlation Coefficients were calculated to explore the relationships between ELI values and joint angles, joint angle excursions and step parameters for each loading condition. Statistical significance was set at $p < 0.05$ in all experimental chapters. Where $p < 0.10$, the results are reported as being close to statistical significance.

As well as SD and CV to assess inter-individual variation, linear multi-level models (MLM), using maximum likelihood estimation, were created for $\dot{V}O_2$, ELI, Cw and gross metabolic data with each method of load carriage. The MLM's were used to estimate the variance between participants (σ^2_u) and the variance between the load masses (σ^2_e) for each load carriage method. Intra-class Correlation Coefficients (ICC) were calculated from the variance components in each MLM to represent the proportion of total variability in the outcome that was attributable to individual differences between participants. The range of percentage change from unloaded walking across participants was also assessed for all joint angles and step parameters.

5.3. Results

There were no significant differences between trial conditions when walking unloaded for any of the physiological or biomechanical variable assessed ($p > 0.05$, Table 16).

Table 16. Mean ± SD differences in $\dot{V}O_2$, $\dot{V}_E$, joint angles, joint angle excursions and step parameters between trial conditions (Head, Back, Back/Front) when walking unloaded.

	Trial Condition			*p* value
	Head	Back	Back/Front	
$\dot{V}O_2$ ($l \cdot min^{-1}$)	0.62 ± 0.14	0.63 ± 0.13	0.63 ± 0.13	0.760
$\dot{V}O_2$ ($ml \cdot kg^{-1} \cdot min^{-1}$)	10.20 ± 1.50	10.35 ± 1.42	10.42 ± 1.18	0.761
$\dot{V}_E$ ($l \cdot min^{-1}$)	17.50 ± 3.79	17.72 ± 3.05	17.38 ± 3.79	0.847
Trunk forward lean (°)	87.9 ± 2.6	87 ± 3.5	87.4 ± 2.9	0.570
Trunk angle excursion (°)	4.1 ± 1.9	3.9 ± 1.5	4.3 ± 1.8	0.767
Hip angle (°)	166.3 ± 10.0	165.9 ± 9.6	166.4 ± 8.6	0.946
Hip angle excursion (°)	19.2 ± 3.4	17.7 ± 4.4	16.6 ± 3.8	0.127
Knee angle (°)	152.2 ± 17.3	152.7 ± 17.7	152.8 ± 17.6	0.927
Knee angle excursion (°)	-33.5 ± 8.1	-34.8 ± 7.3	-36.5 ± 6.0	0.292
Ankle angle (°)	99.9 ± 6.8	100.1 ±7.4	100.2 ± 7.1	0.987
Ankle angle excursion (°)	4.1 ± 3.3	3.9 ± 4.0	4.7 ± 4.1	0.717
Step length (metres)	0.50 ± 0.02	0.50 ± 0.03	0.49 ± 0.03	0.646
Cadence ($steps \cdot sec^{-1}$)	1.67 ± 0.08	1.69 ± 0.09	1.69 ± 0.11	0.659
Step time (seconds)	0.60 ± 0.03	0.60 ± 0.03	0.59 ± 0.04	0.713
DST (seconds)	0.23 ± 0.02	0.23 ± 0.02	0.22 ± 0.02	0.232
SST (seconds)	0.37 ± 0.03	0.37 ± 0.03	0.37 ± 0.03	0.994

* DST = Double stance time; SST = Single stance time

5.3.1. Physiological variables

5.3.1.1. Rate of oxygen consumption ($\dot{V}O_2$)

There were no significant differences in $\dot{V}O_2$ between the three loading methods (main effect of load carriage method, $p = 0.814$, $\eta^2 = 0.012$) but $\dot{V}O_2$ did increase significantly as the mass of the load increased (main effect of load mass, $p < 0.001$, $\eta^2 = 0.743$). Post-hoc analysis indicated that $\dot{V}O_2$ significantly increased from unloaded walking with the 9, 12, 15 and 20 kg loads ($p < 0.05$). Figure 13 shows the interactions between load mass and the three loading methods. The pattern of response was similar between the three load methods and this was confirmed by a lack of interaction effect between load method and load mass ($p = 0.151$, $\eta^2 = 0.089$).

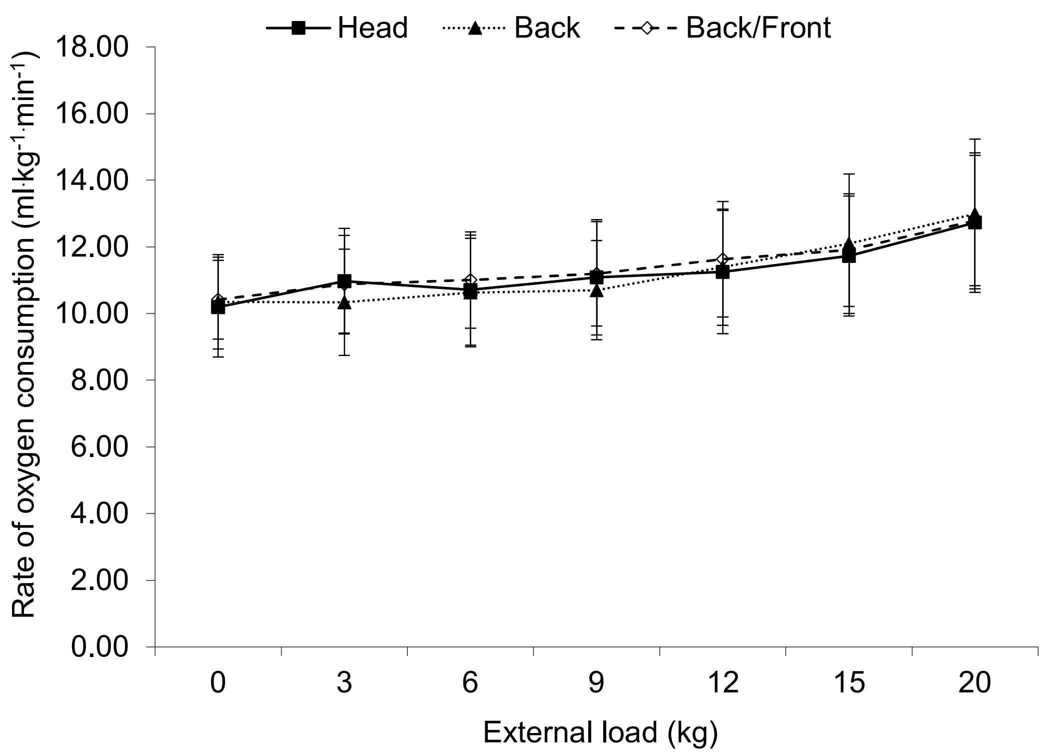


Figure 13. Mean ± SD rate of oxygen consumption (ml·kg^{-1}·min^{-1}) values for each loading condition and load mass.

5.3.1.2. Relative load carriage economy (ELI)

ELI values were not significantly different between loading methods (main effect of load carriage method, $p = 0.483$, $\eta^2 = 0.042$). The average ELI values across all load mass were 0.95 ± 0.11, 0.93 ± 0.08 and 0.94 ± 0.06 for head, back and back/front, respectively. There was a significant difference in ELI between the load masses (main effect of load mass, p = 0.001, η^2 = 0.328). However, there was no significant load method x load mass interaction (p = 0.094, η^2 = 0.107). Figure 14 shows that in the Back condition, economy decreased as the mass of the load increased from 3 kg (ELI = 0.95 ± 0.06) to 9 kg (ELI = 0.90 ± 0.07) and then increased again as the load mass increased from 9 kg to 20 kg (ELI = 0.94 ± 0.11). For Back/Front, the ELI values decreased from 3 kg (ELI = 0.99 ± 0.06) as the load mass increased up to 15 kg (ELI = 0.91 ± 0.07). For Head, ELI was highest with 3 kg (ELI = 1.03 ± 0.08) and lowest with 12 kg (0.92 ± 0.09).

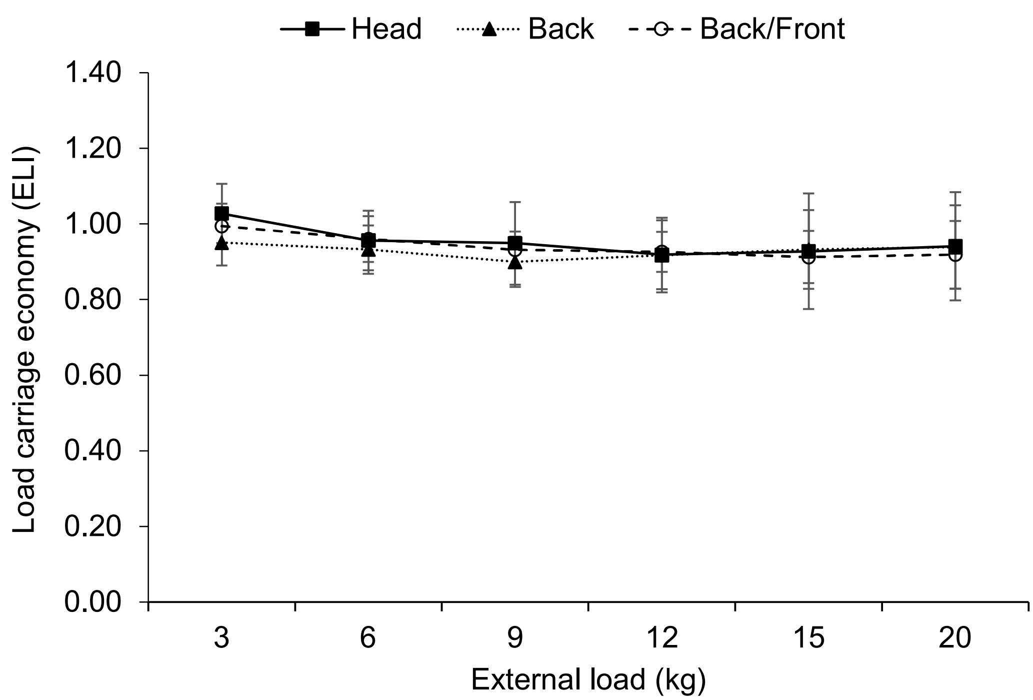


Figure 14. Mean ± SD ELI values for each loading method and load mass.

Figure 15 shows the results for load carriage economy presented as the energy cost of walking per unit distance (Cw). There was no significant difference in Cw between loading methods (main effect of load carriage method p = 0.802, η^2 = 0.013). The Cw was significantly different between the load masses (main effect of load mass, p < 0.001, η^2 = 0.421), with post-hoc analysis revealing a significant decrease in Cw from unloaded to loaded walking (p < 0.05). The largest decrease in *C*w from unloaded walking was for the Back method with 9 kg (-0.021 ml·kg^{-1}·metre^{-1}). For Head and Back/Front, the largest decrease from unloaded was with 12 kg (-0.017 ml·kg^{-1}·metre^{-1}) and 15kg (-0.018 ml·kg^{-1}·metre^{-1}), respectively. There was no significant interaction effect between load method and load mass (p = 0.113, η^2 = 0.096).

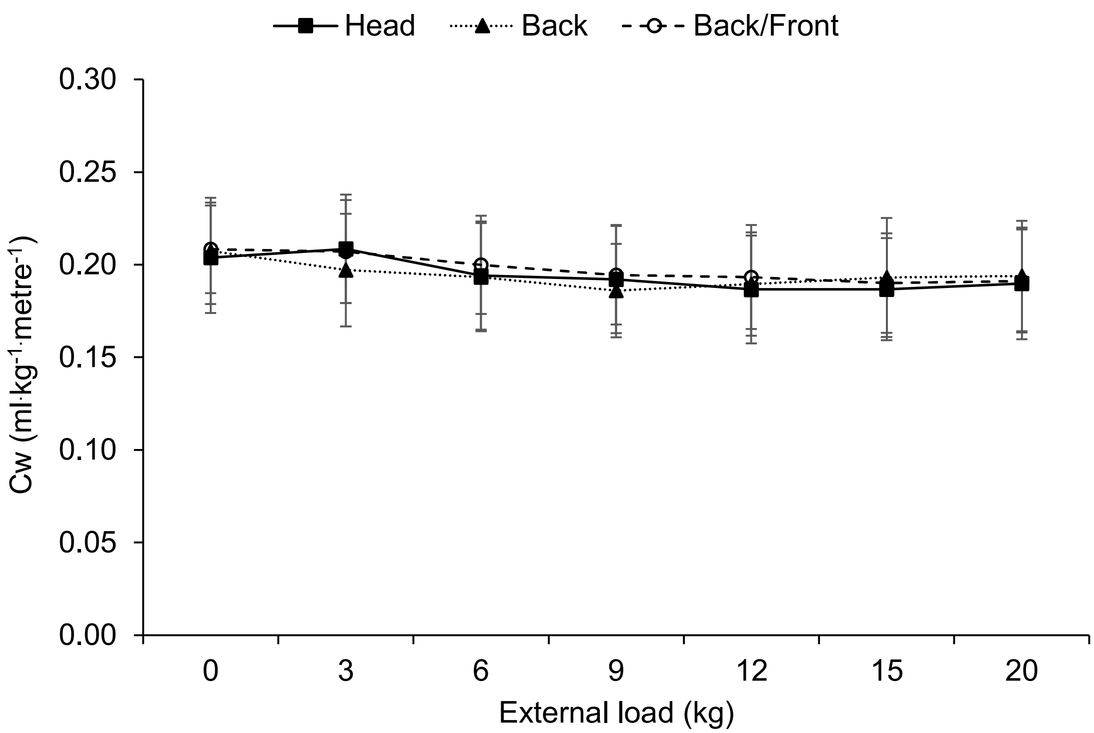


Figure 15. Mean ± SD the energy cost of walking per unit distance (Cw) for each load method and load mass.

5.3.1.3. Gross metabolic rate

The metabolic rate per kilogram body mass (W/kg) was not significantly different between load carriage methods (main effect of load carriage method, p = 0.893, η^2 = 0.005). The metabolic rate increased significantly with an increase in load mass (main effect of load mass, p < 0.001, η^2 = 0.752) with post-hoc analysis indicating that that the metabolic rate increased significantly from the unloaded walking condition with 6, 9, 12, 15 and 20 kg (p < 0.05). There appeared to be a similar pattern of response between each of the load carriage methods, which was confirmed by a lack of interaction between load carriage method and load mass (p = 0.224, η^2 = 0.089).

5.3.1.4. Minute ventilation, breathing frequency and tidal volume

Minute ventilation significantly increased as the mass of the external load increased (main effect of load mass, p < 0.001, η^2 = 0.735; Figure 16) with all loading methods. However, there was no significant difference between the three methods (main effect of load carriage method, p = 0.323, η^2 = 0.062). There was a similar pattern of response to increasing load mass for breathing frequency, with an increase as the mass of the load increased (main effect of load mass p < 0.001, η^2 = 0.641) but no difference between methods (main effect of load carriage method, p = 0.553, η^2 = 0.034). Tidal volume was larger for unloaded walking when carrying 20 kg in all loading methods (main effect of load mass, p = 0.004, η^2 = 0.260), with increases of 0.07 ± 0.10 litres, 0.02 ± 0.06 litres and 0.04 ± 0.10 litres for Head, Back and Back/Front, respectively. There was no significant difference in tidal volume between load carriage methods (main effect of load carriage method, p = 0.076, η^2 = 0.141), although there was a tendency for tidal volume to increase from unloaded in the Head condition compared to Back and Back/Front. The average Δ tidal volume from unloaded walking across all load mass was 0.04 ± 0.08 litres, 0.00 ± 0.06 and 0.02 ± 0.08 for Head, Back and Back/Front, respectively.

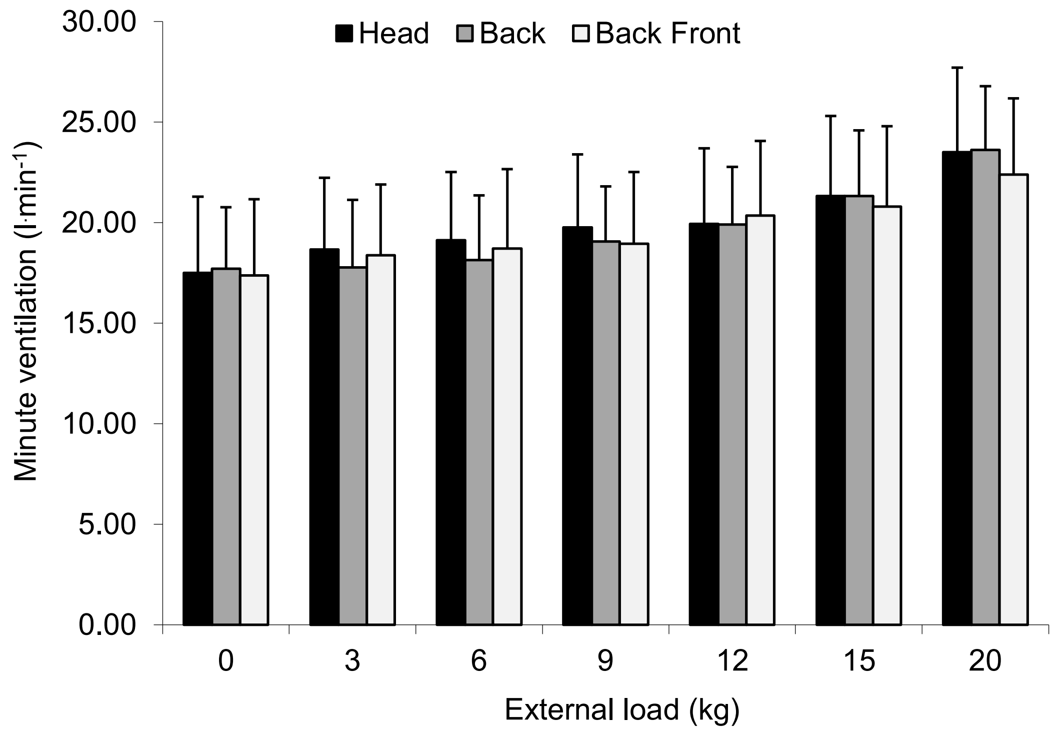


Figure 16. Mean ± SD minute ventilation ($l \cdot min^{-1}$) for each load method and load mass.

5.3.2. Kinematic variables

5.3.2.1. Trunk movement

Figure 17 shows ∆ trunk forward lean with each of the three loading methods and each load mass. The ∆ trunk forward lean was significantly different between loading methods (main effect of load carriage method, $p < 0.001$, $\eta^2 = 0.847$) and load mass (main effect of load mass, $p < 0.001$, $\eta^2 = 0.715$). Post-hoc analysis revealed significant differences in trunk forward lean between all three methods ($p < 0.05$). There was also significant interaction effect between load method and load mass ($p < 0.001$, $\eta^2 = 0.754$). In both the Back and Back/Front methods, ∆ trunk forward lean increased each time the external mass increased. This increase was much greater in the Back method, (10.7° increase from 3 kg to 20 kg) compared to the back/front method (2.4° increase from 3 kg to 20 kg). In the Head method, ∆ trunk forward lean decreased as the load mass increased (-2.2° decrease from 3 kg to 20 kg).

The Δ trunk angle excursion during the stance phase (heel-strike to toe-off) (Figure 18) was significantly different between loading methods (main effect of load carriage method, p = 0.021, η^2 = 0.203) and load mass (main effect of load mass, p = 0.004, η^2 = 0.183). Post hoc analysis showed no significant pairwise comparisons but there was a tendency for reduced trunk angle excursion for Back (all mass pooled = -2.2°) compared to Head (all mass pooled = -1.5°) (p = 0.059). There was also a significant interaction effect between load method and load mass (p = 0.001, η^2 = 0.165). In the Back method, Δ trunk angle excursion decreased as the mass of the load increased (-3.2° from 0 – 20 kg). The Δ trunk angle excursion also decreased with both the Back/Front and Head methods, although there was not a consistent pattern of response for these two methods across the different load masses.

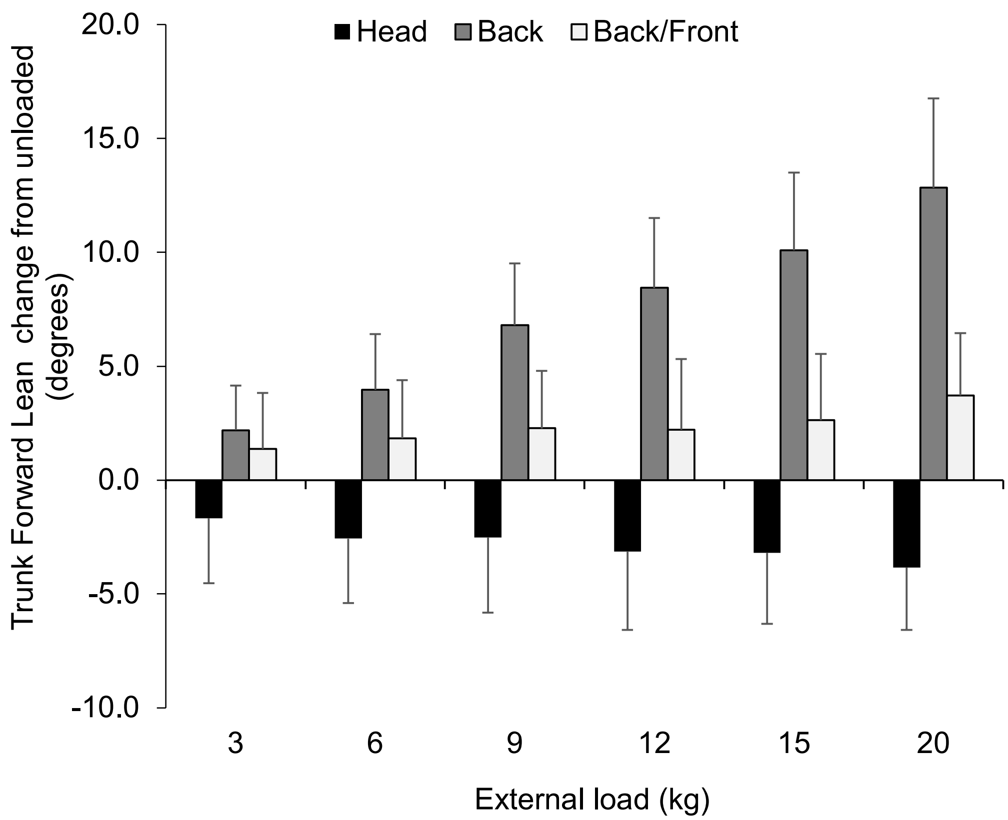


Figure 17. Mean ± SD change in trunk forward lean (degrees) from the unloaded condition for each loading method and each of the load masses. Positive and negative values indicate increased and decreased forward lean, respectively.

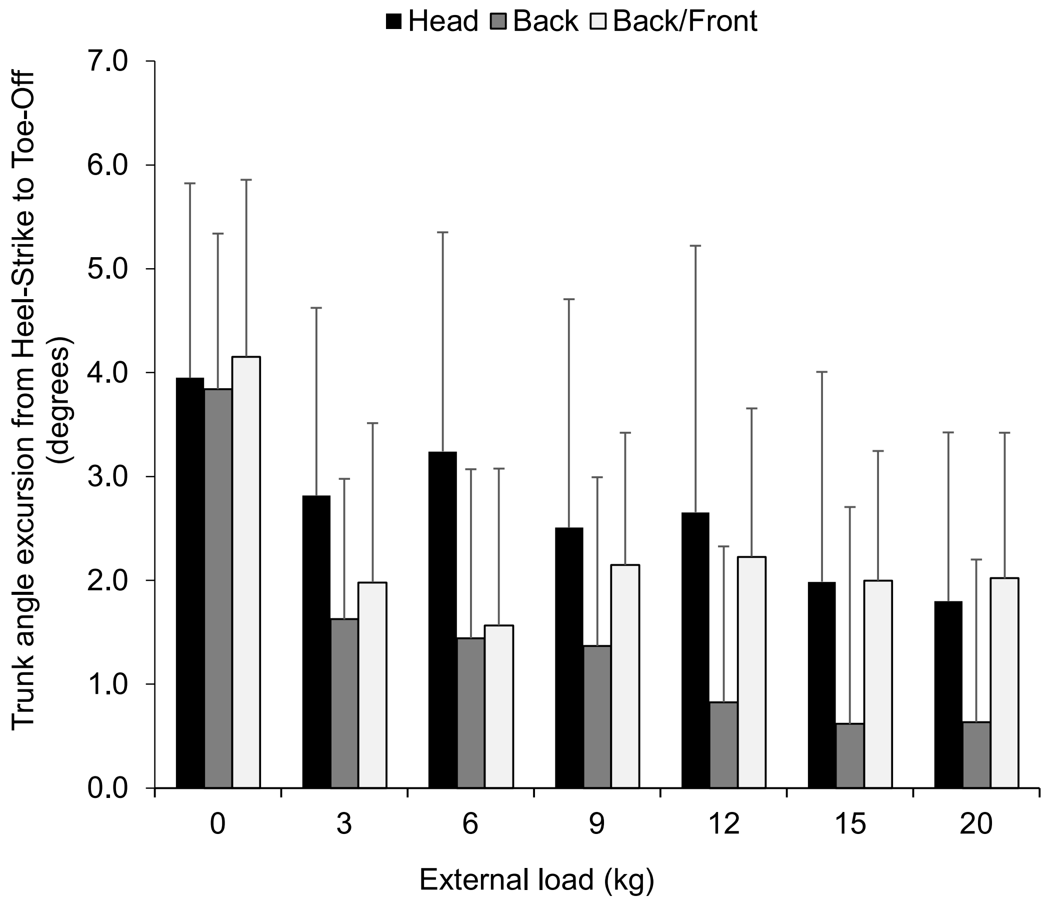


Figure 18. Mean ± SD Trunk angle excursion (degrees) values during the stance phase from heel-strike to toe-off with each loading method and each of the load masses.

5.3.2.2. Hip movement

There was a significant difference in the Δ hip angle from unloaded to loaded walking between methods (main effect of load carriage method, $p < 0.001$, $\eta^2 = 0.754$) and load mass (main effect of load mass, $p < 0.001$, $\eta^2 = 0.808$). In the Back condition, hip angle decreased as the mass of the external load increased with a -2.2° decrease from unloaded walking with 3 kg and a -13.4° decrease with 20 kg. There was also a trend for hip angle to decrease from unloaded walking as the mass of the load increased (Δ hip angle was -1.3° and -5.4° for 3 kg and 20 kg, respectively) in the Back/Front condition, with the exception of the 12 kg load. This pattern of response to additional load mass was not the same across loading methods and in the Head condition, hip angle increased from unloaded to loaded walking but there was little difference between the load masses (the largest difference between load mass was 0.8° between 9 kg and 20 kg). There was a significant interaction effect between load method and load mass (p = 0.001, $\eta^2 = 0.652$).

The ∆ hip angle excursion during the stance phase (Figure 19) was significantly different between loading methods (main effect of load carriage method, $p < 0.001$, $\eta^2 = 0.750$) and load mass (main effect of load mass, $p = 0.001$, $\eta^2 = 0.328$). There was also a significant interaction effect between load method and load mass ($p < 0.001$, $\eta^2 = 0.285$). Post-hoc analysis revealed significant differences in ∆ hip angle excursion between all three methods ($p < 0.05$). In the Back condition, ∆ hip angle excursion increased as the mass of the load increased (∆ hip angle excursion = 4.3° and 13.2° for 3 kg and 20 kg, respectively). There was an increase in hip angle excursion from unloaded to loaded walking in the Back/Front condition, however this did not change with an increase in load mass. With the 3 kg load, ∆ hip angle excursion was similar between Back and Back/Front (4.3° vs 4.4° for Back and Back/Front, respectively). As the mass of the load increased, the ∆ hip angle excursion in the Back/Front condition did not concomitantly increase. There was little difference in ∆ hip angle excursion with any of the loads in the Head condition (largest difference = -2.7° with 6 kg).

5.3.2.3. Knee movement

Knee angle decreased (increased knee flexion) from unloaded walking in all loading conditions (Figure 19) but there was no significant difference in ∆ knee angle from unloaded walking between loading methods (main effect of load carriage method, $p = 0.961$, $\eta^2 = 0.002$). There was large variation in knee angle as indicated by the standard deviation (Figure 19). Knee flexion increased significantly as the mass of the load increased (main effect of load mass, $p < 0.001$, $\eta^2 = 0.440$). The largest increase in knee flexion from unloaded walking occurred with the 20 kg load in all methods (head = -2.7°, back = -3.3°, back/front = -2.4°).

The pattern of response for the ∆ knee angle excursion during the stance phase to increasing load mass was similar to that of the ∆ hip angle excursion in all loading methods (Figure 19). The ∆ knee angle excursion during the stance phase was significantly different between load carriage methods (main effect of load carriage method, $p < 0.001$, $\eta^2 = 0.380$) and load mass (main effect of load mass, $p < 0.001$, $\eta^2 = 0.321$). There was also a significant interaction effect between load method and load mass ($p = 0.028$). In the Back and Back/Front loading methods, there was an increase in knee angle excursion from heel-strike to toe-off compared to unloaded walking. For Head, knee angle excursion tended to decrease from unloaded walking. The largest Δ knee angle excursion occurred with 6 kg in the Head condition (-3.5°), 20 kg in the Back condition (3.4°) and 9 kg in the Back/Front condition (9.8°).

5.3.2.4. Ankle movement

There was no significant difference between load carriage methods for the Δ ankle angle from unloaded walking (main effect of load carriage method, $p = 0.301$, $\eta^2 = 0.065$; Figure 19). There was also no significant difference in the Δ ankle angle between different load mass (main effect of load mass, $p = 0.142$, $\eta^2 = 0.101$). The ∆ ankle angle excursion during the stance phase (Figure 19) was not significantly different between load carriage methods (main effect of load carriage method, $p = 0.198$, $\eta^2 = 0.093$) but there was a significant difference between load mass (main effect of load mass, $p = 0.018$, $\eta^2 = 0.176$). Post-hoc analysis revealed that there was a significant difference in Δ ankle angle excursion between the 3 kg and 12 kg mass ($p = 0.048$), and the 3 kg and 20 kg mass ($p = 0.002$). There was no significant interaction effect between load method and load mass ($p = 0.149$). In the Back and Head methods, the largest Δ ankle angle excursion from unloaded to loaded walking occurred with 20 kg (Back = 4.5°, Head = 2.5°). There was very little change from unloaded walking with the Back/Front method, with the exception of the 9 kg load (∆ ankle angle excursion with 12 kg = 3.2°). Two participants had a large ∆ ankle angle excursion in the Back/Front 12 kg condition, which is responsible for the increase in angle ankle for this loading condition compared to others.

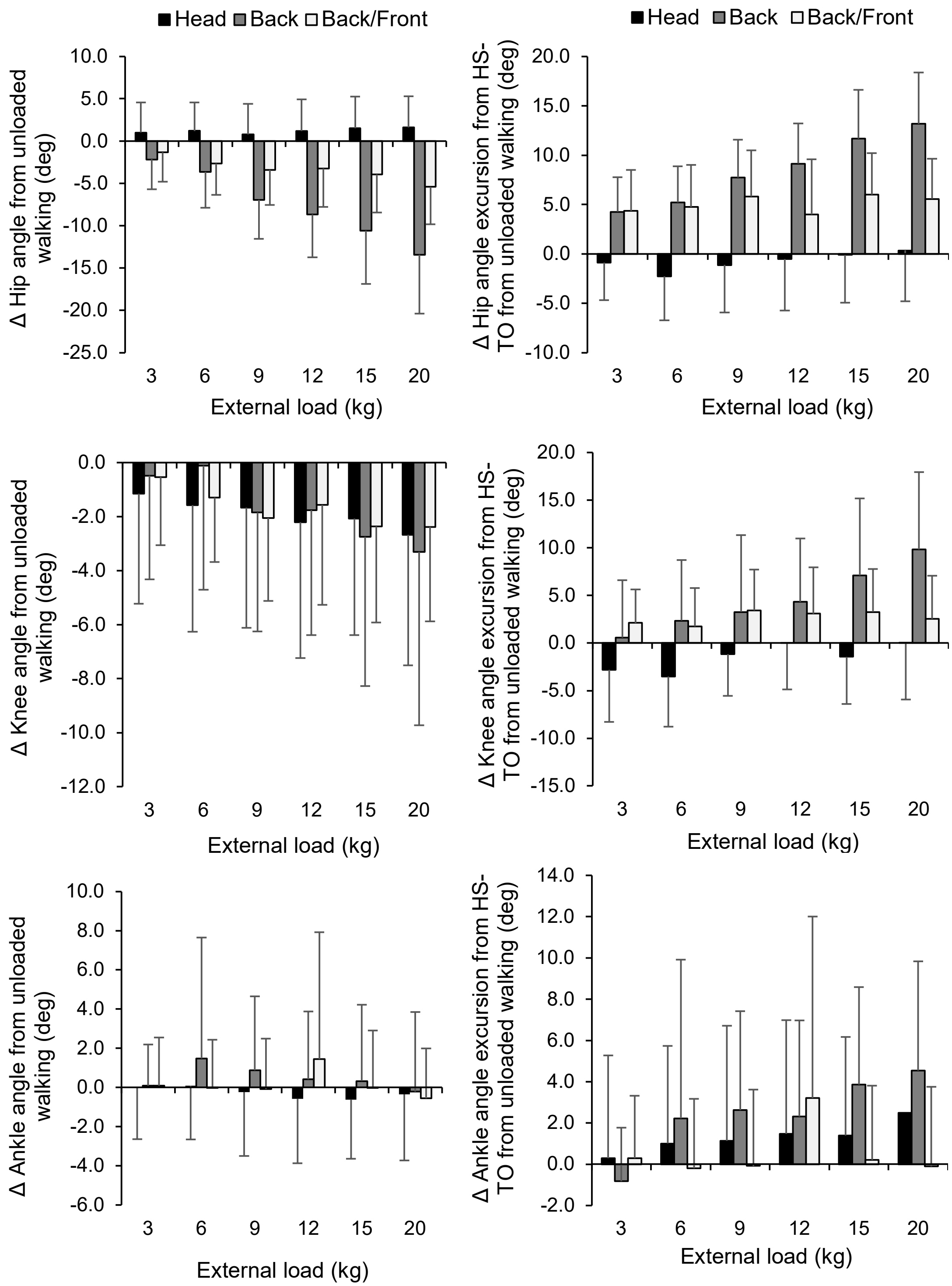


Figure 19. Mean ± SD change in joint angle from unloaded walking for the hip, knee and ankle for each loading condition. For the left column of figure, positive and negative values indicate extension and flexion, respectively.

5.3.2.5. Step length and cadence:

The Δ step length from unloaded walking was significant with load mass (main effect of load mass, $p = 0.028$, $\eta^2 = 0.135$) but not across load carriage methods, although the difference between methods was close to statistical significance (main effect of load carriage method, $p = 0.063$, $\eta^2 = 0.150$). Post-hoc analysis revealed that there were significant differences in Δ step length from unloaded walking between 3 kg and 9 kg loads ($p = 0.041$), 3 kg and 15 kg ($p = 0.022$) and 6 kg and 15 kg ($p = 0.017$). There was no significant interaction effect between load method and load mass ($p = 0.236$, $\eta^2 = 0.077$). The change in step length from unloaded to loaded walking was small in all conditions; the largest change occurred with 15 kg in the Back condition (0.012 ± 0.014 metres). Although the difference in step length was not significant between methods, step length consistently increased from unloaded for back-loading but decreased for Back/Front (Figure 20). The lack of interaction effect between method and mass could be due to the magnitude of the standard deviations in all load carriage conditions. For Head, the Δ step length did not show a consistent pattern with alterations to load mass

Step cadence showed an inverse pattern to step length in all conditions (Figure 20). The Δ step cadence from unloaded walking was not significantly different between loading method, although the difference was close to statistical significance (main effect of load carriage method, $p = 0.061$, $\eta^2 = 0.152$), or load mass (main effect of load mass, $p = 0.121$, $\eta^2 = 0.096$). There was also no significant interaction effect between load method and load mass ($p = 0.225$, $\eta^2 = 0.080$). As with step length, although the difference in cadence was not significant between methods, there were consistent differences in methods across load mass with cadence decreasing from unloaded for Back but increasing for Back/Front.

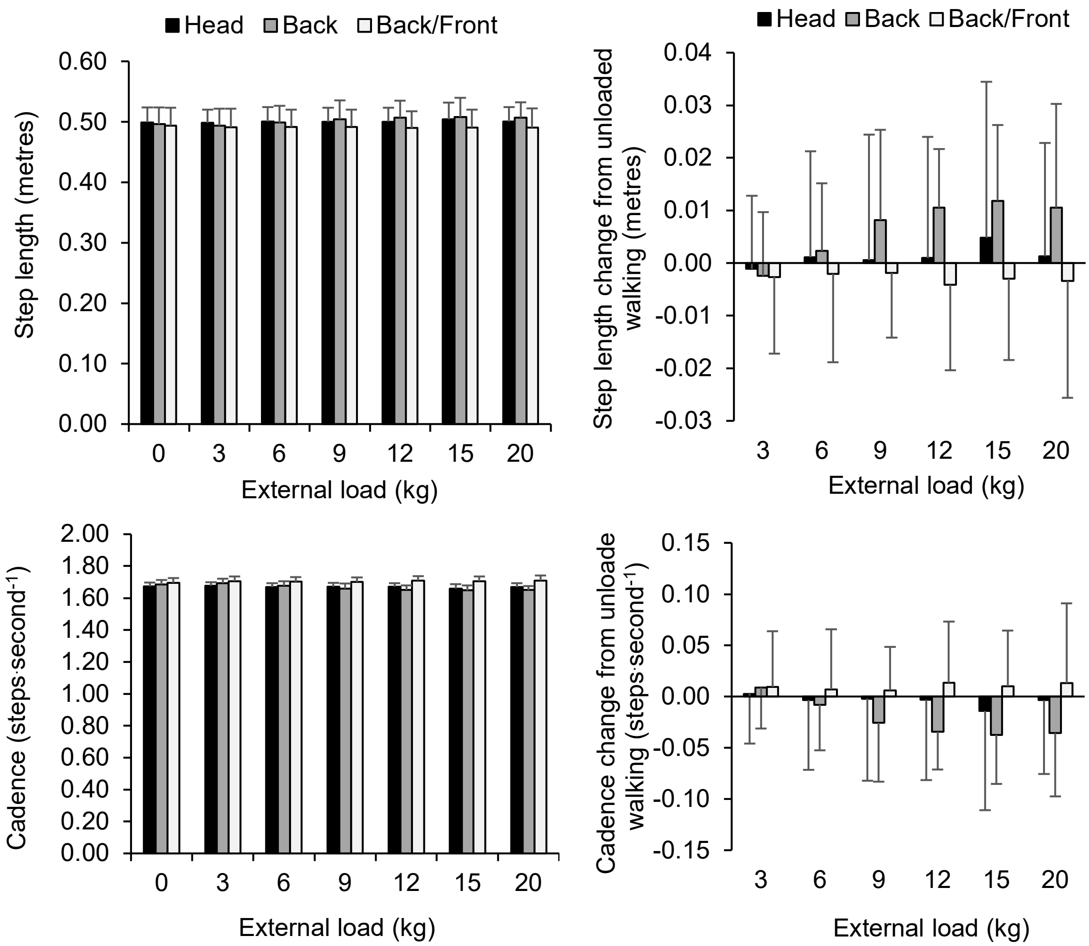


Figure 20. Mean ± SD step length and cadence for each loading condition and change in step length and cadence from unloaded walking for each loading condition.

5.3.2.6. Step time, double stance time and single stance time

There was no significant difference between load carriage methods for the Δ step time from unloaded walking, although the difference was close to statistical significance (main effect of load carriage method, $p = 0.059$, $\eta^2 = 0.153$). There was also no significant difference between load mass for Δ step time from unloaded walking, although the difference was again close to statistical significance (main effect of load mass, $p = 0.061$, $\eta^2 = 0.132$). There was no significant interaction effect between load method and load mass ($p = 0.292$, η^2

= 0.069). There was a tendency for the Back method to result in an increase in step time from unloaded walking and the Head method to result in a decreased step time from unloaded walking (Figure 21).

The Δ double stance time from unloaded walking was significantly different between load carriage methods (main effect of load carriage method, $p = 0.018$, $\eta^2 = 0.210$) and load mass (main effect of load mass, $p = 0.001$, $\eta^2 = 0.666$). Double stance time increased as the mass of the load increased (Figure 21), with a larger increase in the back-loading method compared to both Head and Back/Front. With 20 kg, the Δ double stance time from unloaded walking was 0.028 ± 0.018 s, 0.018 ± 0.012 s and 0.014 ± 0.013 s for Back, Back/Front and Head, respectively. The Δ single stance time from unloaded walking was not significantly different between load carriage methods (main effect of load carriage method, $p = 0.313$, $\eta^2 = 0.066$) but there was a significant difference between load mass (main effect of load mass, $p < 0.001$, $\eta^2 = 0.403$). There was a tendency for single stance time to decrease as the mass of the load increased in all load carriage methods (Figure 21). Post-hoc analysis showed significant differences in Δ single stance time from unloaded walking between 3 kg and 15 kg ($p = 0.044$), 3 kg and 20 kg ($p = 0.002$), 6 kg and 12 kg ($p = 0.030$), 6 kg and 15 kg ($p = 0.003$), 6 kg and 20 kg ($p < 0.001$), and 9 kg and 20 kg ($p = 0.001$).

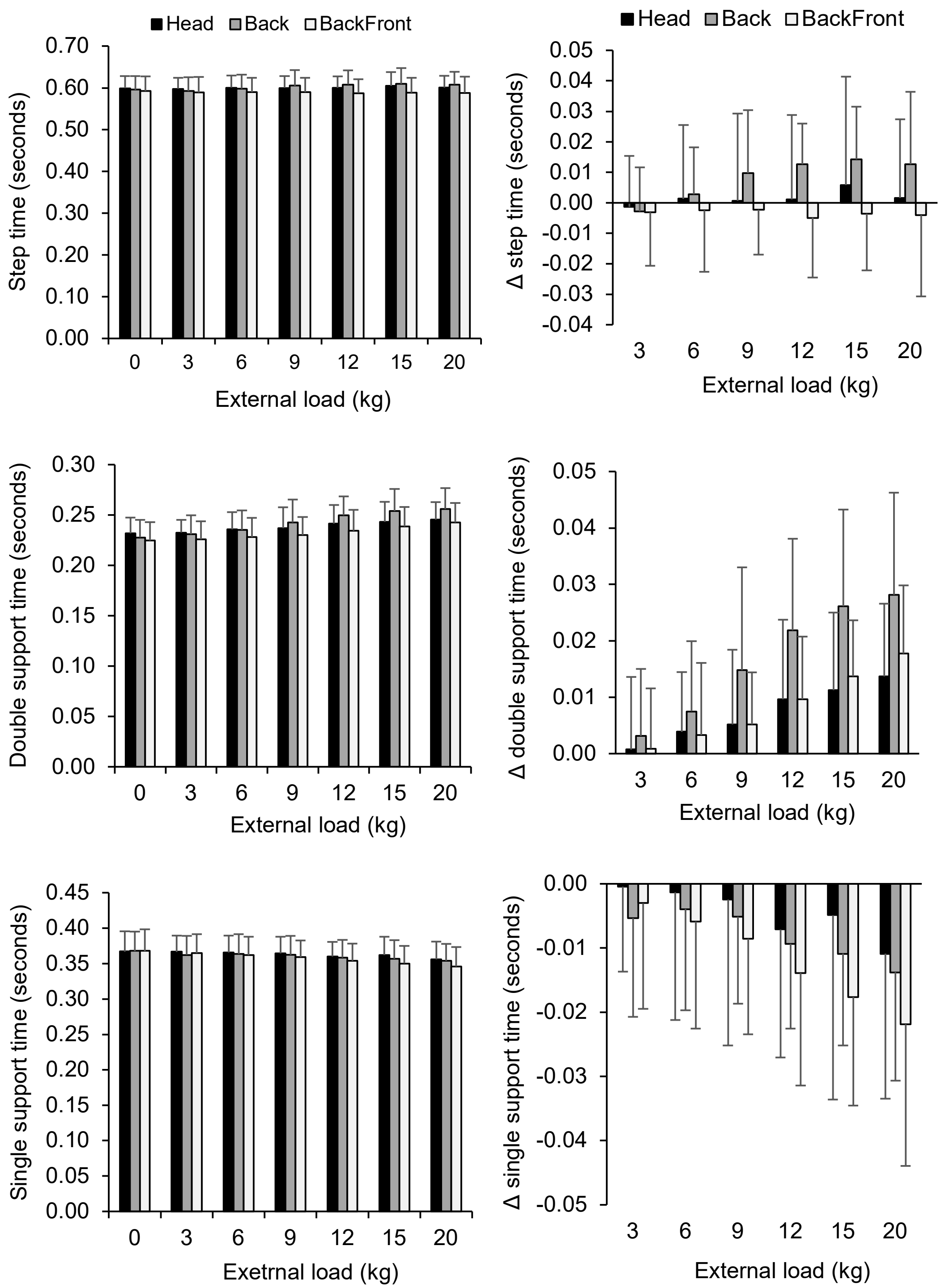


Figure 21. Mean ± SD step time, double stance time, single stance time for each loading condition and change in step time, double stance time and single stance time from unloaded walking for each loading condition.

5.3.3. Relationships

5.3.3.1. Physical characteristics and load carriage economy

There were no significant moderate ($r = 0.4 – 0.7$) or strong relationships ($r > 0.7$) between ELI values and stature, body mass or body mass index (BMI) for any of the methods or load mass. The strongest relationship between ELI and physical characteristics was a negative correlation between ELI and BMI for the Back method with 20 kg (r = -0.319, r^2 = 10.17%, p = 0.196).

5.3.3.2. Joint angles and load carriage economy

Considering relationships between Δ trunk movement and ELI values, there was a significant moderate negative relationship between ELI and Δ trunk angle excursion with the 20 kg load carried using the Back method (r = -0.507, r^2 = 25.70%, p = 0.032). Whereas in the Back/Front method, there was a significant moderate relationship between ELI and Δ trunk forward lean with 9 kg (r = -0.491, r^2 = 24.11%, p = 0.039). In the Head method, there were no moderate-strong relationships between any of the trunk movement variables and ELI (the strongest relationship between ELI and Δ trunk angle excursion was with 3 kg; r = -0.322, r^2 = 10.37%, p = 0.193).

There was a significant strong relationship between Δ hip angle excursion and ELI in the 20 kg Back condition (r = -0.773, r^2 = 59.72%, p = 0.001). There were also moderate relationships between Δ hip angle and ELI in the 6kg and 12 kg Back conditions (r = 0.450, r^2 = 20.25%, p = 0.061 and r = 0.416, r^2 = 17.31%, p = 0.086 for 6kg and 12kg, respectively). Considering the Back/Front method, there was a significant moderate relationship between Δ hip angle and ELI with the 9 kg load (r = 0.534, r^2 = 28.52, p = 0.023).

The only significant relationships between knee movement and ELI was in the 20 kg Back condition. In this condition, there were significant moderate relationships between ELI and change in knee angle from unloaded walking (r = -0.505, r^2 = 25.05, p = 0.032), and ELI and Δ knee angle excursion (r = -0.589, r^2 = 34.69, p = 0.010).

The difference between ankle angle from unloaded to loaded walking was moderately, and sometimes significantly, related to ELI for the Head method with 3 kg (r = -0.405, r^2 = 16.40%, p = 0.095), 6 kg (r = -0.454, r^2 = 20.61%, p = 0.058), 15 kg (r = -0.579, r^2 = 33.52%, p = 0.012) and 20 kg (r = -0.479, r^2 = 22.94%, p = 0.44). In the Back/Front method, the ∆ ankle angle and ∆ ankle angle excursion were also significantly and moderately related to ELI in the 15 kg condition (r = 0.627, r^2 = 39.31%, p = 0.005 and r = - 0.485, r^2 = 23.52%, p =0.042 for ∆ ankle angle and ∆ ankle angle excursion, respectively).

5.3.3.3. Step parameters and load carriage economy

There were significant moderate relationships between ELI and the change in time in double stance from unloaded to loaded walking with 15 kg in the Head condition (r = -0.639, r^2 = 40.83%, p = 0.004) and back-loading condition (r = 0.547, r^2 = 29.92%, p = 0.019). Considering the Back/Front method, there were moderate relationships between ELI and unloaded to loaded walking step time with 9 kg (r = -0.463, r^2 = 21.44%, p = 0.053), and ELI and unloaded to loaded walking step length with 15 kg (r = 0.458, r^2 = 20.98%, p = 0.056).

5.3.4. Subjective perceptions

5.3.4.1. Ratings of perceived exertion (RPE)

RPE scores significantly increased each time the mass of the external load was increased (main effect of load mass, p < 0.001, η^2 = 0.656). The mean RPE for all load mass combined was 10 ± 4, 10 ± 4 and 9 ± 4 for Head, Back and Back/Front, respectively. The difference between methods for change in RPE scores from unloaded walking was close to statistical significance (main effect of load carriage method, p = 0.064, η^2 = 0.149).

5.3.4.2. Pain/discomfort scores:

There was no significant difference between load carriage methods for the change in pain/discomfort scores from unloaded to loaded walking, although the difference between methods was close to statistical significance (main effect of load carriage method, p = 0.345, η^2 = 0.056). As the load mass increased, the change in pain/discomfort from unloaded walking significantly increased (main effect of load mass, $p = 0.009$, $\eta^2 = 0.331$). There was also a significant difference in pain/discomfort between body segments (main effect of body position, $p = 0.012$; $\eta^2 = 0.177$). Table 17 shows the total pain/discomfort scores for each loading condition through the addition of pain/discomfort scores from all body areas. Overall pain/discomfort was similar between methods with the exceptions of the 15 kg Back/Front condition, which was associated with less overall pain/discomfort compared to the other methods (68 ± 13 mm, 81 ± 13 mm and 89 ± 13 mm for Back/Front, Head and Back, respectively), and the 20 kg Back condition, which associated with more overall pain/discomfort than the Head and Back/Front methods (148 ± 20 mm, 113 ± 17 mm and 111 ± 17 mm for Back, Head and Back/Front, respectively).

Table 17. Mean ± SD Sum total pain/discomfort scores (mm) from all body segments combined for each loading condition.

	0 kg	3 kg	6 kg	9 kg	12 kg	15 kg	20 kg
Head	0 ± 0	3 ± 1	9 ± 2	19 ± 4	42 ± 7	81 ± 13	113 ± 17
Back	1 ± 1	3 ± 1	7 ± 2	18 ± 4	40 ± 7	89 ± 13	148 ± 20
Back/Front	0 ± 0	1 ± 1	5 ± 2	15 ± 4	37 ± 7	68 ± 13	111 ± 17

The difference in pain/discomfort scores between body segments is highlighted in Table 18, which shows the difference in scores for each body segment between methods with the 20 kg load. The largest difference between methods occurred at the neck with an increase in pain/discomfort of 21 and 22 mm for the Head method compared to the Back and Back/Front methods, respectively. Table 18 also shows that there were notably lower pain/discomfort scores for Back/Front compared to the other methods at the back of the shoulders, the front of the shoulders and the upper back with 20 kg. Further, the Head method was associated with lower pain/discomfort scores for the lower limbs compared to the other methods.

Table 18. Mean ± SD RPE and pain/discomfort scores (mm) for the 20kg load for each method. Values

	Head	Back	Back/Front
RPE	13 ± 5	13 ± 5	12 ± 6
Neck	24 ± 27	3 ± 3	2 ± 2
Back Shoulders	13 ± 27	19 ± 21	8 ± 15
Front Shoulders	21 ± 32	17 ± 22	7 ± 14
Chest	4 ± 10	1 ± 4	2 ± 8
Upper Back	23 ± 28	17 ± 26	8 ± 17
Abdomen	4 ± 10	2 ± 6	7 ± 15
Lower Back	4 ± 12	12 ± 23	12 ± 20
Hips	2 ± 5	6 ± 22	12 ± 24
Buttocks	1 ± 5	6 ± 16	5 ± 11
Quadriceps	3 ± 7	12 ± 23	11 ± 23
Hamstrings	1 ± 3	12 ± 22	9 ± 21
Knees	5 ± 13	7 ± 21	7 ± 18
Calves	2 ± 6	13 ± 25	7 ± 20
Ankles	1 ± 4	10 ± 23	6 ± 10
Feet	6 ± 14	12 ± 23	7 ± 15
Total	113 ± 17	148 ± 20	111 ± 20

5.3.5. Individual variation

5.3.5.1. $\dot{V}O_2$

The magnitude of standard deviations and coefficients of variation indicates the variability in $\dot{V}O_2$ across the different methods (Table 19).The mean CV for $\dot{V}O_2$ between the three unloaded walking trials was 13%. The MLM analysis showed a significant difference in estimated variance between participants $\dot{V}O_2$ with the Head method (σ^2_u = 2.34, standard error = 0.81, p = 0.004), the Back method (σ^2_u = 2.26, standard error = 0.79, p = 0.004) and the Back/Front method (σ^2_u = 2.00, standard error = 0.67, p = 0.004). The estimated variance in $\dot{V}O_2$ between load mass conditions was also significant for Head (σ^2_e = 0.64, standard error = 0.08, $p < 0.001$), Back (σ^2_e = 0.80, standard error = 0.10, $p < 0.001$) and

Back/Front (σ^2_e = 0.43, standard error = 0.06, p < 0.001). The ICC values for individual differences in $\dot{V}O_2$ as a proportion of the total variance were 0.78, 0.74 and 0.82 for Head, Back and Back/Front, respectively.

Table 19. Mean, standard deviation (SD) and coefficient of variation (CV) for $\dot{V}O_2$ ($ml \cdot kg^{-1} \cdot min^{-1}$) values for each loading method and load mass.

	0 kg	3 kg	6 kg	9 kg	12 kg	15 kg	20 kg
Head							
$\dot{V}O_2$	10.20	10.97	10.71	11.09	11.25	11.73	12.73
SD	1.50	1.59	1.65	1.73	1.85	1.80	2.09
CV (%)	14.71	14.49	15.41	15.60	16.44	15.35	16.42
Back							
$\dot{V}O_2$	10.35	10.34	10.63	10.70	11.39	12.10	12.99
SD	1.42	1.59	1.63	1.49	1.74	2.09	2.25
CV (%)	13.72	15.38	15.33	13.93	15.28	17.27	17.32
Back/Front							
$\dot{V}O_2$	10.42	10.88	11.01	11.19	11.63	11.91	12.79
SD	1.18	1.47	1.45	1.57	1.74	1.69	1.95
CV (%)	11.32	13.51	13.17	14.03	14.96	14.19	15.25

5.3.5.2. ELI

Table 20 shows the mean, standard deviation and coefficients of variation for ELI values with each loading condition. The large magnitude of standard deviations and coefficients of variation indicates the large variability in ELI values across the different methods. The magnitude of the standard deviation and coefficients of variation increased as the mass of the external load increased with all loading methods. Of the three methods, the highest deviation and variation values occurred in the Head method, with the lowest occurring in the Back/Front method. There was significant variance between participants for ELI values with Head (σ^2_u = 0.008, standard error = 0.002, p = 0.006), Back (σ^2_u = 0.003, standard error =

0.001, p = 0.015) and Back/Front (σ^2_u = 0.002, standard error = 0.001, p = 0.013). The estimated variance in ELI between load mass conditions was also significant for Head (σ^2_e = 0.005, standard error = 0.001, p < 0.001), Back (σ^2_e = 0.004, standard error = 0.001, p < 0.001) and Back/Front (σ^2_e = 0.002, standard error = 0.001, p < 0.001). The ICC values for individual differences in ELI as a proportion of the total variance were 0.63, 0.42 and 0.44 for head-, back- and back/front-loading, respectively.

Table 20. Mean, standard deviation (SD) and coefficient of variation (CV) for ELI values for each loading method and load mass.

	3 kg	6 kg	9 kg	12 kg	15 kg	20 kg
Head						
ELI	1.03	0.96	0.95	0.92	0.93	0.94
SD	0.08	0.08	0.11	0.09	0.15	0.14
CV (%)	7.61	8.21	11.49	9.90	16.46	15.21
Back						
ELI	0.95	0.93	0.90	0.92	0.93	0.94
SD	0.06	0.06	0.07	0.10	0.10	0.11
CV (%)	6.39	6.88	7.34	10.76	11.17	11.77
Back/Front						
ELI	0.99	0.96	0.93	0.93	0.91	0.92
SD	0.06	0.06	0.05	0.05	0.07	0.09
CV (%)	6.01	6.32	5.16	5.75	7.55	9.72

There was a difference between methods in the load mass with which the majority of the 18 participants had their lowest ELI value (Figure 22). In the back-loading method, most participants had their lowest ELI value (most economical) with the 9 kg load (n = 7). In the back/front condition, the majority of participants were most economical with the 20 kg load (n = 10). In the head-loading condition, 20 kg was the most economical load (n = 5) but there was little difference between the 20 kg load and 6 kg (n = 4), 12kg (n = 4) and 15kg (n = 3) loads in this condition.

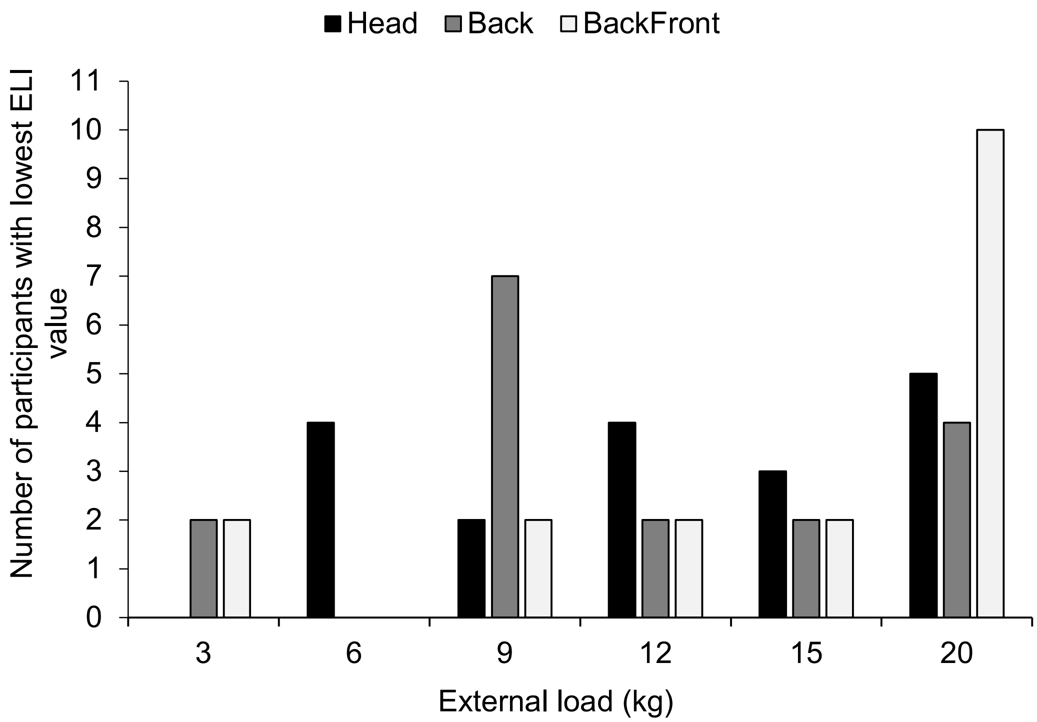


Figure 22. The load mass where participants had their lowest ELI value (most economical) for each method of load carriage.

Considering the most economical method for each load mass, the majority of participants had their best economy for 3 kg (n = 12), 9 kg (n = 9) and 12 kg (n = 7) with the Back method. For 6 kg, eight participants had their best economy with the Head method and eight participants had their best economy with the Back method. For 15 kg and 20 kg, the majority of participants had their best economy with the Head method (n = 10 and n = 7 for 15 kg and 20 kg, respectively).

Figure 23 shows the ELI values for each participant in each method across all load mass (pooled load mass). With load mass pooled, seven participants had their lowest ELI value with the Head method, six had their lowest ELI value with the Back method and five had their lowest ELI with the Back/Front method.

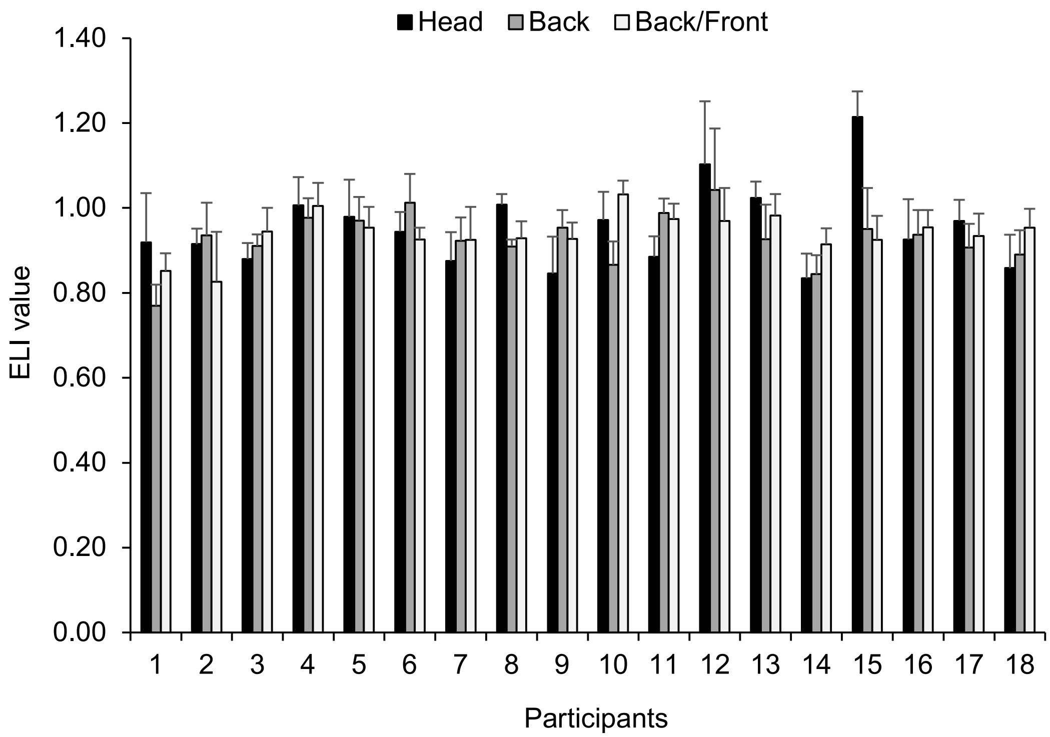


Figure 23. Mean ± SD ELI values for each participant in each condition across loads of 3 - 20 kg.

Figure 24 shows that for each of the loading methods, most participants had their highest ELI values (least economical) with 3 kg. Nine participants had their least economical bout of load carriage (highest ELI) with the Head method, four had their least economical bout with the Back method and five were least economical with the Back/Front method.

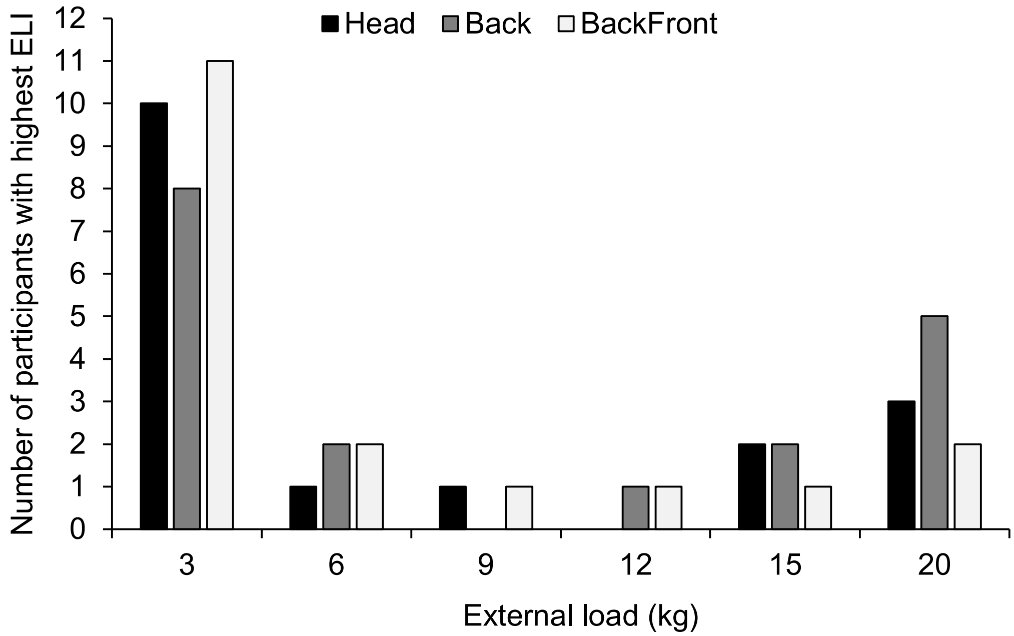


Figure 24. The load mass where participants had their largest ELI (least economical) value for each method of load carriage.

5.3.5.3. Gross metabolic rate

Table 21 shows the mean, standard deviation and coefficients of variation for the metabolic rate across all loading conditions. The variance between participants for metabolic rate was significant with the Head method (σ^2_u = 0.25, standard error = 0.09, $p < 0.01$), the Back method (σ^2_u = 0.25, standard error = 0.01, $p < 0.01$) and the Back/Front method (σ^2_u = 0.21, standard error = 0.07, $p < 0.01$). Between load mass conditions, the estimated variance in metabolic rate was also significant for Head (σ^2_e = 0.08, standard error = 0.01, $p < 0.01$), Back (σ^2_e = 0.09, standard error = 0.01, $p < 0.01$) and Back/Front (σ^2_e = 0.05, standard error = 0.01, $p < 0.01$). The ICC values for individual differences in metabolic rate were 0.77, 0.73 and 0.80 for head-, back- and back/front-loading, respectively.

Table 21. Mean, standard deviation (SD) and coefficient of variation (CV) values for metabolic rate (W/kg) with each load method and mass combination.

	0 kg	3 kg	6 kg	9 kg	12 kg	15 kg	20 kg
Head							
Metabolic rate (W/kg)	3.44	3.70	3.64	3.76	3.82	3.99	4.34
SD	0.48	0.54	0.52	0.57	0.62	0.60	0.69
CV (%)	14.06	14.55	14.35	15.26	16.22	14.99	15.79
Back							
Metabolic rate (W/kg)	3.51	3.51	3.61	3.65	3.88	4.12	4.44
SD	0.48	0.55	0.56	0.50	0.57	0.69	0.75
CV (%)	13.67	15.54	15.55	13.63	14.60	16.74	16.87
Back/Front							
Metabolic cost (W/kg)	3.50	3.68	3.71	3.77	3.95	4.03	4.33
SD	0.39	0.47	0.48	0.52	0.56	0.55	0.63
CV (%)	11.03	12.90	12.93	13.67	14.25	13.58	14.53

5.3.5.4. Kinematic measures

The range of percentage change for kinematic variables from unloaded walking are presented in Table 22 and Table 23. For step parameters (Table 22), the largest range occurred for the change in double stance time from unloaded to loaded walking with all methods. The largest range for change in double stance time from unloaded walking occurred in the Back 20 kg (+39% to +3%). Table 23 shows that the largest range of percentage change for joint angles from unloaded walking. The largest ranges occurred in for the trunk for the Back 12kg (+18% to -5%), 15kg (+20% to -6%) and 20kg (24% to -8%) conditions.

Table 22. Range of percentage change from unloaded walking for step length, cadence, stance time, double stance time and single stance time.

	3 kg	6 kg	9 kg	12 kg	15 kg	20 kg
Head						
Step length (m)	+4% to -8%	+9% to -8%	+12% to -9%	+10% to -8%	+14% to -6%	+10% to -7%
Cadence (steps$\cdot$s^{-1})	+9% to -4%	+9% to -8%	+9% to -10%	+9% to -9%	+7% to -13%	+8% to -9%
Step time (s)	+4% to -8%	+9% to -8%	+12% to -9%	+10% to -8%	+15% to -6%	+10% to -7%
Double stance time (s)	+13% to -11%	+11% to -9%	+15% to -7%	+15% to -11%	+15% to -4%	+18% to -3%
Single stance time (s)	+7% to -4%	+10% to -10%	+11% to -11%	+9% to -13%	+13% to -16%	+8% to -12%
Back						
Step length (m)	+6% to -5%	+5% to -6%	+12% to -3%	+6% to -1%	+7% to -3%	+10% to -4%
Cadence (steps$\cdot$s^{-1})	+5% to -6%	+6% to -4%	+3% to -11%	+2% to -6%	+3% to -7%	+4% to -9%
Step time (s)	+6% to -5%	+5% to -6%	+12% to -3%	+6% to -1%	+7% to -3%	+10% to -4%
Double stance time (s)	+14% to -8%	+15% to -5%	+23% to -5%	+29% to -4%	+29% to 0%	+39% to +3%
Single stance time (s)	+9% to -7%	+6% to -8%	+9% to -6%	+4% to -8%	+5% to -11%	+5% to -11%
Back/front						
Step length (m)	+6% to -7%	+9% to -5%	+4% to -5%	+6% to -7%	+5% to -6%	+8% to -7%
Cadence (steps$\cdot$s^{-1})	+8% to -6%	+6% to -8%	+5% to -4%	+8% to -6%	+6% to -4%	+8% to -8%
Step time (s)	+6% to -7%	+9% to -5%	+4% to -5%	+6% to -7%	+5% to -6%	+8% to -8%
Double stance time (s)	+8% to -6%	+15% to -9%	+8% to -8%	+11% to -4%	+16% to 0%	+22% to +1%
Single stance time (s)	+13% to -10%	+10% to -7%	+10% to -8%	+5% to -11%	+6% to -13%	+6% to -12%

Table 23. Range of percentage change from unloaded walking for trunk, hip, knee and ankle angle during the stance phase.

	3 kg	6 kg	9 kg	12 kg	15 kg	20 kg
Head						
Trunk angle (º)	+4% to -8%	+3% to -7%	+3% to -9%	+3% to -12%	+2% to -10%	-1% to -10%
Hip angle (º)	+5% to -3%	+3% to -2%	+4% to -3%	+6% to -2%	+5% to -2%	+4% to -2%
Knee angle (º)	+2% to -6%	+1% to -9%	+2% to -9%	+3% to -11%	+3% to -8%	-1% to -8%
Ankle angle (º)	+3% to -3%	+4% to -4%	+4% to -6%	+3% to -6%	+2% to -6%	+5% to -6%
Back						
Trunk angle (º)	+5% to -1%	+9% to -1%	+12% to -3%	+18% to -5%	+20% to -6%	+24% to -8%
Hip angle (º)	+1% to -6%	+5% to -5%	-2% to -7%	-3% to -9%	-4% to -10%	-5% to -13%
Knee angle (º)	+3% to -5%	+9% to -3%	+1% to -4%	+3% to -5%	+2% to -4%	+2% to -7%
Ankle angle (º)	+3% to -2%	+18% to -3%	+6% to -4%	+4% to -3%	+5% to -5%	+5% to -5%
Back/Front						
Trunk angle (º)	+8% to -1%	+6% to -2%	+7% to -3%	+8% to -4%	+8% to -1%	+10% to +1%
Hip angle (º)	+2% to -5%	+1% to -5%	+1% to -5%	+2% to -6%	+1% to -7%	+1% to -8%
Knee angle (º)	+1% to -1%	0% to -3%	+1% to -4%	+3% to -3%	+1% to -5%	+2% to -4%
Ankle angle (º)	+4% to -4%	+2% to -3%	+2% to -3%	+15% to -3%	+2% to -5%	+2% to -3%

5.3.6. Summary of the results

Analysis of group data:

- There was no significant difference in ELI values between head- (0.95 ± 0.11 with load mass pooled), back- (0.93 ± 0.08 with load mass pooled), and back/front-loading (0.94 ± 0.06 with load mass pooled) (p = 0.483, η^2 = 0.042; Figure 14).
- There was a significant difference between the three loading methods for the mean joint angle of the trunk (p < 0.001, η^2 = 0.847) and hip (p < 0.001, η^2 = 0.754) across the stance phase. Forward lean increased from 3 to 20 kg for back- (10.7°) and back/front-loading (2.4°), but decreased for head-loading (-2.2°) (Figure 17).
- There was also a significant difference between the three loading methods for joint angle excursion from heel-strike to toe-off at the trunk (p = 0.021, η^2 = 0.203), hip (p < 0.001, η^2 = 0.750), and knee (p < 0.001, η^2 = 0.750). The largest excursions from unloaded walking occurred with 20 kg in the back-loading method which decreased trunk angle excursion (-3.2 ± 0.9°) and increased hip (13.2 ± 5.2°) and knee (9.8 ± 8.1°) angle excursion (Figure 18 and Figure 19).
- There were large effect sizes, but no statistically significant difference, between loading methods for the change in step length (p = 0.059, η^2 = 0.153), cadence (p = 0.061, η^2 = 0.152) or step time (p = 0.059, η^2 = 0.153) from unloaded walking. The change in double support time was significantly different between methods (p = 0.018, η^2 = 0.210) (Figure 20 and Figure 21). The largest change in spatiotemporal variables from unloaded walking occurred with the back-loading method. For this loading method, with load mass pooled, there was increased step length (1.4%), step time (1.4%) and double stance time (7.6%) from unloaded walking, while cadence decreased (-1.3%).
- There were no moderate (r = *0.4 – 0.7*) or strong relationships (r > 0.7) between ELI values and stature, body mass or BMI. For back-loading with 20 kg, ELI significantly correlated with the Δ trunk angle excursion (r = -0.507, p = 0.032), the Δ hip angle excursion (r = -0.773, p = 0.001),

the Δ knee angle (r = -0.505, p = 0.032) and the Δ knee angle excursion r = -0.589, p = 0.010) from unloaded walking. For back/front-loading with 9 kg, there was a moderate relationship between ELI and Δ trunk forward lean (r = -0.491, p = 0.039) and the Δ step time (r = -0.463, p = 0.053) from unloaded walking.

- There was no significant difference in pain/discomfort scores between loading methods. However, there was a notably larger total pain/discomfort (sum of all body segments) for back-loading (148 ± 20 mm) compared to the other methods (Head = 113 ± 17mm; Back/Front = 111 ± 17mm) with 20 kg (Table 18).

Analysis of inter-individual variation:

- The largest CV for $\dot{V}O_2$ occurred with the 20 kg back-loading condition (17%). The largest CV's for the head and back/front methods occurred with 12 kg (16%) and 20 kg (15%), respectively. Inter-individual differences accounted for the largest proportion of the total variance for $\dot{V}O_2$, with ICC values of 0.78, 0.74 and 0.82 for head-, back-, and back/front-loading, respectively.
- The CV's for ELI were larger for the head-loading conditions compared to the other two methods with largest magnitudes of 16%, 12% and 10% for head-, back-, and back/front-loading, respectively (Table 21). The ICC values for individual differences in ELI as a proportion of the total variance were 0.63, 0.42 and 0.44 for head-, back- and back/front-loading, respectively.
- For back-loading, most participants had their lowest ELI with 9 kg (n = 7). For back/front-loading, most participants had their lowest ELI with 20 kg (n = 10). For head-loading, most participants had their lowest ELI with 20 kg (n = 5) (Figure 22).
- Considering spatiotemporal variables, the largest range for the percentage change from unloaded walking occurred for double stance time (3 – 39% for back-loading with 20 kg) (Table 22). Trunk angle had the largest range of response between individuals for the joint angles measured (Table 23). The largest range for the change in trunk angle

from unloaded walking was +24% to – 8% for the back-loading 20 kg condition.

5.4. Discussion

The aims of this study were:

1. To assess the economy and sagittal plane kinematics associated with three methods of load carriage that have all been reported as economical, but all constrain posture differently.
2. To assess the amount of inter-individual variation in economy and sagittal plane kinematics associated with each method of load carriage.

This discussion is split into two parts. The first part is focused on the group data for load carriage economy and the sagittal plane walking gait kinematics associated with back-, back/front- and head-loading (section 5.4.1). The second part is focused on the individual variation in economy and walking gait kinematics for the three methods of load carriage (section 5.4.2).

5.4.1. Group data for load carriage economy and walking gait kinematics

The main findings of the group data in the present study were that load carriage economy was not significantly different between back-, back/front- and head-loading with loads ranging from 3 – 20 kg (Figure 14), despite there being significant differences in sagittal plane kinematics between the three methods.

The pattern of response for load carriage economy was similar between ELI (Figure 14) and *C*w (Figure 15). *C*w was calculated to allow for direct comparisons between the findings of this study and those of Abe et al. (2004), who reported improved economy with 9 kg and 12 kg carried on the back when walking at speeds of 2.4 – 3.6 $km{\cdot}h^{-1}$. The *C*w results for back-loading in this study are similar to those of Abe et al. (2004), with a decrease of -0.02 $ml{\cdot}kg^{-1}{\cdot}metre^{-1}$ from unloaded to loaded walking when 9 kg was carried on the back. As such, the findings of this study support the theory that back-loading is more economical with moderate loads of 9-12 kg than either lighter or heavier loads. In line with the findings of Lloyd and Cooke (2000b), the lowest values for both *C*w and ELI in the back/front-loading method occurred at a heavier load than in the back-loading method. Therefore, although there were no significant differences between back- and back/front-loading, the pattern of response to increasing load mass with these methods lends some support to the suggestion that back-loading is economical with relatively light (up to 12 kg), but not heavy loads, and with heavy loads, back/front-loading appears to be more economical than back-loading. The economy data in this study also show that head-loading was as economical as both back-loading and combined back/front-loading. The head-loading data reported here are consistent with the ELI values reported by Lloyd et al. (2010c) and previous studies that have investigated the metabolic cost of head-loading (Lloyd et al., 2010b, Soule and Goldman, 1969, Nag and Sen, 1979).

Female volunteers with head-loading experience were recruited so that direct comparisons could be made with the research of Maloiy et al. (1986) and Charteris et al. (1989), both of which reported that African women with several years of head-loading experience were able to carry loads of up to 20% body mass with no additional energy expenditure above that required for unloaded walking. Lloyd et al. (2010c) showed that relative load carriage economy is independent of experience, with a similar percentage of experienced and inexperienced head-loaders being more economical at carrying a load on the head than on the back (38.5% vs 36.4% for experienced and inexperienced, respectively). As such, it is unlikely that experience influenced economy in the present study, although this was not controlled for. Increasing the mass of the load resulted in significantly increased $\dot{V}O_2$ with all methods (Figure 13). Therefore, the mean $\dot{V}O_2$ data presented here do not support the existence of an energy-saving phenomenon for experienced head-loaders as suggested by Maloiy et al. (1986) and Charteris et al. (1989). The difference in findings between this study and those of Maloiy et al. (1986) and Charteris et al. (1989) is likely to be explained by differences in sample size. The findings of Maloiy et al. (1986) and Charteris et al. (1989) were based on samples of five women and six women, respectively. Lloyd et al. (2010c) showed that, with a larger sample of participants (n = 24), it is possible to select a subset of women who can achieve remarkable levels of head-loading economy, similar to those reported in earlier studies (Maloiy et al., 1986, Charteris et al., 1989), despite mean group data showing that the energy cost of head-loading rises in proportion to mass of the load carried. Furthermore, Maloiy et al. (1986) The findings of this study support those of Lloyd et al. (2010c) with some women demonstrating better economy when head-loading, while others were more economical at back-loading or back/front-loading (Figure 23), despite there being no difference in economy between methods when comparing the mean data.

Trunk forward lean increased from unloaded walking in the back and back/front methods (Figure 17), with a considerably larger increase for back-loading compared to back/front-loading (8.6 ± 2.5° and 3.5 ± 2.7° when all load masses are combined for back-loading and back/front-loading, respectively). Figure 18 shows a load dependent increase in forward lean in the back-loading condition, with forward lean increasing each time the external mass increased. An increase in Δ trunk forward lean with back-loading compared to evenly distributing the load around the trunk is consistent with previous research comparing backpacks and back/front packs (Kinoshita, 1985, Lloyd and Cooke, 2011). The addition of external mass to the back will have resulted in a greater posterior displacement of the COM of the whole system compared to the back/front condition. Therefore, the increased trunk forward lean when back-loading is likely to have occurred to counter this posterior shift in an attempt to restore the COM of the combined system to the original COM of the body when walking unloaded to improve postural stability (Kinoshita, 1985, Martin and Nelson, 1986, Goh et al., 1998, Harman et al., 2001).

There is a paucity of research examining the postural adjustments associated with transporting a load on the head. The findings of this study show that head-loading causes a decrease in trunk forward lean from unloaded walking. This is likely to be a consequence of the need to balance the load on top of the head requiring individuals to adopt a more upright posture. It was expected that smaller perturbations from the unloaded condition would be associated with an improved economy. However, larger increases in Δ trunk forward lean with the back-loading method were not accompanied by a higher energy expenditure compared to the other conditions. Given the lack of association between trunk forward lean and load carriage economy in this study, it seems unlikely that forward lean alone is directly responsible for any differences in load carriage economy. This is supported by research that has shown relatively low absolute levels of activity in postural muscles associated with forward lean (Motmans et al., 2006, Al-Khabbaz et al., 2008) and suggests that leaning forward to counteract the posterior shift in the position of the COM when back-loading is not a sole determinant for economy with this method of load carriage.

The Δ trunk angle excursion from unloaded walking during single foot contact (heel-strike to toe-off) decreased in all conditions (Figure 18). A decreased trunk angle excursion in the back-loading condition was associated with a concomitant increase in trunk flexion angle each time mass was added, which has been a consistent finding in the literature (Harman et al., 2000, Harman et al., 2001, Attwells et al., 2006, Liew et al., 2016, Yen et al., 2011). With back/front-loading, the trunk angle excursion appeared to be greater than back-loading with 12, 15 and 20 kg loads. Lloyd and Cooke (2000a) demonstrated a requirement for lower peak propulsive force with a back/front load compared to back-loading, which they suggested could represent an energy saving mechanism with back/front-loading, caused by increased momentum associated with a greater joint angle excursion in the trunk. However, in this study, the relationships between Δ trunk angle excursion and ELI for the back/front method with heavier load masses (12, 15 and 20 kg) were weak. In the head-loading condition, trunk angle excursions were largest for most of the loads. This was a surprising finding given that head-loading requires the load to be balanced on top of the head, and it was expected that this would constrain posture in an upright position. Arm movement was not controlled in the present study, with some participants using one or both arms to support the load on the head, while others walked without supporting the load with arms. At first, it was thought that supporting the load with the hands might allow for a greater trunk angle excursion when head-loading. However, there was only a moderate relationship between how the load was supported on the head (no hands, one hand or both hands) and trunk angle excursion (*r*

= -0.465, p = 0.052), and a weak relationship between how the load was supported on the head and ELI (r = 0.316, p = 0.202).

Given the significant differences found in trunk movement between methods, it is not unexpected that sagittal plane hip angle and sagittal plane hip angle excursion were also significantly different between methods. There was a significant difference between methods for both Δ hip and Δ knee angle excursion from unloaded walking. For both variables, the largest change from unloaded walking occurred in the 20 kg back-loading condition (13.2° and 9.8° for Δ hip and Δ knee angle excursion, respectively). The findings from a meta-analysis by Liew et al. (2016) suggest that back-loading is associated with increased sagittal plane hip and ankle angle excursion, with no change in knee angle excursion. In contrast, an increase in knee angle excursion was found in the present study. Previous studies have reported both increased (Harman et al., 2000, Attwells et al., 2006) and unchanged (Majumdar et al., 2010) knee flexion angles in response to back-loading. These equivocal findings are likely, in part, to be caused by differences in study design with differences in walking speed and load mass employed between studies. Individual variation in loaded walking gait kinematics could also be partly responsible for the equivocal findings of previous studies, particularly given the large standard deviations that can be seen in this study (Figure 19).

Back-loading was associated with a small increase in step length (and concomitant decrease in cadence), while back/front-loading was associated with a very small decrease in step length at a set walking speed of 3 $km \cdot h^{-1}$. Similarly, previous studies have indicated that back- and back/front-loading only produce small changes in stride/step length. Back-loading has been the most studied method, with some reporting a slight shortening (Martin and Nelson, 1986, LaFiandra et al., 2003b), no change (Wood and Orloff, 2007, Singh and Koh, 2009) or a slight lengthening of stride length (Lloyd and Cooke, 2011). These equivocal findings could be caused by individual variation in response to load carriage, with Lloyd and Cooke (2011) reporting a change in stride length ranging from +12% to -6% during level walking with 25.6 kg using back- and back/front-loading. In the present study, the change in step length from unloaded walking with 20 kg ranged from +10% to -4% for back-loading, +10% to -7% for head-loading and +8% to -7% for back/front-loading (Table 22). Given that both increases and decreases in stride length have been associated with increases in energy expenditure (Högberg, 1952, Heinert et al., 1988, Cotes and Meade, 1960, Cavanagh and Williams, 1982), Lloyd and Cooke (2011) speculated that fairly large perturbations would have some impact on load carriage economy. However, they found no strong relationships between the change in stride length and economy with load carriage and concluded that perturbations in stride length are insufficient alone to explain differences in load carriage economy between methods. The findings of this study add further support to the suggestion that changes in step/stride length are insufficient in themselves to explain individual differences in economy between back-, back/front- and head-loading.

There was no difference in overall subjective perceptions between head-, back- and back/front-loading. The subjective perception results for back- and head-loading are broadly consistent with those reported by Lloyd et al. (2010d) who also found no significant difference in terms of overall RPE or pain/discomfort scores. However, again similar to (Lloyd et al., 2010d), the present study did find significant differences between the loading methods for the scores of each body segment. In line with the findings of Lloyd et al. (2010d), head-loading was associated with larger pain/discomfort at the neck compared to the other methods, while back-loading was associated with larger pain/discomfort for the shoulders. Table 17 shows the general pattern of response for increasing pain/discomfort with increasing load, which is also a consistent finding in the literature (Mackie and Legg, 2008, Lloyd et al., 2010d). One particularly interesting finding in the subjective perception data was the lower pain/discomfort at the shoulders for back/front-loading with 20 kg compared to the head-loading method. It’s possible that, for head-loading, using the arms to balance the load on the head led to a large magnitude of discomfort at the shoulders, which was equal to that of the pain/discomfort recorded for the back-loading condition with 20 kg. All subjective perception data presented here exhibits a high degree of variability as indicated by the standard deviations. This is likely to be due to several factors including individual pain thresholds as well as individual differences in interpreting the VAS scales.

5.4.2. Individual variation in load carriage economy and walking gait kinematics

Although there was no difference in load carriage economy between methods when comparing the group data, the standard deviations and CV's in Table 20 indicate that there was large inter-individual variation. The highest coefficients of variation were 16%, 12% and 10% for head-, back- and back/front-loading, respectively. The study in Chapter 4 showed that the day-day reliability (CV) for ELI when walking with a rucksack at 3 $km \cdot h^{-1}$ is 4% and 3% for 7 kg and 20 kg, respectively. As such, the variation in load carriage economy found in this study for back-loading cannot be explained by the day-to-day variation in an individual's economy when carrying a load. As such, the individual variation in back-loading economy found in this study cannot be explained by day-to-day variation. This is also likely to be the case for both back/front- and head-loading, particularly given the large coefficients of variation for $\dot{V}O_2$ for both back/front-loading (highest CV = 15%) and head-loading (highest CV = 16%) compared to the day-to-day variation of ~ 5 - 9% previously reported for unloaded walking (de Mendonça and Pereira, 2008, Wergel-Kolmert and Wohlfart, 1999, Blessinger et al., 2009, Darter et al., 2013). Furthermore, the CV in $\dot{V}O_2$ (Table 19) and ELI (Table 20) increased as the mass of the load increased with all methods, indicating that inter-individual variation in load carriage economy increased as the mass of the load is increased. To account for individual differences in substrate oxidation, the metabolic rate (metabolic power per kg body mass) was also measured. Table 21 shows that the group means for metabolic rate displayed a similar pattern of response to the $\dot{V}O_2$ data, with the metabolic rate tending to increase as the mass of the load increased in all loading methods. However, there was little difference in the metabolic rate between the three methods. The CV's for metabolic cost were similar to those for the $\dot{V}O_2$ indicating that the variability in metabolic costs was not related to differences in substrate utilisation.

Only one participant was most economical with the same load mass across all loading methods, which suggests that economy with one method does not predict economy with another. Figure 22 shows that in the back/front condition, the majority of participant's were most economical with the 20 kg load (n = 10). This finding offers some support to studies that have found back/front-loading to be more economical than back-loading when carrying heavier loads (Datta and Ramanathan, 1971, Legg and Mahanty, 1985, Lloyd and Cooke, 2000b, Lloyd and Cooke, 2011). In the back-loading condition, the majority of participants were most economical when carrying the 9 kg load (n = 7). Abe et al. (2004) reported that a load of 9 kg (~15% body mass for their participants) carried on the back yielded a better economy compared to loads of 6 and 12 kg. However, unlike Abe et al. (2004) who selected participants based on their physical characteristics (average body mass in Abe et al. (2004) was 62.1 ± 1.2 kg), the participants in the present study varied in body mass, with a range of 47.9 – 72.6 kg for individuals who were most economical with 9 kg carried on the back. Therefore, the good economy associated with the 9 kg load does not appear to be a consequence of the load representing a particular percentage of body mass. This also appears to be the case for the economy associated with the 20 kg load in the back/front-loading condition, with the body mass of the participant's most economical with this load ranging from 48.8 – 85.4 kg.

The lack of moderate or strong relationships between ELI values and body mass, stature or BMI indicates that individual differences in physical characteristics were not related to the individual differences in relative load carriage economy. This data is in line with the findings of Lloyd et al. (2010a) who showed that ELI is independent of body composition and the magnitude of the external load carried. The lack of significant correlation between ELI and physical characteristics is also likely to explain the difference in interclass correlation coefficients between the ELI data and both the $\dot{V}O_2$ and metabolic rate data. The intraclass correlation coefficients from the MLM's indicate that variance between individuals represented the largest proportion of the total variance in the $\dot{V}O_2$ (ICC = 0.78, 0.74 and 0.82 for head-, back- and back/front-loading, respectively) and metabolic rate data (ICC = 0.77, 0.73 and 0.80 for head-, back- and back/front-loading, respectively). The high proportion of variance assigned to individual differences in $\dot{V}O_2$ and metabolic rate is likely, in part, to be a result of individual differences in body mass (CV = 17.6%). Relative to body mass, the 20 kg load condition represented 23.4% of heaviest participant and 41.5% of the lightest participant, with an average of 33.6% ± 5.6%. It is well established that the energy cost of load carriage increases linearly as the mass of the load increases with both absolute and relative loads (Quesada et al., 2000, Bastien et al., 2005, Christie and Scott, 2005). Therefore, differences in the relative loads between participants is likely to account for some of the large variance in $\dot{V}O_2$ found in this study.

There was an overall trend for the standard deviation and coefficients of variation for relative load carriage economy to increase as the mass of the external load increased, with all loading methods. This finding suggests that the magnitude of individual variation in load carriage economy is dependent on the mass of the load. It is possible that the magnitude of walking gait perturbations, as a consequence of increased load mass, varies between individuals, which could then lead to an increased variance in relative load carriage economy with heavier loads.

There was a large amount of individual variation present in all of the sagittal plane kinematic data, which is indicated by the large standard deviations and large range of percentage change from unloaded walking. Yet, the lack of strong relationships between ELI and Δ joint angles, Δ joint angle excursions and Δ step parameters from unloaded walking indicate that none of these variables alone were associated with determining individual load carriage economy. Given the variability in all kinematic variables, it's possible that several factors might align in individuals to influence economy rather than there be a single set of generalizable factors applicable to all individuals for each method.

A slow speed of 3 $km{\cdot}h^{-1}$ was used in this study to enable comparisons with previous research that have reported an energy saving phenomenon with load carried at slow walking speeds (Maloiy et al., 1986, Lloyd and Cooke, 2000b, Abe et al., 2004). However, not permitting participants to walk at a self-selected speed might have perturbed the individuals normal gait pattern (Martin and Morgan, 1992) and could have contributed to individual variation. The results obtained from controlled laboratory conditions are valuable, but it is important to note that real life load carriage tasks are often performed on uneven terrain at non-constant, self-selected speeds. This may cause additional metabolic costs and biomechanical challenges compared to the laboratory environment and, as such, is a limitation of this research and all laboratory-based load carriage research.

5.5. Conclusion

Based on the mean data presented here, there appears to be no significant difference in load carriage economy between back, back/front and head-loading loading, despite significant differences between the methods in the change in sagittal plane kinematics from unloaded to loaded walking. There was, however, a considerable amount of individual variation in both load carriage economy and sagittal plane kinematics. This study showed no strong correlations between alterations in sagittal plane kinematics caused by load carriage and ELI values. It's likely that biomechanical factors combine to influence load carriage economy rather than there be a single set of generalizable factors, applicable to all individuals for each method.

This chapter highlights the need for a framework to identify and analyse the key biomechanical factors associated with individual load carriage economy. As a single sagittal plane kinematic factor does not appear to determine individual load carriage economy, it is important to understand how mechanical factors interact during load carriage for different individuals. Identifying individual differences in these interactions with different methods of load carriage could be important in discovering the individual determinants of load carriage economy, particularly as modifications in walking mechanics have been shown to influence the energy cost of locomotion (e.g. Cavanagh and Williams, 1982, Donelan et al., 2001).

Chapter 6. The development of a deterministic model to identify the biomechanical determinants of load carriage economy

6.1. Introduction

Chapter 5 highlighted the need for a framework to identify and analyse the key biomechanical factors associated with individual load carriage economy. Theoretical and statistical modelling techniques have been used to identify key biomechanical parameters in performance-related research (Chow and Knudson, 2011, Lees, 2002, Glazier et al., 2006). Systematic models have been used in qualitative analysis to identify the key characteristic of a skill (e.g. Hay and Reid, 1988, Knudson and Morrison, 2002). The most widely used of these is the deterministic model (Hay and Reid, 1988), also known as hierarchical model (Bartlett, 2014) or a factors-results model (Adrian and Cooper, 1995). This type of model was originally introduced to provide a theoretical basis for identifying the mechanical aspects of athletic performance (Hay and Reid, 1988). Hay and Reid (1988) outlined four basic steps:

i. The development of a model (block diagram) showing the relationships between the result and the factors that produce the outcome measure (Figure 25).
ii. Observation of performance.
iii. Evaluation of the relative importance of the factors that predict the outcome measure.
iv. Instruction of the performer in accord with the conclusions reached in the course of the analysis.

The principles that dictate the structure of deterministic models have since been described by many biomechanists (Lees, 2002, Glazier et al., 2006, Chow and Knudson, 2011). The key features that should be adhered to in the creation of a deterministic model are that the top level should be the primary performance outcome measure, it should only incorporate mechanical factors and each factor in the model should be determined by the factors that appear in the level directly below it (Figure 25).

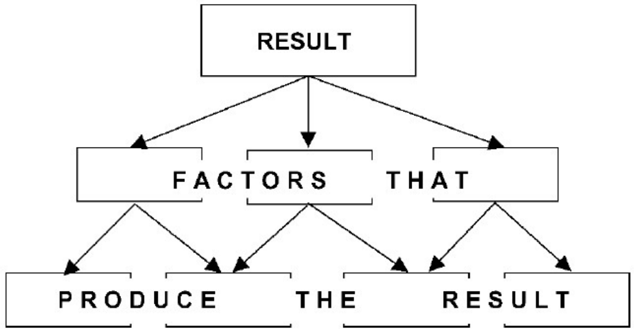


Figure 25. The deterministic model proposed by Hay and Reid (adapted from Hay and Reid, 1988)

Glazier and Robins (2012) suggested that deterministic models are models of performance and not models of technique. Therefore, they can be used to identify factors that are relevant to performance but not necessarily technique. In other words, deterministic models provide information on the performance parameters that are important but not how the performance parameters are generated. Indeed, it is possible to have alternative techniques that can lead to the same performance outcome. However, the conclusions of Glazier and Robins (2012) seem limited because if a deterministic model defines the outcome variables and how they relate to each other row by row, then it would be possible to assess variations in both technique and performance (Hay and Reid, 1988, Lees, 2002).

An advantage of deterministic models is that they can be used to provide a theoretical basis for statistical modelling (Bartlett, 2014, Chow and Knudson, 2011). Partial correlations and multiple regression analysis can be used to define factors that are meaningful in determining the outcome variable. However, a concern when using this method for statistical modelling is that a large sample of participants and trials is required to achieve an acceptable power value, particularly for well-developed models with many levels of factors (Chow and Knudson, 2011). For example, Hay et al. (1981) recruited 194 participants to identify the factors that determine vertical jumping. Recruiting a similarly large number of participants is unfeasible for the research in this thesis given the repeated measures study designs and walking durations required to assess economy.

Several authors have developed deterministic models to provide a theoretical basis for identifying the mechanical aspects of a movement, without examining the strength of relationships between factors in the model (e.g. Sanders and Kendal, 1992, Ham et al., 2007, Paradisis and Cooke, 2001, Hay and Reid, 1988). This approach uses the deterministic model as a framework to understand and quantitatively analyse factors relating to performance and technique. A rigorously developed deterministic model enables performance parameters to be selected and justified based on a theoretical rationale (Glazier et al., 2006, Chow and Knudson, 2011). Therefore, the use of a hierarchical modelling approach could be considered superior to randomly selecting performance parameters because the model helps to ensure all important variables are included while any trivial variables are excluded.

Deterministic models have been developed for a range of activities including sprinting (Hay and Reid, 1988, Hunter et al., 2004, Paradisis and Cooke, 2001), long jump (Hay, 1993, Chow and Hay, 2005, Hay, 1986), vertical jump (Hay et al., 1981), swimming (Guimaraes and Hay, 1985, McLean et al., 2000, Pai et al., 1984), and the discus throw (Leigh et al., 2008). However, no deterministic models have been published for walking or load carriage activities.

6.1.1. Aims

The aim of the work in the chapter was to develop a theoretical deterministic model that can be used as a framework to analyse gait alterations from unloaded walking as a consequence of carrying additional load in different load carriage methods. From this, the principle mechanics of loaded walking could be identified and tested in the subsequent chapter.

6.2. Existing theories for reducing the metabolic cost associated with human walking

Before developing a model for load carriage economy, it is useful to examine existing theories for reducing the metabolic cost of the human walking gait. There are two longstanding predominant theories for minimising the energy cost of walking. These theories are termed the six determinants of gait and the inverted pendulum (Figure 26). The six major determinants in normal gait are kinematic factors (pelvic rotation, pelvic tilt, knee flexion in the stance phase, foot mechanics, knee mechanics and lateral displacement of the pelvis) proposed to minimise the mechanical energy cost of locomotion by reducing the vertical displacement of the COM of the body (Inman and Eberhart, 1953). This is based on the premise that larger vertical displacements of the COM require a greater energetic cost to elevate the COM over the stance leg. In contrast, the inverted pendulum theory states that mechanical energy is reduced if the stance leg is kept relatively straight during the stance phase, acting like a pendulum (Cavagna et al., 1977). The inverted pendulum motion provides a mechanical energy exchange between potential and kinetic forms that is proposed to reduce the metabolic energy cost of walking.

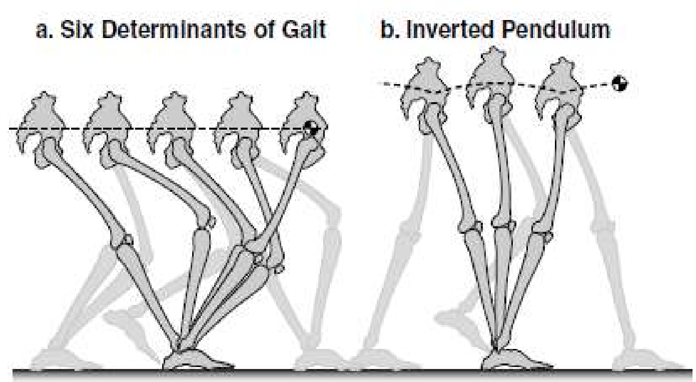


Figure 26. The two predominant theories of minimising the energy cost of human walking. (a) The six determinates of gait (Inman and Eberhart, 1953). (b) The inverted pendulum theory (Cavagna et al., 1977). Figure adopted from (Kuo, 2007).

The six determinants of gait have been included in a number of scientific textbooks (Rose and Gamble, 1994, Whittle, 2014, Perry and Burnfield, 1992). Yet, Gordon et al. (2003), Ortega and Farley (2005) and Wurdeman et al. (2017) all showed that modifying the human gait to reduce the vertical displacement of the COM increases the metabolic cost of walking. Furthermore, knee flexion during the stance phase (Gard and Childress, 1999), pelvic rotation about the vertical axis (Kerrigan et al., 2001) and pelvic tilt (Gard and Childress, 1997) might only provide a negligible contribution to reducing the vertical displacement of the COM.

The inverted pendulum theory might better explain the human gait. As humans walk, the body rises and falls in each stride, gaining and losing potential energy. They body also speeds up and slows down in each stride, gaining and losing kinetic energy (Alexander, 1996). This results in an inverted pendulum like motion of the body's COM. The theory predicts that the inverted pendulum motion of the COM will act conservatively, with an exchange between potential and kinetic energy forms that reduces the metabolic cost of walking (Cavagna et al., 1977). While this is an attractive idea, Cavagna et al. (1977) appears to be the only author that has provided data to support an energy exchange between kinetic and potential forms, with a conservation of energy of up to 65% for walking. Despite the lack of data, the inverted pendulum theory has been accepted by many authors (e.g. Donelan et al., 2002b, Kuo et al., 2005). This is perhaps due to the visible pendular motion of the body's COM during the walking gait.

As the human gait is not a frictionless freely swinging pendulum, consideration of how walking deviates from a pendulum like behaviour might be useful in understanding the economy of walking, as it could be the deviations that cost metabolic energy. One such deviation is the step-to-step transition, with an energy requirement to redirect the COM of the body from one pendular arc to the next in the transition between steps (Donelan et al., 2002a). The step-to-step transitions are unlikely to be the only energy requirement of walking. Both the inverted pendulum and six determinants of gait theories only consider work performed on the COM with massless legs, however, human legs have substantial mass and the forced motion of the legs relative to the torso will require a metabolic cost. Kuo (2001) suggested that the metabolic cost of walking could increase as a function of step frequency and that the cost of forced leg motion could act as a trade-off against the cost of step-to-step transitions, to minimise the overall cost at an intermediate combination of step length and step frequency.

In summary, it appears that attempting to flatten the trajectory of the COM results in a greater metabolic cost than a pendular trajectory, as it requires a greater amount of joint torque and work (Gordon et al., 2003, Ortega and Farley, 2005, Wurdeman et al., 2017). However, a pendular trajectory of the COM requires transitions between pendulum like steps, with the leading and trailing legs performing negative and positive work on the COM, respectively, to redirect its velocity between steps. Reducing the mechanical work required to redirect the COM between steps could reduce the energy cost of walking.

6.3. Theoretical development of a walking deterministic model.

6.3.1. Outcome measure

The first step in the development of the deterministic model was to identify the outcome measure at the top of the block diagram. For many skills/movements, the outcome is an objective measure of the performance. Since load carriage economy is a physiological factor, it is not solely determined by mechanical quantities and therefore could not be the outcome measure at the top of the model. Instead, a suitable outcome measure was identified using mathematical models for predicting load carriage energy expenditure (Givoni and Goldman, 1971, Pandolf et al., 1977). Givoni and Goldman (1971) created a predictive equation for the metabolic rate of carrying additional load (Equation 9) that accounts for body mass, external load, walking speed, walking gradient and terrain. Pandolf et al. (1977) later revised this equation (Equation 10) to enable predictions of metabolic rate during standing and slower walking speeds (< 2.5 km$\cdot$h^{-1}).

$$M = \eta (W + L) [2.3 + 0.32(V - 2.5)^{1.65} + G (0.2 + 0.07(V - 2.5))] \quad \text{Equation 9}$$

$$M = 1.5 W + 2.0 (W + L) (L / W)^2 + \eta (W + L) (1.5 V^2 + 0.35 VG) \quad \text{Equation 10}$$

where M is metabolic rate (watts), W is the mass of the participant (kg), L is the mass of the load carried (kg), V is the walking speed ($m \cdot s^{-1}$), G is the walking gradient (%) and η is the terrain factor ($\eta = 1.0$ for treadmill).

Equations 9 and 10 both identify body mass, external load mass, walking speed, walking gradient and the type of terrain as the key mechanical determinants of metabolic rate during load carriage (Figure 27) with both of these equations designed to predict the metabolic cost of back loading in military personnel. Furthermore, walking speed, walking gradient, body mass, and external load have also been included as key factors in more recent predictive equations for the metabolic cost of load carriage (Ludlow and Weyand, 2017, Santee et al., 2001). It is worth noting that the Pandolf equation has been reported to underestimate the energy expenditure of load carriage (Bach et al., 2017, Drain et al., 2017, Ludlow and Weyand, 2016). Drain et al. (2017) found that the Pandolf equation under-predicted the metabolic cost of load carriage (22.7 and 38.4 kg carried as a combination of a backpack, body armour, webbing and a replica assault rifle) by 12-17% at walking speeds of 4.5 $km \cdot h^{-1}$ and by 21-33% at slower and faster speeds of 2.5 and 6.1 $km \cdot h^{-1}$, respectively. Although the Pandolf equation appears to under-estimate metabolic cost in a laboratory environment, Vine et al. (2020) showed that the Pandolf equation more accurately predicts the metabolic cost of load carriage using military personnel in a field based environment with 40 and 50 kg at 4.8 $km \cdot h^{-1}$ compared to other predictive equations including the Givoni and Goldman (1971) equation and more recent equations by Santee et al. (2001) and Ludlow and Weyand (2017). However, Vine et al. (2020) also found that the Pandolf equation under- and over-predicted metabolic cost for other load-speed combinations, while other predictive equations consistently under-estimated metabolic cost for all of load-speed combinations tested (Givoni and Goldman, 1971, Ludlow and Weyand, 2017, Santee et al., 2001). It is not surprising that the metabolic cost of load carriage is difficult to accurately predict with a single equation given the differences in energy expenditure associated with different load placements (e.g. in the hand, on the feet, close to the body's COM) and the individual variation reported by Lloyd et al. (2010c) and the research in this thesis.

Load placement has also been shown to influence load carriage economy (e.g. Soule and Goldman, 1969, Datta and Ramanathan, 1971), and is another mechanical variable that should be considered as a factor influencing load carriage economy. Exercise economy is measured at a set velocity and, as such, the influence of different load carriage conditions on walking mechanics can be assessed through maintaining a constant walking speed. Therefore, walking speed was selected as the outcome measure at the top of the deterministic model. The model applies to walking at a constant speed by identifying the combination of underlying mechanical variables that produce the constant walking speed. To assess load carriage, the underlying mechanical variables that produce the constant walking speed for unloaded and loaded walking conditions can be compared.

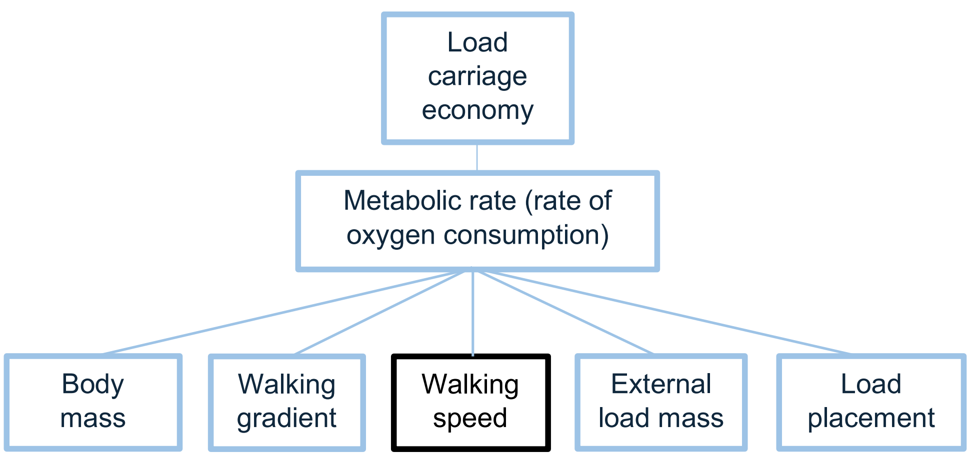


Figure 27. A deterministic model for load carriage economy using factors included in the predictive equation by Pandolf et al. (1977) and load placement to account for changes in metabolic rate with different load placements.

Result/Outcome factor definition:

Walking Speed: The distance covered by the whole body per unit of time. Measured in metres per second ($m{\cdot}s^{-1}$).

The second step in the process of developing the deterministic model was to identify the factors that produce the outcome measure (Hay and Reid, 1988). Where possible, each factor in the model was completely determined by the factors linked to it in the level below. All factors in the model were identified through the application of fundamental mechanics. Previously established models for running (Paradisis and Cooke, 2001, Hunter et al., 2004) (Figure 28), along with suggested mechanical principles of energy expenditure when walking (Cavagna et al., 1977, Alexander, 1991, Donelan et al., 2002a, Kuo, 2007, Inman and Eberhart, 1953) were considered to help identify the factors that should be included in the model. The description of factors in each level and how they were calculated is provided in the following sections of this chapter.

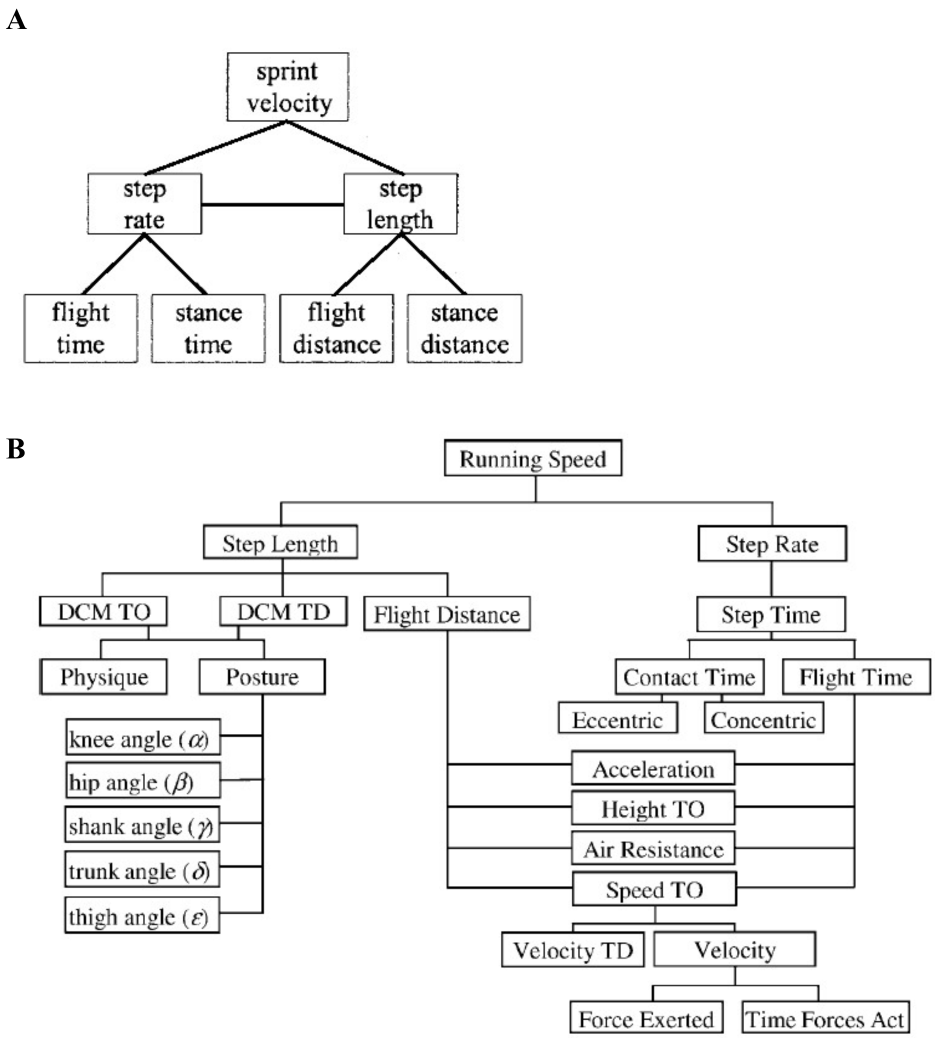


Figure 28. Deterministic models for running speed adapted from (A) Hunter et al. (2004) and (B) Paradisis and Cooke (2001).

6.3.2. Level 2

For a body moving at a constant velocity, speed can be calculated as the distance travelled divided by the time taken. As such, walking speed can be calculated from the horizontal displacement of the COM of the body divided by the time taken. Deterministic models for running speed have used stride/step length and stride/step frequency as the two determining factors of speed (Hay and Reid, 1988, Hunter et al., 2004, Paradisis and Cooke, 2001), with running speed calculated as the product of stride/step length multiplied by stride/step frequency (Figure 28). This principle also applies to other human gaits such as walking. A larger step length will result in an increased horizontal displacement of the COM per step (Kuo and Donelan, 2010)(Figure 29) and, for a given step length, an increased step cadence will result in less time per step. As such, it is unnecessary to include COM displacement and time in level 2 of the model (Figure 30).

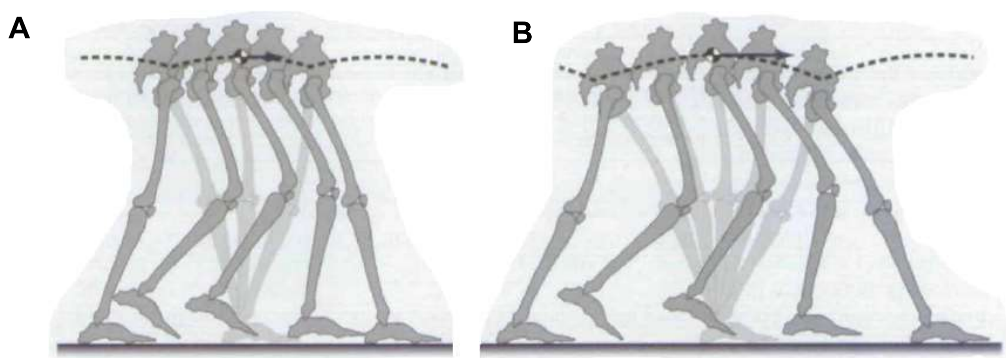


Figure 29. An illustration of centre of mass trajectories (dashed line) with (A) shorter and (B) longer step lengths. Adapted from Kuo and Donelan (2010).

Level 2 factor definitions, calculations and model (Figure 30):

Step length: The linear distance travelled from one heel-strike to the subsequent opposite foot heel-strike. Measure in metres.

Cadence: The number of steps taken every second, measured in steps per second.

Walking Speed = Step Length x Cadence

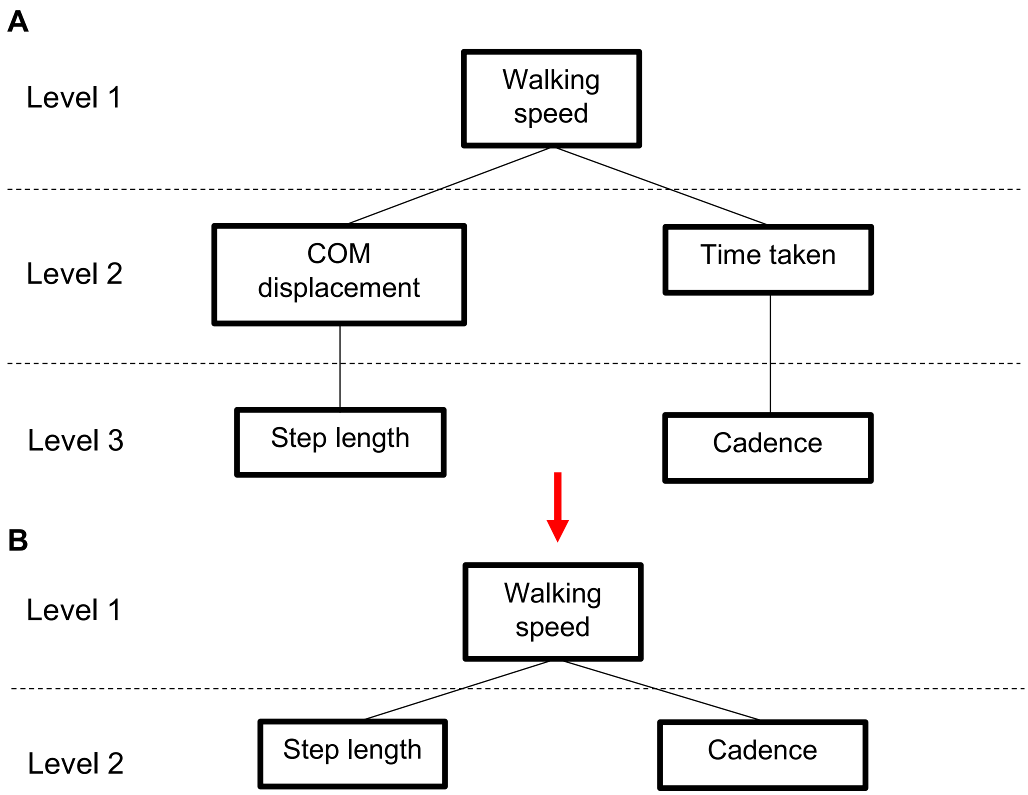


Figure 30. A deterministic model with factors that immediately determine walking speed. (A) Speed determined by displacement and time. Displacement and time are then determined by step length and cadence, respectively. (B) A condensed version of the model.

6.3.3. Level 3

The third level of the model is concerned with the determinants of step length and cadence. Perry and Burnfield (1992) defined step length as the distance between the initial contact by one foot and the subsequent initial contact be the contralateral foot. To identify the determinants of step length and cadence, it is useful to consider the different phases of each gait cycle (or stride), which comprises of two successive steps (Figure 31). Each gait cycle includes one stance and one swing period for each leg. Stance is the period in which the foot is in contact with the ground, starting at initial contact and ending at toe-off. Swing is the period that the foot is off the ground while the limb advances during the gait cycle, starting at toe-off and ending at initial contact. The stance phase is further divided into three phases (Figure 31). The first subdivision of stance is an initial period of double stance, were both legs are in contact with the ground (initial double-limb stance). Initial double-limb stance starts from initial contact with one foot and ends at the subsequent toe-off of the contralateral foot. The second subdivision of stance is single-limb stance, which occurs once the opposite foot is lifted for swing and one leg supports the body. The third, and final, subdivision is a second period of double stance (terminal double-limb stance). Terminal double-limb stance begins when the contralateral leg ends its swing phase by making initial contact with the ground and continues until toe-off of the original stance limb.

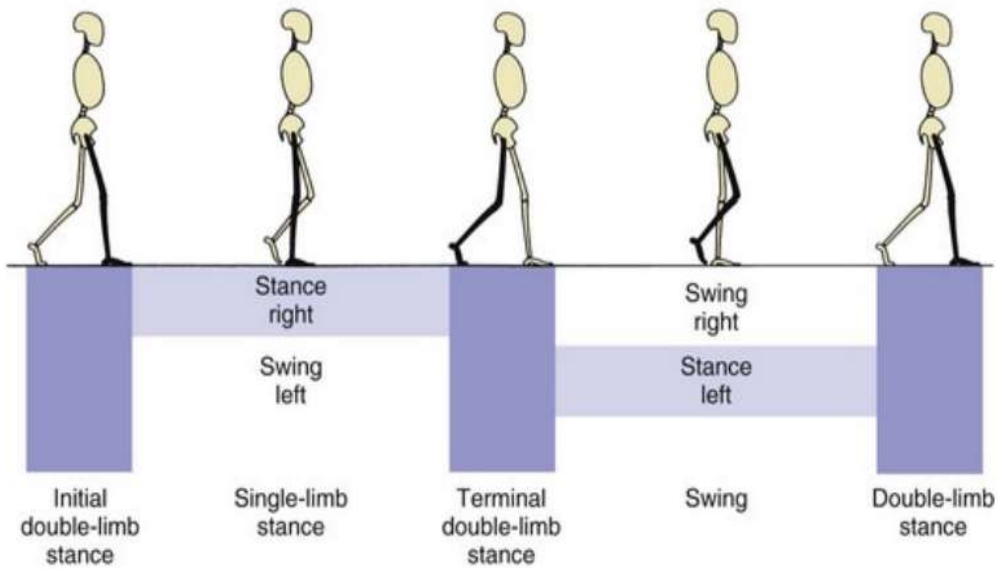


Figure 31. Subdivisions of the gait and their relationship to the pattern of bilateral foot contact. Adapted from (Perry and Burnfield, 1992).

The length of each step during human walking appears to be determined by the distance that the limb advances during the swing phase and movements of the contralateral stance foot while in contact with the ground (most likely from low foot-floor friction coefficients, Figure 32). Indeed, step length is reduced when anticipating slippery floors and lower frictional forces (Cham and Redfern, 2002). For a given walking speed, cadence will be determined by the time duration of each step.

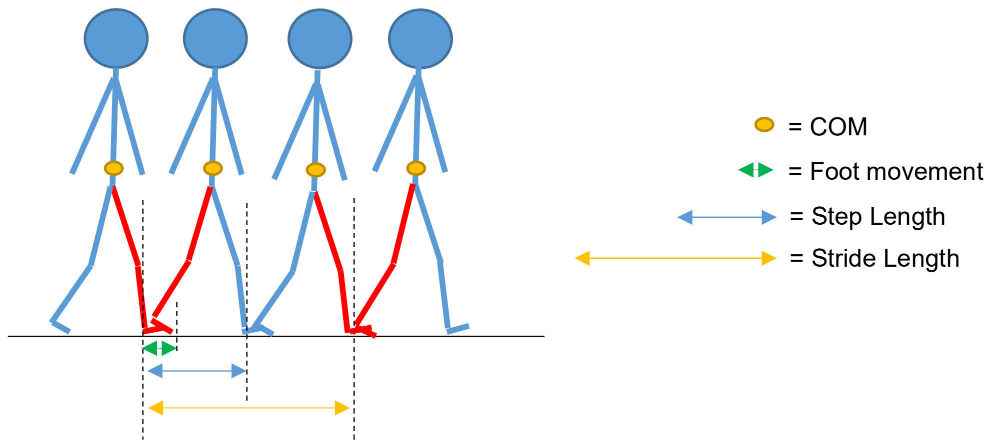


Figure 32. An illustration of walking gait step parameters.

Level 3 factor definitions, calculations and model (Figure 33):

Foot ahead distance: The distance of the leading foot in front of the trailing foot (one heel-strike to the successive foot heel-strike).

Foot movement: Determined by the horizontal distance the stance foot moves while in contact with the ground. This is often minimal but depends on the friction coefficients between the foot and the surface.

Step time: Step time is the time take from one heel-strike to the contralateral heel-strike, measured in seconds.

Step length = Foot ahead distance + Foot movement

Calculation: Cadence = Duration of time taken to complete *n* steps / *n* steps.

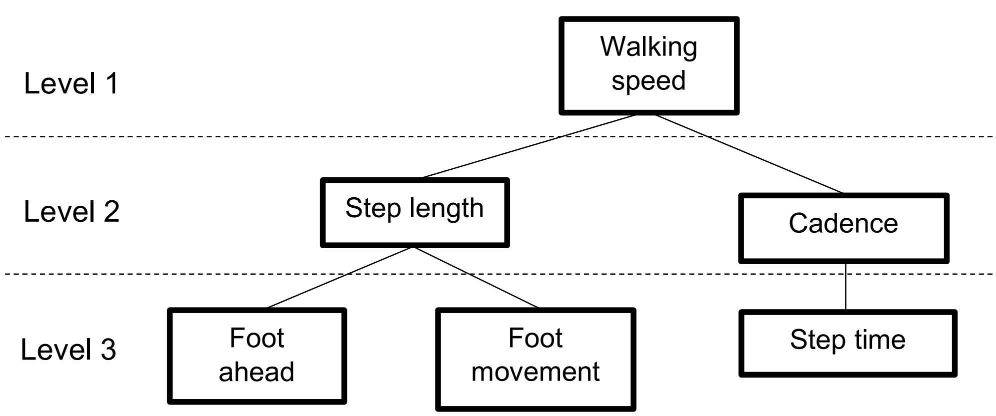


Figure 33. A three-level deterministic model for walking speed.

6.3.4. Level 4

The factors in level four should determine foot ahead distance of the limb in the swing phase, movement of the stance and step time. The determinants of step time were the simplest to identify. Each step involves a period of double stance followed by a period of single stance. In healthy walking gaits for men, Murray *et al.* (1964) found that the stance phase accounts for approximately 60% of the gait cycle, while the swing phase accounts for approximately 40%. The duration of the stance phase appears to have an inverse relationship with walking speed, with the duration decreasing as walking speed increases (Andriacchi et al., 1977). There is a concomitant single stance time to each swing phase that is of equal duration. As such, step time is determined by the sum of double stance time and single stance time.

Determining foot ahead distance is more complex than step time, due to the number of factors that contribute to the movement. In a model for running speed, Hay and Reid (1988) use the terms ‘Physique’ and ‘Body position’ to determine take-off and landing distance for stride length (p. 282, Hay and Reid, 1988). They referred to physique as the anthropometric details of the performer and body position as the position of the limbs. Lees (2002) suggested that Hay and Reid (1988) broke their own rules of deterministic models by introducing the terms physique and body position because they do not fully determine the factor in the level above mathematically. While it is difficult to determine step length mathematically, mechanical relationships can be used, and have been used by Hay and Reid (1988). Mechanically, the foot ahead distance is determined in part by the lengths, masses and the location of the centres of mass of each of the individual’s body segments. These factors will be represented in the model by the term ‘physique’. The foot ahead distance is also determined in part by how the segments of the performer’s body are positioned through the step, particularly the joints angles of the entire lower limb. The factors that describe the position of the segments and the relative angles of the segments are represented in the model by the term ‘change in posture during step’.

Most deterministic models focus on movement in a single plane of motion and models for running speed have focused on motion in the sagittal plane (Hunter et al., 2004, Paradisis and Cooke, 2001), where most movement occurs. The majority of motion also occurs in the sagittal plane for walking, but it could also be important to consider movement in all planes of motion in order to identify the key determinants of load carriage. Indeed, LaFiandra et al. (2002) identified differences in rotational movements of the pelvis and torso between loaded (backpack) and unloaded walking. Step width (Figure 34), a frontal plane motion, is another factor that appears to be influenced by load carriage, with a linear increase in step width variability as load mass increases (Huang and Kuo, 2014). Step width also appears to influence the energy cost of walking with an individual's preferred step width minimising the metabolic cost of unloaded walking (Donelan et al., 2001). In early versions of the deterministic model, step width was incorrectly included on level 5 as a determining factor of change in posture during each step. For a given walking speed and step length, an increase in step width would increase the distance between the feet (Figure 34) and could influence the distance that one foot is placed ahead of another. As such, step width was moved to level 4 as a determining factor of foot ahead distance.

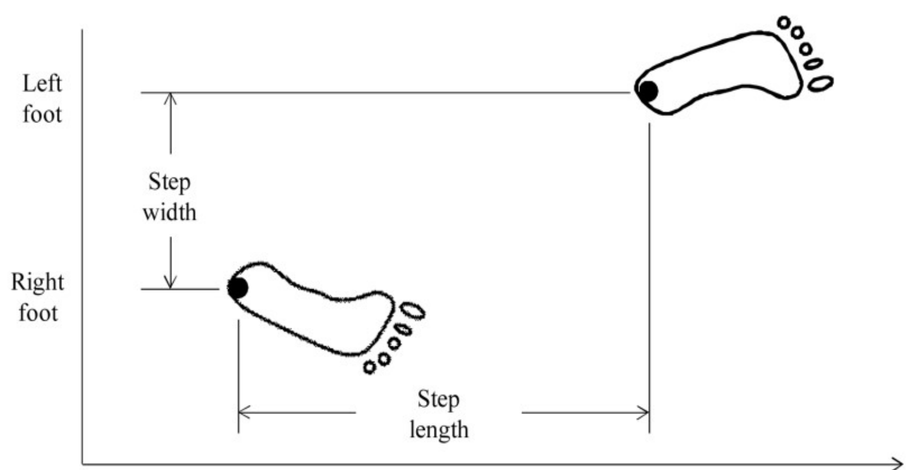


Figure 34. Measurements of step length and step width using initial contact of each foot.

Level 4 factor definitions, calculations and model (Figure 35):

Physique: The lengths, masses and the location of the centres of gravity of each of the individual's body segments.

Change in posture during step: The factors that describe the position of the segments and the relative angles of the segments.

Step width: The medio-lateral separation of the feet. The distance between the heels is used as the point on the feet for the basis of measurement.

Double stance time: The time spent with both feet in contact with the ground during a step, measured in seconds.

Single stance time: The time spent with a single foot in contact with the ground during a step, measured in seconds.

Foot ahead distance is dependent on the mechanical relationships between an individual's physiques, change in posture through each step and the width of the step.

Step Time = Double Stance Time + Single Stance Time

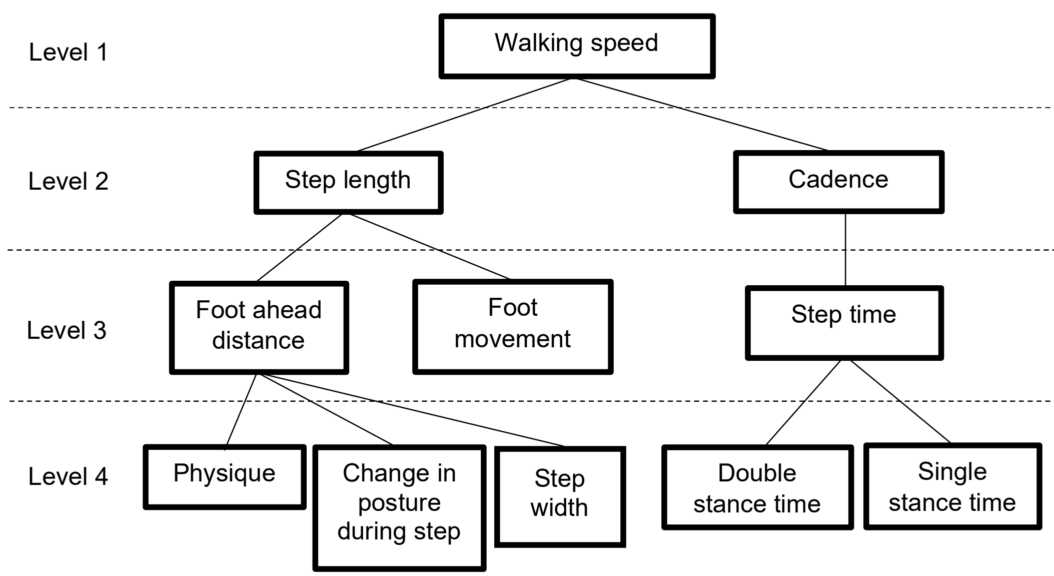


Figure 35. A four-level deterministic model for walking speed.

6.3.5. Level 5

The overall change in posture during each step is determined from the change in position of all body segments. As the aim of this model is to facilitate the analysis of unloaded and loaded walking, the factors included in the level directly below ‘change in posture during step’ will be kinematic factors that are, theoretically, the major determinants of walking posture with and without load. Walking, as with other forms of locomotion, requires angular motion at the joints for linear motion of the COM to occur (Herr and Popovic, 2008). Studies of the mechanics of human walking are useful in considering the key changes in posture that occur during each step. The six determinants, outlined by Inman and Eberhart (1953), to reduce the amplitude of oscillations of the COM along its vertically arced path are:

- Pelvic rotation: The pelvis rotates about a vertical axis, to the right and to the left, relative to the line of progression. Rotation of the pelvis allows the pelvis to contribute to step length.
- Pelvic tilt: Relative to the horizontal plane, the pelvis tilts downward on the opposite side to the weight-bearing limb. Pelvic tilt is largest at the mid-point of a step, when the COM is vertically above the stance foot.
- Knee and hip flexion during stance: The knee joints undergo flexion during stance when body weight passes over the stance leg.
- Knee and Foot mechanics: These two determinants are concerned with the smoothing of the COM pathway when the pendular arcs intersect from step-to-step. Inman and Eberhart (1953) suggested that the angular displacements of the ankle, foot and knee are intimately related (Figure 36).
- Lateral displacement of the pelvis: If the limbs were parallel, there would be excessive lateral displacement of the COM form step-to-step. Tibiofemoral angle and hip joint adduction prevent excessive lateral displacement.

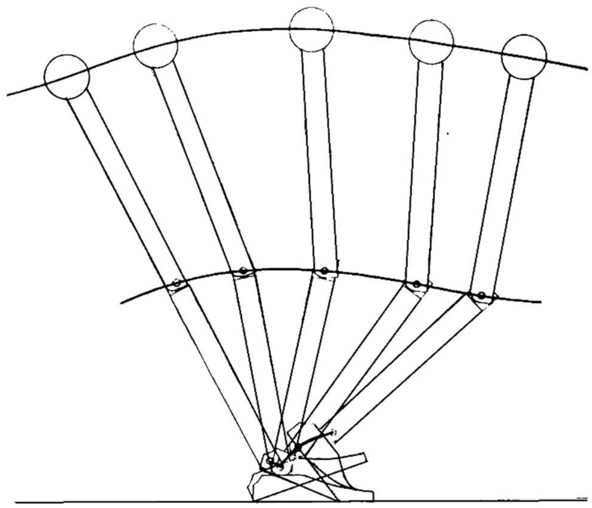


Figure 36. Related motion arcs for the hip, knee and ankle during the stance phase. This figure is adopted from Inman and Eberhart (1953).

While these factors are not determinants of change in posture during each step, they provide a good indication of the primary kinematic factors involved in walking. Inman and Eberhart (1953) started their investigation by using a compass gait model, which is the simplest model to analyse bipedal locomotion (Alexander, 1991). The compass gait model considers lower limb and COM motion in the sagittal plane, which is where the majority of motion in the walking gait occurs. Ortega and Farley (2005) demonstrated that simultaneously increasing hip, knee and ankle flexion in the stance limb reduces the vertical displacement of the COM, however, contrary to the hypothesis of Inman and Eberhart (1953), they found that the combined flexion of these joint angles doubled the metabolic cost of walking. As such, it is clear that sagittal plane movements of the main lower limb joints (hip, knee and ankle) do influence walking economy and it is important to include these factors in the deterministic model to assess changes in lower limb posture during each step with external loads.

Inman and Eberhart (1953) included pelvic rotation as a key determinant in reducing the vertical displacement of the COM when walking. However, Kerrigan et al. (2001) estimated that pelvic rotation only accounts for 12% (2.5

mm) of a reduction in COM vertical displacement. Increasing the amplitude of pelvic rotations clearly increases step length (Nottrodt, 1982, Huang et al., 2010) and, as such, pelvic rotation is included in level 5 of the model. Movement of the hip joint in the frontal plane (abduction/adduction) is also included at this level of the model to account for lateral displacement of the pelvis, which was identified by Inman and Eberhart (1953) as a factor that could lead to increased energy expenditure if the displacement of the body's COM is displaced from the line of progression due to increased muscular effort (Perry and Burnfield, 1992). Lin et al. (2014) quantified the medio-lateral displacement of the COM and indicated that hip adduction contributes significantly to the displacement of the COM in the frontal plane during the walking gait.

To describe the basic functions of gait, Perry and Burnfield (1992) divided the body into two sections; a passenger section and a locomotor section. In this description of the gait, the head, neck, trunk and arms are grouped in the passenger unit because they are carried rather than contributing to walking locomotion. The locomotor unit consists of the two lower limbs and the pelvis with 11 joints involved (lumbosacral, both hips, knees, ankles, subtalars and metatarsophalangeal joints). While motions of the lower limbs are the predominant factors in human locomotion, Chapter 5 clearly demonstrated that trunk motion can be influenced by additional load, particularly with loads placed directly on the trunk. As such, trunk motion has been included in this level of the deterministic model to facilitate the analysis of different load placements.

Ground reaction forces have been used extensively to analyse human locomotion. Increased walking speed coincides with an increase in the magnitude of all three components of ground reaction force (vertical, antero-posterior and medio-lateral) and shorter force periods (Nilsson and Thorstensson, 1989). As such, and as would be expected, ground reaction force appears to influence the periods of double-limb and single-limb stance and the duration of each gait phase appears to be determined by the time that ground reaction forces are applied.

Level 5 factor definitions, calculations and model (Figure 37):

Pelvic rotation: Measured as the internal and external rotation of the hip during each step.

Hip flexion/extension: Measured as the angle at the hip marker between the trunk and the thigh.

Hip adduction/abduction: Frontal plane hip movements. Measured as the medial and lateral alignment of the thigh.

Knee flexion/extension: Measured as the absolute angle between the shank and thigh.

Ankle plantarflexion/dorsiflexion: Measured as the absolute angle between the foot and the shank

Time forces act (both legs in contact): The total amount of time that forces are exerted on the ground during double support.

Time forces act (single leg in contact): The total amount of time that forces are exerted on the ground during single stance.

Change in posture during step is determined by the change in position of all body segments during the step. The relevant kinematic factors, based on previous literature, have been included in the model to analyse the walking gait and load carriage.

Single stance time = Time forces act with a single leg on the ground.

Double stance time = Time forces act with both legs on the ground.

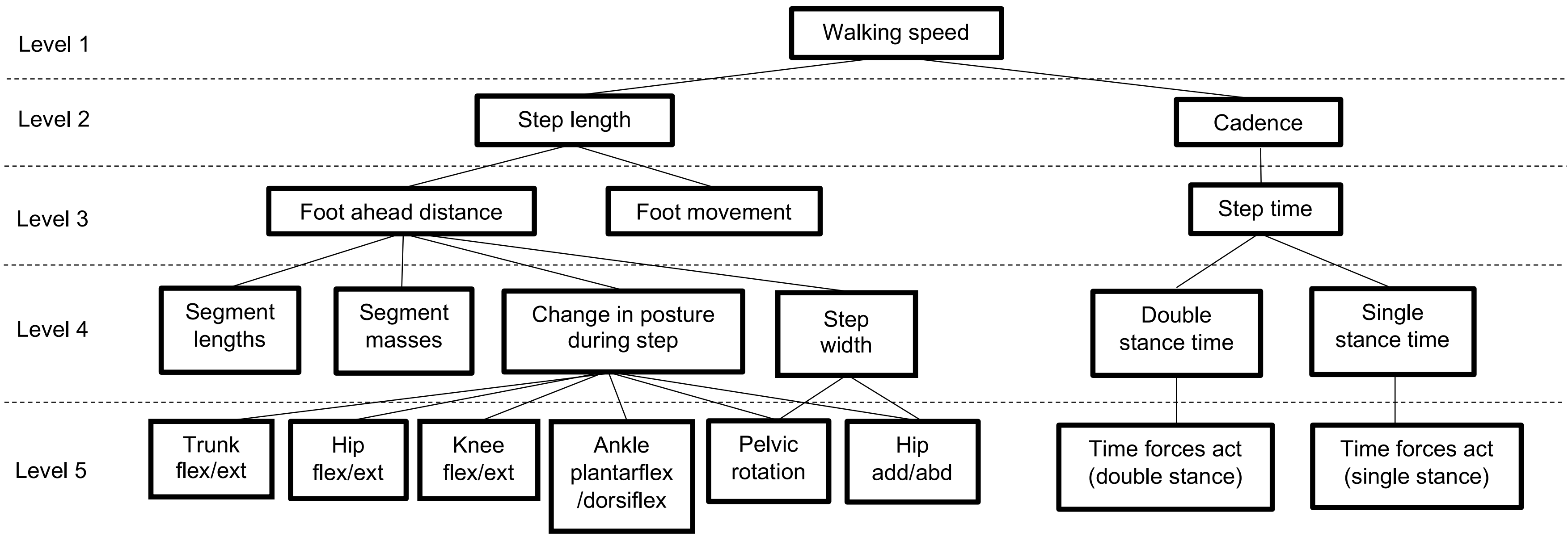


Figure 37. A five-level deterministic model for walking speed.

6.3.6. Level 6

Simple models of dynamic walking (such as Figure 26 in section 6.2) have predicted that work must be performed to redirect the COM velocity for each step-step transition (Alexander, 1991, Kuo, 2002). COM work rate during walking is often assessed as the inner product of ground reaction force of each leg and the COM velocity (Cavagna, 1975, Donelan et al., 2002a, Huang and Kuo, 2014), with COM velocity measured from the integration of ground reaction force (Cavagna, 1975). As such, both the impulse-momentum relationship and the work-energy relationship, when used to assess the walking gait, are built on ground reaction force and COM velocity.

The time that the resultant ground reaction force (GRF) acts for a given walking speed can be determined using the impulse-momentum relationship. According to this relationship, the change in momentum of the body is equal to the impulse that it produces (Equation 11)

$$I = mv_f - mv_i \qquad \text{Equation 11}$$

where I is the impulse, mv_f is the final momentum and mv_i is the initial momentum. This mechanical relationship can be used to determine the time that forces act during the double limb stance and single limb stance of the gait cycle. Impulse is the integral of the resultant force over a time interval (Equation 12)

$$I = \int F \cdot \Delta t \qquad \text{Equation 12}$$

where F is force and t is time. Thus,

$$\Delta t = \frac{mv_f - mv_i}{F} \qquad \text{Equation 13}$$

Walking, like all forms of locomotion, requires angular motion of each body segment for translation of the whole body to occur. Therefore, segmental angular momentums are required to provide linear momentum of the COM.

Whole body angular momentum during the walking gait appears to be small, not deviating substantially from zero, despite large segmental angular momentums (Herr and Popovic, 2008). This indicates that large segmental momentums cancel each other out. Herr and Popovic (2008) found that segmental angular momentums cancelled each other out ~95% in the medio-lateral, ~80% in the vertical and ~70 in the antero-posterior directions in normal unloaded walking. For a set walking speed, the momentum of the body at the end of each single stance period will influence the net external force impulse required during the subsequent double stance period. Using the same premise, the momentum at the end of each double support period will influence the net external force impulse required during the subsequent single stance period.

Level 6 factor definitions, calculations and model (Figure 38):

Net Force Exerted: The amount of time that forces are exerted on the ground. Determined by the net of braking and propulsive force.

Whole body momentum from single stance phase (going into double stance): The momentum of the whole body carried into double stance from the single stance phase.

Whole body momentum from double stance (going into single stance): The linear momentum of the whole body carried into single support from the double support phase.

Time forces act (double stance) = Whole body linear momentum from single stance / Net force exerted during double stance

Time forces at (single stance) = Linear Momentum from double stance / Net force exerted during double stance

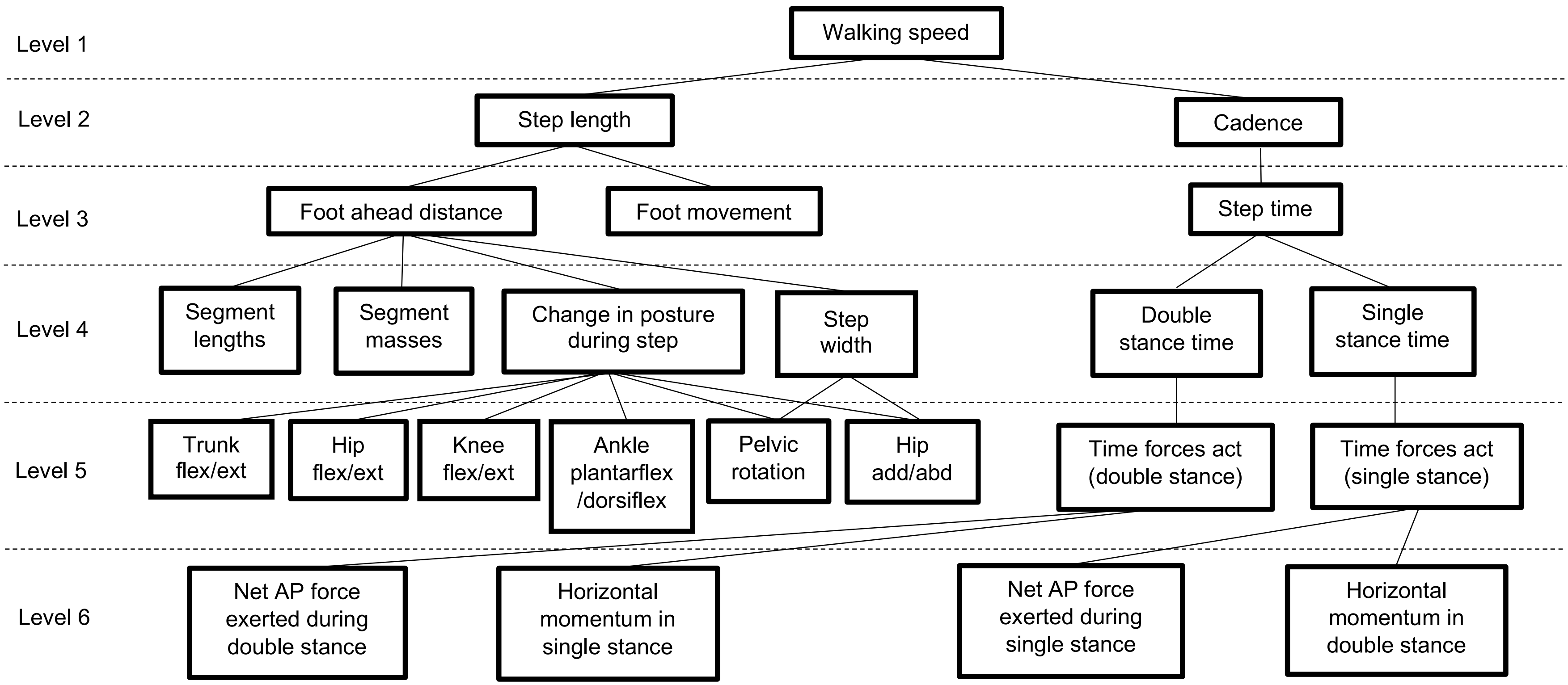


Figure 38. A six-level deterministic model for walking speed.

6.3.7. Level 7

Walking involves accelerations and decelerations, even during steady state walking, to maintain a constant speed (Peterson et al., 2011). Antero-posterior impulses appear to modulate walking speed (Nilsson and Thorstensson, 1989) and both braking and propulsive impulses increase with increases in walking speed (Nilsson and Thorstensson, 1989, Peterson et al., 2011). Peak antero-posterior GRFs and impulses increase with increasing step length for a set walking speed (Martin and Marsh, 1992). As such, presumably increased cadence during a set walking speed reduces the antero-posterior impulses due to the GRF's being applied over a shorter period.

Level 7 factor definitions, calculations and model (Figure 39):

Braking force (front leg): The force applied by the front leg at heel-strike during the braking phase.

Propulsion force (rear leg): The force applied by the rear leg to propel the body forward.

Whole body horizontal linear momentum: Mass multiplied by the change in velocity of the centre of mass.

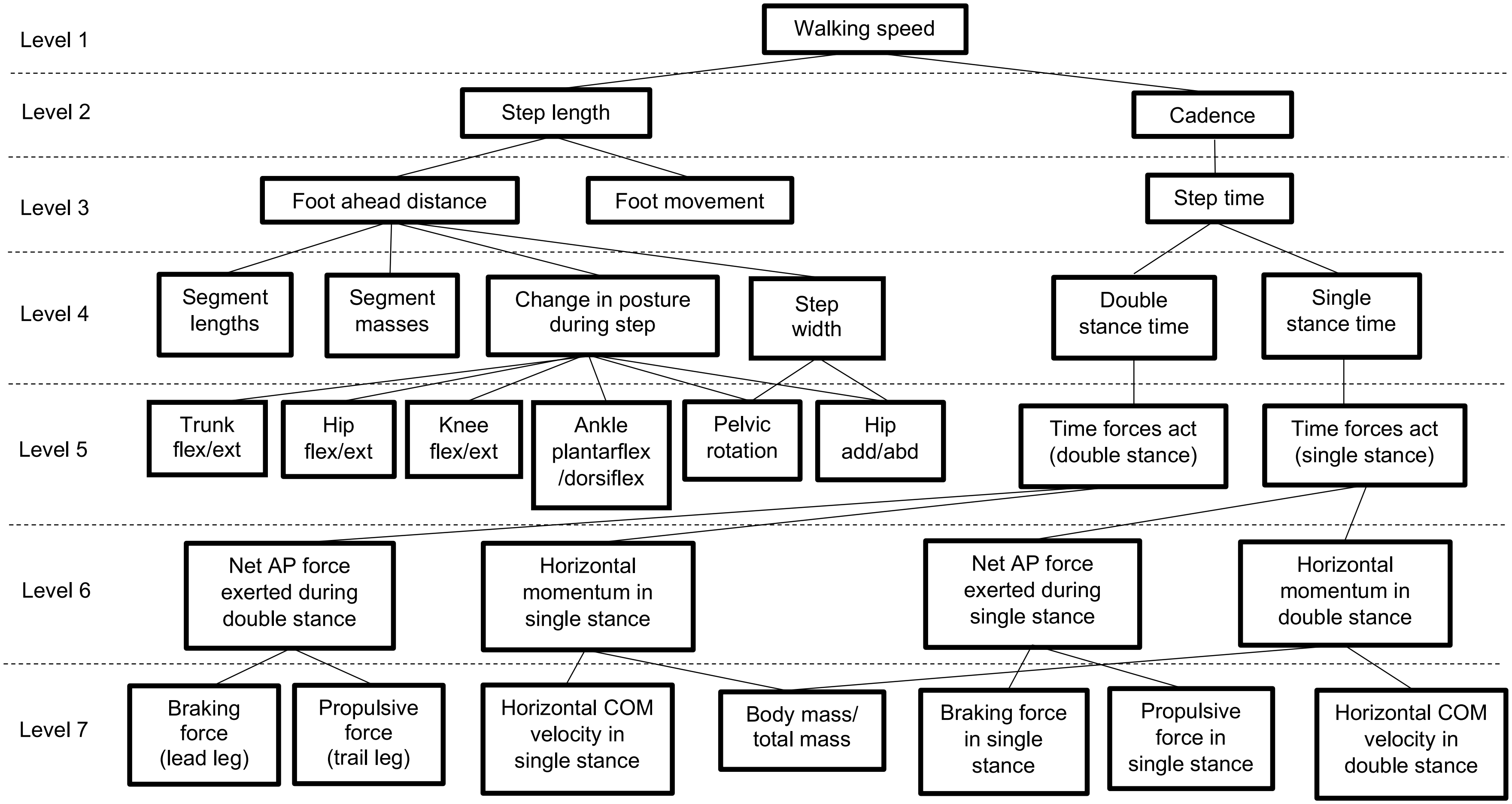


Figure 39. A seven-level deterministic model for walking speed.

6.4. Chapter summary

The work in this chapter describes the development of a novel theoretical deterministic model for walking speed. The model can be used to facilitate the analysis of the mechanical factors involved in loaded and unloaded walking at constant and varying locomotion speeds. Using the created model (Figure 39) as a framework to analyse the mechanical determinants of load carriage economy does not allow for statistical modelling (partial correlation or multiple regression analysis) because a constant value is required for walking speed in order to measure economy. However, the model does provide a more systematic approach for identifying factors that might affect walking economy and ensure that no factors are overlooked. It also enables the analysis of how mechanical factors interact during load carriage.

Factors in the model such as step width, pelvic rotation and net force during different gait cycle phases highlight the need for subsequent experimental chapters to include measures of ground reaction force and three-dimensional motion analysis to analyse the walking gait. The model shows that step length and cadence are important determining factors for walking at a specific speed. Assessing how these factors, and the factors that determine them, change from unloaded walking when an external load is carried could provide an insight into the causes of individual differences in load carriage economy. It's possible that a combination of different magnitudes of different mechanical changes could produce the same outcome measure of walking speed, but account for increases, decreases or no change in economy.

Chapter 7. The biomechanical responses to back-, back/front- and head-loading and their relationship with economy

Part of this work has been accepted for a peer-reviewed conference paper:
Hudson, S. Low, C. Cooke, C. Vanwanseele, B. and Lloyd, R., (2020), The effects of step width control on load carriage economy. *Proceedings of the 38th International Conference of Biomechanics in Sports.* Liverpool, UK: International Society of Biomechanics in Sports

7.1. Introduction

The overarching aim of this PhD research was to identify the key biomechanical factor(s) that determine an individual's load carriage economy with methods that place the load close to, or in alignment, with the centre of mass of the body. The research described in Chapter 5 provides evidence that load carriage economy for head-, back- and back/front-loading is not solely explained by sagittal plane trunk, hip, knee or ankle kinematics. As a result of the findings from Chapter 5, the deterministic model described in Chapter 6 was developed to provide a framework that can be used to identify walking gait adaptations to different load carriage conditions. The research in this chapter uses the deterministic model to identify walking gait adaptations to load carriage and investigate the relationship between these adaptations and relative load carriage economy.

Investigating correlations between load carriage economy and loaded walking gait perturbations can identify potential determinants of load carriage economy but does not show causation. As such, one of the objectives of this PhD research was to conduct cause and effect trials by manipulating variables that are identified as potential candidates for determining load carriage economy. This objective was based on identifying candidate variables that might determine load carriage economy from the research described in Chapter 5. However, none of the variables analysed in that study explained differences in economy between load carriage methods or individual variation in load carriage economy. Several factors described in the deterministic model were not assessed in Chapter 5, these factors include step width, pelvic rotation, braking and propulsive ground reaction forces (GRF) and whole-body horizontal momentum. Out of these factors, step width appears a most likely candidate variable, based on the walking gait literature, to influence individual differences in walking economy. Donelan et al. (2001) showed that young healthy individuals preferred an energetically optimal step width of 0.13 ± 0.03 *L,* where *L* is step width expressed as a fraction of leg length, compared to wider and shorter steps widths which require a greater metabolic cost. Donelan et al. (2001) reported a 45% increase in metabolic cost for a step width of 0.45*L* compared to the preferred step width condition. They also reported an 8% increase in metabolic cost when step width was decreased to 0.00 - 0.10*L.* Wide step widths appear to increase the energy cost of unloaded walking by increasing the mechanical work required to redirect the centre of mass from step-to-step (Donelan et al., 2002a, Donelan et al., 2001). Narrow step widths, when step widths are narrower than the width of the foot (width of the foot in the Donelan et al. (2001) study was 0.11*L*), appear to increase the mechanical work required laterally to move the swing leg to avoid the stance leg which increases the energy cost of unloaded walking (Shorter et al., 2017). Alterations in step width as a consequence of load carriage could therefore lead to alterations in load carriage economy, particularly if load carriage causes an individual to take much wider or narrower steps than their preferred step width when walking unloaded. Previous research on the effect of load carriage on step width has found no difference in step width expressed as a percentage of stature, with weighted vests between 10-30% body mass (Silder et al., 2013). Kinoshita (1985) also reported no differences in step width from unloaded walking with 20% and 40% body mass evenly distributed around the torso, but did find a significant increase in step width of 2.6 cm from unloaded walking with 40% body mass (~ 25 kg) carried in a backpack. To the author's knowledge no studies have assessed the effect of step width on economy for back-loading or head-loading, which could have an increased requirement for lateral stabilisation compared to methods that evenly distribute load around the torso.

The deterministic model includes anteroposterior GRF and whole-body linear momentum as factors that determine the duration of double and single stance time. To date, Lloyd and Cooke (2011) are the only authors that have reported relationships between kinetic variables and load carriage economy. They found a moderate negative relationship between ELI values and maximum braking force (r = -0.661) for back/front-loading (with 25.6 kg), which suggests that lower ELI values (better economy) are associated with smaller braking forces for that method. Furthermore, Lloyd and Cooke (2011) found a strong relationship between ELI and the difference between loaded and unloaded maximum braking force (r = 0.797) for back/front-loading, showing that smaller loaded-unloaded differences for maximum braking force are associated with improved economy. This supports the notion that improved load carriage economy might be due to smaller unloaded to loaded walking gait adaptations, particularly for back/front-loading with heavy loads.

A benefit of using the deterministic model to assess walking gait adaptation to load carriage is that the model provides an opportunity to understand how variables interact for a specific load method and mass combination, and how those interactions might differ between individuals. This could be an important aspect of understanding the determinants of individual load carriage economy, particularly given the large amount of individual variation in sagittal plane kinematics and step parameters that were highlighted in Chapter 5.

There were two main aims for the research in this chapter. The first aim was to use the deterministic model developed in Chapter 6 as a framework to compare the walking gait adaptations to head-, back- and back/front-loading, and assess relationships between the walking gait adaptations and load carriage economy, to try and identify the determinants of load carriage economy. The second aim was to assess the effect of step width control on load carriage economy. There were three hypotheses for this study. The first hypothesis was that there would be no difference in load carriage economy for head-, back-, and back/front-loading. The second hypothesis was that the most economical participants with each load carriage condition would exhibit the smallest walking gait perturbations from unloaded walking. The third hypothesis was that manipulating step width to the preferred unloaded width would improve load carriage.

7.2. Methods

7.2.1. Participants

Fifteen apparently healthy volunteers (10 males, 5 females) took part in this study (age 26 ± 3 years, mass 73.6 ± 10.1 kg, stature 1.78 ± 0.07 metres). Participants were recruited from the student population at KU Leuven, Belgium. An *a priori* power calculation performed using G*Power© software determined that a sample size of 15 was required for 80% power and to detect significance, based on an anticipated medium effect size (Richardson, 2011).

7.2.2. Experimental design

All trials were conducted in the Movement and Posture Analysis Laboratory Leuven, which is part of the Faculty of Movement and Rehabilitation Sciences at KU Leuven. Figure 40 provides an overview of the experimental design. Participants attended the laboratory on two separate occasions in order to complete a familiarisation and three main trial conditions. The first visit involved the familiarisation and one of the trial conditions. The remaining two trial conditions were completed in the second visit. Trial conditions differed by load carriage method, with load carried on the head (Head), back (Back) or evenly distributed between the back and front of the torso (Back/Front). The order in which the trial conditions were completed was randomised (via the picking of a marked piece of paper from a hat). Each trial condition involved eight, four-minute periods of walking at 3 $km \cdot h^{-1}$. The eight periods of walking were split into two blocks of four, separated by 10 minutes of rest. In the first block of four, participants walked unloaded, followed by walking with loads of 3, 12 and 20 kg. Each of these walking periods was separated by 2 minutes of rest. In the second block of four, the unloaded and loaded walking stages (3, 12, 20 kg) were repeated, but this time step width was controlled to match the participants preferred unloaded step width while carrying load. Visits to the laboratory were separated by 3-4 days. In the 24 hours prior to each test participants were asked to maintain a similar diet, refrain from alcohol consumption and refrain from moderate-vigorous exercise. Participants walked barefoot in all trials. This allowed markers to be placed directly on the skin, closer to the underlying bone, reducing the influence of marker movement artefacts.

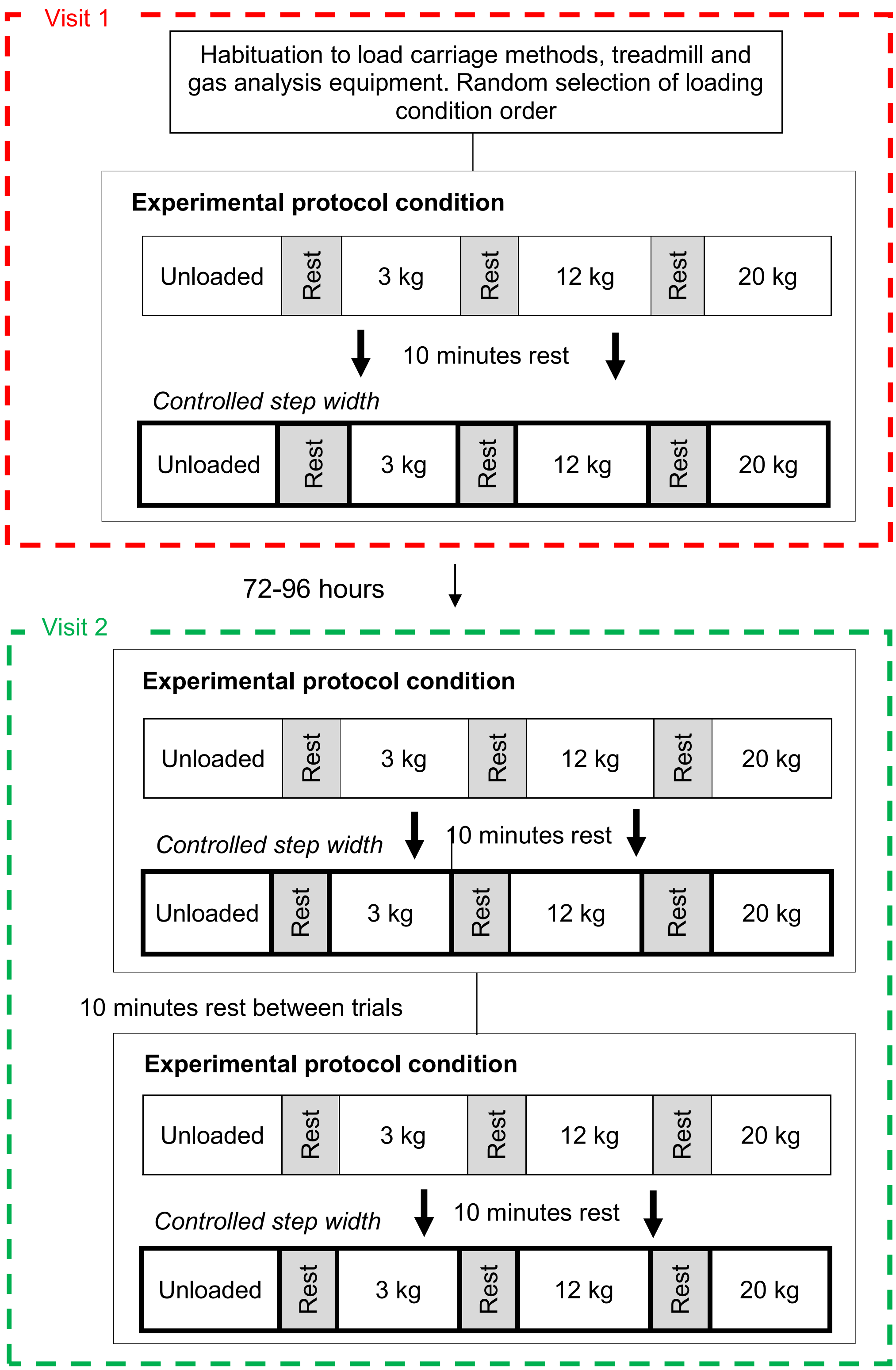


Figure 40. An overview of the experimental design of the study in Chapter 7. Each experimental protocol condition represents one of the three load carriage methods (Head, Back, Back/Front), completed in a randomised order.

7.2.3. Experimental procedures

7.2.3.1. Loading methods

Figure 41 shows the Head, Back and Back/Front methods. Each load carriage condition was described in detail in Chapter 3. The bucket used for the Head condition was attached to the ceiling using a safety harness to ensure that it would not hit the ground if dropped (Figure 41, image A). A portable computerised online gas analysis system (Oxycon Mobile, Jaeger) was worn on the anterior of the trunk and had a total mass (including the housing vest) of 1 kg. This device was worn during the unloaded and loaded walking trials. As such, the additional 1 kg was not included in the calculation of load carriage economy because it does not alter the calculated ELI value for each load carriage condition.

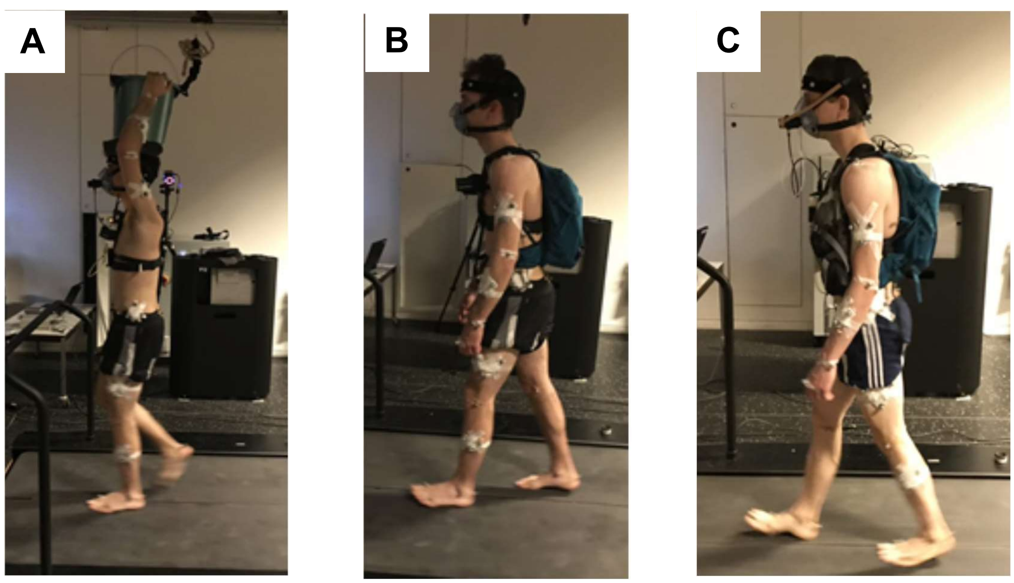


Figure 41. Sagittal plane images of the Head (A), Back (B) and Back/Front (C) load carriage conditions.

7.2.3.2. Main trials

Each trial began with a measurement of the participant's body mass. Participants were then fitted with retroreflective markers, a facemask for the gas analysis system and a heart rate monitor and asked to walk unloaded on the treadmill at 3 $km \cdot h^{-1}$ for four minutes at 0% gradient. After four minutes there was a two-minute rest period during which the participants were fitted with the appropriate loading device for the trial. The initial load was set at 3 kg. At the end of the rest period, participants recommenced walking with the load at a speed of 3 $km \cdot h^{-1}$ for a further four minutes. This pattern of work and rest continued with loads of 12 and 20 kg being carried in the subsequent stages. There was then a 10-minute rest period, after which, the loaded walking stages (3, 12, 20 kg) were repeated (4 minutes of walking followed by 2 minutes of rest), but this time step width was controlled. The rate of oxygen consumption ($\dot{V}O_2$) and heart rate were measured at the end of each rest period to ensure participants had returned to baseline.

7.2.3.3. Expired gas analysis

Expired gas measurements were made continuously throughout each period of exercise using a portable computerised online gas analysis system (Oxycon Mobile, Jaeger). The $\dot{V}O_2$ in the final minute of each unloaded and loaded walking period was used to calculate the ELI for each load carriage condition.

7.2.3.4. Subjective perceptions

Ratings of perceived execution were measured in the final 30 seconds of each walking period. During each rest period, participants were asked to rate pain/discomfort for 15 areas of the body (as described in Chapter 3) by marking visual analogue scales for each body area.

7.2.3.5. Biomechanical data collection

Kinematic and kinetic data were collected to assess the factors in the deterministic model developed in Chapter 6 (Figure 42, repeated here for ease of reference). It was assumed that there would be high frictional coefficients between the foot and treadmill belt so movement during the foot-floor contact would be minimal. As such, foot movement at level 2 of the model was not measured and step length was solely determined by foot ahead distance. Without considering foot movement, foot ahead distance and step length are the same value, so only step length is reported in this study. The time forces act in double stance and single stance are identical to the durations of double stance and single stance, respectively. As such, only double and single stance time are reported in this study.

Whole body motion and ground reaction forces were measured for six consecutive strides during the final minute of each walking period. A similar number of strides have been used to analyse the biomechanics of load carriage in previous research (Lloyd and Cooke, 2011, Harman et al., 2000, Silder et al., 2013, Birrell and Haslam, 2009, Wills et al., 2019, Chow et al., 2005). Whole body motion was measured using a motion capture system (Vicon, Oxford Metrics, UK). Thirteen infra-red cameras (sampling frequency of 100 Hz) were used to capture the trajectories of sixty-five spherical reflective markers (14 mm in diameter), attached to the participant in accordance with the modified full body Plug-in Gait model. A full description of the modifications to the Plug-in Gait model is provided in Appendix O. Markers were attached bilaterally to anatomical landmarks on the head, upper limbs, trunk, pelvis and lower limbs to define joint centres and track body segments. Ground reaction forces for the left and right legs were measured synchronously with the motion capture system using a floor mounted split-belt instrumented treadmill (Forcelink, Motekforce, Netherlands), with two force plates (AMTI, Watertown, US) sampling at 1000 Hz.

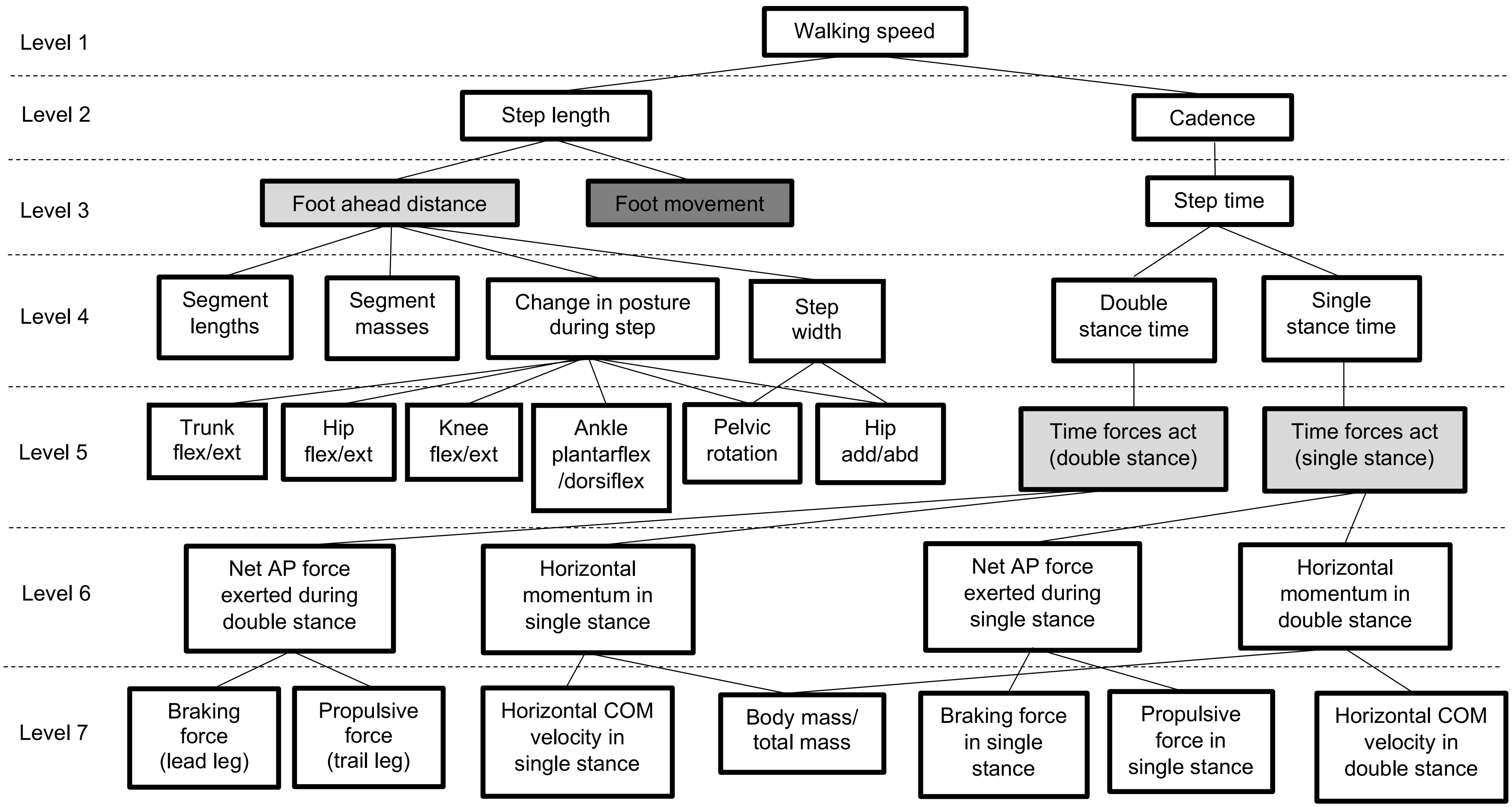


Figure 42. The deterministic model for walking speed developed in Chapter 6. Dark grey boxes indicate variables that were not measured in this study. Light grey boxes indicate variables not reported in this study because the linked factor in the level above is the same value. AP is anteroposterior.

The Vicon system was calibrated prior to each trial using a calibration wand (Figure 43), consisting of four markers mounted in known locations. The calibration wand was used to define the origin and orientation of the GCS and to calibrate the capture volume, which was set at x = 2 m, y = 3 m, z = 4 m over the treadmill. A static trial was recorded for each participant at the beginning of each experimental protocol condition. The static trial involved the capture of a single frame with the participant in a stationary pose.

Figure 43. Images of the calibration wand in place at the centre of the treadmill to define the orientation of the global coordinate system.

7.2.3.5.1. Coordinate systems

The GCS was a Cartesian right-handed orthogonal coordinate system with a fixed origin. The six DOF method was used to define the LCS of each segment, with a minimum of three non-collinear markers used to create each rigid segment. The LCS was also a Cartesian right-handed orthogonal system fixed to each segment, so that it moved with the segment. The orientation of the LCS with respect to the GCS defined the orientation of the segment in the GCS and changed as the segment moved through the capture volume (Zatsiorsky, 1998). This allowed for the calculation of segment displacements and velocities. Using the right-hand rule for the GCS and each LCS, the positive z-axis was vertically upwards, positive y-axis were directed from posterior to anterior and positive x-axis was pointed to the right in a medial to lateral direction.

7.2.3.5.2. Marker set

A modified version of the Vicon full body Plug-in Gait marker set (Vicon, Oxford Metrics, UK) was used to measure 3D whole body kinematics. The modifications included the use of non-collinear marker clusters to improve segment tracking of the upper arm, lower arm, pelvis, thigh and shank, and additional markers on the medial knee and ankle joint axes to improve joint centre location. Further detail of the modifications to the full-body Plug-in Gait are outlined in Appendix O. The marker set can be seen in Figure 44.

Figure 44. Anterior and posterior views of the full body marker set with the participant performing the static calibration pose.

The considerations for defining each body segment and joint centre location for whole-body 3D motion analysis of load carriage, using a modified Plug-in Gait marker set, are described below.

Head segment:

Four markers attached to a headband were used to define the head segment during the calibration and motion trials. The headband was positioned so that two markers were placed on the anterior head over the left (left front head) and right temples (right front head), and the other two markers were placed on the posterior head, on the same transverse plane as the two anterior head markers (left and right back head).

Upper arm segments:

The marker positions outlined for the upper arms were identical for both the left and right arms. To define the upper arm, a marker was placed on the shoulder at the acromio-clavicular joint to define to proximal end and on the lateral epicondyle of the humerus to define the distal end. A three non-collinear marker cluster was also midway between the proximal and distal markers to track the motion of the segments.

Lower arm and hand segments:

The marker positions outlined for the lower arms and hands were identical for both the left and right sides of the body. The marker on the lateral epicondyle of the humerus was used to define the proximal end of the lower arm. To define the distal end of the lower arm, markers were placed on the radius-styloid process and the ulna-styloid process. A three non-collinear marker cluster was placed midway between the proximal and distal points of the lower arm. To define the distal end of the hand segment, a marker was placed on the head of the third metacarpal.

Thorax segment and glenohumeral joint centre location:

To define the thorax segment during the static trials, markers were placed on the left and right acromio-clavicular joints, the 7th cervical vertebrae (spinous process of the 7th cervical vertebrae), the 10th thoracic vertebrae (spinous process of the 10th vertebrae), the clavicle (suprasternal notch where the clavicles meet the sternum), the sternum (xiphoid process of the sternum) and on the right upper back right back (scapula). The marker on the 10th vertebrae, sternum and right upper back were removed for the motion trials to allow for the back- and back/front-load placements.

Due to the need to remove markers on the thorax segment for some loading conditions, the thorax was modelled based on the Rab upper extremity model (Rab et al., 2002, Petuskey et al., 2007), which represents a minimal marker set for the upper extremities that is useful for gait analysis but it is insufficient more complex upper body movements, such as throwing activities. This method of modelling the thorax segment requires markers on the clavicle, 7th cervical vertebrae, left and right acromio-clavicular joint and markers on the pelvic segment (detailed below), which enabled the torso to be modelled with the addition of the portable gas analysis system, and back and back/front loading conditions.

In accordance with the Rab upper extremity model, the glenohumeral joint centre (GHJ) was located as an axial plane offset of -17% of the markers on the left and right acromio-clavicular joint (Rab et al., 2002). Rab et al. (2002) determined the magnitude of this offset from direct measurements of two participants and anatomical data available in the literature, based on seven cadavers (Van der Helm et al., 1992). Campbell et al. (2009) suggested that regression equations based on magnetic resonance imaging from healthy participants might provide a more accurate estimation for the GHJ location than those based on data from cadavers. However, this method is based on a cluster of markers placed on the acromion, which would have interfered with the shoulder straps in the back and back-loading methods.

Pelvis segment:

Markers were placed on the pelvis in accordance with the Plug-in Gait model. This places markers on the, right and left posterior superior iliac spine (PSIS) and sacrum (placed mid-way between the left and right posterior superior iliac spine). Visual3D's (c-motion, USA) CODA pelvis was used to model the pelvis segment. In this model, the origin of the pelvis's local coordinate system is defined as the mid-point between the left and right ASIS markers. Creating a CODA pelvis segment in Visual3D automatically creates landmarks for the hip joint centres (HJC) based on predictive equations from Bell et al. (1989), Bell et al. (1990). These predictive equations define the location of the right and left hip joint centres (RHJC and LHJC, respectively) using the following coordinates:

RHJC = (ML = 0.36*ASIS_Distance; AP = -0.19*ASIS_Distance; Axial = 0.30*ASIS_Distance)
LHJC = (ML = -0.36*ASIS_Distance; AP = -0.19*ASIS_Distance; Axial = -0.3*ASIS_Distance)

where ML is medio lateral, AP is anteroposterior and ASIS_Distance is the 3D distance between the right and left ASIS. Instead of using the automatic HJC locations, the Harrington et al. (2007) predictive equations for HJC were used in this study as they are based on magnetic resonance imaging and have been shown to be more accurate than the predictive equations of Bell et al. (1989) and Davis et al. (1991) when compared to computer tomography (Anderson et al., 2013). The Harrington et al. (2007) predictive equations define the location of the right and left hip joint centres (RHJC and LHJC, respectively) using the following coordinates:

RHJC = (ML = 0.33*ASIS_Distance+0.0073; AP = -0.24*RVP_Depth-0.0099; Axial 0.3*ASIS_Distance– 0.0109)
LHJC = (ML = -0.33*ASIS_Distance+0.0073; AP = -0.24*RVP_Depth-0.0099; Axial-0.3*ASIS_Distance)

where RVP_Depth is the 3D distance between the mid-point of the ASIS and the mid-point of the PSIS. In addition, functional approaches to estimating the hip joint centre such as the geometric fit method (Sangeux et al., 2014) only performs marginally better (3 to 6mm) than the Harrington et al. (2007) equations (Kainz et al., 2015). As such, the Harrington et al. (2007) prediction equations were used to estimate the position of hip joint centre instead of a functional approach, which requires the collection of additional functional calibration trails, and medical imaging techniques such as X-ray, magnetic resonance imaging and computer tomography due to the high financial costs of using these techniques. Furthermore, although pelvis is known to have considerable morphological differences between sexes, Hara et al. (2016)

found that hip joint location is not one of them. Therefore, the same predictive equation was used to calculate the hip joint centre for both males and females.

Thigh segments:

The marker positions outlined for the thighs were identical for both the left and right legs. The estimated RHJC and LHJC locations defined the proximal ends of the right and left thigh segments, respectively. The distal joint centre of the thigh segment was defined as the midpoint between markers placed on the medial and lateral epicondyles of the femur. A three non-collinear marker cluster was placed on the centre region of the thigh for segment tracking.

Shank segments:

The proximal joint centres of the right and left shank segments were defined as the mid-point between the medial and lateral epicondyle markers on the right and left femurs, respectively. The joint centres at the distal end of the left and right shank segments were defined as the mid-point between markers placed on the lateral and medial malleolus of the tibia and fibula, respectively. A three non-collinear marker cluster was placed on the centre region of both shanks to track the motion of the segments.

Foot segments:

Markers were place on the first and the fifth metatarsal heads, and on the calcaneous at the same height above the plantar surface of the foot as the metatarsal markers. The distal joint centre of each foot segments was defined as the mid-point between the metatarsal head markers.

7.2.3.1. Data processing

The static pose was used to create a 15-segment model in Visual 3D (Visual 3D, C-Motion, Inc. Germantown, USA) (Figure 45) by using the marker positions to define body segment coordinate systems, tracking marker locations, joint centres and segment lengths for each participant. The default mass and centre of mass location for each segment in Visual3D is based on regression equations by Dempster (1955) using data from eight cadavers. As outlined in Chapter 3, the default settings in Visual3D were altered to the adjusted values of De Leva (1996) to estimate body segment inertial parameters in this study.

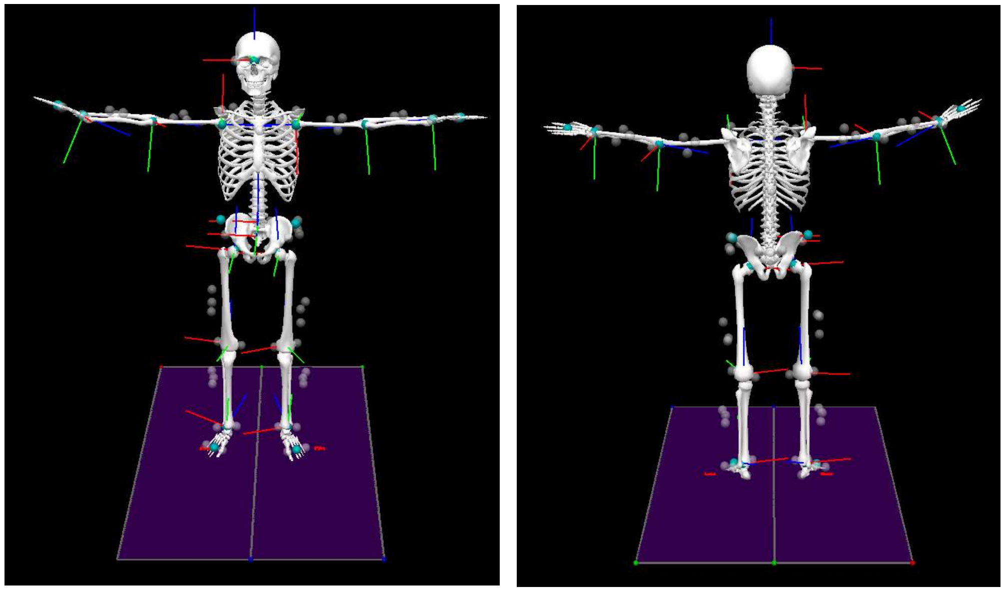


Figure 45. Anterior and posterior views of the three-dimensional, 15-segment model created in Visual3D using a static trial.

Gait events of heel-strike and toe-off where automatically identified in Visual 3D using the vertical GRF data, with detection thresholds set to 20 N, and used to determine spatiotemporal variables. Marker trajectories were low pass filtered at 6 Hz using a 2nd order Butterworth filter. The same filter was also used on the kinetic data. Using different cut-off frequencies for force and position data can cause artefacts, particularly for high impact movements (Bisseling and Hof, 2006, Kristianslund et al., 2012). Kristianslund et al. (2012) suggested that force and movement data should be processed with the same filter and at the same cut-off frequency in order to reduce error in terms of differences between how the two signals are processed.

Joint angles of the trunk, hip, knee and ankle were measured using an x-y-z Cardan rotation sequence in Visual3D software (Visual 3D, C-Motion, Inc. Germantown, USA), in line with the International Society of Biomechanics (ISB) recommendations' for reporting joint motion (Wu and Cavanagh, 1995). The x-y-z Cardan rotation sequence for the hip, knee and ankle was flexion/extension-abduction/adduction-axial rotation. Baker (2001) showed that the sequence of rotations for the Cardan angle sequence of the pelvis relative to the GCS is more accurate as axial rotation-obliquity-tilt. As such, the sequence of rotations would be z-y-x, and this was the Cardan sequence of rotations used for the pelvis segment angles in this study.

Biomechanical variables associated with factors in the deterministic model were also measured. These variables included vertical and mediolateral GRF's, vertical impulse, stance time and trunk axial rotation. Ground reaction force data were presented as absolute values and normalised to total mass ($N \cdot kgTM^{-1}$), which was the combined mass of the participant and the external load carriage device. This allowed for comparisons to be drawn between participants by mitigating any influence of body mass (Birrell et al., 2007, Birrell and Haslam, 2010).

Whole body horizontal linear momentum in the anteroposterior direction was calculated as the product of the mean horizontal velocity of the body's COM and body mass (or total mass for load carriage conditions) in Visual 3D. COM velocity was determined from the ground reaction forces using the method outlined by Cavagna (1975). First, the acceleration of the body's COM in the anteroposterior direction was calculated from the anteroposterior ground reaction force component using the equation:

$$\sum\overline{\boldsymbol{F}} = m\overline{\boldsymbol{a}} \qquad \text{Equation 14}$$

where $\sum\overline{F}$ is the average net force, m is body mass and $\overline{a}$ is average acceleration. The COM velocity was then calculated from the time integral of the COM accelerations. Average acceleration is the change in velocity over time, and as such, the equation for force can be stated as:

$$\sum\overline{\boldsymbol{F}} = m\left(\frac{\Delta \boldsymbol{v}}{\Delta \boldsymbol{t}}\right) \qquad \text{Equation 15}$$

where Δv is change in velocity and Δt is change in time. Multiplying both sides of the equation by Δt provides the impulse-momentum relationship:

$$\sum\overline{\boldsymbol{F}}\ \Delta t = m\ \Delta\mathbf{v} \qquad \text{Equation 16}$$

7.2.3.2. Step width measurement and control procedures

Step width was measured as the average medio-lateral distance between heel-marker positions at heel-strike for 12 steps during the final minute of walking. Step width control was achieved using constant visual feedback, similar to the methods of Arellano and Kram (2011). First, the participants' preferred unloaded walking step width was measured and marked out at the rear of the treadmill using two pieces of tape, with each piece of tape an equal distance from the centre of the treadmill belt (Figure 46, image B). A digital camera (JVC Everio, Japan) was positioned 1.5 metres behind the treadmill, to record the heel markers and the tape at the end of the treadmill belt. The camera was linked to a monitor placed 3.5 metres in front of the participants while walking on the treadmill (Figure 46, image A). The monitor was positioned on an adjustable shelf which was altered in order to place the monitor at the height of each participant's eye-line. Participants were asked to align the heel markers to the taped lines at the back of the treadmill and given additional verbal feedback on their foot placements.

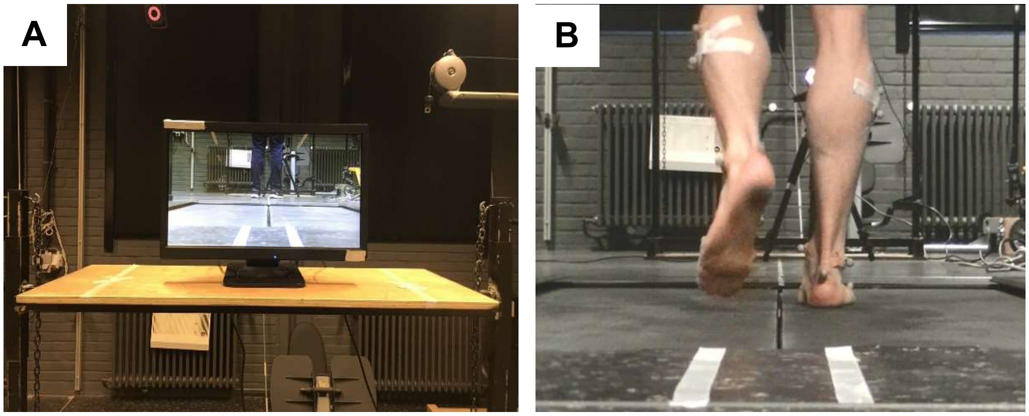


Figure 46. Images of the experimental set-up to control step width. Image A shows the monitor which provided images of the participants foot placements. Image B shows a participant walking with their heel markers aligned with the tape positioned at the rear of the treadmill, which was used to signify the participants required foot placements.

7.2.4. Inter-individual analysis of biomechanical variables

Between-participant standard deviation (SDb) and within-participant standard deviation (SDw) about the mean were calculated for spatiotemporal variables, joint angle kinematics and ground reaction forces. To assess the level of within participant variability, SDw for each variable represents the standard deviation about the mean of six consecutive strides. Where significant relationships were identified between ELI and loaded walking gait adaptations, participants were ranked in order of economy (from lowest ELI to highest ELI) for all adaptations on the same level of the deterministic model to assess for interactions.

7.2.5. Statistical analysis

Descriptive statistics (mean ± SD) were calculated for all outcome measures. A one-way ANOVA with repeated measures was used to test for significant main effects of method for all unloaded walking variables. A two-way ANOVA with repeated measures was used to test for significant main effects and interactions in physiological and biomechanical variables between load carriage methods and load mass (method x mass). A three-way ANOVA with repeated measures was used to assess VAS data (body position x method x mass). Post-hoc tests for significant main effects were conducted using a Bonferroni correction. Statistical significance was set at $p < 0.05$ in all experimental chapters. Where $p < 0.10$, the results are reported as being close to statistical significance. Pearson's product moment correlation coefficients were used to assess relationships between ELI values and the physical characteristics of participants (body mass, stature and BMI). Relationships were also calculated between ELI and the mechanical variables at each level of the deterministic model for walking speed presented in Chapter 6.

To assess inter-individual variation, linear multi-level models (MLM), using maximum likelihood estimation, were created for $\dot{V}O_2$, and ELI with each method of load carriage. The MLM's were used to estimate the variance between participants (σ^2_u) and the variance between the load masses (σ^2_e) for each load carriage method. Intra-class Correlation Coefficients (ICC) were calculated from the variance components in each MLM to represent the proportion of total variability in the outcome that was attributable to individual differences between participants. CV's and SD were also used to assess inter-individual variation for $\dot{V}O_2$, and ELI. Between participant standard deviation (SDb) and within participant standard deviation (SDw) were calculated for spatiotemporal, joint angle and ground reaction force data to assess inter- and intra- individual variation

7.3. Results

7.3.1. Physical characteristics

There was no significant difference between trial conditions for body mass (p = 0.361) or BMI (p = 0.365). As such, there was no significant difference between load carriage methods for the absolute load as a percentage of body mass for 3 kg (p = 0.168), 12 kg (p = 0.394) or 20 kg (p = 0.279). The 3 kg, 12 kg and 20 kg conditions represented 4 ± 1%, 17 ± 2% and 28 ± 4% of the participants' body mass, respectively. The range of body mass was 59.3 – 96.4 kg with the 3 kg, 12 kg and 20 kg loads represented a range of 3 – 5%, 12 – 20% and 21 – 34% of body mass, respectively. On average, the male participants were heavier and taller (Male: mass 76.7 ± 9.5 kg, stature 1.81 ± 0.05 metres) than female participants (Female: mass 67.3 ± 9.1 kg, stature 1.72 ± 0.05 metres).

7.3.2. Rate of oxygen consumption ($\dot{V}O_2$)

There was no significant difference between trials for the three load carriage methods when walking unloaded for absolute $\dot{V}O_2$ (761.8 ± 133.7, 721.3 ± 103.7 and 728.4 ± 99.2 $ml \cdot min^{-1}$ for Head, Back and Back/Front, respectively, p = 0.238) and $\dot{V}O_2$ normalised for body mass (10.3 ± 1.2, 9.9 ± 1.4 and 10.0 ± 1.2 $ml \cdot kg^{-1} \cdot min^{-1}$ for Head, Back and Back/Front, respectively, p = 0.390).

$\dot{V}O_2$ significantly increased as the mass of the load increased (main effect for load mass, $p < 0.001$, $\eta^2 = 0.857$) and there was a significantly larger increase in $\dot{V}O_2$ for the Head method compared to the two trunk loading methods (main effect for load method, $p < 0.001$, $\eta^2 = 0.440$) (Figure 47). The method x mass interaction was significant ($p = 0.001$, $\eta^2 = 0.350$). The largest difference in $\dot{V}O_2$ between loading methods occurred with the heaviest load (20 kg), with an increase in $\dot{V}O_2$ from unloaded walking of 4.14 ± 2.10 $ml \cdot kg^{-1} \cdot min^{-1}$, 2.42 ± 1.14 $ml \cdot kg^{-1} \cdot min^{-1}$ and 1.91 ± 0.93 $ml \cdot kg^{-1} \cdot min^{-1}$ for Head, Back and Back/Front, respectively. There was no significant difference for $\dot{V}O_2$ between the Back and Back/Front methods (p = 1.000).

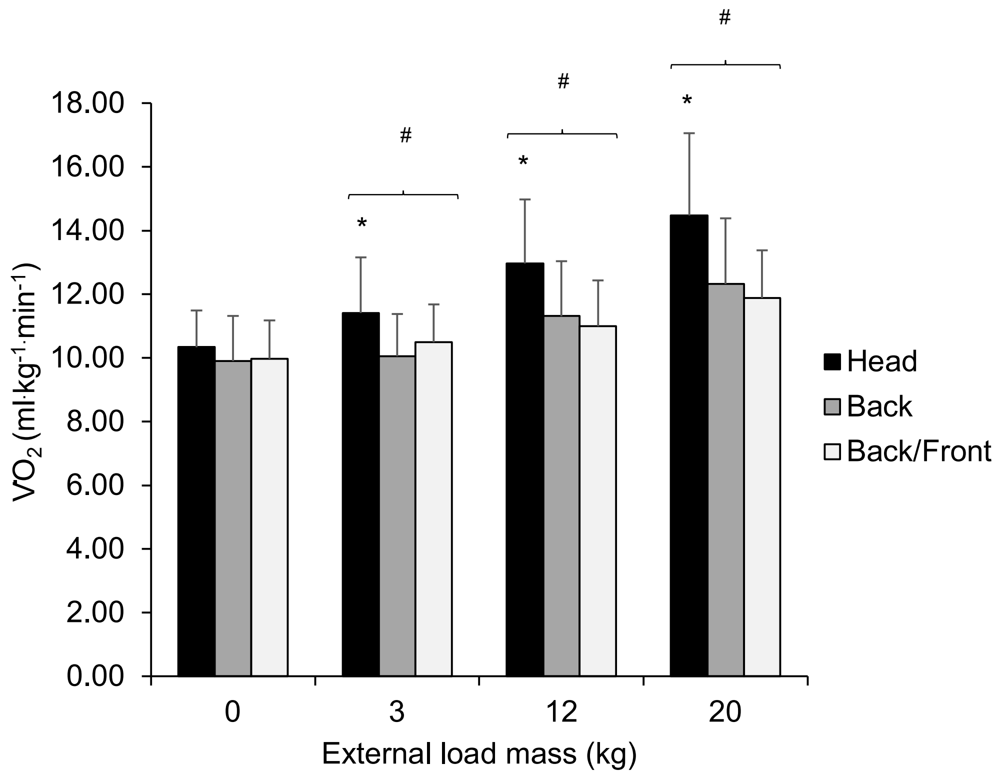


Figure 47. Mean ± SD V̇O2 for each loading method and load mass with preferred step width. * denotes a significant difference compared to the other load carriage methods. # denotes a significant difference compared to the previous load mass.

7.3.3. Relative load carriage economy

There was a significant main effect of load method for ELI values (p = 0.002, η^2 = 0.423) with significantly larger values for Head compared to Back (p = 0.014) and Back/Front (p = 0.010). The largest difference between Head and the two trunk loading methods occurred with the 20 kg mass (ELI = 1.10 ± 0.15, 0.98 ± 0.09 and 0.94 ± 0.08 for Head, Back and Back/Front, respectively). The difference in ELI values between the Back and Back/Front methods was not significant (p = 1.000).

No significant difference was observed for ELI values between load mass (main effect for load mass, p = 0.410, η^2 = 0.054), however there was a significant method x mass interaction effect (p = 0.030, η^2 = 0.211). Figure 48 illustrates that ELI increased as the load mass increased for the Head method

(1.06 ± 0.09, 1.07 ± 0.11 and 1.10 ± 0.15 for 3 kg, 12 kg and 20 kg, respectively) and decreased as the load mass increased for the Back/Front method (1.01 ± 0.06, 0.95 ± 0.07 and 0.94 ± 0.08 for 3 kg, 12 kg and 20 kg, respectively). The ELI for Back remained constant (0.98 ± 0.06, 0.98 ± 0.07 and 0.98 ± 0.09 for 3 kg, 12 kg and 20 kg, respectively). With sex included in the two-way repeated measures ANOVA as a between subjects' factor, there was no significant interaction effect between loading method and sex (p = 0.872) or load mass and sex (p = 0.134). With load mass pooled, males had ELI values of 1.08 ± 0.03, 0.99 ± 0.00 and 0.97 ± 0.04 for head-, back- and back/front-loading, respectively. Females has ELI values of 1.08 ± 0.04, 0.95 ± 0.01 and 0.96 ± 0.01 for head-, back- and back/front-loading, respectively.

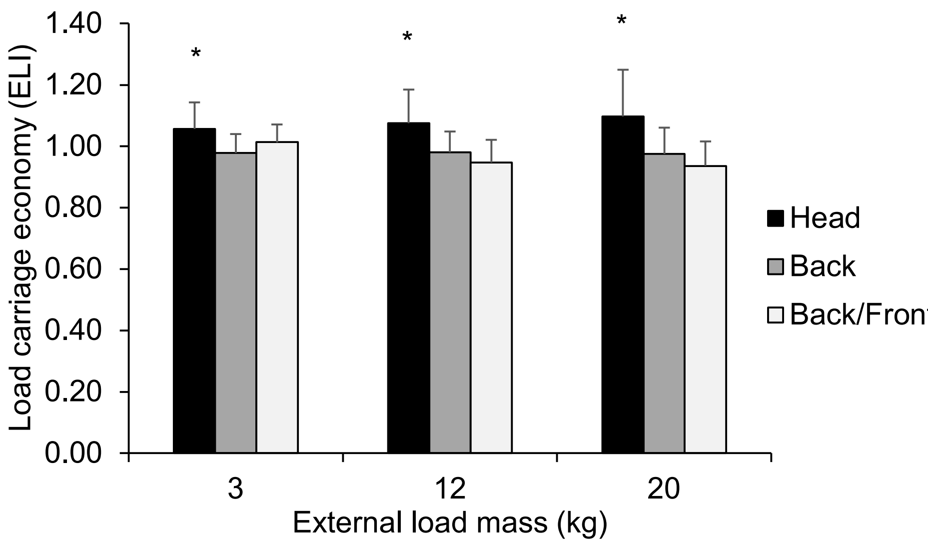


Figure 48. Mean ± SD ELI values for each loading method and load mass with preferred step width. * denotes a significant difference compared to the other load carriage methods.

7.3.4. Spatiotemporal gait parameters

Table 24 shows the spatiotemporal measures for each load carriage condition. No significant differences were observed for any of the measured spatiotemporal gait parameters between unloaded walking trials (p > 0.05).

A significant main effect for load method was observed for the Δ step length (p = 0.045, η^2 = 0.198), Δ cadence (p = 0.001, η^2 = 0.391), Δ step time (p = 0.013, η^2 = 0.268) and Δ single stance time (p = 0.010, η^2 = 0.283) from unloaded walking. Specifically, cadence was significantly slower for Back compared to Head (p = 0.010) and Back/Front (p = 0.032), with Δ cadence decreasing from unloaded walking for Back (-0.02 ± 0.05 $steps \cdot s^{-1}$) and increasing for Head (0.04 ± 0.06 $steps \cdot s^{-1}$) and Back/Front (0.01 ± 0.03 $steps \cdot s^{-1}$). However, post hoc adjustment showed no significant difference in Δ step length from unloaded walking between any of the methods. There were significant method x mass interaction effects for Δ step length (p = 0.008, η^2 = 0.216) and Δ cadence (p = 0.001, η^2 = 0.292). For Head, there was a decrease in step length and concomitant increase in cadence as the mass of the load increased, whilst for Back, step length increased (with a concomitant decrease in cadence) as the load mass increased. The Δ step length and cadence from unloaded walking across load mass was minimal for the Back/Front method (Table 24).

There was a tendency for decreased step times from unloaded walking for Head compared to Back (-0.01 ± 0.02 s vs. 0.01 ± 0.02 s p = 0.058). The Δ single stance time from unloaded was significantly decreased for Head compared to Back (-0.02 ± 0.02 s vs. 0.00 ± 0.02 s, p = 0.026) and there was also a tendency for reduced single stance time for Head compared to Back/Front (Back/Front = -0.01 ± 0.02 s, p = 0.070). Significant main effects for load mass were observed for single stance time (p < 0.001, η^2 = 0.458) and double stance time (p < 0.001, η^2 = 0.808). Post hoc analysis showed that single stance time significantly decreased from unloaded walking by -0.01 s with 12 kg compared to 3 kg (p = 0.049) and by -0.02 s with 20 kg compared to 3 kg (p = 0.002). Double stance time significantly increased from unloaded walking with each increase in load mass (p < 0.05).

Table 24. Mean ± SD magnitudes for spatiotemporal gait parameters unloaded walking and each load carriage condition. Significance values are for the change from unloaded walking for each variable.

Spatiotemporal variable	0kg			3kg			12kg			20kg			*p* - value	
	H	B	B/F	H	B	B/F	H	B	B/F	H	B	B/F	Method	Mass
Step length (metres)	0.54 ± 0.03	0.54 ± 0.03	0.53 ± 0.03	0.53 ± 0.03	0.53 ± 0.03	0.53 ± 0.03	0.53 ± 0.03	0.54 ± 0.04	0.53 ± 0.03	0.52 ± 0.04	0.55 ± 0.04	0.54 ± 0.03	**0.045**	0.374
Cadence ($steps \cdot s^{-1}$)	1.56 ± 0.08	1.56 ± 0.09	1.56 ± 0.08	1.58 ± 0.09	1.56 ± 0.09	1.57 ± 0.08	1.59 ± 0.10	1.54 ± 0.09	1.57 ± 0.09	1.61 ± 0.12	1.53 ± 0.10	1.56 ± 0.07	**0.001**	0.729
Step time (s)	0.64 ± 0.03	0.64 ± 0.03	0.64 ± 0.03	0.64 ± 0.04	0.64 ± 0.04	0.64 ± 0.03	0.63 ± 0.04	0.65 ± 0.04	0.64 ± 0.04	0.62 ± 0.05	0.65 ± 0.04	0.64 ± 0.04	**0.013**	0.591
Single stance time (s)	0.46 ± 0.03	0.45 ± 0.03	0.45 ± 0.03	0.45 ± 0.03	0.46 ± 0.03	0.45 ± 0.03	0.44 ± 0.03	0.46 ± 0.03	0.44 ± 0.03	0.43 ± 0.03	0.45 ± 0.03	0.44 ± 0.02	**0.010**	**< 0.001**
Double stance time (s)	0.18 ± 0.02	0.19 ± 0.02	0.19 ± 0.02	0.19 ± 0.01	0.19 ± 0.02	0.19 ± 0.02	0.20 ± 0.02	0.20 ± 0.02	0.20 ± 0.02	0.20 ± 0.02	0.20 ± 0.01	0.21 ± 0.02	0.743	**< 0.001**

H = Head, B = Back, B/F = Back/Front

7.3.5. Joint angle kinematics

Mean ± SD peak sagittal plane joint angles and joint angles at heel strike and toe off are presented in Table 25 and Table 26, respectively. No significant difference was observed between methods for any of the measured joint angle kinematics for unloaded walking trials ($p > 0.05$).

There was a significant main effect of load method for all of the Δ trunk ($p \leq 0.003$, $\eta^2 \geq 0.347$) and the Δ hip ($p \leq 0.002$, $\eta^2 \geq 0.432$) angle variables from unloaded walking, except for the Δ trunk angle excursion between heel strike to toe off from unloaded walking, although this was close to statistical significance ($p = 0.094$, $\eta^2 = 0.180$). The Head method was associated with a more upright posture compared to the Back and Back/Front methods, with significantly less peak trunk flexion from unloaded walking for Head (-7.42 ± 3.39°) compared to Back (3.76 ± 3.19°, $p < 0.001$) and Back/Front (1.93 ± 1.47°, $p < 0.001$). Furthermore, the Back method was associated with a larger Δ peak trunk flexion angle ($p = 0.003$) and Δ peak hip flexion angle (4.19 ± 3.53° vs. 3.12 ± 2.07°, $p = 0.004$) from unloaded walking compared to the Back/Front method. The largest difference between methods for joint angle kinematics occurred with 20 kg. Figure 49 illustrates the sagittal plane joint angles over the gait cycle for each load carriage method with 20 kg.

Considering the Δ knee and the Δ ankle angles from unloaded walking between load methods, the was a significant main effect of method for the Δ knee angle at toe off ($p < 0.001$, $\eta^2 = 0.465$), the Δ knee angle excursion from heel strike to toe off ($p < 0.001$, $\eta^2 = 0.597$), the Δ peak ankle plantarflexion ($p = 0.021$, $\eta^2 = 0.283$) and the Δ ankle angle at heel strike ($p = 0.009$, $\eta^2 = 0.286$) from unloaded walking. Post hoc analysis revealed significantly greater knee flexion at toe-off for Head compared to Back ($p = 0.004$) and Back/Front compared to Back ($p = 0.010$) (Table 26). The Δ knee angle excursion from heel strike to toe off from unloaded walking was significantly smaller for Back/Front (0.3 ± 2.5°) compared to Head (-1.3 ± 2.6°, $p = 0.004$) and Back/Front compared to Back (2.2 ± 3.0°, $p = 0.012$). The difference in Δ knee angle excursion from unloaded walking was also significant between Head and Back ($p < 0.001$). Post hoc adjustment showed that for the Δ peak ankle plantarflexion from unloaded walking, there was a significant increase in plantarflexion from unloaded for Back (-1.3 ± 2.0°) compared to Back/Front (0.1 ± 1.6°, p = 0.003) and a tendency for increased plantarflexion from unloaded walking for Back compared to Head (0.5 ± 2.7°, p = 0.061). For the Δ ankle angle at heel strike from unloaded walking, there was significantly greater dorsiflexion from unloaded walking for the Head method (0.86 ± 1.01°) compared to Back/Front (0.04 ± 0.74°, p = 0.016).

There was a significant main effect of load mass for the change in all sagittal plane joint angles from unloaded walking at heel strike and toe off ($p \leq 0.046$; $\eta^2 \geq 0.198$) (Table 26). There was also a significant main effect of load mass for the change in all peak joint angle variables from unloaded walking ($p \leq 0.039$; $\eta^2 \geq 0.250$), except for peak hip extension, knee ROM and peak ankle dorsiflexion (Table 25). Flexion angles of the trunk and hip increased as the mass of the load increased for the Back and Back/Front methods and decreased as the load mass increased for the Head method.

There were also significant method x mass interactions for the Δ peak trunk flexion angle from unloaded walking (p = 0.001, η^2 = 894) and the Δ peak trunk extension angle from unloaded walking (p = 0.001, η^2 = 861). Trunk flexion increased as the mass of the load increased for Back-loading while trunk extension increased as the mass of the load increased for Head-loading.

Table 25. Mean ± SD magnitudes for kinematic joint angles for unloaded walking and each load carriage condition. Main effects of load carriage method and load mass are reported for the change from unloaded walking for each variable. Negative values represent extension, positive values represent flexion.

Joint kinematics	0kg			3kg			12kg			20kg			*p* - value	
	H	B	B/F	H	B	B/F	H	B	B/F	H	B	B/F	Method	Mass
Trunk														
Peak Flexion (°)	3.03 ± 2.82	2.75 ± 2.40	2.98 ± 2.57	-3.23 ± 2.80	3.70 ± 3.06	3.71 ± 2.78	-4.58 ± 3.51	6.43 ± 3.10	5.20 ± 2.61	-5.35 ± 3.55	9.39 ± 3.87	5.82 ± 3.50	**< 0.001**	**< 0.001**
Peak Extension (°)	-1.12 ± 2.64	-1.17 ± 2.44	-1.24 ± 2.55	-7.23 ± 2.62	-0.75 ± 3.11	-0.30 ± 2.67	-8.73 ± 3.27	1.65 ± 2.96	0.86 ± 2.53	-9.59 ± 3.43	4.30 ± 3.62	1.22 ± 3.21	**< 0.001**	**0.002**
ROM (°)	4.15 ± 1.17	3.91 ± 1.04	4.22 ± 0.84	4.00 ± 0.76	4.46 ± 1.48	4.01 ± 0.96	4.15 ± 0.90	4.78 ± 0.90	4.34 ± 1.15	4.24 ± 0.92	5.09 ± 0.86	4.61 ± 1.10	**0.003**	**0.001**
Hip														
Peak Flexion (°)	22.72 ± 3.37	23.10 ± 2.91	22.86 ± 3.07	16.42 ± 2.92	24.57 ± 2.94	23.98 ± 3.25	15.45 ± 3.42	28.16 ± 3.26	26.26 ± 2.94	15.28 ± 3.84	31.31 ± 4.18	27.71 ± 3.66	**< 0.001**	**< 0.001**
Peak Extension (°)	-14.74 ± 3.88	-14.06 ± 3.96	-14.86 ± 3.37	-20.88 ± 3.40	-14.23 ± 4.71	-13.94 ± 3.37	-22.22 ± 3.93	-12.72 ± 4.08	-13.17 ± 3.55	-22.90 ± 4.35	-10.88 ± 4.55	-13.32 ± 4.56	**< 0.001**	0.291
ROM (°)	37.46 ± 2.50	37.16 ± 2.49	37.73 ± 2.16	37.30 ± 3.18	38.80 ± 2.97	37.92 ± 2.36	37.67 ± 2.97	40.88 ± 2.42	39.43 ± 2.84	38.18 ± 3.00	42.19 ± 2.56	41.03 ± 3.12	**< 0.001**	**< 0.001**
Knee														
Peak Flexion (°)	54.04 ± 4.51	55.22 ± 4.22	54.75 ± 4.43	54.23 ± 3.66	55.62 ± 4.74	55.15 ± 4.43	55.23 ± 3.25	56.28 ± 4.54	55.70 ± 4.16	55.98 ± 3.49	56.61 ± 4.35	56.35 ± 4.43	0.862	**< 0.001**
Peak Extension (°)	-0.61 ± 2.77	0.27 ± 3.47	-0.13 ± 3.24	-0.47 ± 2.72	0.35 ± 3.42	-0.09 ± 3.14	0.17 ± 2.85	0.71 ± 2.98	0.34 ± 3.34	1.13 ± 2.87	1.19 ± 3.18	0.46 ± 3.18	0.523	**0.039**
ROM (°)	54.65 ± 4.45	54.95 ± 4.64	54.88 ± 4.05	54.70 ± 3.80	55.27 ± 4.94	55.24 3.90	55.06 ± 3.05	55.47 ± 4.72	55.36 ± 3.74	54.85 ± 3.30	55.42 ± 5.10	55.89 ± 3.84	0.824	0.603
Ankle														
Peak Dorsiflexion (°)	9.55 ± 2.52	9.47 ± 2.52	9.69 ± 2.08	9.81 ± 2.60	9.48 ± 2.31	10.26 ± 2.62	10.05 ± 2.66	9.35 ± 2.50	10.00 ± 2.18	10.32 ± 2.54	9.21 ± 2.80	10.12 ± 2.24	0.106	0.776
Peak Plantarflexion (°)	-16.14 ± 6.29	-15.38 ± 5.29	-15.65 ± 4.93	-15.36 ± 6.17	-15.94 ± 5.97	-15.23 ± 4.95	-15.41 ± 5.41	-16.71 ± 5.44	-15.64 ± 4.78	-16.16 ± 5.57	-17.45 ± 5.82	-15.84 ± 5.02	**0.021**	**0.031**
ROM (°)	25.69 ± 6.12	24.85 ± 4.48	25.34 ± 4.16	25.17 ± 4.60	25.43 ± 5.07	25.49 ± 4.20	25.46 ± 3.97	26.05 ± 3.85	25.64 ± 4.41	26.49 ± 4.43	26.66 ± 4.27	25.96 ± 4.53	0.205	**0.024**

H = Head, B = Back, B/F = Back/Front

Table 26. Mean ± SD magnitudes for kinematic joint angles for unloaded walking and each load carriage condition. Significance values are for the change from unloaded walking for each variable. Negative values represent extension, positive values represent flexion.

Joint kinematics	0kg			3kg			12kg			20kg			*p* - value	
	H	B	B/F	H	B	B/F	H	B	B/F	H	B	B/F	Method	Mass
Trunk														
Pose at heel strike (°)	1.92 ± 2.74	1.79 ± 2.41	1.80 ± 2.61	-4.20 ± 2.67	2.36 ± 2.98	2.57 ± 2.71	-5.72 ± 3.42	5.31 ± 3.04	4.02 ± 3.31	-6.46 ± 3.44	8.14 ± 3.73	4.56 ± 3.31	**< 0.001**	**< 0.001**
Pose at toe off (°)	-0.48 ± 2.60	-0.60 ± 2.48	-0.62 ± 2.49	-6.73 ± 2.68	-0.06 ± 3.18	0.33 ± 2.70	-8.15 ± 3.34	2.47 ± 3.04	1.67 ± 2.53	-8.88 ± 3.47	5.20 ± 3.60	1.99 ± 3.25	**< 0.001**	**< 0.001**
Heel strike to toe off excursion (°)	2.40 ± 0.80	2.39 ± 0.56	2.41 ± 0.70	2.53 ± 0.65	2.42 ± 0.79	2.24 ± 0.71	2.43 ± 0.78	2.83 ± 0.56	2.35 ± 0.80	2.42 ± 0.72	2.94 ± 0.40	2.57 ± 0.69	0.094	**0.026**
Hip														
Pose at heel strike (°)	19.45 ± 3.68	19.84 ± 3.98	19.83 ± 3.76	13.21 ± 3.15	21.03 ± 4.62	21.12 ± 3.82	12.17 ± 3.66	24.90 ± 4.86	23.66 ± 3.84	12.09 ± 4.14	28.34 ± 5.36	25.07 ± 4.17	**< 0.001**	**< 0.001**
Pose at toe off (°)	-6.23 ± 3.90	-3.62 ± 5.64	-5.84 ± 3.13	-12.01 ± 3.31	-3.13 ± 6.09	-4.15 ± 3.31	-12.39 ± 3.98	-1.05 ± 5.81	-2.75 ± 3.11	-12.67 ± 4.33	1.53 ± 6.20	-2.36 ± 3.85	**< 0.001**	**0.002**
Heel strike to toe off excursion (°)	25.69 ± 4.04	23.47 ± 6.83	25.67 ± 3.58	25.21 ± 4.34	24.16 ± 7.35	25.29 ± 3.61	24.56 ± 4.62	25.95 ± 7.30	26.41 ± 3.66	24.76 ± 4.07	26.81 ± 7.70	27.43 ± 3.67	**0.002**	**< 0.001**
Knee														
Pose at heel strike (°)	2.50 ± 3.19	3.32 ± 3.90	3.15 ± 3.74	2.79 ± 2.97	3.99 ± 3.52	3.66 ± 3.26	3.57 ± 3.12	5.60 ± 3.59	4.80 ± 3.61	4.75 ± 3.04	6.91 ± 3.05	5.64 ± 3.85	0.192	**< 0.001**
Pose at toe off (°)	37.67 3.95	38.83 4.11	38.76 4.30	38.75 3.67	38.64 4.39	39.69 3.95	40.44 3.99	38.84 3.92	39.85 3.93	41.31 3.54	38.96 3.77	40.54 3.74	**< 0.001**	**0.001**
Heel strike to toe off excursion (°)	35.17 4.46	35.51 4.27	35.61 4.60	35.96 3.99	34.66 4.48	36.03 3.80	36.87 3.91	33.24 3.70	35.05 3.75	36.56 3.11	32.05 3.17	34.90 3.91	**< 0.001**	**0.046**
Ankle														
Pose at heel strike (°)	-4.25 ± 3.08	-4.36 ± 2.73	-4.06 ± 2.88	-3.79 ± 2.96	-4.46 ± 2.87	-4.06 ± 2.67	-3.26 ± 2.96	-4.03 ± 2.71	-4.05 ± 2.91	-3.13 ± 3.12	-3.44 ± 2.60	-3.96 ± 2.80	**0.009**	**0.002**
Pose at toe off (°)	-12.67 5.37	-12.69 4.53	-12.49 4.03	-13.03 5.27	-12.69 4.53	-12.06 4.04	-13.03 5.27	-14.19 4.76	-12.88 4.08	-13.86 5.32	-14.93 5.02	-13.17 4.22	0.337	**0.001**
Heel strike to toe off excursion (°)	8.42 3.75	8.51 4.47	8.43 2.18	8.78 3.66	8.23 2.92	8.00 2.23	9.78 3.60	10.17 2.97	8.83 2.54	10.73 3.76	11.49 3.41	9.21 2.77	0.158	**0.001**

H = Head, B = Back, B/F = Back/Front

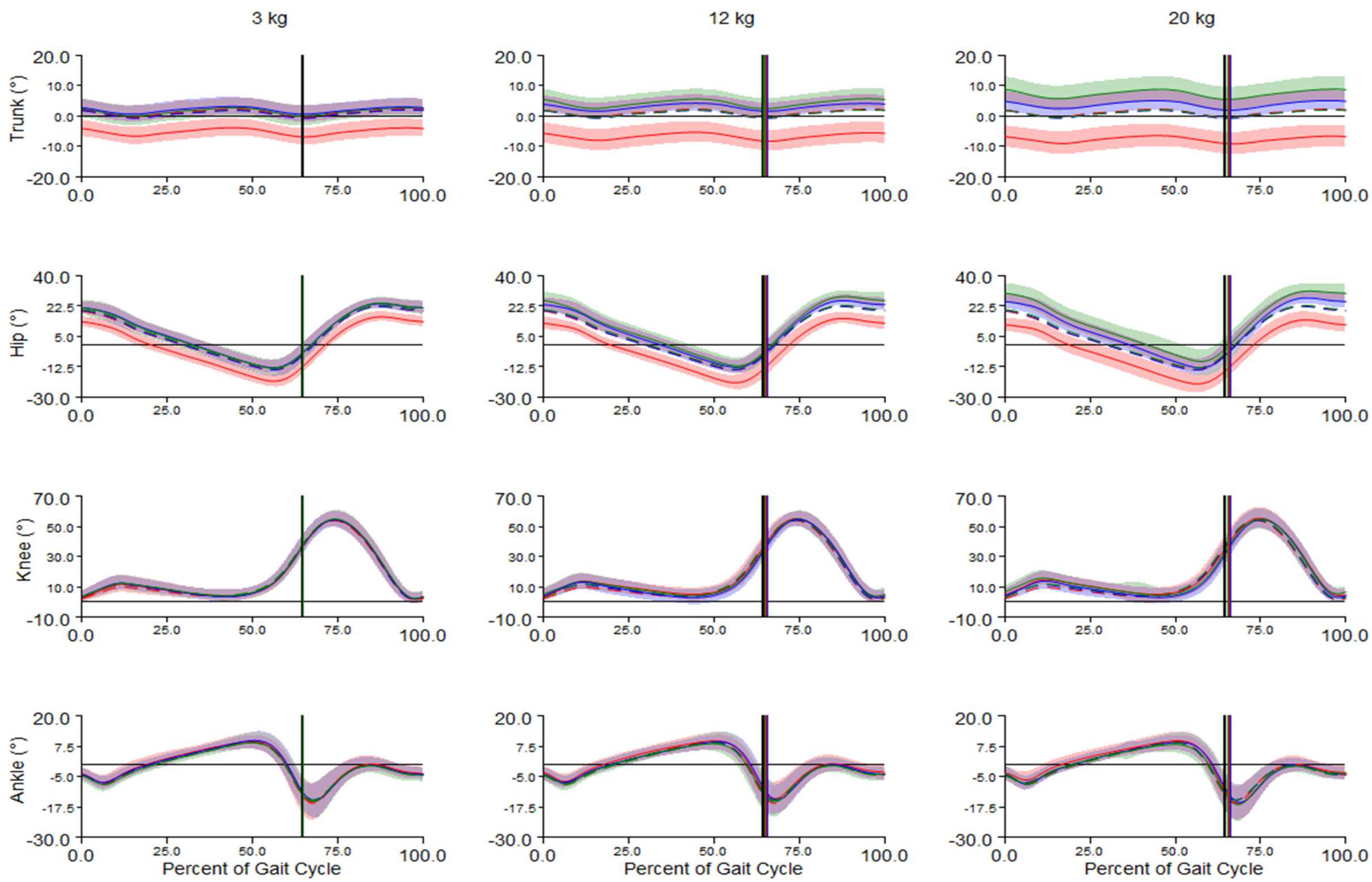


Figure 49. Trunk, hip, knee and ankle sagittal plane kinematics while carrying 3, 12 and 20 kg. Red lines represent the head-loading method, green lines represent the back-loading method and blue lines represent the doublepack method. The shaded areas represent standard deviations. Unloaded walking kinematics for each method are included as dashed lines in each figure. Vertical lines indicate the end of the stance phase.

There was a significant main effect of load method for the Δ trunk axial rotation from unloaded walking ($p < 0.001$, $\eta^2 = 0.621$, Figure 50). There was a significant decrease in trunk axial rotation for Head (-4.14 ± 2.61°) compared to Back (-1.37 ± 1.82°, $p = 0.001$) and Back/Front (-1.00 ± 2.07°, $p < 0.001$). There was also a significant main effect of load mass for the Δ trunk axial rotation from unloaded walking ($p = 0.015$, $\eta^2 = 0258$). However, post hoc adjustment showed no significant difference between any of the load masses, although the 2.71° decrease from 3 to 20 kg was close to achieving significance ($p = 0.061$).

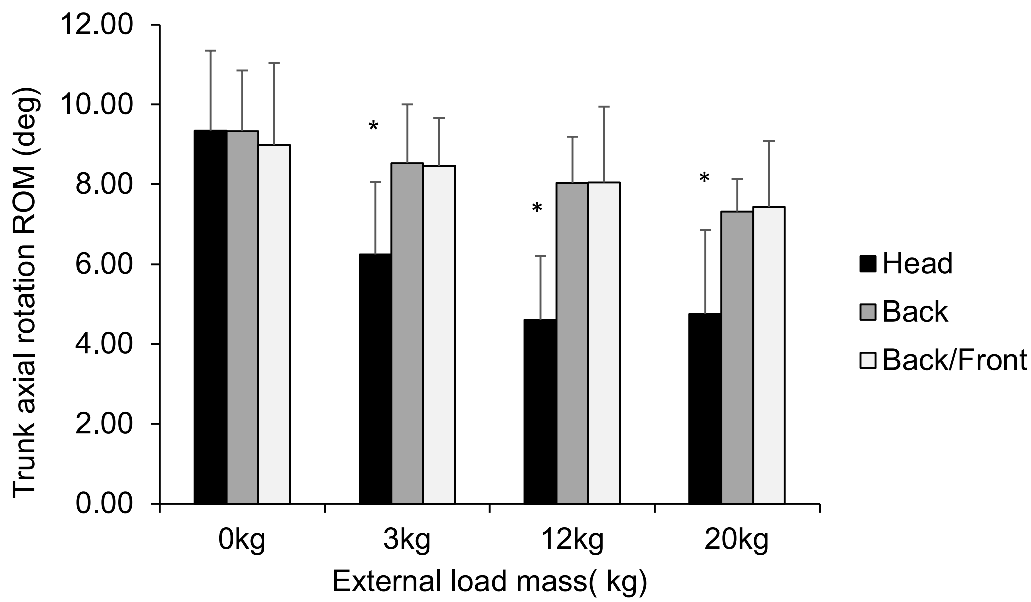


Figure 50. Mean ± SD trunk angle axial rotation for each load carriage condition. * indicates a significant difference from the other methods

The Δ pelvic axial rotation from unloaded walking was significantly different for method (main effect of load method, $p = 0.001$, $\eta^2 = 0.373$, Figure 51). Post hoc analysis revealed that pelvic rotation significantly decreased from unloaded walking for Back (-1.58 ± 1.57°) and Back/Front (-1.67 ± 1.54°) compared to Head (0.10 ± 1.95°) (Back vs. Head, $p = 0.014$; Back/Front vs. Head, $p = 0.015$). There was also a significant main effect of mass on pelvic rotation ($p = 0.013$, $\eta^2 = 0.331$). Post hoc analysis between the load mass conditions showed a significant decrease in pelvic rotation between 3kg and 12kg (-0.64° ± 1.89 vs. -1.25° ± 1.85, p = 0.016).

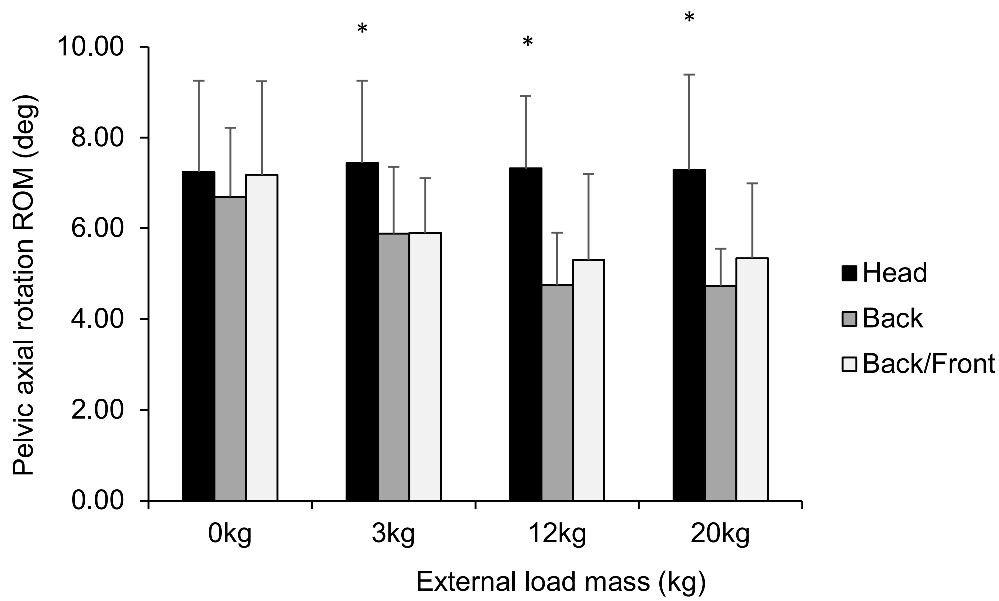


Figure 51. Mean ± SD pelvic axial rotation (degrees) in each load carriage condition. * indicates a significant difference from the other methods.

7.3.6. Ground reaction forces

No significant differences were observed between unloaded walking trials for any of the measured ground reaction force variables (p > 0.05) (Table 27).

Significant main effects of load method were observed for the Δ minimum vertical force ($p < 0.001$, $\eta^2 = 0.536$) and Δ 2nd peak component of vertical force ($p < 0.001$, $\eta^2 = 0.486$) from unloaded walking as absolute values and normalised to total mass (minimum force: $p = 0.001$, $\eta^2 = 0.469$; 2nd peak: $p = 0.003$, $\eta^2 = 0.339$) (Table 27). The difference in the Δ propulsive force between methods was also close to significance ($p = 0.059$, $\eta^2 = 0.183$). Specifically, post hoc analysis of the Δ 2nd peak vertical force relative to total mass revealed a significantly lower magnitude of force for Head compared to Back/Front (p = 0.011) (pooled load mass, Head = 10.24 ± 0.28 $N{\cdot}kgTM^{-1}$ vs Back/Front = 10.43 ± 0.26 $N{\cdot}kgTM^{-1}$). For the Δ minimum vertical force from unloaded walking relative to total mass, post hoc analysis of revealed significantly greater minimum force for Head (pooled load mass = 8.99 ± 0.17 $N{\cdot}kgTM^{-1}$) compared to Back (pooled load mass = 8.88 ± 0.17 $N{\cdot}kgTM^{-1}$) ($p = 0.035$) and Back/Front (pooled load mass = 8.83 ± 0.19 $N{\cdot}kgTM^{-1}$) ($p = 0.002$). Although the difference in propulsive force was not significantly different between methods, there was a tendency for Back-loading to be associated with a larger Δ peak propulsive force from unloaded walking compared to the other methods (pooled load mass of -14.86 ± 11.50 N, -18.93 ± 12.72 N and -15.65 ± 11.58 N for Head-, Back- and Back/Front, respectively).

There were significant main effects of load mass observed for the majority of measured ground reaction force variables ($p < 0.05$, $\eta^2 > 0.295$) (Table 27), except for propulsive force normalised to total mass, lateral force and medial force normalised to total mass. The magnitude of force significantly increased as the mass of the load increased for the Δ peak vertical forces as absolute values from unloaded walking ($p < 0.001$). When normalised to total mass, the Δ 1st peak vertical force significantly increased as the mass of the load increased ($p \leq 0.031$) but for the Δ 2nd peak vertical the only significant difference was between the 3 kg and 20 kg ($p = 0.002$). Post hoc analysis showed that the Δ peak propulsive force from unloaded walking increased with each increase in mass ($p < 0.05$). For peak propulsive force with 20 kg, the absolute increase from unloaded walking was -25.15 ± 10.12 N, -32.23 ± 6.56N and -27.43 ± 7,71 N for Head, Back and Back/Front-loading, respectively. The difference in Δ peak propulsive force from unloaded walking between load mass disappeared when normalised to total mass. Considering the Δ peak medial force from unloaded walking, post hoc analysis revealed significant differences in Δ peak medial force from unloaded walking for 3 kg (1.24 ± 0.76 N) compared to 12 kg (5.93 ± 1.58 N; $p = 0.014$), 3 kg compared to 20 kg (10.44 ± 1.34 N; $p = 0.001$) and 12 kg compared to 20 kg ($p = 0.041$).

Table 27. Mean ± SD magnitudes for ground reaction forces for unloaded walking and each load carriage condition. Significance values are for the change from unloaded walking for each variable.

Peak kinetic variables	0kg			3kg			12kg			20kg			*p* - value	
	H	B	B/F	H	B	B/F	H	B	B/F	H	B	B/F	Method	Mass
Vertical GRF														
1^{st} Peak (N·)	743.7 ± 99.7	740.2 ± 98.4	738.7 ± 95.31	775.2 ± 101.3	772.4 ± 104.8	767.2 ± 96.5	854. ± 95.12	859.2 ± 110.5	850.6 ± 103.8	924.1 ± 109.5	927.0 ± 107.7	929.6 ± 104.3	0.435	**< 0.001**
2^{nd} Peak (N)	772.6 ± 101.3	773.4 ± 103.1	770.8 ± 104.2	792.1 ± 100.5	807.3 ± 104.2	802.3 ± 102.3	878.1 ± 94.9	894.7 ± 101.7	889.6 ± 103.0	948.6 ± 105.0	967.6 ± 104.5	971.3 ± 102.2	**< 0.001**	**< 0.001**
1^{st} Peak ($N{\cdot}kgTM^{-1}$)	10.1 ± 0.2	10.1 ± 0.2	10.1 ± 0.2	10.1 ± 0.2	10.1 ± 0.2	10.0 ± 0.3	10.0 ± 0.3	10.1 ± 0.3	9.9 ± 0.2	9.9 ± 0.3	9.9 ± 0.3	9.9 ± 0.3	0.512	**< 0.001**
2^{nd} Peak ($N{\cdot}kgTM^{-1}$)	10.5 ± 0.3	10.5 ± 0.3	10.5 ± 0.3	10.3 ± 0.3	10.6 ± 0.2	10.5 ± 0.3	10.3 ± 0.3	10.5 ± 0.2	10.4 ± 0.3	10.1 ± 0.3	10.4 ± 0.2	10.4 ± 0.2	**0.003**	**0.001**
AP GRF														
Braking (N)	76.4 ± 14.9	75.6 ± 14.0	78.3 ± 16.9	83. 8 ± 17.4	80.4 ± 14.8	82.5 ± 17.4	95.0 ± 17.9	93.5 ± 17.0	93.1 ± 18.8	104.4 ± 17.7	106.5 ± 17.4	103.9 ± 18.2	0.163	**< 0.001**
Propulsive (N)	-97.8 ± 13.3	-99.5 ± 12.3	-99.5 ± 12.4	-102.6 ± 14.4	-104.6 ± 11.8	-102.6 ± 13.7	-112.5 ± 11.5	-119.0 ± 12.4	-115.9 ± 12.6	-123.0 ± 14.6	-131.7 ± 12.0	-127.0 ± 13.5	0.059	**< 0.001**
Braking ($N{\cdot}kgTM^{-1}$)	1.0 ± 0.2	1.0 ± 0.2	1.1 ± 0.2	1.1 ± 0.2	1.1 ± 0.2	1.1 ± 0.2	1.1 ± 0.2	1.1 ± 0.2	1.1 ± 0.2	1.1 ± 0.1	1.1 ± 0.2	1.1 ± 0.2	0.253	**0.008**
Propulsive ($N{\cdot}kgTM^{-1}$)	-1.3 ± 0.2	-1.4 ± 0.2	-1.4 ± 0.2	-1.3 ± 0.2	-1.4 ± 0.1	-1.4 ± 0.2	-1.3 ± 0.2	-1.4 ± 0.2	-1.4 ± 0.2	-1.3 ± 0.1	-1.4 ± 0.2	-1.4 ± 0.2	0.147	0.645
ML GRF														
Medial (N·)	50.1 ± 14.2	51.4 ± 14.3	47.0 ± 14.5	52.5 ± 12.5	51.1 ± 14.6	48.6 ± 14.5	59.4 ± 13.6	53.5 ± 15.4	53.4 ± 16.9	62.7 ± 13.0	61.3 ± 16.4	55.8 ± 16.1	0.133	**< 0.001**
Lateral (N)	12.3 ± 5.4	10.5 ± 3.7	15.9 ± 10.4	14.9 ± 6.5	13.0 ± 4.8	16.8 ± 10.9	13.9 ± 5.3	12.8 ± 5.2	16.9 ± 8.5	13.5 ± 4.2	12.0 ± 5.0	19.5 ± 11.8	0.958	0.829
Medial ($N{\cdot}kgTM^{-1}$)	0.7 ± 0.1	0.7 ± 0.1	0.6 ± 0.1	0.7 ± 0.1	0.7 ± 0.1	0.6 ± 0.1	0.7 ± 0.1	0.6 ± 0.1	0.6 ± 0.1	0.7 ± 0.1	0.7 ± 0.1	0.6 ± 0.1	0.135	0.357
Lateral ($N{\cdot}kgTM^{-1}$)	0.2 ± 0.1	0.1 ± 0.1	0.2 ± 0.1	0.2 ± 0.1	0.2 ± 0.1	0.2 ±0.1	0.2 ± 0.1	0.2 ± 0.1	0.2 ± 0.1	0.1 ± 0.1	0.1 ± 0.1	0.2 ± 0.1	0.649	**0.004**

H = Head, B = Back, B/F = Back/Front, GRF = Ground reaction forces, AP = Anteroposterior, ML = Mediolateral, TM = Total mass (combined body mass and external load)

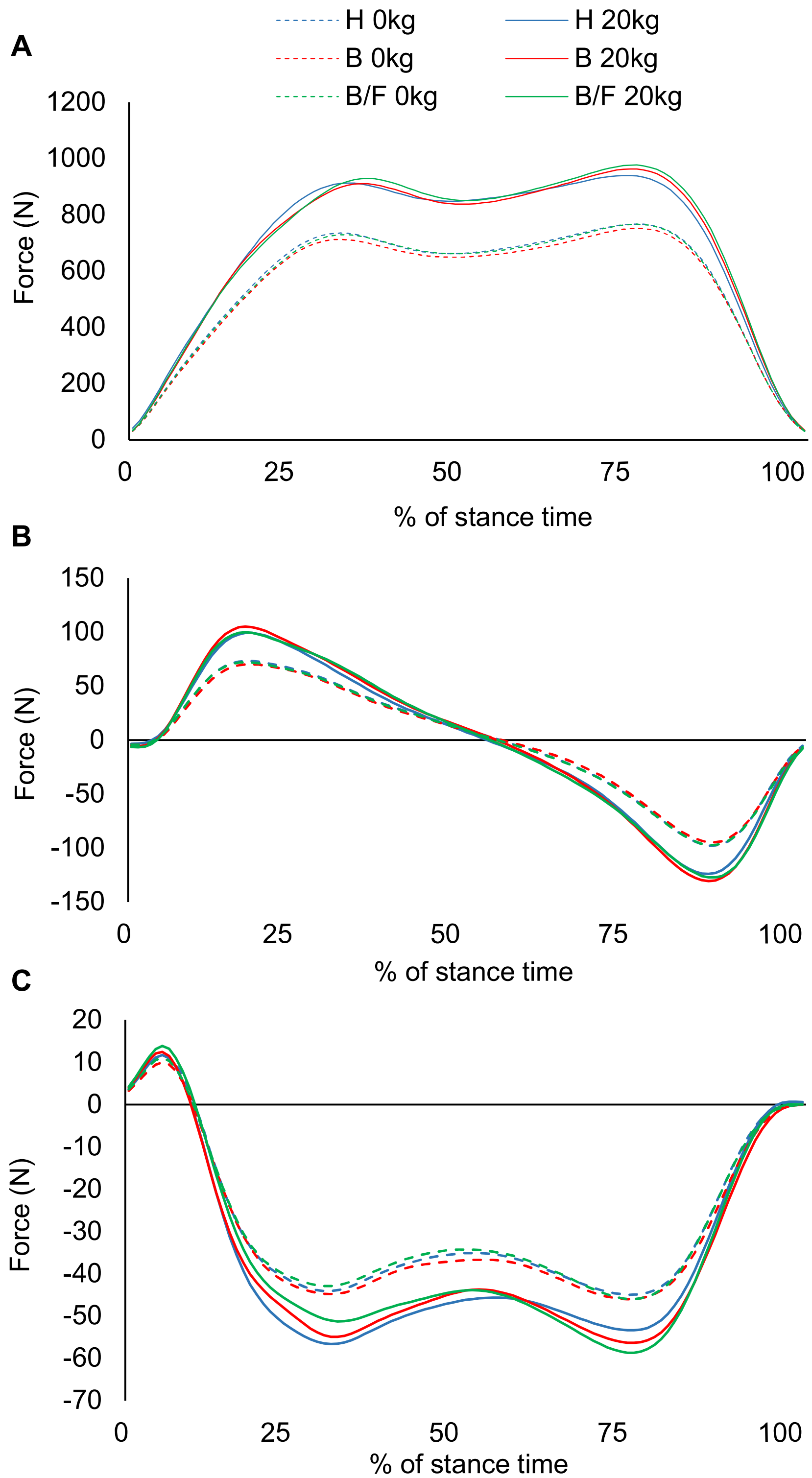


Figure 52. Vertical (A), anteroposterior (B) and mediolateral (C) forces for unloaded walking and 20kg in each load carriage method.

7.3.6.1. Force and momentum in double and single stance

Table 28 shows the mean ± SD values for anteroposterior force, impulse and momentum in double and single stance for all load carriage conditions. The Δ net antero-posterior GRF from unloaded walking for double support time was significantly different for method (main effect of load method, $p = 0.049$, $\eta^2 = 0.214$) with post-hoc analysis showing a significant difference for Head-loading compared to Back/Front-loading ($p = 0.012$). This difference was the result of an increase in propulsive force in the Back/Front-loading method from unloaded walking (load mass pooled, Back/Front = -2.33 ± 5.19 N) in contrast to an increase in braking force from unloaded walking in the Head-loading condition (load mass pooled, Head = 0.22 ± 3.83 N). The increase in the Δ net whole body horizontal momentum was also significantly different between methods (main effect of load method, $p = 0.037$, $\eta^2 = 0.244$). Post hoc analysis indicated that there was a smaller Δ momentum from unloaded walking in double stance for the Head compared to Back (load mass pooled, 0.23 ± 0.75 $kg{\cdot}m/s^{-1}$ vs. 0.76 ± 0.73 $kg{\cdot}m/s^{-1}$, $p = 0.029$), and Head compared to Back/Front (load mass pooled for Back/Front, 0.73 ± 1.13 $kg{\cdot}m/s^{-1}$, $p = 0.012$).

During single stance, the only significant difference between methods was for the Δ mean braking impulse from unloaded walking, which significantly reduced for Head loading compared to the two trunk loading methods (main effect of load mass, $p = 0.004$, $\eta^2 = 0.322$).

The Δ anteroposterior force, impulse and momentum variables for double stance from unloaded walking significantly increased as the mass of the load increased for all variables (main effect of load mass, $p \leq 0.017$, $\eta^2 \geq 0.303$). For single stance, mean braking and propulsive force variables significantly increased as the mass of the load increased (main effect of load mass $p < 0.001$, $\eta^2 \geq 0.762$). There was a significant main effect of load mass for the Δ single stance form unloaded for braking impulse ($p < 0.001$, $\eta^2 = 0.618$) and propulsive impulse ($p < 0.001$, $\eta^2 = 0.673$) with significant increases in both impulse variables each time load mass increased (all $p < 0.05$).

Table 28. Mean ± SD magnitudes for force, impulse and momentum in double stance and single stance gait phases for unloaded walking and each load carriage condition. Significance values are for the change from unloaded walking for each variable.

Variable	0kg			3kg			12kg			20kg			*p* - value	
	H	B	B/F	H	B	B/F	H	B	B/F	H	B	B/F	Method	Mass
Double stance phase														
Mean braking force (N)	42.4 ± 11.5	41.3 ± 11.3	42.9 ± 12.2	46.9 ± 12.5	43.7 ± 11.8	46.1 ± 12.6	54.1 ± 12.0	52.6 ± 11.3	51.9 ± 12.1	60.1 ± 12.6	60.9 ± 10.5	59.4 ± 13.0	0.391	**< 0.001**
Mean propulsive force (N)	-69.6 ± 10.2	-70.7 ± 8.9	-70.4 ± 9.4	-73.6 ± 10.7	-74.0 ± 10.2	-72.9 ± 11.0	-80.1 ± 7.3	-84.3 ± 9.4	-82.8 ± 9.2	-88.1 ± 10.4	-92.8 ± 9.0	-91.4 ± 9.5	0.345	**< 0.001**
Net AP force (N)	-27.1 ± 5.6	-29.3 ± 6.2	-27.5 ± 7.2	-26.7 ± 5.9	-30.3 ± 5.9	-26.8 ± 7.5	-26.1 ± 7.6	-31.7 ± 6.4	-30.8 ± 6.9	-28.0 ± 7.3	-31.9 ± 6.1	-32.0 ± 7.5	**0.034**	**0.017**
Mean braking impulse ($N{\cdot}s^{-1}$)	7.9 ± 2.4	7.6 ± 2.3	8.0 ± 2.5	8.8 ± 2.6	8.2 ± 2.4	8.7 ± 2.7	10.7 ± 2.9	10.5 ± 2.5	10.3 ± 2.8	12.1 ± 3.0	12.5 ± 2.4	12.2 ± 3.0	0.287	**< 0.001**
Mean propulsive impulse ($N{\cdot}s^{-1}$)	-12.8 ± 2.3	-13.0 ± 2.1	-13.1 ± 2.1	-13.8 ± 2.4	-13.9 ± 2.3	-13.6 ± 2.4	-15.7 ± 2.3	-16.7 ± 2.4	-16.3 2.4	-17.7 ± 2.9	-19.0 ± 2.2	-18.7 ± 2.6	0.121	**< 0.001**
Mean momentum ($kg{\cdot}m/s^{-1}$)	4.97 ± 0.96	5.4 ± 1.1	5.1 ± 1.2	5.0 ± 1.0	5.7 ± 1.1	5.0 ± 1.2	5.0 ± 1.2	6.3 ± 1.3	6.0 ± 1.2	5.6 ± 1.4	6.5 ± 1.2	6.5 ± 1.4	**0.037**	**< 0.001**
Single stance phase														
Mean braking force (N)	34.2 ± 4.8	34.0 ± 5.3	34.2 ± 4.5	35.2 ± 6.2	36.1 ± 4.2	35.0 ± 5.4	38.1 ± 4.5	39.2 ± 5.3	38.3 ± 5.4	40.4 ± 5.6	42.8 ± 6.2	42.7 ± 6.6	0.321	**< 0.001**
Mean propulsive force (N)	-30.5 ± 6.3	-29.4 ± 7.2	-30.4 ± 6.1	-31.5 ± 6.1	-30.1 ± 7.1	-31.5 ± 6.6	-34.4 ± 5.8	-35.2 ± 7.4	-34.7 ± 6.6	-36.8 ± 6.7	-39.7 ± 6.5	-38.6 ± 7.3	0.325	**< 0.001**
Net AP force (N)	3.7 ± 5.2	4.6 ± 5.3	3.8 ± 4.8	3.7 ± 5.1	6.0 ± 4.8	3.6 ± 3.3	3.6 ± 5.4	3.9 ± 5.2	3.6 ± 4.7	3.6 ± 5.2	3.1 ± 6.8	4.0 ± 4.7	0.986	0.447
Mean braking impulse ($N{\cdot}s^{-1}$)	9.4 ± 1.6	9.5 ± 1.8	9.2 ± 1.5	9.1 ± 1.7	9.7 ± 1.4	9.1 ± 1.6	9.4 ± 1.4	10.7 ± 1.6	10.0 ± 1.7	9.7 ± 1.7	11.3 ± 1.5	10.9 ± 1.8	**0.004**	**< 0.001**
Mean propulsive impulse ($N{\cdot}s^{-1}$)	-5.7 ± 1.4	-5.4 ± 1.8	-5.8 ± 1.7	-5.9 ± 1.5	-5.7 ± 1.6	-5.9 ± 1.7	-6.6 ± 1.4	-6.5 ± 1.8	-6.4 ± 1.6	-6.8 ± 1.6	-7.3 ± 2.0	-7.1 ± 1.7	0.442	**< 0.001**
Mean momentum ($kg{\cdot}m/s^{-1}$)	-1.7 ± 2.3	-2.1 ± 2.4	-1.7 ± 2.1	-1.6 ± 2.2	-2.7 ± 2.1	-1.6 ± 1.4	-1.6 ± 2.3	-1.8 ± 2.3	-1.6 ± 2.1	-1.6 ± 2.2	-1.4 ± 2.9	-1.8 ± 2.0	0.977	0.364

H = Head, B = Back, B/F = Back/Front. Negative anteroposterior force values indicate propulsion, positive values indicate braking.

7.3.7. Relationships between economy and walking gait variables

7.3.7.1. Physical characteristics and ELI

There were no significant strong correlations ($r > 0.7$) between ELI values and stature, body mass or BMI for any of the load carriage methods or load mass. There was a moderate positive correlation for ELI and body mass ($r = 0.409$, $r^2 = 16.72\%$, $p = 0.082$), and ELI and BMI ($r = 0.401$, $r^2 = 16.08\%$, $p = 0.095$) in the 12 kg Back condition.

7.3.7.2. Variables in the deterministic model and ELI

All significant ($p < 0.05$) or close to significant ($p < 0.1$) relationships between factors included in the deterministic model and ELI for each load method are presented in Table 29, Table 30 and Table 31.

Table 29. Relationships between the change in model factors from unloaded and ELI for each load carriage method with 3 kg.

		3 kg		
Model level	Independent variable	Head ELI	Back ELI	B/F ELI
5	Δ Trunk angle TO		0.467	
5	Δ Peak hip extension		0.447	
5	Δ Knee angle HS		0.472	
5	Δ Knee ROM HS-TO		-0.522*	
5	Δ Ankle ROM		0.507	
5	Δ Trunk rotation	-0.609*		
5	Δ Pelvic rotation	0.539*		
6	Δ Momentum in DS		-0.488	
6	Δ Momentum in SS	-0.449		
7	Δ COM velocity in DS		-0.500	
7	Δ Propulsive force in SS	-0.459		

* indicates a significant relationship, HS = heel-strike, TO = Toe off, SS = Single stance phase, DS = Double stance phase

Table 30. Relationships between the change in model factors from unloaded and ELI for each load carriage method with 12 kg.

Model level	Independent variable	12 kg Head ELI	Back ELI	B/F ELI
2	Δ Step length	-0.650*		
2	Δ Cadence	0.574*		
3	Δ Step width		0.484	
4	Δ DST	-0.651*	0.648*	
5	Δ Trunk ROM HS-TO	0.465		0.462
5	Δ Knee ROM HS-TO		-0.528*	
5	Δ Trunk ROM	-0.561*		
5	Δ Ankle ROM		0.475	
6	Δ Momentum in SS	-0.485		
6	Δ Net AP force in SS	-0.460		
7	Δ COM velocity in SS	-0.533*		

* indicates a significant relationship. DST = Double stance time, HS = heel-strike, TO = Toe off, AP = Anteroposterior, SS = Single stance phase

Table 31. Relationships between the change in model factors from unloaded and ELI for each load carriage method with 20 kg.

Model level	Independent variable	20 kg Head ELI	Back ELI	B/F ELI
4	Δ DST		0.644*	
5	Δ Peak trunk flexion			0.578*
5	Δ Peak trunk extension			0.535*
5	Δ Trunk angle HS			0.593*
5	Δ Trunk angle TO			0.647*
5	Δ Peak hip extension			0.506
5	Δ Hip angle TO			0.507
5	Δ Knee angle HS			0.571*
5	Δ Knee ROM HS-TO			-0.519*
5	Δ Ankle angle HS	-0.487		
5	Δ Ankle angle TO	-0.523*		
6	Δ Net AP force in DS	-0.572*		
6	Δ Momentum in DS	-0.520*		
7	Δ Propulsive force in DS (trail leg)			0.450
7	Δ COM velocity in DS	-0.457		

* indicates a significant relationship, DST = Double stance time, HS = heel-strike, TO = Toe off, AP = Anteroposterior, DS = Double stance phase

For Head-loading with 3 kg, there was a significant positive correlation for ELI and Δ Pelvic ROM (r = 0.539, r^2 = 29.05%, p = 0.038) and a significant negative correlation for ELI and Δ Trunk ROM (r = -0.609, r^2 = 37.09%, p = 0.016). For head-loading with 12 kg, there were significant moderate negative correlations for ELI and Δ step length (r = -0.650, r^2 = 42.25%, p = 0.009), Δ double stance time (r = -0.650, r^2 = 42.25%, p = 0.009), Δ COM velocity in single stance (r = -0.533, r^2 = 28.40%, p = 0.041) and Trunk ROM (r = -0.561, r^2 = 31.47%, p = 0.029). There was also a significant moderate positive correlation for the Δ cadence from unloaded walking with 12 kg carried on the Head (r = 0.574, r^2 = 32.94%, p = 0.023. For head-loading with 20 kg, there were significant moderate negative correlations for ELI and Δ ankle angle at toe off (r = -0.523, r^2 = 27.35%, p = 0.045), Δ net anteroposterior force in double stance (r = -0.572, r^2 = 32.72%, p = 0.026) and Δ momentum in double stance (r = -0.520, r^2 = 27.04%, p = 0.047).

For Back-loading, with 3 kg there was a significant moderate relationship between ELI and Δ knee angle excursion from unloaded walking (r = -0.519, r^2 = 26.94% p = 0.047). There were significant moderate positive correlations between ELI and Δ double support time from unloaded walking with 12 kg (r = 0.648, r^2 = 41.99% p = 0.009) and 20 kg (r = 0.644, r^2 = 41.47% p = 0.010). There were no significant relationships between ELI and the change in step width from unloaded walking, however there was a moderate positive relationship for the Back method with 12 kg (r = 0.484, r^2 = 23.42% p = 0.067). With 12 kg there was a significant moderate correlation for ELI and the Δ knee angle excursion from unloaded walking (r = 0.530, r^2 = 29.09%, p = 0.042).

For the Back/Front method, there was a moderate negative correlation between ELI and Δ step length from unloaded walking (r = -0.485, r^2 =23.52% p = 0.067) with the 12 kg load. All significant relationships for ELI and joint angle kinematics in the Back/Front condition occurred with the 20 kg load. There was a significant positive correlation for ELI and Δ peak trunk flexion (r = 0.578, r^2 = 33.41%, p = 0.024), and Δ peak trunk extension (0.535, r^2 = 28.62%, p = 0.040) from unloaded walking. There were also significant positive relationships between ELI and Δ forward lean from unloaded at heel strike (r

= 0.593, r^2 = 35.16%, p = 0.019) and toe off (r = 0.647, r^2 = 41.86%, p = 0.009). ELI and the Δ knee angle at heel strike from unloaded walking were significantly related (r = 0.571, r^2 = 32.60%, p = 0.026) as well as ELI and the Δ knee angle excursion from unloaded walking (r = -0.519, r^2 = 26.94%, p = 0.048).

7.3.7.3. Additional biomechanical measures and ELI

All significant ($p < 0.05$) or close to significant ($p < 0.1$) relationships between additional biomechanical measures and ELI for each load method are presented in Table 32 and Table 33. There were no relationships between the Δ the additional measures from unloaded walking and ELI for any load method with the 3 kg mass.

Table 32. Relationships between the additional biomechanical measures and ELI for each load carriage method with 12 kg.

	12 kg		
Independent variable	Head ELI	Back ELI	B/F ELI
Δ 1st Peak Vertical N·kgTM^{-1}			-0.497
Δ Peak Propulsive N·kgTM^{-1}		-0.457	
Δ Peak Medial		0.539*	
Δ Peak Medial N·kgTM^{-1}		0.531*	
Δ Mean Braking Impulse DS	-0.522*		
Δ Mean Propulsive Impulse DS	-0.600*		

* indicates a significant relationship, DS = double stance

Table 33. Relationships between the change in model factors from unloaded and ELI for each load carriage method with 20 kg.

	20 kg		
Independent variable	Head ELI	Back ELI	B/F ELI
Δ 2nd Peak Vertical			-0.629*
Δ 2nd Peak Vertical N·kgTM^{-1}			-0.603*
Δ Peak Lateral			0.484

* indicates a significant relationship

Considering the Head loading 12 kg condition, there were significant moderate negative relationships between ELI and the Δ mean braking impulse (r = -0.522, r^2 = 27.24% p = 0.046) and mean propulsive impulse (r = -0.600, r^2 = 36.00% p = 0.018) during the double stance phase. For the Back method, there was a significant positive relationship for ELI and the Δ peak medial force from unloaded walking with 12 kg (r = 0.539, r^2 = 29.05% p = 0.038). There was also significant correlation between ELI and the Δ in medial force relative to total mass ($N \cdot kgTM^{-1}$) from unloaded walking with 12 kg (r = 0.531, r^2 = 28.19%, p = 0.042). For the Back/Front method with 20kg, there was a significant negative relationship for ELI and the Δ 2nd peak of vertical force from unloaded walking (r = - 0.629, r^2 = 39.56% p = 0.012). There was a significant correlation between ELI and the Δ 2nd peak vertical force relative to total mass ($N \cdot kgTM^{-1}$) from unloaded walking (r = -0.603, r^2 = 36.36%, p = 0.017) with 20 kg.

To further explore the multiple significant correlations, present in the Head 12 kg and Back/Front 20 kg conditions, participants were ranked in order of economy for the variables significantly related to ELI. The ranking figures for the Head method with 12 kg and the Back/Front method with 20 kg are shown in Figure 53 and Figure 54, respectively.

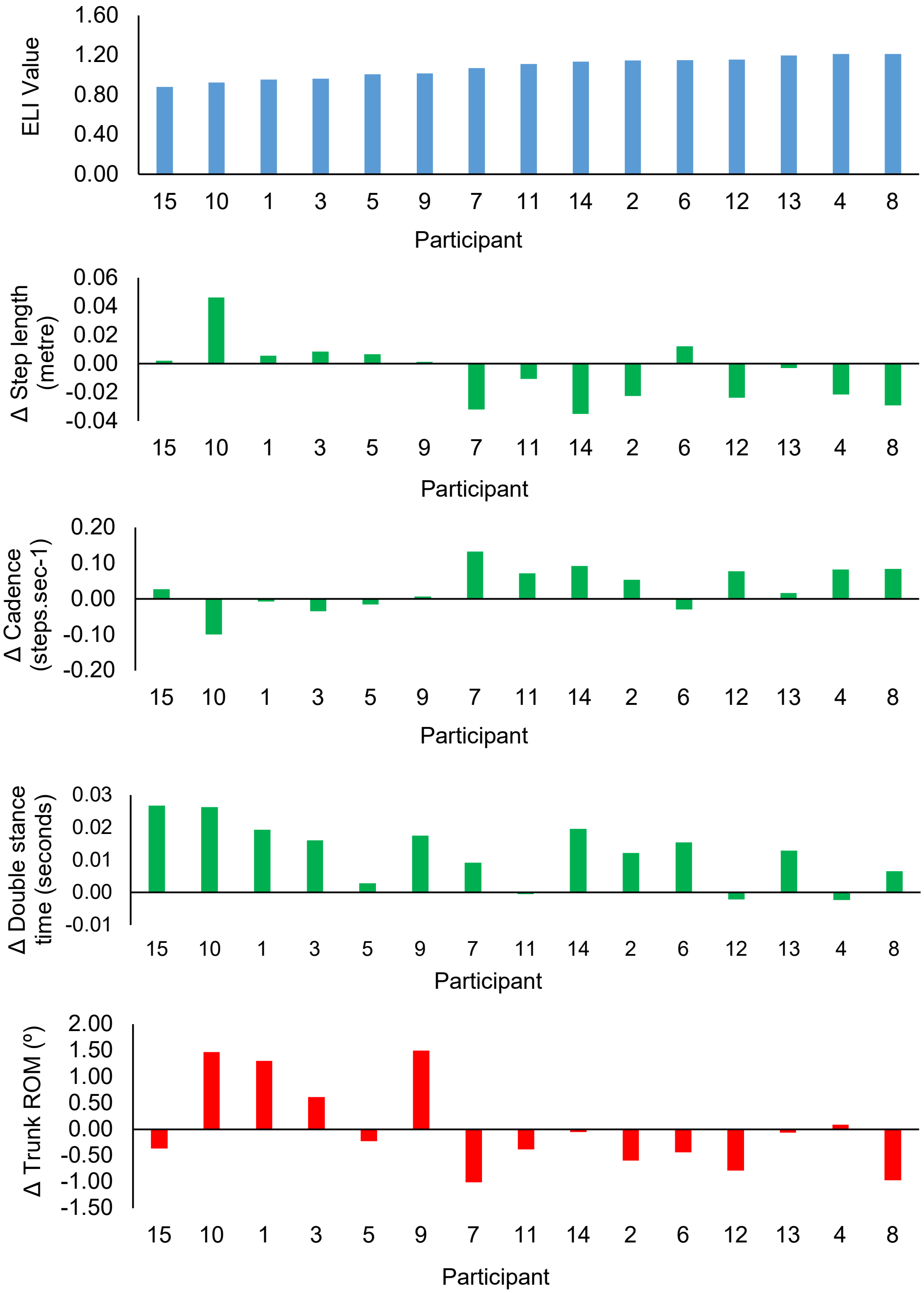


Figure 53. Step length, cadence, double stance time and trunk range of motion for head-loading with 12 kg ranked in order of the most of least economical participants.

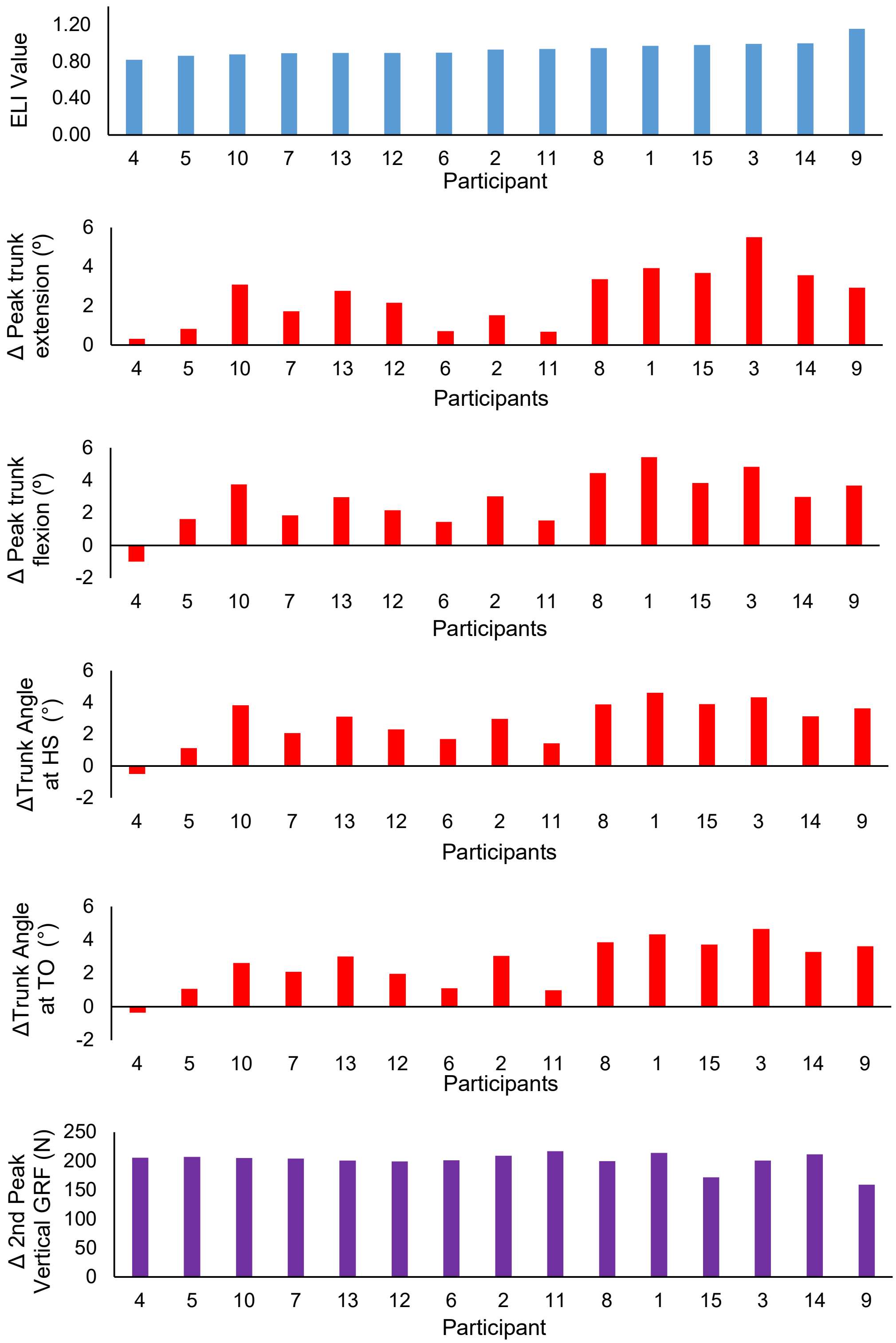


Figure 54. Step length, peak trunk flexion, peak trunk extension, trunk angle at heel strike (HS) and toe off (TO), and 2nd peak vertical force for back/front-loading with 20 kg ranked in order of the most of least economical participants.

7.3.8. Subjective perceptions

7.3.8.1. Ratings of perceived exertion (RPE)

RPE significantly increased as the mass of the load was increased (main effect load mass, $p = 0.001$, $\eta^2 = 0.798$). There was also a significant difference in RPE between the methods of load carriage (main effect load method, $p = 0.006$, $\eta^2 = 0.307$). Post hoc analysis showed that the Head method was associated with significantly higher RPE compared to Back/Front (10 ± 6 vs. 9 ± 4, $p = 0.013$) and there was a tendency for higher RPE for the Head method compared to the Back method (10 ± 6 vs 9 ± 5 $p = 0.059$).

7.3.8.2. Pain/discomfort scores

Table 34 shows the total pain/discomfort scores for each load carriage condition (all body areas combined). Pain/discomfort scores significantly increased as the mass of the load increased for all methods (main effect of load mass, $p = 0.003$, $\eta^2 = 0.425$). There was a significant main effect of load carriage method on the change in pain/discomfort scores from unloaded walking ($p = 0.007$, $\eta^2 = 0.296$). Specifically, there was significantly less pain/discomfort for the Back/Front method compared to the Head method ($p = 0.026$). Table 35 show the difference in pain discomfort scores between methods with 20 kg, which is where the largest difference in overall pain/discomfort exists between methods occurred. The largest difference between methods occurred at the neck with 20 kg (19 ± 17mm, 4 ± 6mm and 4 ± 6mm for Head, Back and Back/Front, respectively). Most pain/discomfort occurred in the upper body, closer to the position of the load. There was a significant difference in pain/discomfort between body positions (main effect of body position, $p = 0.002$, $\eta^2 = 0.406$), with significantly higher pain scores for the neck compared to the chest ($p = 0.047$) and the neck compared to the quadriceps ($p = 0.05$).

Table 34. Mean ± SD total pain/discomfort (mm) scores from all body segments combined for each loading condition. Less pain/discomfort is indicated by a lower score.

	0 kg	3 kg	12 kg	20 kg
Head	11 ± 3	22 ± 3	57 ± 9	75 ± 12
Back	8 ± 2	12 ± 2	35 ± 6	73 ± 11
Back/Front	13 ± 2	14 ± 2	35 ± 6	49 ± 8

Table 35. Mean ± SD RPE and pain/discomfort scores (mm) for the 20 kg load. Less pain/discomfort is indicated by a lower score.

	Head	Back	Back/Front
RPE	14 ± 3	13 ± 3	12 ± 3
Neck	19 ± 17	4 ± 6	4 ± 6
Back Shoulders	17 ± 26	16 ± 19	13 ± 17
Front Shoulders	15 ± 21	16 ± 17	14 ± 17
Chest	1 ± 2	5 ± 11	2 ± 4
Upper Back	7 ± 11	11 ± 18	7 ± 13
Abdomen	1 ± 2	4 ± 9	0 ± 1
Lower Back	5 ± 9	4 ± 7	4 ± 5
Hips	1 ± 3	3 ± 4	1 ± 2
Buttocks	1 ± 3	2 ± 5	0 ± 1
Quadriceps	1 ± 2	3 ± 6	1 ± 2
Hamstrings	1 ± 2	1 ± 2	1 ± 2
Knees	3 ± 6	1 ± 2	1 ± 3
Calves	1 ± 3	1 ± 2	0 ± 1
Ankles	1 ± 4	1 ± 2	0 ± 1
Feet	2 ± 4	2 ± 3	1 ± 2
Total	75 ± 12	73 ± 11	49 ± 8

7.3.9. Individual variation

7.3.9.1. $\dot{V}O_2$

The SD's and CV's shown in Table 36 indicate the magnitude of inter-individual variation in $\dot{V}O_2$ across the different methods. The mean CV for $\dot{V}O_2$ between the three unloaded walking trials was 13%. The inter-individual variation increased as the mass of the load increased and the magnitude of increase was smaller for the Back/Front method (change in CV of 0.47% from 0 – 20 kg) compared to the Head (change in CV of 6.67% from 0 – 20 kg) and Back (change in CV of 2.42% from 0 – 20 kg) methods.

Table 36. Mean, standard deviation (SD) and coefficient of variation (CV) for $\dot{V}O_2$ ($ml \cdot kg^{-1} \cdot min^{-1}$) values with each load method and load mass.

	0 kg	3 kg	12 kg	20 kg
Head				
$\dot{V}O_2$	10.34	11.41	12.97	14.48
SD	1.15	1.75	2.01	2.58
CV (%)	11.12	15.38	15.47	17.79
Back				
$\dot{V}O_2$	9.90	10.05	11.31	12.33
SD	1.41	1.33	1.73	2.06
CV (%)	14.26	13.26	15.28	16.68
Back/Front				
$\dot{V}O_2$	9.97	10.50	10.99	11.88
SD	1.21	1.19	1.44	1.50
CV (%)	12.11	11.32	13.14	12.58

The MLM analysis showed a significant difference in estimated variance between participants $\dot{V}O_2$ with the Head (σ^2_u = 2.31, standard error = 0.95, p = 0.016), the Back (σ^2_u = 2.22, standard error = 0.85, p = 0.009) and the Back/Front loading conditions (σ^2_u = 1.34, standard error = 0.52, p = 0.010). The estimated variance in $\dot{V}O_2$ between load mass conditions was also significant for Head (σ^2_e = 1.22, standard error = 0.25, p < 0.001), Back (σ^2_e = 0.41, standard error = 0.08, p < 0.001) and Back/Front (σ^2_e = 0.34, standard error = 0.07, $p < 0.001$). The ICC values for individual differences in $\dot{V}O_2$ as a proportion of the total variance were 0.65, 0.84 and 0.79 for Head, Back and Back/Front, respectively.

7.3.9.2. ELI

The inter-individual differences in ELI (SD and CV) for all loading conditions are presented in Table 37. The inter-individual variation in ELI follow a similar pattern of response to the $\dot{V}O_2$ data with the magnitude of inter-individual variation increasing as the mass of the load increase for all load methods. However, unlike the $\dot{V}O_2$ data, for ELI the magnitude of increase from unloaded walking to 20 kg was similar between the Back (CV increase of 2.5%) and Back/Front (CV increase of 2.8%) methods. Inter-individual variation in ELI was larger for the Head method (CV increase of 5.7%) compared to the Back and Back/Front methods.

Table 37. Mean, standard deviation (SD) and coefficient of variation (CV) for ELI values with each load method and load mass.

	3 kg	12 kg	20 kg
Head			
ELI	1.06	1.07	1.10
SD	0.09	0.11	0.15
CV (%)	8.18	10.24	13.92
Back			
ELI	0.98	0.98	0.98
SD	0.06	0.07	0.09
CV (%)	6.29	6.97	8.79
Back/Front			
ELI	1.01	0.95	0.94
SD	0.06	0.07	0.08
CV (%)	5.69	7.80	8.53

There was significant variance between participants for ELI values with head-loading (σ^2_u = 0.005, standard error = 0.003, p = 0.050) and back-loading (σ^2_u = 0.003, standard error = 0.001, p = 0.015). The variance between participants for and back/front-loading was close to significance (σ^2_u = 0.002, standard error = 0.001, p = 0.068). The estimated variance in ELI between load mass conditions was also significant for Head (σ^2_e = 0.007, standard error = 0.001, p < 0.001), Back (σ^2_e = 0.001, standard error = 0.001, p < 0.001) and Back/Front (σ^2_e = 0.002, standard error = 0.001, p < 0.001). The ICC values for individual differences in ELI as a proportion of the total variance were 0.45, 0.72 and 0.41 for Head, Back and Back/Front, respectively.

Figure 55 shows the most economical load mass with each load carriage method. Most participants had their best economy with 3 kg for the Head method (n = 8), with 12 kg for the Back method (n = 7) and with 20kg for the Back/Front method (n = 8).

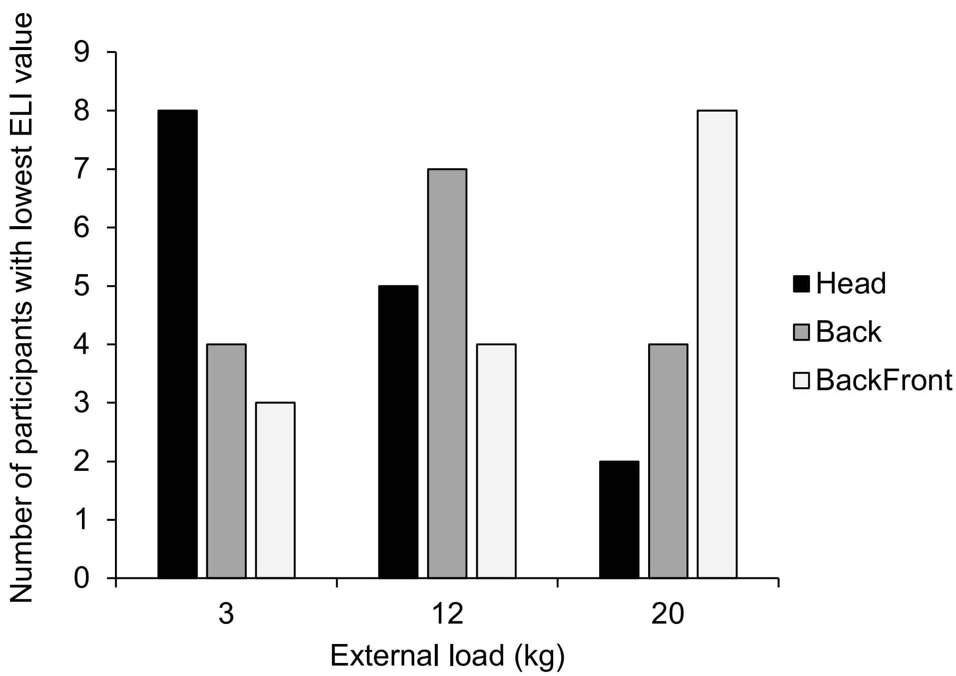


Figure 55. The most economical load mass for individuals for each load carriage method.

7.3.9.3. Within- and between- participant variability for walking gait adaptations

For spatiotemporal variables, single stance time had a larger magnitude of SDw (within participant standard deviations) than SDb (between participant standard deviations) for all load masses in the head- and back- loading methods, indicating that with-participant variability was greater than inter-individual variability. The largest difference between SDw and SDb for single stance occurred for back-loading with 12 kg (mean = 0.2 s, SDw = 0.04 s, SDb = 0.02 s). Overall, the largest difference between SDw and SDb for spatiotemporal variables occurred for cadence with the head-loading method and 20 kg (mean = 1.61 steps·s^{-1}, SDw = 0.18 steps·s^{-1}, SDb = 0.12 steps·s^{-1}). Step length was also associated with larger SDw than SDb for all methods with the 20 kg load. Step length was also associated with larger SDw than SDb for back-loading with the 20 kg load and had the same magnitude for SDw and SDb with 20 kg for head- and back/front-loading (Head: SDw = 0.04 m, SDb = 0.04 m; Back: SDw = 0.06 m, SDb = 0.04 m Back/Front: SDw = 0.03 m, SDb = 0.03 m).

For ground reaction forces, lateral force had greater SDw than the SDb for the head- and back-loading methods. The largest increases in SDw compared to SDb for lateral force occurred with 20 kg for head-loading (mean = 13.53N, SDw = 7.22N, SDb = 4.23N) and 12 kg for back-loading (mean = 12.82N, SDw = 7.16N, SDb = 5.23N). Posterior force also had greater SDw than SDb for head-loading with 12 kg (mean = -112.5N, SDw = 13.51N, SDb = 11.49N). and back-loading with 20 kg (mean = -131.7N, SDw = 12.53N, SDb = 11.99N). Considering vertical force, the magnitude of SDb was greater than SDw indicating that inter-individual variability was greater than within-participant variability.

For all joint angle kinematics, the magnitude of SDb was greater than SDw. The largest magnitude of SDb in the joint angle data occurred for ankle plantarflexion with 0 kg (peak plantarflexion angle = -16.14°, SDb = 6.29°, SDw = 3.31°). SDb and SDw for all spatiotemporal variables, joint angle kinematics and ground reaction forces for each load carriage condition are shown in Appendix P.

The largest differences for the magnitude of change in walking gait perturbations between participants occurred for the spatiotemporal variables, with the largest difference occurring with 20 kg for double and single stance times. With 20 kg the range of change in double stance time from unloaded walking between participants was +18% to -4%, +22% to -15% and +17% to -3% for head-, back- and back/front- loading, respectively. For single stance with 20 kg, the range was +2% to -15%, +17% to -11% and +10% to - 10% for head-, back- and back/front- loading, respectively. For step length, the largest range of change from unloaded walking occurred for head-loading with 20 kg (+7% to -13%). This was also the case for cadence, with a largest range of change from unloaded of +10% to -6% for head-loading with 20 kg. Considering trunk angle, the range between participants increased as the mass of the load increased. With 20 kg, the change in trunk angle at heel-strike from unloaded walking ranged from -3% to -18%, +3% to +11% and -1% to +5% for Head, Back and Back/Front, respectively. The change in trunk angle at toe off from unloaded walking with 20 kg from -4% to -18%, +1% to +10% and 0% to +5% for Head, Back and Back/Front, respectively.

7.3.10. Step width and load carriage economy

Table 38 shows the difference between preferred step width and controlled step width with each of the load carriage conditions. The change in step width from unloaded walking with preferred step width was significantly different between load carriage methods (main effect of load carriage method, p = 0.004, η^2 = 0.330) but there was no significant difference for load mass (main effect of mass, p = 0.563, η^2 = 0.040). Post-hoc analysis revealed significant differences for Head versus Back (p = 0.013) and Head versus Back/Front (p = 0.015). The largest difference between methods for Δ step width from unloaded walking occurred with 20kg (0.02 ± 0.02 m, 0.00 ± 0.02 m and 0.00 ± 0.02 m for Head, Back, and Back/Front, respectively). There was a tendency for a method x mass interaction effect (p = 0.059, η^2 = 0.147), with step width increasing for heavier load in Head-loading but not in the other two methods.

With step width controlled, there was no significant difference in step width for loading method (main effect of load carriage method, p = 0.506, η^2 = 0.051) or load mass (main effect of mass, p = 0.211, η^2 = 0.113).

Table 38. Mean ± SD and coefficients of variation (CV) for step width in the preferred step width and controlled step width conditions with each load method and mass combination.

	Uncontrolled Step Width				Controlled Step Width			
	0 kg	3 kg	12 kg	20 kg	0 kg	3 kg	12 kg	20 kg
Head								
Step width (m)	0.14	0.15	0.16	0.16	0.15	0.15	0.15	0.15
SD	0.02	0.03	0.03	0.02	0.02	0.02	0.02	0.02
CV (%)	11.93	18.60	16.96	14.83	11.19	12.02	11.33	9.85
Back								
Step width (m)	0.15	0.14	0.14	0.15	0.15	0.14	0.14	0.14
SD	0.02	0.02	0.02	0.02	0.02	0.02	0.02	0.02
CV (%)	12.46	13.08	13.09	10.50	14.48	12.65	10.43	10.58
BF								
Step width (m)	0.14	0.14	0.14	0.14	0.15	0.15	0.15	0.15
SD	0.01	0.01	0.02	0.02	0.02	0.02	0.01	0.02
CV (%)	7.84	10.29	12.96	16.98	14.72	14.44	9.86	10.64

The $\Delta\dot{V}O_2$ from unloaded walking with step width controlled displayed a similar pattern of response to the uncontrolled step width condition. The difference between preferred and controlled step width for $\dot{V}O_2$ was not significant for load carriage method (main effect of load carriage method, p = 0.345) or load mass (main effect of load mass, p = 0.939).

Figure 56 shows ELI values for preferred and controlled step width for each load carriage condition. With step width controlled, ELI values were significantly different for loading method (main effect of load carriage method, p = 0.001, η^2 = 0.507). Head-loading was associated with larger ELI values with every load mass compared to the other methods. The largest difference in ELI values between head-loading and the other methods was with 20 kg (1.11 ± 0.14, 0.94 ± 0.09 and 0.94 ± 0.07 for head, back and back/front, respectively). The difference between preferred and controlled step width for ELI was not significant between methods (main effect of load carriage method, p = 0.301, η^2 = 0.088) or load mass (main effect of load mass p = 0.872, η^2 = 0.011). The largest change in ELI values between the two step width conditions occurred when carrying 12 kg (change in ELI from preferred to modified step with = -0.05 ± 0.09) and 20 kg (change in ELI from preferred to modified step with = -0.04 ± 0.10) on the back.

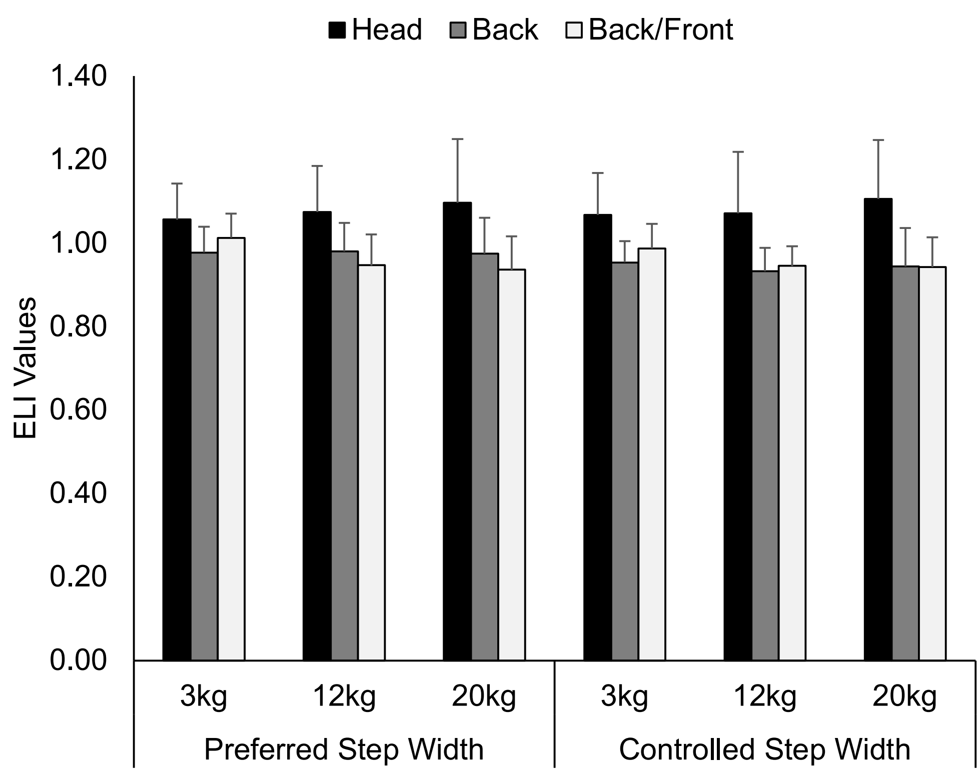


Figure 56. Mean SD ELI values for each load carriage condition with preferred and controlled step width

7.3.11. Summary of the results

Analysis of group data:

- ELI values were significantly higher for head-loading (1.08 ± 0.12 with load mass pooled) compared to back- (0.98 ± 0.07 with load mass pooled) and back/front-loading (0.97 ± 0.07 with load mass pooled) (main effect for load method, $p < 0.001$, $\eta^2 = 0.440$; Figure 48).
- There was a significant difference between methods for the Δ step length ($p = 0.001$, $\eta^2 = 0.391$), Δ cadence ($p = 0.045$, $\eta^2 = 0.198$), Δ step time ($p = 0.013$, $\eta^2 = 0.268$) and Δ single support time ($p = 0.010$, $\eta^2 = 0.283$) from unloaded walking (Table 24). Specifically, head-loading was associated with an increase in cadence and decrease in step length, step time and single support time compared to back- and back/front-loading.
- There were several significant differences in kinematics between the load carriage methods, with the largest differences occurring at the trunk and hip joints (Table 25, Table 26 and Figure 49):
    - Considering trunk angle, there was a significant difference between methods for all measured variables ($p \leq 0.003$, $\eta^2 \geq 0.347$), except for the Δ trunk angle excursion, which was close to statistical significance and had a large effect size ($p = 0.094$, $\eta^2 = 0.180$). Specifically, head-loading was associated with increased trunk extension from unloaded walking compared to the other loading methods. With pooled load mass, peak trunk forward lean values were -7.42 ± 3.39°, 3.76 ± 3.19° and 1.93 ± 1.47° for head-, back-, and back/front-loading, respectively. Trunk ROM significantly increased from unloaded walking for back-loading (0.86 ± 0.91° with pooled load mass) compared to head- (-0.02 ± 0.72° with pooled load mass) and back/front-loading (0.10 ± 0.76° with pooled load mass) (Back vs. Head, $p = 0.028$; Back vs Back/Front, $p = 0.002$).
    - Hip angle followed a similar pattern to the trunk, with significantly increased hip extension from unloaded walking for head-loading

compared to the back and back/front methods. With pooled load mass, peak hip extension values were -7.26 ± 4.40°, 1.44 ± 2.91° and 1.39 ± 1.39° for head-, back-, and back/front-loading, respectively (main effect for load method, $p < 0.001$, $\eta^2 = 0.760$).

  - Trunk axial rotation was significantly reduced from unloaded walking for head-loading (-4.14 ± 2.61° with pooled load mass) compared to back- (-1.37 ± 1.82° with pooled load mass, $p = 0.001$) and back/front-loading (-1.00 ± 2.07° with pooled load mass, $p < 0.001$) (Figure 50). Pelvic rotation significantly decreased from unloaded walking for back- (-1.58 ± 1.57° with pooled load mass) and back/front-loading (-1.67 ± 1.54° with pooled load mass) compared to head-loading (0.10 ± 1.95° with pooled load mass) (Back vs. Head, $p = 0.014$; Back/Front vs. Head, $p = 0.015$; Figure 51).

- There was a significant difference for the increase in minimum vertical force ($p = 0.001$, $\eta^2 = 0.469$) and 2nd peak vertical force ($p = 0.003$, $\eta^2 = 0.339$) from unloaded walking between the loading methods (Table 27 and Figure 52). Specifically, head-loading produced a greater minimum force (with pooled load mass, Head = 8.99 ± 0.17 $N{\cdot}kgTM^{-1}$ vs. Back = 8.88 ± 0.17 $N{\cdot}kgTM^{-1}$ vs. Back/Front = 8.83 ± 0.19 $N{\cdot}kgTM^{-1}$) and lower 2nd vertical peak (with pooled load mass, Head = 10.24 ± 0.28 $N{\cdot}kgTM^{-1}$ vs. Back = 10.47 ± 0.23 $N{\cdot}kgTM^{-1}$ vs. Back/Front = 10.43 ± 0.26 $N{\cdot}kgTM^{-1}$) compared to the other methods.
- Head-loading was associated with significantly lower net anteroposterior force (main effect for load method, p = 0.034, $\eta^2 = 0.214$) and horizontal momentum (main effect for load method, p = 0.037, $\eta^2 = 0.244$) during the double stance phase compared to the two trunk loading methods (Table 28).
- There was a significant difference in pain/discomfort scores between the loading methods ($p = 0.007$, $\eta^2 = 0.296$), with lower total pain/discomfort (sum of all body segments) for back/front-loading (49 ± 8 mm) compared to the other methods (Head = 75 ± 12 mm; Back = 73

± 11 mm) with 20 kg. Head-loading also had higher total pain/discomfort scored compared to the other the methods with 3 and 12 kg (Table 34)

Relationships between ELI and walking gait adaptations:

- ELI significantly correlated with the Δ cadence (r = 0.574, p = 0.025), the Δ step length (r = -0.650, p = 0.009), Δ double support time (r = -0.651, p = 0.009), Δ COM velocity in single stance (r = -0.533, p = 0.041) and the Δ trunk angle ROM (r = -0.560, p = 0.030) from unloaded walking with 12 kg carried on the Head (Table 30).
- Figure 53 show that a smaller adjustment from unloaded-loaded walking for these variables appears to be beneficial for Head loading economy with 12 kg.
- ELI for Back/Front carrying 20 kg was significantly correlated with the Δ maximum (r = 0.578, p = 0.024) and the Δ minimum (r = 0.535, p = 0.040) sagittal plane trunk angles from unloaded walking, and the 2nd peak of vertical force (r = -0.600, p = 0.018) from unloaded walking (Table 31). It appears from the relationships and ranking figure (Figure 54) that smaller unloaded-loaded adjustments for upper body movements in the sagittal plane are beneficial for economy Back/Front economy with 20 kg.
- For back-loading, ELI was significantly correlated with the Δ double stance time from unloaded walking with 12 kg (r = 0.648, p = 0.009) and 20 kg (r = 0.644, p = 0.010).

Analysis of inter-individual variation:

- The largest CV for $\dot{V}O_2$ occurred with the 20 kg head-loading condition (17.79%). The largest CV's for the back and back/front methods occurred with 20 kg (16.68%) and 12 kg (13.14%), respectively (Table 36). Inter-individual differences accounted for the largest proportion of the total variance for $\dot{V}O_2$, with ICC values of 0.65, 0.84 and 0.79 for head-, back-, and back/front-loading, respectively.
- The CV's for ELI were larger for the head-loading conditions compared to the other two methods with largest magnitudes of 13.92%, 8.79%

and 8.53% for head-, back-, and back/front-loading, respectively (Table 37). The ICC values for individual differences in ELI as a proportion of the total variance were 0.45, 0.72 and 0.41 for head-, back- and back/front-loading, respectively.

- For back-loading, most participants had their lowest ELI with 9 kg (*n* = 7). For back/front-loading, most participants had their lowest ELI with 20 kg (*n* = 8). For head-loading, most participants had their lowest ELI with 3 kg (*n* = 8) (Figure 55).
- Considering biomechanical variables, the largest range for the percentage change from unloaded walking occurred for double stance time (-15 to +22% for back-loading with 20 kg). SDw was larger than SDb for single stance time with all load mass in the head- and back-loading conditions (SDw = 0.04 s vs. SDb = 0.02 s, for Back 12 kg). The largest difference between SDw and SDb occurred for cadence with 20 kg using the head-loading method (mean = 1.61 $steps \cdot s^{-1}$, SDw = 0.18 $steps \cdot s^{-1}$, SDb = 0.12 $steps \cdot s^{-1}$).

The effect of step width on load carriage economy:

- There was no difference in ELI values between the preferred and controlled step width conditions (p = 0.301, η^2 = 0.088) (Figure 56).
- Preferred step width was significantly larger for head-loading compared to back- (p = 0.013) and back/front-loading (p = 0.015), with a largest difference between methods of 0.01 m with the 20 kg load (Table 38).

7.4. Discussion

The aims of this chapter were:

1. To compare the economy and walking gait adaptations associated with head-, back-, and back/front-loading, and assess relationships between the walking gait adaptations and economy.
2. To investigate the effect of step width control on load carriage economy.

This discussion is split into three parts. The first part is focused on discussing the group data for load carriage economy and walking gait adaptations (section 7.4.1). The second part is focused on the individual variation in economy and walking gait adaptations (section 7.4.2). The final part of this discussion is on the effect of manipulating step width on load carriage economy (section 7.4.3).

7.4.1. Group data for load carriage economy and walking gait adaptations.

This is the first study to assess both the kinematic and kinetic differences associated with head-, back-, and back/front-loading and their relationship to economy. The main findings of the group data were that head-loading was associated with significantly worse economy compared to the two trunk loading methods. With load mass pooled, the ELI values were 1.08 ± 0.12, 0.98 ± 0.07 and 0.97 ± 0.07 for head-, back- and back/front-loading, respectively. There were also significant differences in the walking gait adaptations associated with each method. The most noticeable difference occurred for joint angle kinematics at the trunk and hip joints (Table 25, Table 26 and Figure 49). The significant relationships found between ELI and walking gait adaptations to load carriage show that smaller unloaded-loaded changes in the sagittal plane motion of the trunk is beneficial for back/front-loading economy when carrying 20 kg. Further, smaller unloaded-loaded changes in step patterns, with accompanying increases in double stance time and trunk motion, appear to be beneficial for head-loading economy when carrying 12 kg. However, these relationships were not evident across all method/load combinations.

The significantly higher ELI values for head-loading compared to the two trunk loading methods was unexpected given that no change in ELI was found between these methods in Chapter 5. One of the main methodological differences between this study and the research in Chapter 5 was the difference in participant population. The participants in Chapter 5 were a group of healthy females (*n* = 18) with a minimum of 5 years' head-loading experience, whilst the participants in this study were healthy males (*n* = 10) and females (*n* = 5) with no head-loading experience. Few studies have investigated head-loading economy in individuals with no experience of using the method. Of those that have, Maloiy et al. (1986) found that inexperienced head-loaders had the same economy when carrying a load on the head, on the back or on the back and head combined. They reported that $\dot{V}O_2$ increased in proportion to the mass of the additional load for all methods (ELI = 1.00). The study by Maloiy et al. (1986) was underpowered with only three inexperienced participants, but their findings are consistent with earlier studies that reported data on head-loading economy for inexperienced individuals. Soule and Goldman (1969) found a proportional increase in $\dot{V}O_2$ in respect to the load mass carried on the head for a group of ten inexperienced head-loaders. Datta and Ramanathan (1971) found no significant difference in $\dot{V}O_2$ between head-loading (both direct and indirect methods) and back-loading for seven inexperienced head-loaders. In agreement with these studies, Lloyd et al. (2010b) found no difference in $\dot{V}O_2$ between head- and back-loading with 10-30% body mass (10% and 30% represented 6.74 kg and 20.22 kg for the average body mass, respectively) in nine female participants from the British Territorial Army with no head-loading experience. Based on this existing evidence, the significantly larger ELI values found for head-loading compared to back- and back/front-loading in the present study are unlikely to be a consequence of load carriage experience. Interestingly, the ELI values calculated from the data provided by Lloyd et al. (2010b) for women from the British Territorial Army were 1.07, 1.10 and 1.15 for 10% (~ 6.7 kg), 20% (~ 13.5 kg) and 30% (~ 20.2 kg) of body mass, respectively. These values are similar to the ELI values in the present study (1.06 ± 0.09, 1.07 ± 0.11 and

1.10 ± 0.15, for 3, 12 and 20 kg, respectively) and show the same pattern of response, with worsening economy as the mass of the load increased.

It is possible that the poor head-loading economy in the present study was, in part, a result of some individuals being very uneconomical when head-loading, with five participants having ELI values of >1.20, yet no participants had ELI values of that magnitude with the back- or back/front-loading methods. There was also a general tendency for most individuals to be less economical when head-loading, with ten participants having an ELI >1.10 in one or more of the head-loading conditions. In the back- and back/front loading methods, only two participants had an ELI >1.10. Inter-individual variation in ELI was also larger for head-loading (10% and 14% for 12 and 20 kg, respectively) compared to back-loading (7% and 9% for 12 and 20 kg, respectively) and back/front-loading (8% and 9% for 12 and 20 kg, respectively). Notably, the three least economical participants with the head-loading method were in the top five for highest ratings of perceived pain/discomfort in the neck region and the least economical head-loader also had the highest pain scores for the front and back of the shoulders with the head method. It's possible that an increased discomfort with the head-loading method could be linked to the higher energy expenditure with this method.

The physiological and biomechanical data from the male and female participants in the present study were combined and analysed together. On average, male participants were heavier and taller than the female participants. However, there was no interaction effect between sex and load carriage method or sex and load mass for the ELI data. This supports the findings of Lloyd et al. (2010a), who reported that the ELI is independent of body mass and stature. The lack of difference in relative load carriage economy between males and females in the present study is also in agreement with the recent work of Prado-Nóvoa et al. (2019), who found no difference in load carriage economy between males and females for load mass relative to body mass. Further, Godhe et al. (2020), provided evidence that the dominant factor in the $\dot{V}O_2$ required to carry heavy load (≥ 20 kg) is body mass, not sex difference, and Vickery-Howe et al. (2020) have found no difference between sexes for $\dot{V}O_2$ relative to body mass when carrying 20% and 40% body mass. Similar walking gait adaptations to load carried carriage have also been reported for males and females with load (22 kg) carried on the back (Krupenevich et al., 2015) and load (10 – 40% body mass) evenly distributed around the torso (Silder et al., 2013, Vickery-Howe et al., 2020). While none of these studies directly investigated male and female responses to head-loading, evidence from trunk-loading studies consistently shown that body mass, not sex specific characteristics, is responsible for differing responses between males and females to load carriage with absolute load masses. As such, the combined use of males and females in the present study is unlikely to have influenced the different findings for head-loading economy between this study and the study in chapter 5.

In agreement with the findings of Chapter 5, the relationship between body mass and ELI was not strong in the present study, which provides further evidence that body mass alone does not determine an individual's economy when carrying an absolute load. Furthermore, Lloyd et al. (2010a) has shown that the ELI is independent of body composition and, as such, it is unlikely that muscle and fat mass explain the difference in head-loading economy between Chapters 5 and 7. However, body composition was not assessed directly in the present study. The present study also did not account for the participants prior physical condition, which Wills et al. (2019) demonstrated can effect load carriage task performance when wearing a weighted vest. As such, it is possible that the physical condition of the participants could account for some of the difference in head-loading economy between the two studies. However, head-loading economy does not appear to be determined by any physical conditioning that occurs through regularly carrying load on the head (Lloyd et al., 2010b, Lloyd et al., 2010c).

As with the economy data, there were also contrasting findings for trunk angle kinematics between the present study and the results from Chapter 5. Specifically, there was a noticeable difference in trunk angle extension with the head-loading method between studies. For the 20 kg condition, the Δ trunk angle from unloaded walking at heel strike was -7.6 ± 3.4° in the present study and -2.4 ± 3.4° for the study in Chapter 5. Based on this finding, it appears that the group of inexperienced head-loaders in the present study exhibited a greater magnitude of trunk extension when head-loading compared to the group of experienced head-loaders that took part in the research in Chapter 5. The difference in measurement techniques between studies could account for some of the variance in trunk angle. However, Schurr et al. (2017) reported good agreement for joint displacement between 2D and 3D in the sagittal plane, with a mean difference of 1.68° for the trunk. To date, no studies have investigated the effect of load carriage experience on walking gait kinematics and it appears, based on a comparison of the findings in the present and those of Chapter 5, that experienced individuals have a posture closer to that of unloaded walking when head-loading compared to those with no experience. It is possible that this could partly account for the difference in head-loading economy between this study and the work in Chapter 5. However, based on the results of Chapter 5, it is unlikely that the differences in sagittal plane trunk movements alone explain the difference in head-loading economy found between the studies in this thesis. This is supported by the lack of moderate or strong relationships between trunk flexion/extension and head-loading economy in the present study.

The magnitude of trunk forward lean for back-loading was also smaller in the present study (6.36 ± 2.82° at heel-strike with 20 kg) compared to the results from Chapter 5 (14.2 ± 3.8° at heel-strike with 20 kg). The different backpack systems used between the studies is unlikely to account for the 8° difference between studies, as this is a magnitude of forward lean that would be expected between different load carriage methods or load masses. For example, Kinoshita (1985) and Lloyd and Cooke (2011) reported differences in forward lean of 7° and 9°, respectively, between back- and back/front-loading methods. The difference in forward lean for the back-loading method between the studies in this thesis might be partly explained by the difference in physical characteristics between the participant groups. In the present study, the 20 kg load was equivalent to, on average, 27.7% body mass. For the study in Chapter 5, the same load was equivalent to 33.8% body mass, on average. Wood and Orloff (2007) reported 10° of forward lean whilst carrying 15% body mass on the back. It’s also possible that the portable gas analysis system, positioned on the anterior of the trunk, could have influenced the reduced forward lean found for back-loading in the present study, although the total mass of the was only 1 kg (including the housing vest).

This study was the first to assess three-dimensional kinematics for head-loading. As such, the reduction in trunk axial rotation for head-loading compared to the two trunk loading methods is a novel finding and is likely to be a consequence of the need to balance the load on the head. This decrease in trunk rotation for head-loading was accompanied by increased pelvic rotation compared to the two trunk loading methods. LaFiandra et al. (2003b) has previously suggested that the coordination between the relative rotations of the torso and pelvis combine to reduce the net angular momentum of the body. This would explain the pattern of response observed between the axial rotation of the pelvis and trunk in the present study. LaFiandra et al. (2003b) also suggested that a key factor in decreased stride length during load carriage is a decrease in pelvic rotation, with an increase in step cadence to compensate when walking at a set speed. In line with this, head-loading was associated with a decrease in step length and single stance time, and an increase in cadence compared to the other methods.

The shorter step length and increased cadence found for head-loading in the present study again differ to those from Chapter 5, which showed little change in step length (0.00 ± 0.02 m with load mass pooled) and cadence (0.00 ± 0.02 steps·sec^{-1} with load mass pooled) from unloaded walking for the head-loading method. The difference in gait event detection methods between the studies could account for some of the difference in spatiotemporal variables between these studies. The vertical ground reaction force method used in the present study is considered as a gold standard measure (Zeni Jr et al., 2008) and is likely to have been more accurate than the technique of visually inspecting video files used in Chapter 5, although no data exists for the level of agreement between these methods. Maloiy et al. (1986) suggested that stride frequency is not altered by head-loading for either trained (n = 5) or untrained head-loaders (n = 3) at a walking speed of 3 km·h^{-1}. However, the present study contradicts the evidence of Maloiy et al. (1986) by showing that head-loading alters cadence from unloaded walking for inexperienced head-loaders. Individual changes in cadence from unloaded ranged from +10% to -6% when head-loading, with five participants having a decreased cadence and ten participants increasing their cadence. As such, it's possible that the results from Maloiy et al. (1986), could have been influenced by individual responses to head-loading. Furthermore, the increase in cadence and decrease in step length found for head-loading compared to unloaded walking in the present study is consistent with the stride patterns observed for head-loading while walking over ground (Charteris, 1985, Charteris et al., 1986).

Despite head-loading being more uneconomical than the other two load carriage methods, a larger Δ step length and a smaller Δ cadence from unloaded walking significantly correlated with improved load carriage economy for head-loading with 12 kg (ELI and Δ cadence, r = 0.574; ELI and Δ step length, r = -0.650). Cooke et al. (1991) suggested that shorter stride lengths may improve economy and stability with vertical loading through reduced vertical oscillations of both the COM and the added load While an increased cadence and decreased step length combination might improve stability and reduce vertical COM oscillations, an increased cadence might also increase the metabolic cost of walking, as the motion of swinging the leg has been estimated to account for ~30% of the overall energy cost of locomotion (Doke et al., 2005, Umberger, 2010). As such, it's possible that there could be an optimal step length and cadence combination that allows for an overall improvement in energy cost through reducing the vertical oscillations of the COM without substantially increasing cadence, and the associated metabolic cost of swinging the legs. Figure 53 shows that the most economical participants for the head-loading 12 kg condition tended to have a step length and cadence closer to that of unloaded walking compared to less economical participants. This would suggest that the optimal step length/cadence combination to reduce energy cost might be close to that of unloaded walking. This adds support to the theory that the unloaded walking gait might provide an optimal strategy for an individual when carrying load. Research on unloaded walking has established that the stride length-stride frequency combination freely chosen for a given speed is close to optimal in terms of economy and that acute perturbations in stride length/stride frequency result in an increase in $\dot{V}O_2$ (Högberg, 1952, Cotes and Meade, 1960, Knuttgen, 1961, Cavanagh and Williams, 1982).

An improved economy for head-loading with 12 kg was also associated with a greater in trunk ROM from unloaded walking. The role of trunk movement for head-loading economy has not been previously reported. It's possible that an increased freedom of movement in trunk for head-loading with a moderate load of 12 kg would provide the same mechanism for improved load carriage economy as that suggested for the back-loading method (Abe et al., 2004, Lloyd and Cooke, 2011). An increase in double support time was also associated with improved economy (r = -0.651; Figure 53) for head-loading with 12 kg. This suggests a possible contribution of improved stability when head-loading on improved head-loading economy, as an increased double stance time shows that more economical individuals spent longer periods of each stance phase with both feet in contact with the ground. Based on the deterministic model, the increased double stance time associated with improved economy was determined by an increased velocity of the body's COM in the single stance phase, which was also significantly related to improved economy (r = -0.533).

The small improvement in relative load carriage economy found for back/front-loading (ELI = 0.94 ± 0.08) compared to back-loading (ELI = 0.98 ± 0.09) with the heaviest load (20 kg) is consistent with the previous work in this thesis and the work of others (Lloyd and Cooke, 2000b). The improved relative load carriage economy for back/front-loading with increased load mass in the present study (ELI of 1.01 ± 0.06 to 0.94 ± 0.08 for 3 to 20 kg) is also consistent with the findings of Chapter 5 (ELI of 0.99 ± 0.06 to 0.92 ± 0.09 for 3 to 20 kg). Furthermore, most individuals had their best relative economy for the back/front-loading method with 20 kg (8 out of 15) compared to the other load masses. It has been suggested that a greater freedom of movement of the trunk, with load evenly distributed around the trunk compared to load carried on the on the back alone, could be responsible for this improved economy with heavy load (Lloyd and Cooke, 2011). In comparison of the sagittal kinematics between methods in the present study, Figure 49 shows that both trunk and hip motions where closer to that of unloaded walking for back/front-loading with 20 kg, compared to back-loading and head-loading. In support of this, improved load carriage economy for back/front-loading with 20 kg was associated with a smaller change in peak trunk flexion (r = 0.578), peak trunk extension (r = 0.535) and trunk angle at heel strike (r = 0.535) and toe off (r = 0.535) in the present study. This provides support for the theory that smaller unloaded-loaded walking gait adaptations might be beneficial for improved load carriage economy, particularly for movements in the upper body where walking gait perturbations appear largest for methods that place the load close to, or in direct alignment, with the body COM.

Back-loading was associated with a 30% increase in trunk angle ROM over the gait cycle compared to unloaded walking and produced the largest trunk ROM of the methods assessed in this study (Table 25). Given that an increased freedom of movement of the trunk has been suggested to improve load carriage economy, it might be expected that a larger trunk ROM would be beneficial for economy with this method. However, ELI remained unchanged across the load mass for back-loading (0.98 ± 0.06, 0.98 ± 0.07 and 0.98 ± 0.09 for 3, 12 and 20 kg, respectively). A such, it appears that and increased movement in the upper body, above that associated with unloaded walking, when carrying load over the stride does not lead to improved economy. It’s possible that increasing rotational movements in the upper body with heavy loads could have a negative impact on load carriage economy by requiring an increase in muscular effort to counteract the movements and redirect the COM between steps (LaFiandra et al., 2002). A greater number of individuals had their best relative economy with 12 kg (n = 7) in the back-loading method compared to the other load masses. Abe et al. (2004) reported an energy saving phenomenon for load mass of 9 -12 kg (10-15% of body mass) carried on the back compared to lighter and heavier loads. They suggested that a possible explanation could be that this moderate load might add to the momentum of body through the stance phase. Theoretically, an increase in momentum would reduce the propulsive force that the muscles need to generate for a given walking speed, reducing the metabolic energy cost. However, this energy saving phenomenon was not found in the group data in the present study. Furthermore, the present study did not find any significant and/or strong relationships between improved load carriage economy and propulsive force or linear momentum during double and single stance.

The deterministic model for walking speed identified anteroposterior ground reaction forces as a key determinant of stance phase durations to achieve a given speed of walking. The increased magnitude of peak anteroposterior forces found in the present study as a result of increased additional load is consistent with previous load carriage research (Kinoshita, 1985, Harman et al., 2000, Lloyd et al., 2011, Huang and Kuo, 2014). The tendency for peak propulsive force to be greater than peak braking force in this study is also consistent with previous literature on back- and back/front-loading methods (Harman et al., 2000, Lloyd et al., 2011) and has been suggested to be a protective mechanism to reduce potentially harmful impact peaks with additional load (Harman et al., 2000). The present study found no significant difference between load carriage methods for peak anteroposterior forces and mean anteroposterior forces (during double and single stance). There was a tendency for the back-loading method to be associated with higher propulsive force compared to the head and back/front method (Table 27), with an increase in propulsive force from 0 - 20 kg of 32% for back-loading compared to 26% and 28% for the head and back/front methods. However, when force was normalised to total mass, there was no tendency for back-loading to be associated with greater propulsive force compared to the other methods. Lloyd and Cooke (2000a) have previously found that peak propulsive force increased by 68% and 40% for back- and back/front-loading, respectively, with 25.6 kg compared to unloaded walking. They speculated that a greater freedom of movement of the trunk for back/front-loading might be associated with greater momentum which would contribute to a lower peak propulsive force. However, this in not supported by the momentum data in the present study, which did not find a greater horizontal momentum of the body's COM for back-loading compared to the other methods.

There is a paucity of research on head-loading kinetics and this is only the second study to compare the kinetics associated with head-loading and other load carriage methods. The first authors to do so were Lloyd et al. (2011) who compared the kinetics associated with head-loading and back-loading. In line with the findings of the present study, they also reported no difference in peak anteroposterior forces between head-loading and back-loading when forces were normalised to total mass. Analysis of the mean anteroposterior forces and horizontal momentum over the double stance phase shows that there was a lower net anteroposterior force and whole-body horizontal momentum from unloaded walking during double stance for head-loading compared to the other methods back/front-loading compared to head-loading. Based on the deterministic model, the reduced momentum for head-loading during double stance is likely to explain the significantly reduced single stance time also found for head-loading compared to the back- and back/front- methods.

The strongest significant relationship found between load carriage economy and the kinetic variables was between ELI and the Δ 2nd peak of vertical force (r = -603) in the back/front method with 20 kg. This shows that an improved load carriage economy was associated with a larger Δ 2nd peak of vertical force from unloaded walking for this loading condition. Birrell and Haslam (2010) reported a significantly smaller 2nd peak of vertical force with load placed on the back compared to load more evenly distributed around the trunk. The authors suggested that the smaller vertical force at toe-off with a backpack was likely to be a result of the increased forward lean associated with this method. However, this is not a consistent finding in the literature with Kinoshita (1985) and Lloyd and Cooke (2000a) both finding no difference in the 2nd peak vertical force between back- and back/front-loading. Furthermore, the findings of the present study show that the smallest vertical force at toe-off occurred for head-loading, which suggests that changes in this kinetic variable are not explained by forward lean. As such, it is unclear why a larger Δ 2nd peak of vertical force was associated with improved economy for back/front-loading with 20 kg in this study. The finding is likely, in part, to be a result of a smaller Δ 2nd peak of vertical force for participants 15 and 9, who were uneconomical in this loading condition, although it's unclear why these participants had a smaller Δ 2nd peak of vertical force compared to others (Figure 54). A smaller Δ peak medial force from unloaded walking was also associated with improved economy for back-loading with 12 kg (r = 531). The magnitude of medial forces has been linked to stability, with increased medial forces indicating less stability when carrying load (Birrell et al., 2007). As such, it's possible that improved economy for back-loading with 12 kg was linked to better stability, indicated by smaller peak medial force from unloaded walking. However, mediolateral forces are associated with large variability (Lloyd et al., 2011) and a rarely reported in the load carriage literature. Indeed, the SDb and SDw for mediolateral forces in the present study showed that intra-individual variation in lateral force was often larger than the magnitude of inter-induvial variation for head and back-loading.

7.4.2. Individual variation in load carriage economy and walking gait biomechanics

The level of variation in relative load carriage economy was large with all loading methods. While the group data showed that, on average, head-loading was the least economical method of load carriage (ELI = 1.06 ± 0.09, 1.07 ± 0.11 and 1.10 ± 0.15 for 3, 12 and 20 kg, respectively), the range of ELI values for head-loading were 0.86 – 1.15 for 3 kg, 0.88 – 1.21 for 12 kg and 0.80 – 1.29 for 20 kg. This shows that, although most individuals were uneconomical head-loaders, some were very economical. A similar pattern of variability was also present for the other methods, with the largest range of ELI values occurring with the 20 kg load for back-loading (0.86 - 1.15) and back/front-loading (0.82 – 1.16) with 20 kg. The largest CV's for $\dot{V}O_2$ and ELI also occurred with the heaviest load in each method. The CV for these variables increased with all methods as the mass of the load increased (Table 37). The highest CV for ELI was 14%, 9% and 9% for head-, back-, back/front- loading, respectively, which all occurred with the 20 kg load. This supports the findings of Chapter 5 and provides further support for the individual variation in load carriage economy observed by Lloyd et al. (2010c).

The increased variation in economy with increasing load mass might be explained by the individual variation in walking gait perturbations from unloaded walking (identified via standard deviations), which also increased with the increase in load mass in the present study. Furthermore, the range of individual responses in spatiotemporal and joint angle variables also increased as the mass of the load increased. The largest differences occurred for head-loading with 20 kg, with a range of +7 to -13%, +10 to -6% and -3% to -18% for the change in step length, cadence and trunk angle from unloaded walking, respectively. Future research might benefit from using heavier loads than those used in the present study, in order to produce larger variance between participants for both economy and walking gait adaptations. Large variance might help to elucidate the key determinants for improved economy with different load carriage methods. Indeed, the use of inexperienced head-loaders in the present study might have resulted in the larger changes in trunk motion and spatiotemporal variables from unloaded walking in this study compared to the research in Chapter 5. These larger perturbations, in turn, might have contributed to the significant relationships found between improved economy and specific spatiotemporal variables for head-loading with 12 kg.

The lack of significant relationships between ELI values and body mass, stature or BMI indicate that individual differences in physical characteristics were not related to the individual differences in load carriage economy. The ICC's indicate that the individual differences in ELI as a proportion of the total variance were 0.45, 0.72 and 0.41 head-, back- and back/front-loading, respectively. As such, the variability in ELI values for each method due to the different load mass was 0.55, 0.28 and 0.59 for head-, back- and back/front-loading, respectively. The smaller variance in ELI for back-loading attributable to the load mass compared to the other methods is unsurprising given the consistency of the mean ELI values for back-loading from 3 – 20 kg. Similar to the research in Chapter 5, the variance between individuals represented the largest proportion of the total variance in the $\dot{V}O_2$ data (ICC = 0.65, 0.84 and 0.79 for head-, back-, and back/front-loading, respectively). The higher proportion of variance assigned to individual differences in $\dot{V}O_2$ compared to

ELI is likely to be a result of inter-individual differences in body mass (CV of 13.78%).

Much of load carriage research has focused on discrete measures at specific events in the gait cycle (e.g. heel strike and toe off) to assess the spatial-temporal characteristics, joint kinematics and ground reaction forces associated with loaded walking (e.g. Liew et al., 2016). Discrete measures at specific points in the gait have enabled the identification of different gait adaptations between different methods of load carriage and different load mass carried in the same load carriage system (e.g. Kinoshita, 1985, Lloyd and Cooke, 2000a, Lloyd et al., 2011). For example, it appears that carrying heavy loads (≥ 20kg) on the back increases forward lean (decreased trunk angle) at heel-strike and toe-off events compared to carrying the same load evenly distributed between the anterior and posterior of the trunk (Lloyd and Cooke, 2011) or carried on the head. Indeed, this study showed significant differences in spatiotemporal characteristics (Table 24), joint kinematics (Table 25) and ground reaction forces (Table 27) between load carriage systems and load mass. However, the findings of this study, along with the finding of Chapter 5, demonstrate that the biomechanical adaptations that clearly distinguish individual load carriage economy with a particular method appear less consistent. Using the deterministic model developed in Chapter 6 as a framework to identify the key determinants of load carriage economy has not provided a clear set of adaptations to load carriage that align with a better or worse load carriage economy for each load method and mass combination. It is possible that focusing on measurements at specific events in the walking gait does not effectively capture the complexity of the coordinated motion of the body during loaded walking and as a result some of the individual differences in gait adaptations to load carriage that lead to individual difference in economy could have been omitted from the analysis. In line with this possibility, this study found that step parameters and medio-lateral ground reactions forces, particularly step length, single stance time and lateral force, were subject to within participant variability, assessed via standard deviations, that was close to or greater than the between participant variability (Appendix P). This level of within participant step-to-step variability might have influenced the individual differences found in load carriage economy, particularly as increased step variability has been linked to an increased energetic cost of up to 9% when walking unloaded walking (O'Connor et al., 2012).

Techniques to measure coordination variability, such as dynamical systems theory (Hamill et al., 1999), have been used to detect skill-dependant changes in movement execution (Bartlett et al., 2007, Wilson et al., 2008, Preatoni et al., 2010) and could be useful in future loaded walking analysis to assess coordinative synergies between elements of the walking gait that are key to performance. The use of techniques such as dynamical systems theory in future load carriage research might elicit whether individual differences in movement variability in response to load results in the individual variation in load carriage that have been found in this thesis and reported in some of the literature (Lloyd and Cooke, 2011, Lloyd et al., 2010c). The number of gait cycles analysed in the present study are probably not appropriate to accurately assess kinematic variability using techniques such as dynamical systems theory, with a 400-step minimum being identified as suitable for such assessments by Owings and Grabiner (2003). The aim of this study was not to assess within participant gait variability, however future studies on the variability of spatial and temporal step kinematics with different load carriage conditions and the associated economy appear to be warranted.

7.4.3. The effect of step width control on load carriage economy

Load carriage has been suggested to place an increased demand on balance, indicated by an increase in step width variability (Huang and Kuo, 2014). O'Connor et al. (2012) suggested that variability in walking gait step width and step length could increase the energy expenditure in unloaded walking gaits. Increasing step width above and below an individual's preferred step width has also been shown to increase the metabolic cost of unloaded walking (Donelan et al., 2001). As such, it was hypothesised that individual differences in step width from the preferred unloaded width, as a consequence of carrying additional load, could align to individual differences in load carriage economy. However, this study found no difference in relative load carriage economy between the preferred and controlled step width conditions.

The lack of significant difference in ELI between the two step width conditions (preferred and controlled) suggests that alterations in step width and medial-lateral stability due to load carriage does not solely explain differences in load carriage economy between head-, back- and back/front-loading. In the uncontrolled condition, step width was wider for head-loading with all load mass compared to the other methods. An increase in step width has been associated with an increased requirement for medio-lateral stability (Young and Dingwell, 2012), which suggests that individuals required wider steps to maintain stability when head-loading. However, the pattern of response for ELI was similar in both step width conditions, which suggests that the alterations in step width when head-loading were not solely responsible for the increased ELI values.

Step width was controlled to each participant's preferred width when walking unloaded because, for unloaded walking, both widening and narrowing step width from an individual's preferred width appears to increase the energy cost of walking (Donelan et al., 2001, Shorter et al., 2017). Furthermore, it has been suggested that an individual's normal walking gait may represent an optimal solution for that individual in relation to their economy (Martin and Morgan, 1992). As such, it was hypothesised that larger adjustments from the preferred unloaded walking gait, as a consequence of load carriage, could worsen economy. It's likely that the small difference in step width between the controlled and preferred conditions seen in this study were not large enough to influence economy. Donelan et al. (2001) showed a 45% increase in metabolic cost for unloaded walking when step width was increased from preferred (0.14 m) to 0.42 m. The largest difference found in this study between controlled and uncontrolled step width was 0.01 metres. As such, it's likely that when walking on even terrain, the alterations in step width induced by load carriage are not large enough to cause an alteration in relative load carriage economy. This is also likely to explain for the lack difference in medial and lateral ground reaction force between the conditions.

The method used to control step width in this study relied on the successful alignment of heel-markers with markers at the rear of the treadmill. While most participants were able to consistently align their heels with the markers, two participants produced narrower step widths in the back 12 kg condition (0.03 metres from the desired width in the controlled condition). As such, the average step width for this load carriage condition was lower than other conditions. Similar real-time visual feedback methods have also reported slightly narrower than target step widths when running (Arellano and Kram, 2011).

7.5. Conclusion

This study found that head-loading was less economical, for a group of inexperienced head-loaders, compared to back- and back/front-loading methods. There was, however, large individual variation for the biomechanical responses and load carriage economy associated with head-, back-, and back/front- loading methods.

Using the deterministic model described in Chapter 6 as a framework to analyse load carriage economy, this study found that smaller unloaded to loaded walking adjustments in step length and cadence, along with increased sagittal trunk range of motion and double stance time, were beneficial for participants with improved economy when carrying a moderate load on the head (12 kg). This study also found that smaller unloaded to loaded walking adjustments in trunk and hips from unloaded walking are beneficial for an improved economy when carrying 20 kg using the Back/Front method.

Load carriage induced alterations in step width from unloaded walking are not large enough to influence load carriage economy when walking on even terrain and, as such, step width does not solely explain individual differences in the economy associated with head-, back-, or back/front-loading.

This study was unable to find a common strategy of walking gait alterations from unloaded to loaded walking that can account for improved economy or worse economy for each load carriage condition. While the deterministic model was effective for comparing and understanding the walking gait adaptations that occur for different load carriage methods, this approach has not been able to fully explain the individual determinants of load carriage economy for each load method and load mass combination in this study. It's possible that only considering the unloaded to loaded adaptations of the body to load carriage with discrete measures has limitations for fully explaining the determinants of load carriage economy. Given the individual variation found in load carriage economy and unloaded to loaded walking gait adaptations, future research might benefit from assessing the role of individual differences in movement variability and coordinative synergies between elements of the walking gait in response to load, to see if there is a link to energetic cost. Furthermore, it might also be useful for future load carriage research to assess the motions of the body and load carriage device separately, as well as a combined system, to understand how different gait adaptations might alter the load carriage devices motion and the link this could have to improved load carriage economy.

Chapter 8. General discussion

8.1. Introduction

The aim of this thesis was to investigate the determinants of individual load carriage economy, with a focus on walking gait biomechanics and load carriage methods that position the load close to the COM of the body, or in the case of head-loading, in vertical alignment with the body's COM. To achieve this aim, the objectives of this thesis were: (i) To assess the suitability of the ELI as a measure of relative load carriage economy; (ii) To establish the extent of individual variation in load carriage economy and walking gait alterations as a consequence of load carriage; (iii) To identify factor(s) that could determine an individual's load carriage economy; (iv) Demonstrate cause and effect by manipulating the identified potential determinant(s) to influence load carriage economy. To achieve the aim and objectives, a total of three studies were conducted, along with the creation of a theoretical deterministic model. This chapter provides a synthesis and interpretation of the research presented in this thesis, as well as discussion of the limitations of the thesis and directions for future research.

8.2. Summary of main findings and original contributions to the research area

The review of literature in Chapter 2 identified the ELI as the most suitable method for measuring load carriage economy. The advantage of using the ELI over other widely used approaches, such as assessing the rate of oxygen consumption (e.g. Legg and Mahanty, 1985, Quesada et al., 2000) or the energy cost of walking (e.g. Abe et al., 2004, Bastien et al., 2005), is that the ELI accounts for $\dot{V}O_2$ when walking unloaded and provides a single unitless value. The ELI has been shown to be a valid measure (Lloyd et al., 2010a) but the research in Chapter 4 was the first study to investigate its reliability. Overall, the ELI demonstrated good test-retest reliability for 7 kg and 20 kg carried on the back at 3 $km{\cdot}h^{-1}$, 6 $km{\cdot}h^{-1}$ and a self-selected walking speed (4.4 ± 0.7 $km{\cdot}h^{-1}$). The systematic bias, LoA, CV and SEM were small in all trial conditions with the largest LoA (± 0.11), highest CV (4.17%) and SEM (0.04) recorded when walking at 3 $km \cdot h^{-1}$ with the 7 kg load. As such, a novel finding of this study was that the ELI is a reliable measure of relative load carriage economy for light (7 kg) and heavy (20 kg) loads carried on the back at a range of walking speeds (3 $km \cdot h^{-1}$ – 6 $km \cdot h^{-1}$). From a practical perspective, evidence of the ELI's reliability could be useful for researchers and developers of load carriage systems, particularly given the ELI's utility in allowing for simple comparisons of economy between different load carriage systems and study designs.

The research in Chapter 5 was designed to assess potential determinants of the energy saving phenomena previously reported for loads carried on the head (Maloiy et al., 1986), back (Abe et al., 2004) and evenly distributed between the back and front of the torso (Lloyd and Cooke, 2000b) when walking at a speed of 3 $km \cdot h^{-1}$. The findings presented in Chapter 5 provide novel evidence that sagittal plane trunk movements, previously postulated to be a determinant of load carriage economy for back- and back/front-loading (Lloyd and Cooke, 2000b, Abe et al., 2004), do not solely explain load carriage economy. Despite finding no difference in economy between load carriage methods, this research did show a considerable degree of inter-individual variation for both ELI and sagittal plane kinematics and is the first study to report this variation for head-loading compared to back- and back/front-loading methods. A novel finding from the measures of inter-individual variation is that increasing the mass of the load appears to increase the magnitude of inter-individual variation for economy, which is likely to be caused by the increase in inter-individual variability in gait perturbations with increasing load mass. Assessing individual differences highlighted that the majority of participants were most economical when carrying 9 kg (7 out of 18 participants) for the back-loading condition. This supports the work of Abe et al. (2004), who reported that a load of 9 kg carried on the back yielded a better economy compared to lighter or heavier loads. In the back/front-loading condition, the majority of participant's were most economical with 20 kg (10 out of 18 participants). This finding provides further evidence to support studies that have found back/front-loading to be more economical than back-loading when carrying heavier loads (Datta et al., 1973, Legg and Mahanty, 1985, Lloyd and Cooke, 2000b).

The development of the deterministic model in Chapter 6 provides a theoretical understanding of the biomechanical factors that interact when walking at a given speed. The model provides a framework that can be used by researchers and developers to assess walking gait perturbations produced by load carriage compared to unloaded walking. Step width, pelvic rotation, antero-posterior ground reaction force and horizontal momentum were all identified, using the model, as factors not measured in Chapter 5 that could be important factors in determining individual load carriage economy.

The research in Chapter 7 used the deterministic model developed in Chapter 6 as a framework to assess walking gait perturbations in response to head-, back- and back/front-loading at a walking speed of 3 $km \cdot h^{-1}$. Relationships were assessed between ELI and gait perturbations from unloaded walking for each load carriage method to try and identify determinants of the energy saving phenomenon previously reported for these methods of load carriage (Maloiy et al., 1986, Abe et al., 2004, Lloyd and Cooke, 2000b). The study provided evidence that reduced gait perturbations from unloaded walking appear to be beneficial for improving economy, Specifically, improved relative load carriage economy when carrying a moderate load (12 kg) on the head was significantly related to a smaller change in step length (r = -0.650, p = 0.009) and cadence (r = 0.574, p = 0.025) from those associated with unloaded walking, along with a larger change in sagittal plane trunk range of motion (r = -0.560, p = 0.030), double stance time (r = -0.651, p = 0.009) and mean COM velocity in single stance (r = -0.533, p = 0.041) from unloaded walking. This study also provided evidence that reduced alterations in trunk movement from unloaded walking appear to be beneficial for an improved economy for back/front-loading with heavier load mass (20 kg). For this load carriage condition, smaller changes in peak trunk extension (r = 0.535, p = 0.040), peak trunk flexion (r = 0.578, p = 0.024), trunk angle at heel strike (r = 0.593, p = 0.020) and trunk angle at toe off (r = 0.647, p = 0.009) from those associated with unloaded walking were all related to improved load carriage economy.

The research in Chapter 7 also showed that the changes in step width associated with different load carriage methods at a range of loads do not provoke changes in step width large enough, on their own, to be an explanatory factor for load carriage economy.

8.2.1. Summary of the original contributions to knowledge and understanding of load carriage

- The research in this thesis provides novel evidence that the ELI is a reliable measure of relative load carriage economy with light (7 kg) and heavy (20 kg) loads at a range of walking speeds (3 $km{\cdot}h^{-1}$ – 6$km{\cdot}h^{-1}$). As such, the ELI can be used with confidence by researchers, developers and manufacturers as a useful tool for comparing the relative economy of different load carriage systems.

- The research in this thesis is the first to compare the magnitude of inter-individual variability in load carriage economy and walking gait adaptations for head-, back- and back/front-loading across a range of load masses. A larger level of inter-individual variation for ELI values was found for a group of experienced head-loaders when carrying (largest CV = 16.%, 12% and 10% for head-, back-, and back/front-loading, respectively) compared to a group of inexperienced head-loaders (largest CV= 14%, 9% and 9% for head-, back-, and back/front-loading, respectively), however this should be interpreted with caution as the data was collected in separate studies. The large inter-individual variation found in two of the experimental studies builds on the findings of Lloyd et al. (2010c) and Lloyd and Cooke (2011), who first showed the existence of large inter-individual variation for economy and step parameters with a heavy load (25.6 kg) carried on the back and evenly distributed around the torso (back/front). The evidence presented in this thesis also provides evidence to show that the magnitude of this inter-individual variation in economy in relation to gait perturbations increases as the mass of the load increases.

- Novel evidence shows that load carriage economy is not solely determined by an increase in the freedom of movement in the trunk in the sagittal plane, which has previously been postulated to be a prominent factor in determining load carriage economy for back- and back/front-loading conditions. This novel evidence was extended by showing that individually self-selected reduced perturbations from unloaded walking for sagittal plane trunk motion are associated with an improved economy when carrying heavy load (20 kg) using the back/front method. Further, the final study provides novel evidence that an improved head-loading economy with a moderate load (12 kg), for a group in inexperienced head-loaders, is associated with self-selected reduced perturbations in step length and cadence from unloaded walking, along with increased double stance time and range of motion of the trunk in the sagittal plane.

- A deterministic model for walking speed has been developed that can be used as a framework to assess changes from unloaded to loaded walking gait for a given walking speed. The model can be used to systematically analyse the biomechanics of load carriage, which should be useful to researchers and developers in further exploring the walking gait adaptations to different load carriage systems and designs.

- The experimental work from the final study also provides novel evidence that self-selected step width adjustments to head-, back- and back/front-loading are not large enough to effect load carriage economy in a general and predictable way and as such, the difference in economy between methods is not solely determined by step width.

8.3. Optimal walking gait adaptations for improved load carriage economy.

The research in this thesis provides evidence that smaller perturbations from unloaded walking for sagittal plane trunk and hip movements when carrying external load might be beneficial for improved load carriage economy. This is particularly evident when carrying a heavy load (20 kg) using the back/front method, for which an improved load carriage economy was associated with smaller perturbations in trunk and hip movements from unloaded walking. This finding builds on the work of Lloyd and Cooke (2011), who found that reduced forward lean for back/front-loading compared to back-loading with 25.6 kg was correlated significantly with improved economy. Abe et al. (2004) showed that carrying a light load on the back (10 - 15% body mass) at slow walking speeds (2.4 - 3.6 $km \cdot h^{-1}$) was more economical than carrying heavier loads and/or walking at a faster pace. They suggested that an increase in rotative torque of the trunk in the sagittal plane with light loads compared to heavy loads was likely to be the mechanism responsible, as heavy load on the back constrains posture in a position of greater forward lean. The findings from this thesis show that an increase in trunk movement (i.e. increased rotative torque) is not generally beneficial for load carriage economy, as an increased trunk range of motion for back-loading in the final study, above that associated with unloaded walking, was not related with improved economy. It's possible that an increase in rotative torque, above that associated with unloaded walking, could lead to an increase in the muscular effort required to counteract the torque in the upper body in order to transition from one step to the next, particularly with heavy loads. This would increase energy expenditure with a concomitant negative impact on economy.

It was surprising that no difference in economy was found in Chapter 5 between head-, back- and back/front-loading given that these methods perturb the posture of the trunk differently. Despite the lack of significant differences in economy between load carriage methods in Chapter 5, there was a tendency for improved economy with back/front-loading compared to back-loading with heavy loads in both Chapter 5 and Chapter 7. The heaviest load used in this research project (20 kg) was lighter than the load employed by Lloyd and Cooke (2000b), (2000c) (25.6 kg) and it's possible that the difference in economy between back- and back/front-loading might be more visible with heavier loads than those used in the research in this thesis. As such, future research comparing the economy of different trunk loading methods, such as back- and back/front-loading, might benefit from employing heavier load masses than those used in the present research in order to identify the advantages of one method compared to another. This might be particularly beneficial in an occupational setting, such as the military services, were loads masses in excess of 20 kg are routinely carried (Knapik et al., 2012).

In Chapter 7, head-loading for a group of inexperienced head-loaders was associated with a reduction in step length and a concomitant increase in cadence from unloaded walking. For vertical loading, shorter strides have been suggested to improve stability and may improve economy through reducing the vertical oscillations of both the COM of the body and added load (Cooke et al., 1991) (Figure 57).

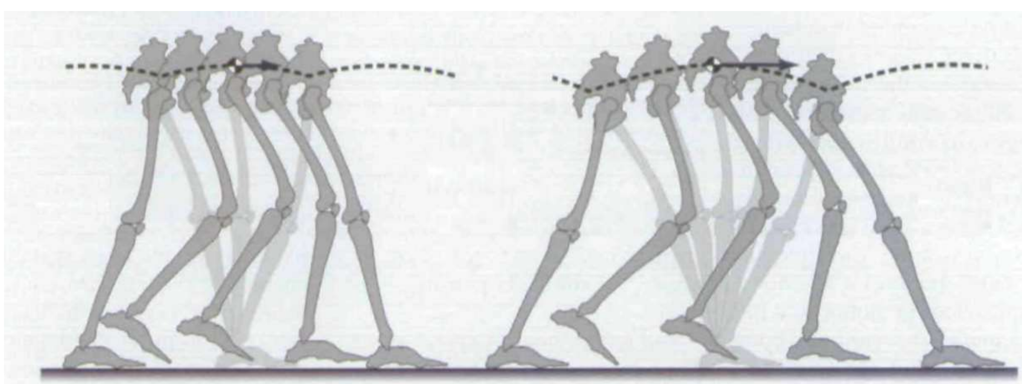


Figure 57. Illustration of increased vertical oscillation of the centre of mass with an increase in step length. Adopted from Kuo and Donelan (2010).

While an increased cadence/decreased step length combination might be beneficial for improving stability and vertical COM oscillations, Kuo (2001) suggested that the metabolic cost of walking might increase as a function of step frequency because the legs are not massless and the forced motion of the legs relative to the torso will require a metabolic cost. This idea is supported by Doke et al. (2005) who suggested that the swinging motion of the legs might account for one third of the net energy required to walk at 1.3 $m \cdot s^{-1}$ (4.68 $km \cdot h^{-1}$). In a comparison of the energetic costs associated with the swing phase and the stance phase, Umberger (2010) suggested that leg swing represented 29% of the muscular cost, while step-to-step transitions, where reduced vertical oscillations of the COM would be beneficial, accounted for 37% of the muscular cost. It's possible that there could be an optimal step length and cadence combination that allows for an overall improvement in energy cost through reducing the vertical oscillations of the COM without substantially increasing cadence, and the associated metabolic cost of swinging the legs. The findings from Chapter 7 of this thesis show that, for a group of inexperienced head-loaders, improved load carriage economy for is associated with smaller changes in step length and step cadence with unloaded walking. It could be that for an individual, the optimal step length-step cadence combination for reduced load carriage economy is close the preferred combination when walking unloaded. This would fit the established theory that, for unloaded walking, the stride length-stride frequency combination freely chosen for a given speed is close to optimal in terms of economy and that acute perturbations in stride length/stride frequency result in an increase in $\dot{V}O_2$ (Högberg, 1952, Cotes and Meade, 1960, Knuttgen, 1961, Cavanagh and Williams, 1982).

In Chapter 5 and Chapter 7, it was possible to select a sub-set of participants that demonstrated a similar level of load carriage economy to the free-ride phenomenon (ELI = ~0.80) (Maloiy et al., 1986) with either back-, back/front or head-loading, despite the group data often demonstrating a more proportional increase in energy expenditure relative to the added mass. This adds further evidence to show that the 'free-ride' hypothesis is not a generalisable finding. Future research investigating load carriage economy and any potential energy saving phenomena, should employ larger samples than those employed in the early studies of Maloiy et al. (1986) (*n* = 5) and Charteris et al. (1989) (*n* = 6). This is further supported by the analysis in Chapters 5 and 7 on the variation in load carriage economy attributable to differences between individuals and the variation attributable to differences between the load mass conditions, for each load carriage method. For example, for the head-loading method in 20 kg, the study in Chapter 5 found a CV of 16% for the ELI data with individual differences accounting for 63% of the variance in load carriage economy for a group of experience head-loaders (47% attributable to the mass of the load).

The work in Chapter 5 showed that individual load carriage economy is not determined by a single sagittal plane kinematic variable for load carried on the head, on the back and evenly distributed between the back and front of the torso, for a group of African women with head-loading experience. The study highlighted that biomechanical factors are likely to act in combination to influence load carriage economy. The ranking figures created in Chapter 7 (Figure 53 and Figure 54) to explore the associations between economy and walking gait adaptations showed that the most economical individuals for head-loading with 12 kg, in a group of inexperienced head-loaders, had the smallest change in step patterns from unloaded walking and the largest increase in double stance time and sagittal plane trunk ROM. While step patterns and joint angle motions of the sagittal plane trunk and hip angle movements from unloaded walking appear beneficial for load carriage economy, a common combination/interaction of biomechanical factors for the most economical load carriers for each load carriage condition was not found.

The perturbations in stride pattern were relatively small with the loads used in this research. This might explain why the determinants of individual differences in load carriage economy for some of the light loads (3 – 9 kg) could not be determined by reduced perturbations in walking gait from unloaded walking. Furthermore, the relationships between improved head-loading economy and biomechanical factors found in Chapter 7, but not Chapter 5, might also be partly explained by the difference in the magnitude of the walking gait perturbations, on average, found between the studies. For example, for head-loading there was, on average, an increase in trunk angle extension (Chapter 5: -2.0° ± 0.6 with pooled mass; Chapter 7: -7.3 ± 3.7° with pooled mass) and larger step cadence (Chapter 5: 0.00 ± 0.02 steps$\cdot$sec^{-1} with pooled mass; Chapter 7: 0.04 ± 0.06 steps·sec^{-1} with pooled mass) from unloaded walking for the group of inexperienced head-loaders in Chapter 7 compared to the experienced head-loaders in Chapter 5. Closer examination of the standard deviations for the change in trunk extension and cadence from unloaded walking between the studies also shows a greater magnitude of variability between participants for these variables in the study in Chapter 7 compared to that of Chapter 5. It is possible that the larger perturbations and greater variability in a group of less economical participants, on average, made the walking gait adaptations that lead to improved economy easier to identify. The role of long-term experience and/or training on walking gait perturbations in response to load carriage does not appear to have been previously investigated. The research in this thesis shows that, for head-loading, long-term (a minimum of 5 years) experience appears to reduce the variability of response for joint angle kinematics of the trunk and spatiotemporal variables between participants. As such, future research investigating the biomechanical determinants of improved load carriage economy for a particular loading method might benefit from assessing, and making comparisons between, relatively experienced and inexperienced participant groups with the load carriage method in question.

The research in this thesis focused on unloaded to loaded walking gait perturbations to identify the determinants of relative load carriage economy. This approach is beneficial for understanding the perturbations that relate to the ELI, which is also an assessment of load carriage that uses unloaded locomotion as a point of reference. This approach has shown that self-selected reduced gait alterations from unloaded walking, that are closer reduced gait perturbations from unloaded walking tend to have improved load carriage economy, particularly when considering trunk movement and stride patterns associated with carrying a heavy load. However, this approach has not revealed a common combination/interaction of biomechanical factors for the most economical individuals with each load carriage condition. It's possible that examining the role of movement variability, might help to tease out the determinants of individual differences in economy across different load carriage conditions, particularly for light loads that elicit smaller changes to the walking gait from unloaded walking. Indeed, the larger magnitude of intra-individual variation, assessed via standard deviations, for single support time, step length and cadence compared to the standard deviations between participants in the final study shows that, along with large inter-individual variation, there was also large within participant movement variability for some of the spatiotemporal variables. This particularly evident for single support time which has been linked to decreased gait stability in elderly populations (Hollman et al., 2007). Reduced step-to-step variability has been shown to reduce the metabolic cost of unloaded walking by up to 9% (O'Connor et al., 2012, Donelan et al., 2004). Although these studies involved forced perturbations to effect step variability, it's possible that individual differences in movement variability could account for some of the variance in load carriage economy between individuals that has been found in this thesis. The use of techniques such as dynamical systems theory in future load carriage research might elicit if individual differences in movement variability, in response to load carriage, contributes to the individual variation in load carriage economy found in this thesis and reported in the literature (Lloyd and Cooke, 2011, Lloyd et al., 2010c).

An approach to examining load carriage energetics and biomechanics that considers the movement of the load, separately to movements of the body, might also be beneficial to fully understand the determinants of load carriage. Research on the walking gait has focused on identifying the source of energy loss during a stride to explain the requirement for the addition of energy, which has led to a focus on the energy cost of step to step transitions (Kuo et al., 2005, Ruina et al., 2005, Adamczyk and Kuo, 2009, Kuo, 2007). Much of this research has focused on the energy required to perform mechanical work in these transitions to redirect the COM velocity from a forward and downward trajectory to a forward and upward trajectory. Theoretically, reducing the directional changes in COM velocity would be beneficial in reducing the amount of work required for step to step transitions and reduce energy cost (Inman and Eberhart, 1953). However, this would require gait adaptations to achieve the reduction in directional changes of the COM, which incur additional energetic costs above those of normal walking (Gard and Childress, 1997, Gard and Childress, 1999, Kerrigan et al., 2001). Although gait adaptations to flatten the COM trajectory when walking unloaded have been shown to worsen economy, Usherwood and Bertram (2016) demonstrated, using data from published head-loading studies (Heglund et al., 1995, Maloiy et al., 1986, Charteris et al., 1989) that reducing the trajectory path of the load (i.e. reduced collision geometry between step-to-step transitions) could reduce the collision cost for step-to-step transitions and improve load carriage economy. As such, it seems reasonable to speculate that load carriage strategies that flatten the arc of the trajectory for the external load during the gait cycle, without incurring a large metabolic penalty as a result of walking gait perturbations required to flatten the arc, might be beneficial for load carriage economy. Rome et al. (2006) found a reduced metabolic cost of 6.2% when walking with 27 kg compared to 0 kg using a backpack designed to reduce the vertical oscillations of the load relative to the body. This reduction in metabolic cost is closer to the 6-9% decrease in $\dot{V}O_2$ reported by Lloyd and Cooke (2000b) for back/front-loading compared to back-loading, than the larger magnitude of reduced metabolic cost purported for the free-ride hypothesis (Maloiy et al., 1986). Using the unloaded and loaded walking data from Rome et al. (2006), estimated ELI values for the locked backpack and spring backpack (reduced vertical oscillation) were 1.05 and 0.98, respectively. As such, the specially designed backpacks from Rome et al. (2006) appear to save a similar level of energy as back/front-loading compared to back-loading with a traditional backpack. To fully understand the role of load and body COM movements, future load carriage research should consider the motion of the load carriage device and motion of the body separately, and in combination, to better understand interactions between the body and different load carriage methods.

8.4. Practical implications of the research presented

The advantage of using the ELI to assess load carriage economy over other widely used approaches, such as assessing the rate of oxygen consumption (e.g. Legg and Mahanty, 1985, Quesada et al., 2000) or the energy cost of walking (e.g. Abe et al., 2004, Bastien et al., 2005), is that the ELI accounts for unloaded walking energy expenditure and provides a single unitless value allowing for simple comparisons between different load carriage systems and study designs. However, the ELI has not been widely adopted by researchers investigating load carriage economy since its development (Lloyd et al., 2010a), with only Godhe et al. (2017) employing the measure directly and Prado-Nóvoa et al. (2019) employing a similar measure which they referred to as the Carrying Cost Index. Load carriage research often involves a single method and the ELI is perhaps deemed unnecessary by authors working on such studies that do not make comparisons between load carriage conditions, as an extra measure of $\dot{V}O_2$ for unloaded walking is required. It's also possible that the ELI has not been widely adopted due to a lack of knowledge on the measure's reliability. Chapter 4 provides a robust assessment of ELI reliability with light and heavy loads across a range of walking speeds. This evidence that the ELI is a reliable measure of relative load carriage economy may contribute to its increased use in scientific literature might increase, particularly for studies comparing the economy associated with different load carriage systems or different walking parameters (e.g. different walking speeds).

The magnitude of individual variation in load carriage economy and walking gait perturbations suggest that a 'one size fits all' approach to load carriage design is not appropriate. For commercial products, increasing the ability to customise a load carriage system appears to be beneficial, particularly if the individual customisation allows load to be carried in a way that reduces gait perturbations from unloaded walking, particularly in relation to trunk movements in the sagittal plane. While bespoke designs might not be appropriate for the commercial market, individually fitted systems might be appropriate for specialized markets (e.g. military and emergency services), perhaps with additional gait analysis for personnel during the initial fitting process to assess individual gait perturbations from unloaded walking and ensure adjustable components of a load carriage system are fitted to minimizes these gait perturbations. Designing load carriage systems to be more customizable is not a new concept. Load carriage systems currently exist that include front balance pockets that can be attached to the front of a backpack and allow load to be more evenly distributed around the trunk compared to the back alone (e.g. AARN pack, New Zealand). Duluth backpacks (Duluth Pack, USA) also include a tumpline that can be used to provide additional support a via the head and neck.

The level of inter-individual variation in load carriage economy and gait perturbations found in this work might be an important consideration for recruitment and employment standards in occupations that require load carriage. When evaluating the impact of load carriage on job related performance in these roles, it might be important to understand how a particular load condition perturbs an individual's gait, rather than assuming that the same gait perturbations will occur for all individuals. This could be particularly relevant from an injury prevention perspective given the individual variation in postural adjustments to load carriage found in the research in this thesis, which could lead to different types of load carriage related injury and prevalence of injuries between individuals. For example, inter-individual variability in knee angle motion in response to load carriage is likely to lead to individual variation in knee moments. Increased total knee joint moments have been suggested as a causative factor of knee osteoarthritis among military personal (Krajewski et al., 2020).

8.5. Strengths and Limitations

There were some strengths and limitations to the research in this thesis, which are outlined in the section.

A strength of this research was the development of a deterministic model to facilitate the analysis of loaded and unloaded walking in Chapter 7. The use of a rigorously developed deterministic meant that the selected performance parameters in Chapter 7 were based on a theoretical rationale, which could be considered superior to randomly selecting performance parameters. Another strength of the research in this thesis was the use of the ELI to assess load carriage economy because it accounts for the energy cost of unloaded walking. Lloyd et al. (2010a) conceptualized the energetic cost of load carriage as: the energy cost of unloaded walking at a given speed + the energy cost required to support and move a given external load ± the net change in the energy cost of movement due to changes in the kinematics and kinetics of movement as a result of the interaction between the load mass, speed and load carriage method. Based on this concept, the ELI appears to be a more appropriate measure of load carriage economy compared to measures that do not account for the energy cost of unloaded walking, particularly for research investigating the determinants of load carriage economy. Furthermore, all load carriage data in this thesis was presented as the change from unloaded to loaded walking, which is more appropriate than reporting absolute values because it accounts for the individual variation in the walking gait biomechanics associated with unloaded walking (Whittle, 2014).

Although the free-ride hypothesis was first identified over 30 years ago for African women with several years of head-loading experience (Maloiy et al., 1986), there is a paucity of load carriage research for this population. A strength of this thesis was the analysis of head-loading economy and walking gait kinematics in a sample of experienced head-loading women, which provided a robust investigation on the determinants of head-loading economy in a sample of participants from the population for which the free-ride hypothesis was first reported. Although it was not feasible to also assess experienced head-loaders in the research in Chapter 7, the use of a force instrumented treadmill alongside a motion capture system could also be considered a strength of the research in this thesis. Few studies investigating the biomechanical determinants of load carriage economy have assessed the associated kinematics and kinetics, and of those that have (Lloyd and Cooke, 2011, Huang and Kuo, 2014), only Huang and Kuo (2014) measured the kinematics and kinetics in the same walking trials.

One potential limitation to this body of work is the controlled walking speed employed in experimental studies 2 and 3 (Chapter 5 and 7). A speed of 3 $km \cdot h^{-1}$ was selected to make comparisons with research that has previously demonstrated energy saving phenomenon for load carriage methods (Maloiy et al., 1986, Charteris et al., 1989, Lloyd and Cooke, 2000b, Abe et al., 2004). However, using a set speed, rather than a self-selected speed, might have affected some individuals' natural walking gait pattern more than others (depending on an individuals preferred unloaded walking speed) and this could have contributed to some of the high levels of individual variation noted in this thesis for both load carriage economy and loaded walking gait characteristics. This also limits the relevance of the findings for the determinants of improved economy for certain populations, such as those in the military and emergency services, who are regularly required to walk at speeds in excess of 5 $km \cdot h^{-1}$ when carrying loads (Knapik et al., 2012).

While the results from experiments in controlled laboratory conditions are valuable for improving the knowledge and understanding of load carriage performance, it is important to note that load carriage tasks are often performed on uneven terrain at non-constant, self-selected speed. This will lead to additional metabolic costs and biomechanical challenges compared to a laboratory environment. As such, the lack of field assessments could be deemed a limitation to this body of work and is an under researched area when considering the biomechanics of load carriage. Load masses of 3 – 20 kg were used for the research in this thesis as energy saving phenomenon have been reported for similar loads within this range. However, some populations such as those in the military services (Knapik et al., 2012) regularly carry loads in excess of 20 kg, and, as such, the findings of this research are not directly applicable to those populations. Future research seems warranted to assess the magnitude of inter-individual variation for economy and walking gait perturbations to load carriage for increased load mass, faster walking speeds and longer walking durations, along with mechanisms to improve load carriage economy under these conditions.

Perhaps another limitation to this research project was the mixed use of experienced and inexperienced head-loaders in Chapters 5 and 7, respectively, without a direct comparison between experienced and inexperienced load carriers. Lloyd et al. (2010c) showed no difference between experienced and inexperienced head-loaders. However, this thesis showed no difference in mean ELI values between methods in Chapter 5 with experienced head-loaders, but a significant difference between mean head-loading economy compared to economy with the other methods for the inexperienced head-loaders in Chapter 7. A study including a direct comparison between experienced and inexperienced head-loaders was not deemed unfeasible for this research project, given the difficulty of recruiting individuals with experience of head-loading in the location of the investigation for Chapter 7 (Leuven, Belgium).

8.6. Directions for future research

The results from the present research project give rise to a number of important research questions, the most pertinent of which are outlined below.

Further work is required to identify the determinants of inter-individual variation in load carriage economy. The research in this thesis identified a large level of individual variation in loaded walking gait characteristics but there was a lack of strong relationship between economy and biomechanical variables. O'Connor et al. (2012) found that increased variability in walking gait step width and step length increased the metabolic energy cost of unloaded walking. The findings of Chapter 7 suggest that variability in loaded walking step width is unlikely to solely explain individual differences in load carriage economy. It is possible, however, that individual variation in the amount of movement in a number of walking gait characteristics could explain some of the individual variation in load carriage economy. This is an avenue of research that appears worthy of exploration through techniques that measure coordination variability such as dynamical systems theory to assess movement variability.

Most load carriage activities occur on uneven terrain and surfaces that are not smooth (e.g. grass, sand, snow). Walking and running unloaded on uneven terrain have been shown to be more energetically costly than walking on smooth ground (Voloshina et al., 2013, Voloshina and Ferris, 2015). As might be expected, Voloshina et al. (2013) demonstrated that walking unloaded on even terrain significantly increased step length and step width variability. The research in this thesis has quantified the inter-individual variation in load carriage economy and loaded walking gait perturbations on smooth, even surfaces. The next step would be to quantify individual variation in these parameters on uneven terrain, where most load carriage activities occur. This could provide useful information for optimal load carriage strategies and load carriage system designs for outdoor environments.

Many recreational and occupational load carriage activities occur over longer distances and durations of time than examined in this thesis. Both load carriage and fatigue have been shown to influence gait characteristics (Qu and Yeo, 2011). Given the level of individual variation reported in this thesis for both load carriage economy and loaded walking gait characteristics, the influence of experience and fatigue on individual differences in the physiological and biomechanical variables associated with load carriage appears to warrant further attention. This could provide useful information for the design of future load carriage systems. Occupational load carriage (e.g. military and emergency services) also regularly occurs at faster walking speeds than those explored in this thesis. To achieve faster walking speeds, individuals with a shorter stature and shorter leg length could be forced to alter their gait to a less efficient movement pattern which could increase individual differences in the energy cost associated with load carriage. To date, large individual variation in load carriage economy has only been reported at walking speeds of ~3 $km \cdot h^{-1}$. Understanding individual differences in the energy cost of load carriage at fast walking speeds, and the potential influence of individual physical characteristics on the magnitude and effect of individual differences, might be of interest to military populations who are required to march at faster speeds.

The research in this thesis, and much of the load carriage literature, has focused on acute responses to load carriage for both walking perturbations and economy. As such, the effect of chronic changes in response to load carriage training are less well known. Future research could build on the recent work of Wills et al. (2019) who found a decrease in the work performed at the knee joint (4.2%) and an increase in the work perform at the ankle (3.7%) during a 5 km loaded march following a targeted 10 week resistance training programme. The change in walking gait perturbations to chronic load carriage activities is not well reported in the literature and research on this and the associated load carriage economy may be of interest to individuals who are required to carry load regularly as part of their occupation.

8.7. Conclusions

This research project has made considerable progress in identifying factors that determine individual load carriage economy and excluding factors that do not. Of the walking gait perturbations measured in this thesis, the freedom of movement of the trunk with back- and back/front-loading conditions was considered to be the most likely candidate for reduced energy expenditure based on the research of Abe et al. (2004) and Lloyd and Cooke (2000b), (2000c). The findings of this thesis show that individual perturbations alone, such as differences in trunk movement or differences in step width, do not solely explain economy for loads ≤ 20 kg at a walking speed of 3 $km \cdot h^{-1}$. An economical advantage for back/front-loading with heavy loads was a consistent finding in this research. The most economical individuals for this load carriage condition self-selected reduced perturbations of the upper body in the sagittal plane from unloaded walking. It is likely that the mechanical advantage for back/front-loading with heavy load is due to movements of the trunk that closely resemble those of unloaded walking. Other instances were individuals were more economical with a particular load carriage condition are also likely to be a result of reduced perturbations from unloaded walking for posture (particularly at the trunk and hip) and step patterns. This was particularly evident for head-loading with a moderate load for individuals with no previous experience of head-loading.

The research in this thesis was not able to identify a combination or interaction of gait perturbations that improved economy across all load conditions and/or individuals. Given the large magnitude of inter-individual variability for load carriage economy and gait perturbations found in this research, it's possible that load carriage economy is not determined by a consistent set of biomechanical factors for each method of loading, are generalizable to all individuals. The large difference between step variability (within-participant variability larger than between participant variability) found for some of the spatiotemporal variables in Chapter 7 suggest that future research might benefit from investigating the role that step to step variability and coordination variability have on load carriage economy.

Chapter 9. References

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Chapter 10. Appendices

Appendix A: Chapter 4 Leeds Trinity ethical approval letter

Leeds Trinity University

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PRIVATE & CONFIDENTIAL

Dr John Perry
Chair of Departmental Ethics Committee for Sport, Health and Nutrition
Tel: 0113 283 7175
E-mail: j.perry@leedstrinity.ac.uk

Date: 4th June 2015

Dear Sean Hudson

Re: ethics application 'The reliability of the Extra Load Index'

Thank you for your recent application for ethical approval for the above named project. The committee received the application along with the following documents attached:

i. Participant information sheet
ii. Participant consent form
iii. Leeds Trinity consent form
iv. Sport, Health and Nutrition screening questionnaire

After reviewing the application I am pleased to confirm that your application for ethical approval has been successful. Good luck with your research.

Yours sincerely

Dr John Perry
Chair of Departmental Ethics Committee for Sport, Health and Nutrition

Cc: Ethics committee file

Appendix B: Chapter 7 Leeds Trinity ethical approval letter

Leeds Trinity University

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Dr Mark Russell
Chair of SSHS Ethics Committee
Tel: 0113 283 7100 ext 649
E-mail: m.russell@leedstrinity.ac.uk

Date: 16th January 2018

Dear Sean

Re: SSHS/2017/070 – The determinants of individual load carriage economy.

Thank you for your recent application for ethical approval for the above named project.

After reviewing the application it has been resolved that the research project is granted ethical approval.

I wish you well in your study,

Yours sincerely

Dr Mark Russell
Chair of School of Social and Health Sciences Ethics Committee

Appendix C: Chapter 7 KU Leuven ethical approval letter

UZ LEUVEN

Leuven, 23 januari 2018

Ethische Commissie
Onderzoek UZ/KU Leuven
Herestraat 49
B 3000 Leuven (Belgium)
tel +32 16 34 86 00
website: www.uzleuven.be/ec/onderzoek
email: ec@uzleuven.be

prof. Benedicte Vanwanseele

Ons kenmerk: S61103	EudraCT-nr:	Belg. Regnr: B322201835069

The determinants of individual load carriage economy.

DEFINITIEF GUNSTIG ADVIES

Geachte College,

De Ethische Commissie Onderzoek UZ/KU Leuven heeft vermeld protocol onderzocht en besproken op haar vergadering van 27 december 2017.

Na inzage van de bijkomende informatie en/of aangepaste documenten met betrekking tot dit dossier is de EC van oordeel dat de voorgestelde studie, zoals beschreven in het protocol, wetenschappelijk relevant en ethisch verantwoord is. Ze verleent dan ook een gunstig advies over deze studie.

> De EC wenst de hoofdonderzoeker/promotor van de studie te wijzen op zijn/haar verantwoordelijkheid betreft de privacy van de persoons-/patiëntgegevens bij contact met de patiënt en/of inzage in het elektronisch medisch dossier, inclusief de correcte implementatie hiervan door medewerkers en studenten. De EC verwijst naar de richtlijnen van ICH/GCP hierover op de website, en benadrukt dat een GCP-opleiding van elke hoofdonderzoeker verwacht wordt. De EC verwijst tevens naar de Belgische wetgeving (Wet van 8/12/1992 ter bescherming van de persoonlijke levenssfeer en Wet van 22/8/2002 betreffende de rechten van de patiënt).

Bij het beoordelen van dit dossier werd rekening gehouden met de documenten en informatie gerelateerd aan deze studie, ingediend op 15 december 2017, 19 december 2017 en 18 januari 2018.

Dit gunstig advies betreft:

Protocol:
  v2 dd 18/01/2018
Informatie en toestemmingsformulier:
  Informed consent v1 dd 19/12/2017 E

*Indien de studie niet binnen het jaar beëindigd is, vereist de ICH-GCP dat een **jaarlijks vorderingsrapport** aan de EC wordt bezorgd.*

*Gelieve tenslotte het (vroegtijdige of geplande) stopzetten van een studie binnen de door de wet vastgestelde termijnen mee te delen en een **Clinical Study Report** aan de EC te bezorgen.*

Met vriendelijke groet,

Prof. Dr. Minne Casteels
Voorzitter
EC Onderzoek UZ/KU Leuven

Cc:
FAGG (Federaal Agentschap voor Geneesmiddelen en Gezondheidsproducten)

CTC (Clinical Trial Center UZ Leuven)

Appendix D: Chapter 4 participant information sheet

Participant Information Sheet

Study Title: The reliability of the Extra Load Index

Introduction

You are being invited to take part in a research study. Before you decide whether or not you would like to take part in the study it is important for you to understand why the research is being done and what it will involve. This information sheet should give you a basic idea of what the research is about and what your participation will involve.

Background Information

Comparing the amount of energy an individual expends when carrying an item in different ways is useful when deciding on the best way to manually carry a load for a long period of time. However, it is currently difficult to compare how much energy it takes to carry a load using different methods (e.g. in a backpack compared to in the hands). This is because previous research has used a variety of different calculations to measure energy expenditure along with a variety of different testing protocols (e.g. different walking speeds or different weight carried)

Identifying a method that requires less energy to carry a load is beneficial as it may allow an individual to carry a load for longer without feeling discomfort and may also allow for more to be carried. Additionally, a method that requires less energy may also reduce the risk of injury occurring as a consequence of carrying a load for long period of time.

In this study, we will test the reliability of a measure that is designed to allow for direct comparisons between how much energy is used to carry a load in different ways. This measure is called the Extra Load Index. If we find it reliable, the Extra Load Index could prove useful for companies that design load carrying devices such as backpacks and consumers who carry items for long periods of time, such as recreational hikers.

The Extra Load Index is a calculation shown below.

$$ELI = \frac{mlO_{2L}/\text{kg total mass per min}}{mlO_{2U}/\text{kg body mass per min}}$$

In this calculation, mlO_{2U} refers to oxygen consumption when walking without carrying anything and mlO_{2L} refers to oxygen consumption when carrying a load. Oxygen consumption is used as a way to assess the amount of energy that has been used.

As shown by the calculation above, the Extra Load Index takes into account the energy used when walking without carrying anything which is an important consideration when identifying the energy it takes to carry a load. The Extra Load Index has the potential to become a standard measure when comparing different methods of load carriage.

Study Aim

The aim of this study is to assess the reliability of the Extra Load Index in order to determine its practicality as a means of comparing different load carrying methods.

Study Requirements

You will be a healthy male or female volunteer aged between 18-55 years. You will be asked to attend the human performance laboratory at Leeds Trinity University on seven separate occasions. You will complete an initial visit to familiarise yourself with the main trial protocol and the equipment that will be used. This session will last approximately 30 minutes. The next 6 visits will be to complete the main trials, with each visit lasting no more than 45 minutes.

You will be able to withdraw from the study at any time during the testing, without having to provide an explanation.

Location

Sport and Exercise Science laboratory (Sports Laboratory) in the Department of Sport, Health & Nutrition at Leeds Trinity University, Brownberrie Lane, Horsforth, Leeds, West Yorkshire, LS18 5HD

Restrictions During Testing

You will be asked to maintain similar training patterns throughout testing. You will also be asked to maintain a similar diet and to refrain from moderate-vigorous exercise and alcohol consumption in the 24 hours before each test. We will confirm these points with you verbally.

Testing Protocol

Initial Visit

In the initial visit, you will be asked to walk on the treadmill until you feel comfortable with the equipment. You will then be asked to repeat this while also carrying a rucksack and wearing the face mask used to analyse the air you breathe out. This visit will also be used to establish a self-selected walking speed that you are most comfortable with.

Main Trials
You will be asked to attend the laboratory on 6 more occasions in order to complete the main trials, with 48 hours between each visit. The first 3 main trials will be to complete each of the 3 experimental conditions in a randomised order. The 3 experimental conditions will then be repeated in the final 3 visits to test for reliability. For each trial, you will be fitted with a face mask (to collect the air you breathe out) and walk on the treadmill while carrying different amounts of weight. To begin you will be asked to walk without carrying anything for 4 minutes followed by a 5-minute rest period. You will then be asked to repeat this process of 4 minutes exercise followed by 5 minutes of rest, carrying a light load (7kg in backpack), a heavy load (35kg in a backpack) and finally, again without carrying anything. The speed you walk at in each trial will either be a slow speed ($3km \cdot h^{-1}$), a fast speed ($6km \cdot h^{-1}$) or a self-selected speed, this be depends on the trial condition.

Potential Benefits to You
You will be provided with information on your energy expenditure when walking and when walking while carrying both light and heavy loads.

Potential Risks to You
There are always potential risks associated with performing any physical activity. However, you will be physically active and you are reminded that you should cease exercise immediately if you begin to feel unwell or unduly distressed.

Contacts

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Appendix E: Chapter 7 participant information sheet

Participant Information Sheet

Determining the energy required for different load carrying methods

Introduction

You are being invited to take part in a research study. Before you decide whether or not you would like to take part in the study it is important for you to understand why the research is being done and what it will involve. This information sheet should give you an idea of what the research is about and what your participation will involve. Ask us if there is anything that is not clear or if you would like more information. Take time to decide whether or not you wish to take part.

Thank you for reading this.

Purpose of the study

Companies around the world are trying to design ways of carrying loads that are more comfortable and more energy efficient than existing methods whilst scientists are trying to find out what makes certain methods of carrying loads better than others. In Africa, and some other parts of the world, women regularly carry heavy loads on their heads. This was thought to be a very energy efficient way of carrying loads but this now doesn't seem to be the case. In the western world, loads are regularly carried on the back and around the front and back of the torso. Again, some have suggested that this is a very energy efficient way of carrying loads, however there appears to be a substantial amount of variation between individuals. In this study we want to compare carrying loads on the head, on the back and spread between the back and front of the torso to see which requires the least energy and find out why some individuals appear to be more energy efficient with certain methods compared to others.

Why have I been chosen?

We are seeking volunteers for this study. We want to look at the difference in energy expenditure and how the body moves when carrying loads in three different ways. To do this we need a group of about 20 individuals who are similar in age (between 18 and 40). Since you fit into such a group you could be included in the study if you wish.

Do I have to take part?

No - it is up to you to decide whether or not to take part. If you do decide to take part you will be given this information sheet to keep and be asked to sign a consent form. If you decide to take part you are still free to withdraw at any time and without giving a reason.

What will I have to do?

To be part of the study you will need to visit the Movement and Posture Analysis Laboratory Leuven twice. Each time you come you will have to walk on a treadmill, carrying loads either in a backpack, on your head or in a backpack that also has pockets at the front. We will place markers on you so that we can capture your movements on the treadmill using our 3D motion capture system. We will also ask you to wear a face mask so that we can measure how much energy you are using. The first visit will last approximately 2 hours and the second visit will last approximately 2.5 hours. Each time you visit you will do the same thing, the only difference will be how you carry the loads.

When you first arrive at the lab, we will weigh you and then attach the markers using double sided sticky tape that peels off easily. It is best if you can bring with you shorts and a T-shirt to wear during the test as it is easier to mark the joints if you are not wearing too many clothes.

We will then check the method of carrying load that you are to use. If it is in a backpack or backpack with front balance pockets we will put the pack on you and adjust it until it fits properly and is comfortable. We will then take it off until it is needed. You will then take your place on the treadmill and have the face-mask fitted. The speed of the treadmill will be set at 3 km/h (a slow walking speed) and you will walk for four minutes. At the end of 4 minutes you will have 2 minute rest. During this time you will be asked to complete some questions about discomfort and your first load will be fitted. The first load will be 3 kg. Once this is fitted you will walk for 4 minutes as before and again there will be a 2 minute rest while you answer the questions and the next load is prepared. The second load will be 12 kg. Once again you will walk for 4 minutes followed by a 2 minute rest when the load will be changed and you will answer the questions about any discomfort you feel. This process will be repeated with a 20 kg. If you are finding it too difficult to carry any load you may stop at that time and we will end the test. You will then have a 10 minute rest before repeating the load carrying procedure but this time we will slightly modify how you walk when carrying the load to see if there is any difference in your energy consumption. For example, we might ask you to put on a brace to change the angles of your hip, knee or ankle joints. We might ask you to change the speed at which you swing your arms or the number of steps you complete, or we might ask you to alter the width of your steps or how far you lean forward when walking.

When you return 3-4 days later the whole process is repeated but you will carry the load in the two remaining ways.

If you are interested in taking part you can spend some time today looking at the equipment we will use and try it out. You will have the opportunity to get used to walking on the treadmill, as well as wearing a face mask, and we will show you the information that is recorded. You will also be able to try out carrying loads in the backpack, the front and back pack and the bucket for your head. Finally, if you are still interested in the study, we will ask you to fill out some forms to make sure it is ok for you to take part in the study and to find out how much experience you have of carrying loads on your back and your head. You will be free to ask any questions you wish.

Restrictions during testing

You will be asked to maintain similar training patterns throughout testing. You will also be asked to maintain a similar diet and to refrain from moderate-vigorous exercise and alcohol consumption in the 24 hours before each test. We will confirm these points with you verbally.

Potential Benefits to you

Most of the benefits of this study relate to increasing our knowledge of what makes one way of carrying a load better than other ways for each individual. You will be able to see your own results for all the experiments. You will be provided with information on your energy expenditure when walking and when walking while carrying both light and heavy loads.

Potential Risks to you

Whenever you take part in any physical activity there are some risks but for this study they are quite small and we have made every effort to reduce them even further. The risks include muscular injury caused by walking with a load and cardiovascular complications caused by exercise. The risk of muscular injury is most likely when you are carrying relatively heavy loads on your head or back, especially if you are not used to doing this. The heaviest load used in this experiment is 20kg, which is the same as a bucket of water. The risk of cardiovascular complications is relatively small in healthy adults and we will only let you take part in the study if you have no history of such difficulties and are feeling well on the day of the test.

Confidentiality and Anonymity

All participant information will remain confidential and you will not be identified in any of the outputs associated with the research project. Data will be recorded and stored on a computer requiring password entry for up to 10 years. Following the completion of the research and potential publication, any documents containing your personal details will be destroyed by permanently deleting files from the computer. Any of your data recorded on paper will be shredded immediately after being transferred onto a computer requiring password entry.

Thanks for taking the time to read this. If you wish to take part in the study you will need to sign the consent form and you will be given a copy of both this information sheet and the consent form to keep.

Contacts

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Appendix F: Health screen questionnaire

Health screening questionnaire

Name:____

Risk factors	Risk factor	No risk factor
1. Age= ___________ years male	≥45	
female	≥55	
2. Have any of your parents, brothers or sisters had a heart attack, bypass surgery, angioplasty, or sudden death prior to 55 years (male relatives) or 65 years (female relatives) *	Yes	No
3. Do you currently suffer from lower back pain or have a history of lower back pain regularly occurring?	Yes	No
4. Do you currently suffer from neck pain or have a history of neck pain regularly occurring?	Yes	No
5. Have you recently suffered from any musculoskeletal injury or have a history of musculoskeletal injuries regularly occurring?	Yes	No
6. Are you currently a smoker- have you quit within the past 6 months- are you exposed to environmental tobacco smoke?	Yes	No
7. In the past 3 months have you performed at least 30 minutes of moderate intensity physical activity or equivalent on at least 3 days of the week?	No**	Yes
8. Do you take blood pressure medication?	Yes	No
Resting blood pressure: SBP=_________mmHg, DBP=____________mmHg	**≥ 140/90*** **	**< 140/90**
Total number of risk factors If yes to early sudden death in family history advise pre-participation screening Avoid maximal testing if currently sedentary If BP ≥140/90mmHg treat as high risk and advise pre-participation screening		

Signs or symptoms		
9. Do you ever have pain or discomfort in your chest or surrounding areas (neck, jaw, arms or other areas)?	Yes	No

10. Are you short of breath at rest or with mild exertion?	Yes	No
11. Have you ever experienced dizziness or loss of consciousness during or shortly after exercise?	Yes	No
12. Do your ankles ever become swollen (other than as a result of an injury)?	Yes	No
13. Do you ever suffer from cramp-like pains in your legs, brought on by exertion and relieved after 1-2 minutes of rest?	Yes	No
14. Do you ever have palpitations (= the unpleasant awareness of the heart beating in your chest) or an unusual period of rapid heart rate?	Yes	No
15. Have you ever been short of breath at rest in the recumbent position or had an attack of breathlessness in the middle of the night which was relieved by sitting up?	Yes	No
16. Has a doctor ever said you have a heart murmur?	Yes	No
17. Do you feel unusually fatigued or find it difficult to breathe with usual activities?	Yes	No
SIGNS OR SYMPTOMS OF DISEASE	Yes/No	

Personal History of disease		
Heart disease	Yes	No
Peripheral vascular disease	Yes	No
Cardiovascular disease	Yes	No
Chronic obstructive pulmonary (emphysema/chronic bronchitis)	Yes	No
Asthma	Yes	No
Interstitial lung disease	Yes	No
Cystic fibrosis	Yes	No
Diabetes mellitus	Yes	No
Thyroid disorder	Yes	No
Renal disease	Yes	No
Liver disease	Yes	No
HISTORY OF DISEASE	Yes/No	

Other Conditions		
Do you have any bone or joint problems such as arthritis or a past injury that might get worse with exercise and/or carrying an additional load?	Yes	No

Do you have any other problem that might make it difficult for you to carry an additional load and/or do strenuous exercise?	Yes	No
Are you or have you been recently pregnant?	Yes	No
Are you on any prescription medications?	Yes	No

I confirm that the above information which I have provided to Leeds Trinity University is true and accurate to the best of my knowledge and belief and I understand that I must notify promptly of any changes to the information.

I understand that the information I have provided above may be used as part of an anonymised dataset by staff or students from SHN for competition of coursework or for research or audit purposes (with the appropriate ethical approval in place)

Signature:_____________________________

Date:________________

Witness signature:__________________________

Date:_____________________________

Appendix G: Chapter 4 written informed consent

Leeds Trinity University

Statement of consent to participate in the investigation entitled:

The reliability of the Extra Load Index

1. I, ………………………………………………………… agree to partake as a subject in the above study.

2. I understand from the participant information sheet, which I have read in full, and from my discussion(s) with Sean Hudson that this will involve me completing an initial visit to familiarise myself with the study protocol and the equipment followed by 4 further visits to complete the 4 main trial conditions, as detailed in the subject information sheet.

3. It has also been explained to me by Sean Hudson that the risks and side effects which may result from my participation are as follows: There are potential risks associated with performing moderate-vigorous exercise, although I am familiar with performing some type of moderate-vigorous exercise and the investigators will remind me to cease exercise immediately should I begin to feel unwell or unduly distressed.

4. I confirm that I have had the opportunity to ask questions about the study and, where I have asked questions, these have been answered to my satisfaction

5. I understand that I must abide by University regulations and the advice of researchers regarding safety.

6. I am aware that I can withdraw my consent to participate in the study at any time and for any reason, without having to explain my withdrawal.

7. I understand that any personal information regarding me, gained through my participation in this study, will be treated as confidential and only handled by individuals relevant to the performance of the study and the storing of information thereafter. Where information concerning myself appears within published material, my identity will be kept anonymous.

8. I confirm that I have been informed of the University's policy relating to the storage and subsequent destruction of sensitive information. I understand that sensitive information I have provided through my participation in this study will be handled in accordance with this policy.

9. I confirm that I have completed the health questionnaire and I know of no reason, medical or otherwise that would prevent me from partaking in this research.

Subject signature: ……………………………………………………………… Date:_____________

Independent witness signature: …………………………………………… Date:_____________

Appendix H: Chapter 7 written informed consent

<u>Statement of consent to participate in the investigation entitled:</u>

The determinants of individual load carriage economy

10. I, ______________________________ agree to partake as a participant in the above study.

11. I understand from the participant information sheet, which I have read in full, and from my discussion(s) with Sean Hudson that this will involve me completing an initial visit to familiarise myself with the study protocol and the equipment followed by 3 further visits to complete the 3 main trial conditions, as detailed in the participant information sheet.

12. It has also been explained to me by Sean Hudson that the risks and side effects which may result from my participation are as follows: There are potential risks associated with performing moderate-vigorous exercise, although I am familiar with performing some type of moderate-vigorous exercise and the investigators will remind me to cease exercise immediately should I begin to feel unwell or unduly distressed.

13. I confirm that I have had the opportunity to ask questions about the study and, where I have asked questions, these have been answered to my satisfaction

14. I understand that I must abide by University regulations and the advice of researchers regarding safety.

15. I am aware that I can withdraw my consent to participate in the study at any time and for any reason, without having to explain my withdrawal.

16. I understand that any personal information regarding me, gained through my participation in this study, will be treated as confidential and only handled by individuals relevant to the performance of the study and the storing of information thereafter. Where information concerning myself appears within published material, my identity will be kept anonymous.

17. I confirm that I have been informed of the University's policy relating to the storage and subsequent destruction of sensitive information. I understand that sensitive information I have provided through my participation in this study will be handled in accordance with this policy.

18. I confirm that I have completed the health questionnaire and I know of no reason, medical or otherwise that would prevent me from partaking in this research.

Subject signature: ____ Date: ____________

Independent witness signature: ______________________________ Date: ____________

Primary Researcher signature: _______________________________ Date: _____________

Appendix I: Consent to scientific illustration

FOR USE WHEN STILL OR MOVING IMAGES WILL BE RECORDED

Consent to scientific illustration
I hereby confirm that I give consent for photographic and/or videotape and sound recordings (the 'material') to be made of me. I confirm that the purpose for which the material would be used has been explained to me in terms which I have understood and I agree to the use of the material in such circumstances. I understand that if the material is required for use in any other way than that explained to me then my consent to this will be specifically sought.
1. I understand that the material will form part of my confidential records and has value in scientific assessment and I agree to this use of the material. Signed.. Date.. Signature of Parent / Guardian in the case of a minor ....
2. I understand the material has value in teaching and I consent to the material being shown to appropriate professional staff for the purpose of education, staff training and professional development. Signed.. Date.. Signature of Parent / Guardian in the case of a minor ....

I hereby give consent for the photographic recording made of me on....................... to be published in an appropriate journal or textbook. It is understood that I have the right to withdraw consent at any time prior to publication but that once the images are in the public domain there may be no opportunity for the effective withdrawal of consent.

Signed .... Date ....

Signature of Parent / Guardian in the case of a minor

....

Appendix I: Load carriage experience questionnaire

Load Carriage Experience

Participant Number: _______
Date of Birth: _____ /_____ / _____
Height:_______cm
Weight:_______kg
Telephone: ______________________

1) Do you have experience of carrying loads on your head? Yes No

If yes, please answer questions 2 – 8. If no, please move to question 9.

2) How old were you when you first carried loads on your head? ____________________

3) How old were you when last carried loads on your head? ________________________

4) How often did you carry loads on your head as a young child (under 5 years of age) and what sort of loads did you carry?

____
____
____
____
____

5) How often did you carry loads on your head as a young child (Age 6-10) and what sort of loads did you carry?

____
____
____
____
____

6) How often did you carry loads on your head between the ages of 11 and 17 and what sort of loads did you carry?

____
____
____
____
____

7) If you are still carrying loads on your head, how often do you do this and what sort of loads do you carry?

____
____
____
____
____

8) What are the heaviest loads you have carried on your head?

____
____
____
____
____

9) Do you have experience of carrying loads in a backpack? Yes No

If yes, please answer questions 10 – 16. If no, please move to question 17.

10) How old were you when you first carried loads in a backpack? ____________________

11) How old were you when last carried loads in a backpack? _________________________

12) How often did you carry loads in a backpack as a young child (under 5 years of age) and what sort of loads did you carry?

____
____
____

____

13) How often did you carry loads in a backpack as a child (age 6-10) and what sort of loads did you carry?

14) How often did you carry loads in a backpack between the ages of 11 and 17 and what sort of loads did you carry?

15) If you are still carrying loads in a backpack, how often do you do this and what sort of loads do you carry?

16) What are the heaviest loads you have carried in a backpack?

17) Do you have experience of carrying loads in a doublepack? Yes No

If yes, please answer questions 17 – 16. If no, you have finished the questionnaire. Thank you for your assistance.

18) How old were you when you first carried loads in a doublepack? ____________________

19) How old were you when last carried loads in a backpack? ________________________

20) How often did you carry loads in a doublepack as a young child (under 5 years of age) and what sort of loads did you carry?

____
____
____
____
____

21) How often did you carry loads in a doublepack as a child (age 6-10) and what sort of loads did you carry?

____
____
____
____
____

22) How often did you carry loads in a doublepack between the ages of 11 and 17 and what sort of loads did you carry?

____
____
____
____
____

23) If you are still carrying loads in a doublepack, how often do you do this and what sort of loads do you carry?

____
____
____
____
____

24) What are the heaviest loads you have carried in a doublepack?

____
____
____
____
____

Thank you for your assistance

Appendix J: Treadmill verification data for the study in Chapter 4

Unloaded (nobody walking)

	Time (s)					
Speed (km/h)	Trial 1	Trial 2	Trial 3	Mean	Actual Speed (m/s)	Actual Speed (km/h)
3	79.09	78.97	78.87	78.98	0.84	3.04
4	58.82	58.79	58.84	58.82	1.13	4.08
5	47.07	47.06	47.09	47.07	1.41	5.09
6	39.09	39.09	39.03	39.07	1.70	6.14

Loaded

Total mass = 84kg body mass + 20kg Backpack

	Time (s)					
Speed (km/h)	Trial 1	Trial 2	Trial 3	Mean	Actual Speed (m/s)	Actual Speed (km/h)
3	82.14	81.72	81.44	81.77	0.81	2.93
4	59.71	59.87	59.62	59.73	1.11	4.01
5	47.38	47.42	47.49	47.43	1.40	5.06
6	40.03	40.03	40.01	40.02	1.66	5.99

Unloaded Repeated (nobody walking)

	Time (s)					
Speed (km/h)	Trial 1	Trial 2	Trial 3	Mean	Mean Actual Speed (m/s)	Actual Speed (km/h)
3	79.06	79.09	79.13	79.09	0.84	3.03
4	59.93	58.87	58.94	59.25	1.12	4.05
5	47.09	47.07	47.12	47.09	1.41	5.09
6	39.16	39.09	39.12	39.12	1.70	6.13

Loaded Repeated

Total mass = 84.3kg body mass + 20kg Backpack

	Time (s)					
Speed (km/h)	Trial 1	Trial 2	Trial 3	Mean	Actual Speed (m/s)	Actual Speed (km/h)
3	81.63	81.53	81.53	81.56	0.82	2.94
4	59.56	59.72	59.93	59.74	1.11	4.01
5	47.71	47.52	47.54	47.59	1.40	5.04
6	39.88	39.71	39.75	39.78	1.67	6.03

Appendix K: Backpack device specifications used for the research in Chapter 4 and Chapter 7

Chapter 4

FEATHERLITE FREEDOM S, M

FABRICS

- Main 100D nylon Robic ripstop
- Contrast 210D nylon Ripstop
- Bottom 500D nylon Kodura
- Liner 40D nylon Ripstop

CM: W 37 D 25 H 55 (S) 65 (M)

LITRES: S 50+12 M 55+12

KILOGRAMS
Balance Pack S 1.58 (No liners) 1.73 (+ liners),
Balance Pack M 1.65 (No liners) 1.82 (+ liners),
Backpack without Balance Pockets S 1.23 (No liners) 1.31 (+
Backpack without Balance Pockets M 1.29 (No liners) 1.37 (+

LOAD RANGE: 18 kg

COLOUR: Green / Grey

CODE: FF-S, FF-L

Product Info | Returns

Karrimor Jura 35 Rucksack

The Karrimor Jura 35 Rucksack with a main compartment, lower compartment and fixed side pockets and lid pockets for plenty of storage, whilst the WindTunnel allows for a cooler and comfortable fit.

> Rucksack
> Capacity: 35L
> Back System: Windtunnel
> Contact mesh: Coolmesh
> Access: Zip access to base compartment
> Dividers: Zip out compartment divider
> Pockets: Fixed side pockets, lid pocket
> Hydration system compatible
> Fabric: KS p300BRS
> Weight: 850g
> Raincover: included
> H53 x W28 x D14cm (approx)
> Karrimor branding
> Wipe clean with a damp cloth

For our full range of View All Outdoor visit Karrimor

Specifications

Aeon ND25	Size: 25
Back Length	14-18" / 36-46cm
Dimensions	51 x 26 x 21cm
Volume	25lt / 1525cu.in
Weight	0.84kg / 1lb 14oz

Appendix L: Residual analysis data

Table 39. Estimated cut-off frequencies from a residual analysis conducted for the research in Chapter 4 on data from two participants with each load mass at 3 km·h^{-1} (0 kg, 7 kg and 20 kg).

		Cut-off frequency (Hz)					
Segment	Axis	P1 0 kg	P2 0 kg	P1 7 kg	P2 7 kg	P1 20 kg	P2 20 kg
Head Centre	X	6	6	5	6	6	6
	Y	6	6	6	5	6	6
Left Shoulder	X	6	6	5	6	6	5
	Y	5	6	6	6	5	7
Left Hip	X	6	6	6	6	6	6
	Y	6	6	6	6	6	6
Left Knee	X	6	5	6	6	5	6
	Y	6	6	6	6	6	6
Left Ankle	X	6	6	6	6	5	6
	Y	7	6	6	6	6	6
Left Foot Tip	X	6	6	6	7	6	6
	Y	7	7	6	7	7	6

* P = Participant

Table 40. Estimated cut-off frequencies from a residual analysis conducted for the research in Chapter 5 on data from three participants with each loading method and a range of loads (3 kg, 9 kg and 20 kg).

		Back cut-off frequency (Hz)			Back/front cut-off frequency (Hz)			Head cut-off frequency (Hz)		
Segment	Axis	P13 3 kg	P8 9 kg	P4 20 kg	P13 3 kg	P8 9 kg	P4 20 kg	P13 3 kg	P8 9 kg	P4 20 kg
Head Centre	X	5	5	6	5	5	6	5	5	5
	Y	6	6	6	6	6	6	6	6	6
Right Shoulder	X	5	6	6	6	6	6	6	6	6
	Y	5	6	6	6	6	6	6	6	6
Right Elbow	X	6	6	5	6	6	6	6	6	6
	Y	6	6	6	6	6	6	6	6	6
Right Wrist	X	6	6	6	7	6	6	6	6	6
	Y	6	6	6	6	6	6	6	6	6
Right Finger	X	6	6	6	7	6	6	6	6	5
	Y	7	6	6	7	6	6	6	6	6
Right Hip	X	6	6	6	5	6	5	5	6	5
	Y	6	6	6	6	6	6	6	6	5
Right Knee	X	6	6	6	5	6	5	5	6	6
	Y	6	6	6	5	6	6	6	6	6
Right Ankle	X	6	6	6	5	6	6	5	6	6
	Y	6	6	7	6	6	6	6	7	7
Right Foot Tip	X	6	6	6	5	6	6	6	6	6
	Y	7	6	7	6	6	6	6	7	7

* P = Participant

Table 41. Estimated cut-off frequencies from a residual analysis conducted for the research in Chapter 7 on data from three participants with each loading method and each load mass (3 kg, 12 kg and 20 kg).

		Unloaded walking cut-off frequency (Hz)			Back cut-off frequency (Hz)			Back/front cut-off frequency (Hz)			Head cut-off frequency (Hz)		
Segment	Axis	P1	P2	P3	P1 3 kg	P2 12 kg	P3 20 kg	P1 3 kg	P2 12 kg	P3 20 kg	P1 3 kg	P2 12 kg	P3 20 kg
Head	X	5	5	6	6	6	6	6	5	6	5	5	6
	Y	5	6	6	6	6	6	6	6	6	6	6	6
Right upper arm	X	6	7	6	6	5	6	6	6	5	6	6	6
	Y	6	6	5	6	6	6	6	6	6	6	6	6
Right lower arm	X	6	6	6	6	6	6	6	7	6	6	6	6
	Y	6	6	6	6	6	6	7	6	6	6	6	6
Right hand	X	6	6	5	6	4	6	5	6	5	6	6	6
	Y	6	6	6	7	6	6	6	6	6	7	6	5
Left upper arm	X	6	7	6	6	6	6	6	6	7	5	6	6
	Y	6	6	6	6	5	6	6	6	6	6	6	6
Left lower arm	X	5	6	6	6	6	6	6	6	6	6	6	6
	Y	6	6	7	6	6	6	6	7	7	6	6	6
Left hand	X	6	6	6	6	6	6	6	6	6	5	6	6
	Y	5	6	7	6	6	6	7	7	7	6	6	6
Thorax	X	6	6	5	6	6	6	4	6	6	4	5	5
	Y	6	6	6	6	5	6	6	6	6	6	6	6

Pelvis	X	5	6	5	5	6	6	6	6	6	6	6	6
	Y	6	6	6	5	6	6	6	6	6	6	5	6
Right thigh	X	6	6	6	5	6	6	6	6	6	6	6	6
	Y	6	6	6	6	6	6	6	6	6	6	6	6
Right shank	X	6	6	6	6	6	6	5	6	6	6	5	6
	Y	6	6	7	6	6	6	6	7	6	8	6	6
Right foot	X	6	6	6	6	6	5	6	6	5	6	6	6
	Y	7	6	7	6	6	6	7	7	6	7	6	6
Left thigh	X	6	6	6	6	4	6	6	6	6	5	6	6
	Y	6	6	6	6	6	6	6	6	6	6	6	6
Left shank	X	6	6	6	7	6	6	6	6	6	7	6	6
	Y	6	6	6	7	6	7	6	6	6	8	6	6
Left foot	X	6	6	6	6	6	6	6	6	6	6	5	6
	Y	7	6	7	7	7	6	7	6	6	8	6	6

* P = Participant

Appendix M: Chapter 4 intra-observer digitising reliability tables

Table 42. Intra-observer reliability for manual digitisation of trunk angle at heel-strike and toe-off for a single participant walking with 20 kg.

	Measurement number									
	1	2	3	4	5	6	7	8	9	10
Trunk angle at heel strike (˚)										
1st measurement	80.2	80.4	81.0	80.2	79.1	80.2	80.4	80.7	79.8	78.9
2nd measurement	81.0	80.3	79.8	79.9	80.6	80.3	80.6	81.1	79.9	80.0
Deviations	-0.8	0.1	1.2	0.3	-1.5	-0.1	-0.2	-0.4	-0.1	-1.1
$(\text{Deviations})^2$	0.6	0.0	1.4	0.1	2.3	0.0	0.0	0.2	0.0	1.2
$\sum(\text{Deviations})^2$	5.9									
Absolute TEM	0.5									
VAV	80.2									
Relative TEM %	0.7									
Trunk angle at toe off (˚)										
1st measurement	81.3	80.8	80.5	81.2	81.5	80.9	81	80.7	81.5	81
2nd measurement	80.9	80.2	79.7	80.5	81.1	80.6	80.3	81	80.7	80.4
Deviations	0.4	0.6	0.8	0.7	0.4	0.3	0.7	-0.3	0.8	0.6
$(\text{Deviations})^2$	0.16	0.36	0.64	0.49	0.16	0.09	0.49	0.09	0.64	0.36
$\sum(\text{Deviations})^2$	3.5									
Absolute TEM	0.4									
VAV	80.8									
Relative TEM %	0.5									

TEM = Technical error of measurement; ∑ = summation; VAV = Variable average value

Table 43. Intra-observer reliability for manual digitisation of hip angle at heel-strike and toe-off for a single participant walking with 20 kg.

	Measurement number									
	1	2	3	4	5	6	7	8	9	10
Hip angle at heel strike (˚)										
1st measurement	155.1	155.8	155.7	155.7	155.3	155	155.1	155.8	155.6	155.6
2nd measurement	155.3	155.8	155.1	155.9	155.8	154.9	155.6	155.7	155.9	155.9
Deviations	-0.2	0	0.6	-0.2	-0.5	0.1	-0.5	0.1	-0.3	-0.3
$(\text{Deviations})^2$	0.04	0	0.36	0.04	0.25	0.01	0.25	0.01	0.09	0.09
$\sum(\text{Deviations})^2$	1.1									
Absolute TEM	0.2									
VAV	155.5									
Relative TEM %	0.2									
Hip angle at toe off (˚)										
1st measurement	165.6	165.8	165.9	166.4	166.2	166.8	166.3	166.5	165.9	166.1
2nd measurement	166.2	166.3	166.6	165.8	166	166.4	166	165.9	166.6	166.8
Deviations	-0.6	-0.5	-0.7	0.6	0.2	0.4	0.3	0.6	-0.7	-0.7
$(\text{Deviations})^2$	0.36	0.25	0.49	0.36	0.04	0.16	0.09	0.36	0.49	0.49
$\sum(\text{Deviations})^2$	3.1									
Absolute TEM	0.4									
VAV	166.2									
Relative TEM %	0.2									

TEM = Technical error of measurement; ∑ = summation; VAV = Variable average value

Table 44. Intra-observer reliability for manual digitisation of knee angle at heel-strike and toe-off for a single participant walking with 20 kg.

	Measurement number									
	1	2	3	4	5	6	7	8	9	10
Knee angle at heel strike (˚)										
1st measurement	162	162	161.8	161.5	162	162.3	161.7	162.1	162.3	161.9
2nd measurement	162.8	162.3	162.3	162.4	162.8	162.1	162.9	162.5	162	162.8
Deviations	-0.8	-0.3	-0.5	-0.9	-0.8	0.2	-1.2	-0.4	0.3	-0.9
$(\text{Deviations})^2$	0.64	0.09	0.25	0.81	0.64	0.04	1.44	0.16	0.09	0.81
$\sum(\text{Deviations})^2$	4.97									
Absolute TEM	0.50									
VAV	162.2									
Relative TEM %	0.31									
Knee angle at toe off (˚)										
1st measurement	116.1	116.4	116.7	116.8	116.8	116.4	116.9	116	117.1	116.1
2nd measurement	115.8	115.9	115.7	116.2	116.9	116.8	116.7	116.8	116.3	116.2
Deviations	0.3	0.5	1	0.6	-0.1	-0.4	0.2	-0.8	0.8	-0.1
$(\text{Deviations})^2$	0.09	0.25	1	0.36	0.01	0.16	0.04	0.64	0.64	0.01
$\sum(\text{Deviations})^2$	3.2									
Absolute TEM	0.4									
VAV	116.4									
Relative TEM %	0.3									

TEM = Technical error of measurement; ∑ = summation; VAV = Variable average value

Table 45. Intra-observer reliability for manual digitisation of Ankle angle at heel-strike and toe-off for a single participant walking with 20 kg.

	Measurement number									
	1	2	3	4	5	6	7	8	9	10
Ankle angle at heel strike (˚)										
1st measurement	118.1	118.4	117.7	117.8	118	118.5	118.4	117.7	118.2	118.1
2nd measurement	117.9	117.7	117.7	117.7	118.1	118.4	118.2	118.3	118.2	118.1
Deviations	0.2	0.7	0	0.1	-0.1	0.1	0.2	-0.6	0	0
$(\text{Deviations})^2$	0.04	0.49	0	0.01	0.01	0.01	0.04	0.36	0	0
$\sum(\text{Deviations})^2$	1.0									
Absolute TEM	0.2									
VAV	118.1									
Relative TEM %	0.2									
Ankle angle at toe off (˚)										
1st measurement	124.6	124.8	124.7	124.9	125.1	125	124.7	124.5	124.9	125
2nd measurement	125.3	125.5	124.8	124.7	124.8	125.1	124.9	124.8	124.9	125.1
Deviations	-0.7	-0.7	-0.1	0.2	0.3	-0.1	-0.2	-0.3	0	-0.1
$(\text{Deviations})^2$	0.49	0.49	0.01	0.04	0.09	0.01	0.04	0.09	0	0.01
$\sum(\text{Deviations})^2$	1.3									
Absolute TEM	0.3									
VAV	124.9									
Relative TEM %	0.2									

TEM = Technical error of measurement; ∑ = summation; VAV = Variable average value

Appendix N: Chapter 5 intra-observer digitising reliability

Table 46. Intra-observer reliability for manual digitisation for the research in Chapter 5. Measurement reliability data is for trunk angle at heel-strike and toe-off for participant 1 in the back-loading method with 20 kg.

	Measurement number									
	1	2	3	4	5	6	7	8	9	10
Trunk angle at heel-strike (˚)										
1st measurement	82.1	82.5	82.5	82.5	82.5	82.5	82.1	82.1	82.1	82.1
2nd measurement	83.2	82.8	82.7	83.2	83.2	83.2	83.2	83.2	82.8	82.8
Deviations	-1.1	-0.3	-0.2	-0.7	-0.7	-0.7	-1.1	-1.1	-0.7	-0.7
$(\text{Deviations})^2$	1.21	0.09	0.04	0.49	0.49	0.49	1.21	1.21	0.49	0.49
$\sum(\text{Deviations})^2$	6.21									
Absolute TEM	0.56									
VAV	82.67									
Relative TEM %	0.67									
Trunk angle at toe off (˚)										
1st measurement	83	83	82.9	83.3	83.4	83.4	83.4	83.4	83.5	83.5
2nd measurement	83.2	82.9	83	83	83	82.8	82.9	82.9	82.9	83
Deviations	-0.2	0.1	-0.1	0.3	0.4	0.6	0.5	0.5	0.6	0.5
$(\text{Deviations})^2$	0.04	0.01	0.01	0.09	0.16	0.36	0.25	0.25	0.36	0.25
$\sum(\text{Deviations})^2$	1.78									
Absolute TEM	0.30									
VAV	83.12									
Relative TEM %	0.36									

TEM = Technical error of measurement; ∑ = summation; VAV = Variable average value

Table 47. Intra-observer reliability for manual digitisation for the research in Chapter 5. Measurement reliability data is for knee angle at heel-strike and toe-off for participant 1 in the back-loading method with 20 kg.

	Measurement number									
	1	2	3	4	5	6	7	8	9	10
Knee angle at heel-strike (˚)										
1st measurement	161.7	160	160.7	161.7	162.2	162	162.6	161.6	161.6	161.7
2nd measurement	162.2	162.2	161.6	161.6	161.1	161.6	161.6	161.4	161.6	161.6
Deviations	-0.5	-2.2	-0.9	0.1	1.1	0.4	1	0.2	0	0.1
$(\text{Deviations})^2$	0.25	4.84	0.81	0.01	1.21	0.16	1	0.04	0	0.01
$\sum(\text{Deviations})^2$	8.33									
Absolute TEM	0.65									
VAV	161.62									
Relative TEM %	0.40									
Knee angle at toe off (˚)										
1st measurement	132.8	133.1	133.5	133.8	133.3	134	133.7	133.2	133.2	133.1
2nd measurement	133.5	132.9	132.5	133.2	133.2	133.6	133.2	133.6	133.6	133.2
Deviations	-0.7	0.2	1	0.6	0.1	0.4	0.5	-0.4	-0.4	-0.1
$(\text{Deviations})^2$	0.49	0.04	1	0.36	0.01	0.16	0.25	0.16	0.16	0.01
$\sum(\text{Deviations})^2$	2.64									
Absolute TEM	0.36									
VAV	133.31									
Relative TEM %	0.27									

TEM = Technical error of measurement; ∑ = summation; VAV = Variable average value

Table 48. Intra-observer reliability for manual digitisation for the research in Chapter 5. Measurement reliability data is for ankle angle at heel-strike and toe-off for participant 1 in the back-loading method with 20 kg.

	Measurement number									
	1	2	3	4	5	6	7	8	9	10
Ankle angle at heel-strike (˚)										
1st measurement	111.1	111.8	111.8	110.8	111.2	111.5	110.7	111.3	110.9	111.3
2nd measurement	110.7	110.9	111.1	110.7	111.5	111.1	111.4	110.2	111.6	111.4
Deviations	0.4	0.9	0.7	0.1	-0.3	0.4	-0.7	1.1	-0.7	-0.1
$(\text{Deviations})^2$	0.16	0.81	0.49	0.01	0.09	0.16	0.49	1.21	0.49	0.01
$\sum(\text{Deviations})^2$	3.92									
Absolute TEM	0.44									
VAV	111.15									
Relative TEM %	0.40									
Ankle angle at toe off (˚)										
1st measurement	128.4	127.8	127.8	128	128.1	128.4	128.2	128.4	128.4	127.9
2nd measurement	128.7	128.5	128.5	128.5	128.1	127.9	128.3	128	128.5	128.2
Deviations	-0.3	-0.7	-0.7	-0.5	0	0.5	-0.1	0.4	-0.1	-0.3
$(\text{Deviations})^2$	0.09	0.49	0.49	0.25	0	0.25	0.01	0.16	0.01	0.09
$\sum(\text{Deviations})^2$	1.84									
Absolute TEM	0.30									
VAV	128.23									
Relative TEM %	0.24									

TEM = Technical error of measurement; ∑ = summation; VAV = Variable average value

Appendix O: Modifications to the full body plug-in gait used for the research in Chapter 7

Table 49. A summary of the modifications made to the full body plug-in gait marker set

Marker position	Full body Plug-in Gait Model	Modification	Justification
Upper Arm	A single marker on the upper lateral surface of the right and left upper arms.	A three marker non-collinear cluster on the upper lateral surface of the right and left upper arms.	To improve segment tracking
Lower Arm	A single marker on the lower lateral surface of the right and left forearm	A three marker non-collinear cluster on the upper lateral surface of the right and left forearms.	To improve segment tracking
Pelvis	Markers on the ASIS, PSIS and Sacrum	Markers on the ASIS, PSIS, sacrum and a three marker non-collinear cluster on the iliac crest.	To improve segment tracking with load carriage devices that include a hip belt
Thigh	A single marker on the lower lateral surface of the right and left thigh	A three marker non-collinear cluster on the lower lateral surface of the right and left thigh	To improve segment tracking

Tibia	A single marker on the lower lateral surface of the right and left shank	A three marker non-collinear cluster on the lower lateral surface of the right and left shank	To improve segment tracking
Knee	A marker on the lateral flexion-extension axis of the left and right knee.	A marker on the lateral and medial flexion-extension axis of the left and right knee.	To improve joint centre location identification
Ankle	A marker on the lateral malleolus of the left and right ankle.	A marker on the lateral and medial malleoli of the left and right ankle.	To improve joint centre location identification
Toe	A marker on second metatarsal head	A marker on the first and fifth metatarsal heads	To improve segment definition

Appendix P: Between- and within- participant variation for spatiotemporal, joint angle and ground reaction force variables from the research in Chapter 7

Table 50. Mean, between-participant standard deviation (SDb) and within-participant standard deviation (SDw) for step parameters for each load carriage condition (* indicates were SDw values were greater than SDb)

		Head				Back				Back/Front			
Step Parameter		0kg	3kg	12kg	20kg	0kg	3kg	12kg	20kg	0kg	3kg	12kg	20kg
Step length	Mean	0.54	0.53	0.53	0.52	0.54	0.53	0.54	0.55	0.53	0.53	0.53	0.54
	SDb	0.03	0.03	0.03	0.04	0.03	0.03	0.04	0.04	0.03	0.03	0.03	0.03
	SDw	0.02	0.03	0.03	**0.04***	0.03	**0.04***	0.04	**0.06***	0.03	**0.03***	0.03	**0.03***
Cadence	Mean	1.56	1.58	1.59	1.61	1.56	1.56	1.54	1.53	1.56	1.57	1.57	1.56
	SDb	0.08	0.09	0.10	0.12	0.09	0.09	0.09	0.10	0.08	0.08	0.09	0.07
	SDw	0.03	0.03	0.04	**0.18***	0.03	0.03	0.03	0.05	0.03	0.03	0.03	**0.10***
Step width	Mean	0.14	0.15	0.16	0.16	0.15	0.14	0.14	0.15	0.14	0.14	0.14	0.14
	SDb	0.02	0.03	0.03	0.02	0.02	0.02	0.02	0.01	0.01	0.01	0.02	0.02
	SDw	0.01	0.01	0.01	0.01	0.01	0.01	0.01	**0.02***	0.01	0.01	0.01	0.01
Step time	Mean	0.64	0.64	0.63	0.62	0.64	0.64	0.65	0.65	0.64	0.64	0.64	0.64
	SDb	0.03	0.04	0.04	0.05	0.03	0.04	0.04	0.04	0.03	0.03	0.04	0.04
	SDw	0.02	0.02	0.02	0.03	0.02	0.02	0.02	**0.06***	0.02	0.02	0.02	0.02
Double stance time	Mean	0.46	0.45	0.44	0.43	0.45	0.46	0.46	0.45	0.45	0.45	0.44	0.44
	SDb	0.03	0.03	0.03	0.03	0.03	0.03	0.03	0.03	0.03	0.03	0.03	0.02
	SDw	0.02	0.02	0.03	0.03	0.02	0.02	0.03	0.03	0.02	0.02	0.02	**0.03***
Single stance time	Mean	0.18	0.19	0.20	0.20	0.19	0.19	0.20	0.20	0.19	0.19	0.20	0.21
	SDb	0.02	0.01	0.02	0.02	0.02	0.02	0.02	0.01	0.02	0.02	0.02	0.02
	SDw	**0.02***	**0.02***	**0.02***	**0.02***	0.02	**0.02***	**0.04***	**0.02***	0.02	**0.02***	0.02	0.02

Table 51. Mean, between-participant standard deviation (SDb) and within-participant standard deviation (SDw) for peak sagittal plane joint angles of the trunk, hip, knee and ankle for each load carriage condition.

Peak Joint Angles		Head				Back				Back/Front			
		0kg	3kg	12kg	20kg	0kg	3kg	12kg	20kg	0kg	3kg	12kg	20kg
Peak Trunk Flex (º)	Mean	3.03	-3.23	-4.58	-5.35	2.75	3.70	6.43	9.39	2.98	3.71	5.20	5.82
	SDb	2.82	2.80	3.51	3.55	2.40	3.06	3.10	3.87	2.57	2.78	2.61	3.50
	SDw	0.70	0.70	0.88	0.83	0.74	1.01	0.71	0.94	0.70	0.70	0.68	0.72
Peak Trunk Ext (º)	Mean	-1.12	-7.23	-8.73	-9.59	-1.17	-0.75	1.65	4.30	-1.24	-0.30	0.86	1.22
	SDb	2.64	2.62	3.27	3.43	2.44	3.11	2.96	3.62	2.55	2.67	2.53	3.21
	SDw	1.01	0.83	0.97	1.11	0.89	0.88	0.91	1.01	0.89	0.80	0.83	0.99
Peak Hip Flex (º)	Mean	22.72	16.42	15.45	15.28	23.10	24.57	28.16	31.31	22.86	23.98	26.26	27.71
	SDb	3.37	2.92	3.42	3.84	2.91	2.94	3.26	4.18	3.07	3.25	2.94	3.66
	SDw	1.37	1.18	1.41	1.57	1.24	1.63	1.33	1.48	1.35	1.51	1.50	1.64
Peak Hip Ext (º)	Mean	-14.74	-20.88	-22.22	-22.90	-14.06	-14.23	-12.72	-10.88	-14.86	-13.94	-13.17	-13.32
	SDb	3.88	3.40	3.93	4.35	3.96	4.71	4.08	4.50	3.37	3.94	3.55	4.56
	SDw	1.67	1.64	1.95	2.27	1.96	1.73	1.84	2.39	1.69	1.61	1.73	2.19
Peak Knee Flex (º)	Mean	54.04	54.23	55.23	55.98	55.22	55.62	56.18	56.61	54.75	55.15	55.70	56.35
	SDb	4.51	3.66	3.25	3.49	4.22	4.74	4.54	4.35	4.43	4.43	4.16	4.43
	SDw	2.14	2.01	2.40	2.41	2.12	2.11	2.24	2.28	2.14	2.25	2.44	2.42
Peak Knee Ext (º)	Mean	-0.61	-0.47	0.17	1.13	0.27	0.35	0.71	1.19	-0.13	-0.09	0.34	0.46
	SDb	2.77	2.72	2.85	2.87	3.47	3.42	2.98	3.18	3.24	3.14	3.34	3.18
	SDw	1.60	1.62	1.80	1.88	1.46	1.75	1.77	2.03	1.39	1.45	1.68	2.18
Peak Ankle Flex (º)	Mean	9.55	9.81	10.05	10.32	9.47	9.48	9.35	9.21	9.69	10.26	10.00	10.12
	SDb	2.52	2.60	2.66	2.54	2.52	2.31	2.50	2.80	2.08	2.62	2.18	2.24
	SDw	1.51	1.48	1.51	1.75	1.31	1.44	1.48	1.57	1.64	1.58	1.70	1.82
Peak Ankle Ext (º)	Mean	-16.14	-15.36	-15.41	-16.16	-15.38	-15.94	-16.71	-17.45	-15.65	-15.23	-15.64	-15.84
	SDb	6.29	6.17	5.41	5.57	5.29	5.97	5.44	5.82	4.93	4.95	4.78	5.02
	SDw	3.31	2.60	3.07	3.35	3.13	3.20	3.02	3.20	3.15	3.23	3.80	3.35

Table 52. Mean, between-participant standard deviation (SDb) and within-participant standard deviation (SDw) for antero-posterior, vertical and medio-lateral ground reaction forces (GRF) for each load carriage condition (* indicates were SDw values were greater than SDb).

		Head				Back				Back/Front			
GRF variable		0kg	3kg	12kg	20kg	0kg	3kg	12kg	20kg	0kg	3kg	12kg	20kg
Braking (N)	Mean	76.44	83.79	95.04	104.38	75.55	80.36	93.46	106.48	78.34	82.53	93.10	103.91
	SDb	14.90	17.41	17.90	17.69	14.01	14.80	17.02	17.41	16.88	17.40	18.82	18.15
	SDw	6.88	7.27	10.95	12.49	9.34	7.83	10.64	11.68	8.10	10.20	11.65	10.90
Propulsive (N)	Mean	-97.84	-102.62	-112.51	-122.99	-99.52	-104.57	-119.02	-131.74	-99.52	-102.69	-115.87	-126.96
	SDb	13.32	14.40	11.49	14.62	12.33	11.83	12.39	11.99	12.42	13.72	12.60	13.50
	SDw	8.43	9.45	**13.51***	13.92	7.07	10.41	11.23	**12.53***	6.97	7.79	9.36	12.23
1st Vertical Peak (N)	Mean	743.68	775.24	854.24	924.11	740.15	772.37	859.02	926.99	738.71	767.17	850.57	929.59
	SDb	99.69	101.29	95.12	109.45	98.40	104.76	110.45	107.66	95.31	96.48	103.84	104.30
	SDw	13.00	12.90	16.02	15.62	13.94	14.91	17.43	20.78	12.92	14.63	18.17	22.15
Force minimum (N)	Mean	657.32	686.57	775.72	839.62	654.90	677.98	759.45	830.57	654.93	674.50	753.09	828.55
	SDb	96.24	89.73	86.95	92.21	91.35	89.03	88.59	88.67	88.81	88.57	88.78	83.79
	SDw	8.61	8.67	12.97	13.78	10.32	11.01	13.04	19.49	10.56	9.67	13.98	18.31
2nd Vertical Peak (N)	Mean	772.55	792.12	878.09	948.56	773.40	807.27	894.68	967.55	770.81	802.30	889.66	971.31
	SDb	99.58	100.48	94.91	104.94	103.11	104.21	101.69	104.51	104.21	102.25	103.01	102.19
	SDw	13.95	15.18	16.83	20.50	12.06	14.11	18.07	24.40	17.22	14.80	18.76	23.05
Medial (N)	Mean	50.07	52.48	59.38	62.68	51.44	51.12	53.45	61.27	46.95	48.56	53.41	55.84
	SDb	14.18	12.47	13.58	13.01	14.25	14.55	15.44	16.41	14.48	14.47	16.91	16.12
	SDw	3.08	3.45	4.51	4.43	3.25	3.40	3.24	3.78	3.18	3.47	3.88	4.02
Lateral (N)	Mean	12.28	14.91	13.92	13.53	10.54	12.97	12.82	12.04	15.99	16.84	16.95	19.55
	SDb	5.36	6.51	5.26	4.23	3.72	4.76	5.23	5.01	10.43	10.87	8.48	11.76
	SDw	4.73	4.95	**5.85***	**7.22***	**4.08***	**4.78***	**7.16***	**6.73***	4.55	5.44	5.31	6.02