UNIVERSITY OF SOUTHAMPTON

FACULTY OF HEALTH SCIENCES

Impact of Anterior Load Carriage on Muscle Fatigue, Gait Kinematics and Pelvis-Trunk Coordination

by

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Thesis for the degree of Doctor of Philosophy

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ABSTRACT

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Background: Carrying is one of the most frequently performed activities of daily living, particularly in industrial settings. However, as the carrying activity involves a direct exposure to several biomechanical mechanisms that can potentially affect the musculoskeletal function of the lower back, a prolonged use of the activity may lead to low back pain (LBP). Therefore, the aim of this doctoral study was to investigate the impact of anterior load carriage on muscle fatigue, spatiotemporal parameters and 3D kinematics of gait, and pelvis-trunk coordination. **Methods:** This cross-sectional study involved 37 healthy people; 20 sedentary individuals and 17 manual workers. All participants were instructed to perform an isometric back endurance test and two gait conditions: 1) standard gait and 2) carrying gait whilst carrying a safe-maximum load (max-kg gait). The spatiotemporal parameters and 3D kinematics of gait, as well as the pelvis-trunk coordination during the activity were measured using the Vicon Motion Analysis System. The muscle fatigue during the activity was measured based on the slope of median frequency (MFslope) of surface electromyography (EMG). The differences between the sedentary and the manual groups in all parameters were examined. **Results**: During the Ito test, there was no significant difference in the isometric back endurance and muscle fatigue between the groups. During the carrying activity, the manual group was able to carry 4 kg heavier maximum load compared to the sedentary group. For the gait parameters, there was a significant effect of load condition (standard gait to max-kg gait) on cadence (increased by 10 steps/minute), stride length (reduced by 4 cm) and stride time (reduced by 0.09 seconds). For the gait kinematics, there was a significant effect of load condition during stance phase on the range of motion (ROM) of ankle (flexion-extension), left hip (flexion-extension, with significant interaction effect), pelvic tilt, left pelvic axial rotation and all trunk movements. For the pelvis-trunk coordination, there was a significant increase in the percentage of in-phase coordination across the activity in flexion-extension (11% increase in the manual group only), lateral flexion (10% increase in the manual group, 5% increase in the sedentary group but on the right side only) and axial rotation (23% increase on the right side only). **Conclusion**: The results suggested that both manual and sedentary groups demonstrated a similar pattern of changes in spatiotemporal parameters regardless of their maximum carrying load. However, the differences in gait kinematics and pelvis-trunk coordination between the groups may indicate common body strategies to adapt with a safe-maximum load limit. Therefore, this study had established a biomechanical baseline of anterior load carriage among healthy population that can be used to guide further research investigation on specific group of people or patients as a part of functional capacity evaluation.

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DECLARATION OF AUTHORSHIP

I, Hanif Farhan Bin Mohd Rasdi, declare that this thesis and the work presented in it are my own and has been generated by me as the result of my own original research.

Title: Impact of Anterior Load Carriage with Progressive Loads on Muscle Fatigue, Gait Kinematics and Pelvis-Trunk Coordination

I confirm that:

1. This work was done wholly or mainly while in candidature for a research degree at this University;
2. Where any part of this thesis has previously been submitted for a degree or any other qualification at this University or any other institution, this has been clearly stated;
3. Where I have consulted the published work of others, this is always clearly attributed;
4. Where I have quoted from the work of others, the source is always given. With the exception of such quotations, this thesis is entirely my own work;
5. I have acknowledged all main sources of help;
6. Where the thesis is based on work done by myself jointly with others, I have made clear exactly what was done by others and what I have contributed myself;
7. Parts of this work have been published as:

   Hanif Farhan, M.R., White, P.J., Warner, M.B. & Adams, J.E. 2015. The relationship between carrying activity and low back pain: A critical review of biomechanics studies. *13*(2). 1-25. Malaysian Journal of Health Sciences.

Signed: ....

Date: 3rd April 2018

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DEFINITIONS AND ABBREVIATIONS

AP	Anti-phase
BOS	Base of support
C7	The spinous process of 7^{th} cervical vertebrae
CM	Centimetre
CMC	Coefficient of multiple correlation
COM	Centre of mass
CVA	Cerebrovascular Accident
D	Standardized mean difference (Cohen's *d)*
D	Mean difference
Df	Degree of freedom
EMG	Electromyography
FCE	Functional capacity evaluation
IJ	Incisura jugularis (suprasternal notch)
IP	In-phase
IqR	Interquartile range
KG	Kilogram
LBP	Low back pain
M	Mean
MATLAB	Matrix Laboratory
Mdn	Median
Min.	Minutes
MMH	Manual material handling
OT	Occupational therapy
P	Value of type-I error
PO	Pelvis only

PX	Processes xiphoidens (xiphoid process)
R	Correlation coefficient
R^2	Coefficient of determination
$R5^{th}MET/L5^{th}MET$	Right/left 5^{th} metatarsal (lateral, base of 5^{th} metatarsal)
RANK/LANK	Right/left ankle (lateral malleolus)
RASI, LASI	Right/left anterior suprailiac spine
RATHI/LATHI	Right/left anterior thigh (lower, anterior 1/3 of the thigh)
RHEE/LHEE	Right/left heel (calcaneous)
RILC/LILC	Right/left iliac spine (most lateral aspect)
RKNE/LKNE	Right/left knee (lateral femur epicondyle)
RLTHI/LLTHI	Right/left lateral thigh (lower lateral 2/3 of the thigh)
RMANK/LMANK	Right/left medial ankle (medial malleolus)
RMKNE/LMKNE	Right/left medial knee (medial femur epicondyle)
RMTHI/LMTHI	Right/left medial thigh (lower medial 1/3 of the thigh)
ROM	Range of motion
RPSI/LPSI	Right/left posterior suprailiac spine
RPTHI/LPTHI	Right/left posterior thigh (lower, posterior 1/3 of the thigh)
RTHI/LTHI	Right/left thigh (lower lateral 1/3 of the thigh)
RTIB/LTIB	Right/left tibia (lower lateral 1/3 of the shank)
RTOE/LTOE	Right/left toe (over 2^{nd} metatarsal head)
RTUB/LTUB	Right/left tibia tuberosity
RTW	Return-to-work

SD	Standard deviation
SPSS	Statistical Package for Social Science
T8	The spinous process of 8th thoracic vertebrae
TO	Trunk only
WHO	World Health Organization
WSSD	Within-subject standard deviation

Chapter 1: INTRODUCTION TO PHD THESIS

1.1 CARRYING ACTIVITY AND LOW BACK PAIN

Low back pain (LBP) had been reported to pose a large socio-economic impact on many countries (Collins et al. 2010). In the United Kingdom, it was estimated that the cost of informal care and work production losses was around £11 billion annually (Maniadakis and Gray 2000). Work absenteeism among workers with LBP had been reported to be approximately three times more than workers without LBP (Widanarko et al. 2012). In the industrial environment, manual material handling (MMH) was reported to be a serious contributing factor to LBP (Waters et al. 2006). Carrying objects is an activity within MMH and is one of the most frequently performed activities of daily living, particularly in the working environment. Commonly in industrial settings, the carrying activity is performed in conjunction with other physical activities, such as lifting, lowering, pushing and pulling, to produce a meaningful job circuit. These activities along with carrying have contributed to various types of back disorders (Kuiper et al. 1999). Studies had reported that carrying activities can possibly lead to low back pain. The possible mechanisms that may lead to LBP are consistent anterior force on the lower back as exerted by a back pack (LaFiandra et al. 2003), modification in spinal proprioception after load carriage (Hung-Kay Chow et al. 2011), decreased coordination variability in load carriage (Seay et al. 2011a) and increased paraspinal muscle activity (Healey et al. 2005b).

The main rehabilitation aim for workers with LBP is their return-to-work (RTW) in a timely and safe manner. One of the major assessments to determine the physical readiness for return-to-work is Functional Capacity Evaluation (FCE). The FCE can be defined as 'a systematic method of measuring an individual's ability to perform meaningful tasks on a safe and dependable basis' (Matheson 2003). It is regarded as the gold standard of vocational assessment (McFadden et al. 2010) and the major role of this assessment is to analyse the consistency between a patient's performance in work-related physical activities and the relevant job demands. In the FCE, each physical activity is commonly tested separately as different test protocols, the activities being chosen as a set of protocols related to the workers' relevant work demand. Being listed in the Dictionary of Occupational Title (United States Department of Labor 1991) as a musculoskeletal work demand, a carrying activity is primarily included as a test protocol in most of the FCE systems. In the RTW programs, the FCE is utilized as the most important assessment in order to evaluate the degree of physical functional limitation among the injured workers (McFadden et al. 2010). However, the readiness for RTW is a multifaceted concept in which it can also be influenced by psychosocial factors (Franche and Krause 2002). Although FCE cannot predict a sustained RTW, a high performance in FCE was reported to associate with a faster RTW (Gross and Battié 2004; Gross et al. 2004; Gross and Battié 2005). Furthermore, rather than being measured as a mainly physical assessment, it can also be conceptualized as a 'behavioural test' due to the fact that the performance in FCE should reflect not only the physical capabilities, but also self-efficacy (defined as 'the confidence in being able to carry out a set of specified activities') (Gross and Battié 2004). Therefore, FCE should be geared to be as relevant as possible to a person's job description (Chen 2007).

According to Isernhagen (1992), in general, there are two types of FCE: psychophysical and kinesiophysical. The psychophysical FCE is determined when the LBP patient is given the control to decide the 'maximum function' that can be performed during the FCE. Whereas, the administering therapist controls the decision in the kinesiophysical FCE. The maximum function can be described as 'the greatest safe ability of a client, either in repetitions or weight capacities' (Isernhagen 1992). In the kinesiophysical FCE, the signs of maximal function can be observed when there is a significant increase in heart rate, movement patterns become more deliberate, evidence of accessory muscle recruitment and/or changes in biomechanics. In reality, both FCE types can be combined in order to get the most accurate result to reflect the actual maximum function of a patient (Valpar International Corporation 2007). Both subjective input from the patient and objective input from the therapist are complementary, thus, are important in determining a 'valid' maximum function in FCE (often being referred as sincerity of effort) (Reneman et al. 2005). For instance, when any FCE protocols are terminated according to the psychophysical approach, it is hard to decide whether the maximum function is truly a maximal effort that can be handled by the worker. This is because maximal effort can be influenced by many factors such as pain, fear-avoidance beliefs, ambiguity in test instruction, lack of understanding in test importance, secondary financial gain and secondary emotional gain (Lechner et al. 1998). Therefore, both psychophysical and kinesiophysical approaches are equally important to determine a valid FCE performance before making any legal decision related to RTW.

In reality, the level of physical exertion during carrying activities among professions varies according to the nature of job requirement. For instance, in a cross-sectional survey among newly employed workers from 12 different occupational groups, Nahit et al. (2001) reported that among other occupational exposures (i.e. lifting, pulling, pushing, standing, driving, stretching below knee, bending and squatting), carrying activity had the strongest association with LBP (OR = 2.4, 95%CI = 1.5 to 3.8), as well as shoulder pain (OR = 3.1, 95% = 1.9 to 4.8) and knee pain (OR = 3.5 , 95%CI = 2.2 to 5.5). Commonly in the industrial setting, occupational carrying is performed in conjunction with other MMH activities as well to produce a meaningful job circuit. Although in FCE, each of different MMH activities are tested separately as different test protocols, the activities are chosen according to the relevant job requirement. For instance, an injured driver may undergo a set of FCE protocols which are not the same as a teacher or a soldier because of their different work demands. Therefore, each protocol should be standardized to ensure a good measurement quality across various types of professions. Most of the FCE systems rely on the construct of physical demands within the Dictionary of Occupational Title (DOT) (United States Department of Labor 1991) for its construct validity (King et al. 1998).

Being listed in DOT as one of the musculoskeletal work demands, carrying activity is primarily included as a test protocol in most of FCE systems such as Isernhagen Works System FCE (WorkWell^{TM}), ERGOS Work Simulator (ERGOS^{TM}), Physical Work Performance Evaluation (ErgoScience^{R}), Ergo-Kit FCE (Ergo-Kit^{R}), Blankenship System (Blankenship, Inc.), and Joule (VALPAR, Inc.) (Gouttebarge et al. 2004). For instance, according to the Isernhagern Work System FCE, for a short two-handed carrying protocol, respondents were instructed to lift a container at waist level, turn 90º, walk 1.2 meters, turn 90º, put the container on another table, and then return the container to its original place. The load within the container was increased progressively. Reneman et al. (2002) reported that this protocol had a good level of test-retest reliability (ICC = 0.77). Furthermore, Brouwer et al. (2003) reported that other than the short protocol, they had also tested the test-retest reliability for long two-handed, long left-handed and long right handed carrying protocols using the same FCE system. For long carrying protocols, this system also applied the same sequence as the short carrying protocol, but with an exception of a longer carrying distance (i.e. 20 meters). As for the result, they reported that all the carrying protocols had good reliability (i.e. ICC > 0.7 for all carrying protocols). Other than that, Tuckwell et al. (2002) also reported a good test-retest (Kappa = 0.75) and inter-rater reliability (Kappa = 0.62) for Physical Work Performance Evaluation. However, they did not specify the distance between each carrying circuit, which may have impact on the outcome of the FCE as the standardised protocols may not have been followed.

1.2 MUSCULOSKELETAL ADAPTATION TO FATIGUE

Across the literature, carrying activities are reported to associate with many neuro-musculoskeletal problems such as LBP, rucksack palsy, knee pain, foot blisters, and local fatigue (Knapik et al. 1996). One of the possible mechanism that can connect carrying activity with such problems is muscle fatigue. In general, muscle fatigue can be described as the failure to maintain the required or expected force (Enoka and Duchateau 2008). Across the literature, time to exhaustion and surface electromyography (EMG) are two common methods to measure muscle fatigue. For instance, the Biering-Sorensen test is one of the most common isometric back endurance tests (Biering-Sørensen 1984). In order to perform the test, the subject is instructed to perform back extension and hold the position as long as possible. During the test, the holding time is taken to indicate the level of back extension endurance, while the EMG is recorded in order to examine muscle fatigue according to the EMG's electrode placement over any specific muscle belly. The EMG indication of muscle fatigue is based on the gradient of the EMG's median frequency slope. When the muscle is fatiguing, the motor unit firing rate decreases (De Luca 1993). Based on the EMG power spectrum over time, the muscle fatigue can be observed according to the shift of power density to a lower frequency. Both mean and median frequencies also decrease throughout the spectrum. According to De Luca (1993), the median frequency was a better indicator over the mean frequency because it was less sensitive to noise and signal aliasing, but more variable at lower frequency. Moreover, when a muscle is fatiguing, the EMG amplitude also increases because additional muscle fibres were recruited in order to generate the same force for the contraction. This explains why short-term fatigue is important to ensure muscle growth in fitness exercise.

Fatigue and pain are interrelated. To explain the relationship between pain and fatigue at the cellular level, it had been reported that there was a biological link between muscular fatigue and pain through the role of ASIC3 channels (an acid-activated ion channel protein) (Sluka and Rasmussen 2010). Repeated muscle activation can cause increased extracellular K^+, which in turn decreases muscle membrane excitability, resulting in a decrease in muscle force production (Allen et al. 2008). Due to metabolic by-products and lactic acid accumulation, these can cause a decrease in pH of muscle tissues (Burnes et al. 2008). Moreover, the acidic extracellular environment (i.e. tissue acidosis) was reported to induce pain and also mechanical hyperalgesia (Kubo et al. 2012). Furthermore, the physical impact of both fatigue and pain on musculoskeletal adaptation share common similarities. For instance, in occupational therapy, it can be very hard to differentiate between the signs of pain and fatigue because in some conditions such as chronic fatigue syndrome, both pain and fatigue usually present together. However, this will not affect the quality of the therapy as the main purpose of the therapy is to find the way to enable the patient to perform the activity independently by exploring the most comfortable and the safest way of performing the activity, regardless of what clinical manifestation does the patient has. Both pain and fatigue can be treated as important signposts to reflect the biomechanical impact (e.g. kinetics and kinematics) of physical activity on musculoskeletal function. Therefore, the knowledge on the mechanism of musculoskeletal compensation due to muscle fatigue is crucial in order to determine physical activity performance, as well as to establish more accurate description about one’s maximum function.

During physical activity, when the body is subjected to fatigue, compensatory mechanisms will be activated in order to keep the physical activity going while coping with the symptoms. This phenomenon is also known as guarded movement, which can be described as ‘the abnormalities in muscle action during physical activity’ (Main and Watson 1996). In low back pain patients, studies had shown an increase in the lumbar muscle activity as compared to healthy controls (Hodges and Moseley 2003; Van der Hulst et al. 2010). For instance, Van der Hulst et al. (2010) compared the muscle activity of erector spinae between chronic low back pain patients and healthy controls while walking on a treadmill at 3.8 km/h. In that study, they found that the muscle activity of erector spinae had increased up to 38% during the activity, which was higher compared to the healthy controls. Particularly, the high erector spinae activity can be found during double support period where the trunk movements in sagittal (i.e. flexion-extension) and frontal planes (i.e. lateral flexion) were controlled by the muscles, and this phenomenon was an example of guarding movement among low back pain patients in order to protect the painful site.

A theory of motor adaptation to fatigue was suggested by Hodges and Tucker (2011) to explain possible changes in motor control due to the pain. In their article, vicious cycle or pain-spasm-pain (Roland 1986) and pain adaptation theories (Lund et al. 1991) were criticized. Both theories have different views on muscle excitability, assuming that the changes in motor activity are predictable. According to the vicious cycle theory, muscle activity increases in accordance to pain. Contrariwise, the pain adaptation theory explained that the muscle activity is rather inhibited. Hodges and Tucker (2011) summarized that the motor control adaptations were not always stereotypical, and there was a non-uniform motor adaptation to pain from single motor neuron towards whole-muscle behaviour. There were four main assumptions in the theory of motor adaptation's theory. Firstly, muscle fatigue can lead to redistribution of activity within-muscle (e.g. activation of other motor unit to maintain the same force) and between-muscle (e.g. change in recruitment pattern, agonist-antagonist reciprocal action). Secondly, the adaptation can change the mechanical outcome of muscle contraction (e.g. in-phase vs. anti-phase coordination). Thirdly, the adaptation can lead to protection from further fatigue or injury on the affected muscle. Fourthly, the adaptation has short-term benefit, but with potential long-term consequences. For instance, although the increase in co-contractions of lumbar muscles is an adaptation strategy to increase the spinal stability, this action can lead to an increase in the spinal loading (Marras et al. 2004). Therefore, the aim of this study was to investigate the impact of fatigue on body movements during a carrying activity. The findings from this study hope to provide a baseline of biomechanical characteristics for carrying activity to assist clinicians in making an evidence-based clinical decision, as well as to guide researchers to further investigate possible biomechanical mechanism that could lead to musculoskeletal disorders.

1.3 STRUCTURE OF THESIS

In total, there are nine chapters in this thesis (Table 1.1). The first three chapters present the introduction, literature review and general methodology of the thesis. Subsequently, the following four chapters are experimental chapters that are designed to follow general format of a journal manuscript. These experimental chapters will be submitted for publication once the PhD has been completed. Each experimental chapter contains introduction, methodology, findings, discussion and conclusion pertaining to the title of the chapter. Following the experimental chapters, the findings from each experimental chapter are discussed collectively as a general discussion chapter. Finally, a conclusion is made at the end of this thesis to summarize all the findings, as well as to suggest future recommendations.

Table 1.1. Arrangement of chapters

Chapter	Title
1	Introduction to PhD Thesis
2	Literature Review
3	General Methodology
4	Reliability of 3D Gait Analysis, Isometric Back Endurance and Muscle Fatigue
5	Comparing Muscle Fatigue during Ito Test and Anterior Load Carriage
6	Spatiotemporal Parameters and 3D Kinematics of Anterior Load Carriage
7	Changes in Pelvis-Trunk Coordination during Anterior Load Carriage
8	General Discussion
9	Conclusion to PhD Thesis

1.4 CURRENT WORK AND ACHIEVEMENTS

This section summarizes the current work and achievement related to this PhD study with regards to scientific publications and conferences.

i. **Hanif Farhan, M.R., White, P.J., Warner, M.B. & Adams, J.E. 2015. The relationship between carrying activity and low back pain: a critical review of biomechanics study. Malaysian Journal of Health Sciences, pp. 1-25. (Published)**

ii. **Hanif Farhan, M.R. 2017. Biomechanical Strategies *in* Curtin, M., Egan, M. Adams, J.E. Occupational Therapy for People Experiencing Illness, Injury Or Impairment: Enabling Occupation, Promoting Participation. 7th ed. Elsevier. (Published)**

iii. **Hanif Farhan, M.R., White, P.J., Warner, M.B. & Adams, J.E. Reliability of 3D Gai.t Analysis, Isometric Back Endurance and Muscle Fatigue. (Article in preparation)**

iv. **Hanif Farhan, M.R., White, P.J., Warner, M.B. & Adams, J.E. Comparing Muscle Fatigue during Ito Test and Anterior Load Carriage. (Article in preparation)**

v. **Hanif Farhan, M.R., White, P.J., Warner, M.B. & Adams, J.E. Spatiotemporal Parameters and 3D Kinematics of Anterior Load Carriage. (Article in preparation)**

vi. **Hanif Farhan, M.R., White, P.J., Warner, M.B. & Adams, J.E. Changes in Pelvis-Trunk Coordination during Anterior Load Carriage. (Article in preparation)**

vii. **The Reliability of Ito test and 3D Kinematics of Normal Gait. Postgraduate Research Conference, School of Health Sciences, University of Southampton, 2015. (Oral presenter)**

viii. **Hanif Farhan, M.R., White, P.J., Warner, M.B. & Adams, J.E. Movement Analysis of a Standardized Carrying Activity with Progressive Loads: A Musculoskeletal Biomechanics Study. Malaysian Occupational Therapy National Conference 2016. (Oral presenter)**

Chapter 2: LITERATURE REVIEW

2.1 INTRODUCTION

The aim of this chapter was to explore the underlying musculoskeletal biomechanics and related parameters of carrying from the literature, as well as to examine its potential relationship with low back pain (LBP). The outcome from this review will assist in determining the objectives for this PhD study later in this chapter. Carrying activities are known to associate with many medical problems such as LBP, stress fractures, rucksack palsy, knee pain, foot blisters, metatarsalgia, local discomfort and local fatigue (Knapik et al. 1996). As one of the most common work-related musculoskeletal disorders, LBP had been reported to pose a large socio-economic impact on many countries (Collins et al. 2010). While the direct healthcare cost of LBP was estimated to be around £1.6 billion annually, the cost of informal care and production losses were estimated to be £10.7 billion in total (Maniadakis and Gray 2000). Furthermore, workers with LBP had been reported to have approximately three times the likelihood for work absenteeism as compared to non-LBP workers (Widanarko et al. 2012). In the industrial setting, manual material handling was reported to be a major contributing factor to LBP (Waters et al. 2006). Kuiper et al. (1999) found that manual material handling activities such as lifting, carrying, pushing, pulling and combined MMH were the risk factors for various types of back disorders. Across the literature, Heneweer et al. (2011) had concluded that there were moderate to strong risk factors of LBP for heavy workload and the accumulation of loads or frequency of load carriage.

Although many studies had attempted to explore the associations between manual material handling and LBP, most were epidemiological studies rather than the examination of biomechanical mechanism. For instance, Eriksen et al. (2004) had reported that the frequency of lifting, carrying, and pushing heavy objects statistically predicted LBP-related sick leave of longer than eight weeks. However, according to a systematic review by Wai et al. (2010), with the exception of the findings from Eriksen et al. (2004), a causal relationship between occupational carrying and LBP could not be confirmed within other high quality epidemiological studies. Furthermore, although there were studies that had been using a video recording to record the activity, those studies were not considered as including robust biomechanical analysis by Wai et al. (2010). Whilst the severity of the exposure to LBP was described, none of the biomechanical parameters (i.e. kinetics or kinematics) were reported. Therefore, an in-depth biomechanical investigation is needed to complement these epidemiological findings in order to understand the mechanism that may eventually lead to the development of LBP.

2.2 ASOCIATION BETWEEN CARRYING AND LOW BACK PAIN

2.2.1 Search Strategy

The Cochrane database was reviewed to ensure that there were no biomechanically focussed literature reviews on the association between carrying activity and LBP. After revealing that there was no such study, this literature review was carried out using AMED, CINAHL, Compendex and MEDLINE online databases based on three main keywords. These keywords were biomechanics (i.e. kinetics, kinematics, posture, and motion analysis and muscle fatigue), low back pain (i.e. low back pain, back injury and back disorders) and carrying (i.e. carrying or weight bearing or moving or load carriage) (Figure 1). To ensure that the most contemporaneous paper was selected, articles published from 2004 to 2017 were selected. English-language publication and peer-reviewed articles only were selected and duplicates across the databases were removed. Each article had to incorporate at least one biomechanical (i.e. kinetics or kinematics) or other related musculoskeletal parameters related to biomechanics.

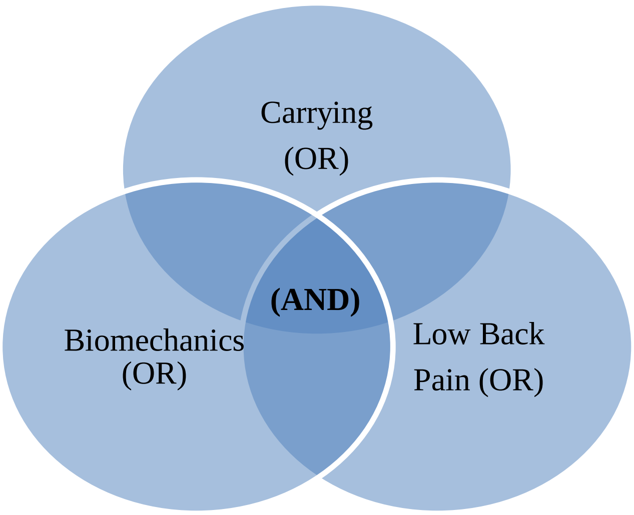


Figure 1. Boolean logic for search strategies

To establish a standardized concept, kinetics was defined as '*the study of the effects of forces on the motion*', whereas kinematics was defined as '*the study of motion without reference to mass or force*' (Knudson 2003). For this review, the carrying activity was defined as moving from one place to another while manually holding a certain load. Studies were considered if the weight of the load carried was specified. Only carrying activities with a posterior, anterior, central, and/or lateral load were included because it was assumed that those variants were commonly performed in various work settings. At least one completed gait cycle must be performed during the activity. The title and abstract of papers identified were screened to clarify the suitability according to the aforementioned inclusion and the exclusion criteria. The full texts were then retrieved to assess the methodological quality and level of evidence.

No study designs were excluded. This review did not include secondary studies (i.e. narrative literature review, systematic review and meta-analysis). Grey literature was also excluded. Any article was excluded if either carrying distance or carrying period was not reported. Dependent carrying such as having assistance from any mechanical device or other individuals to perform the carrying activity was also considered as an exclusion criterion. As this review aimed to investigate the aforementioned association among working population, any studies with participants aged less than 16 years old were excluded. Any biomechanics studies that met the inclusion criteria were considered. The level of evidence and methodological quality were determined according to the guideline provided in the Health Evidence Bulletins: Wales Project (Weightman et al. 2004). This guideline provides a series of critical appraisal tools (CAT) to assess the methodological quality across various types of study design. The CAT contains a series of questions to assist with the critical appraisal that includes background, trustworthiness, finding, and relevance.

2.2.2 Identification of studies

Figure 2 illustrates the method by which the relevant studies were included and excluded from the databases. After combining the major keywords (i.e. '*carrying*', '*low back pain*' and '*biomechanics*'), a total of 836 studies from 2004 to 2017 were found from AMED, CINAHL, Compendex, and MEDLINE. This number was then reduced to 694 after removal of duplicates. After screening for review articles, grey literature, and non-English articles, this number was further reduced to 564. Of these, 413 studies were biomechanics studies.

Among those, only 32 studies were the studies of carrying activity. However, after reviewing the full texts, 16 studies were excluded because using non-targeted participants or there was insufficient information about the carrying activity. Finally, 16 studies had met all of the inclusion criteria and were critically reviewed.

2.2.3 Summary of Studies

The carrying activities were assessed according to several criteria, namely the target population (job specific or non-job specific), load position (anterior, posterior, bilateral, central, or mixed), load weight (percentage of body mass or standardized) and carrying speed (self-selected pace or standardized). After screening, sixteen studies met all inclusion criteria and were critically reviewed (Figure 2). Among all, six studies had included comparative groups to test their research questions. For instance, Krupenevich et al. (2015) had compared gait kinetics and kinematics between genders. Both chronic LBP and healthy respondents were recruited in order to compare the impact of chronic LBP on the variability of stature loss, stature recovery and/or paraspinal muscle activity while carrying a weighted vest, (Healey et al. 2005a; Healey et al. 2005b; Healey et al. 2008) and trunk and pelvic coordination whilst carrying an anterior load carriage at various speed (Kim and Chai 2015). Likewise, Rodacki et al. (2005) used both obese and non-obese respondents to examine the impact of obesity on stature changes and stature recovery while both hands carried hand-loads.

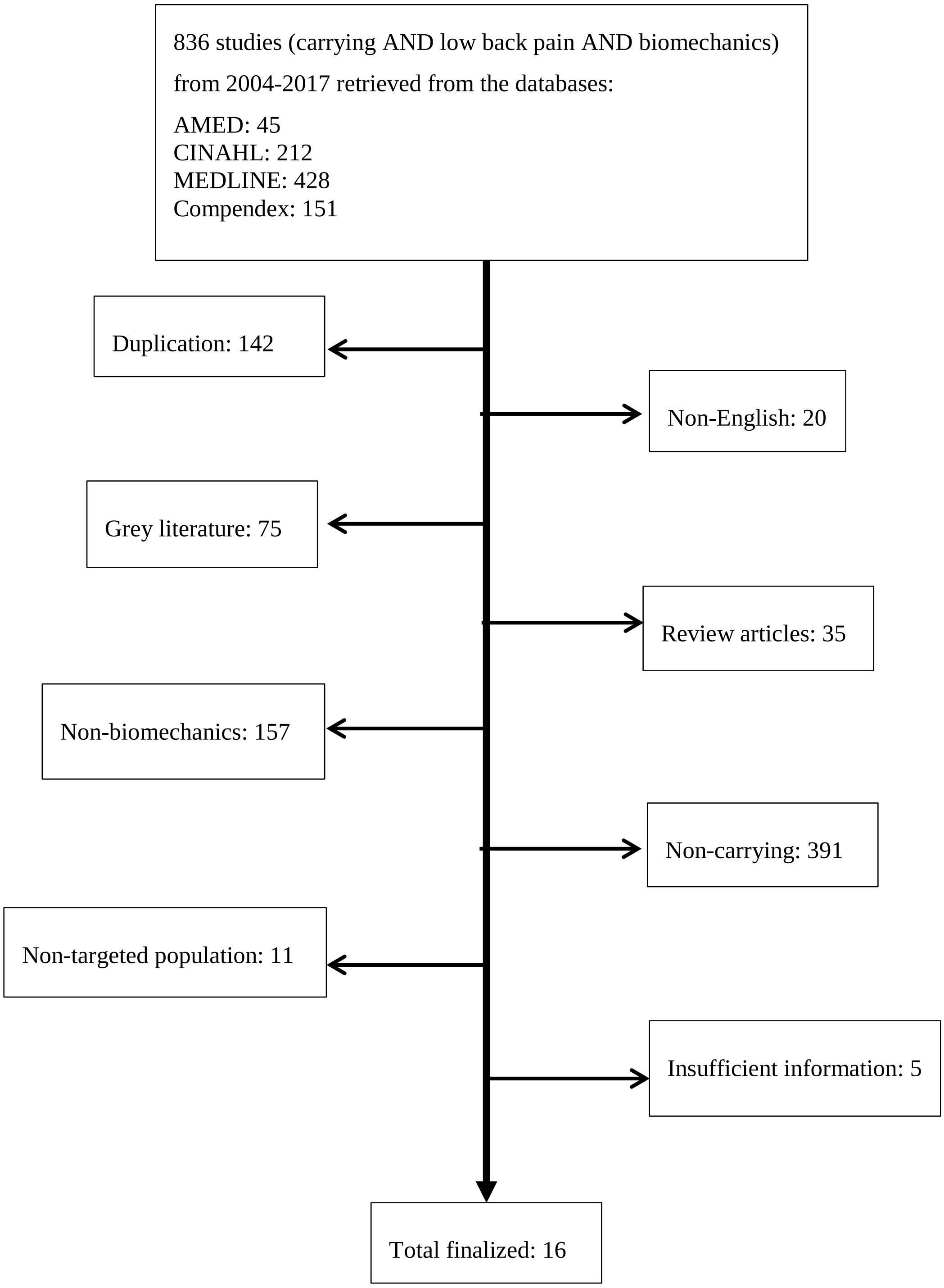


Figure 2 Flowchart of study selection

Table 2.1. Summary of studies

Author (year)	**Participant**	**Carrying characteristics**	**Measurements and outcome parameters**
Dahl et al. (2016)	24 healthy young adults.	Carrying two different backpacks (i.e. posterior load vs. bilateral loads) with no load and with 15% and 25% of body mass (BM) on a treadmill at the speed of 1.4 m/s for 6 minutes.	Vicon motion analysis system to measure posture and kinematics. AMTI force instrumented treadmill to measure ground reaction forces.
Mallakzadeh et al. (2016)	9 healthy male college students.	Carrying two different backpacks (i.e. with vs. without strap, 10% BM) on a treadmill with velocity was changed from 1.5 m/s to 2.5 m/s for 15 minutes, then 2m/s for another 15 minutes.	Three cameras (Basler Industrial Cameras, Pilot Series, piA640-210gc, 210 fps, Japan) with SIMI motion analysis software to capture kinematics data to measure head, neck, and trunk flexion or extension, lateral displacement, and velocity.
Muslim and Nussbaum (2016)	9 healthy participants.	Carrying posterior load using their hands at the upper two corners of carriage bag with three load masses (20%, 35% and 50% of BM) and three load sizes (small, medium and large) on a walkway (5m) and treadmill (90s), both using self-selected walking pace.	6-camera Vicon motion analysis system to capture torso kinematics. Force platform to measure ground reaction forces (GRF). Electromyography (EMG) equipment to measure the activity of paraspinal (L1 & L3), rectus abdominis and external oblique muscles.
Kim and Chai (2015)	10 healthy subjects and 10 patients.	Carrying anterior load (10% BW) at walking speeds of 3.5, 4.5, or 5.5 km/h.	6-camera Vicon motion analysis system to capture kinematics data for measuring trunk and pelvic coordination in transverse plane.
Krupenevich et al. (2015)	11 males and 11 females	Carrying a 22 kg backpack with three load conditions: unloaded, low-back placement, and mid-back placement.at 1.5 m/s speed over level ground.	8-camera infrared digital camera system (Qualisys) to capture 3D kinematics (i.e. stride length; walking velocity; trunk angular position; and joint angular positions and angular velocities at the right hip, knee, and ankle). Force platform to measure ground reaction forces.
Smallman et al. (2013)	13 healthy males	Carrying a hand-held anterior and posterior load carriage (15% BW) with and without the assistance of the Mover's Assistive Device (MAD) using preferred walking speed for 55 right-foot strides.	6-camera Vicon motion analysis system to determine the intersegment coordination between trunk-pelvis and trunk-box in transverse plane.

Author (year)	Participant	Carrying characteristics	Measurements and outcome parameters
Hung-Kay Chow et al. (2011)	13 healthy adults.	Carrying a backpack (10% BM) for 30 minutes, followed by 30 minutes of unloaded walking (speed = 1.1 m/s). (BM = body mass)	Electrogoniometric system with four gravitational-referenced accelerometers to measure changes in spinal curvature and trunk posture.
Simpson et al. (2011)	15 female recreational hikers.	Carrying a backpack with 4 mass conditions (no backpack, 10% BM, 20% BM, 30% BM and 40% BM) in an 8 km circuit at self-selected pace.	OPTOTRAK 3020 motion analysis system to measure sagittal plane peak trunk flexion angle relative to the horizontal and range of motion during stance phase.
(Seay et al. 2011a)	11 male soldiers.	Walking (1.34 m/s) and running (2.46 m/s) with and without holding a rifle (rifle weight = 2.4kg) for on treadmill for a total 16 minutes.	ProReflex motion analysis system to measure 3D segmental angles of pelvis and trunk
Majumdar et al. (2010)	10 male soldiers.	Walking while carrying 9 military load conditions at self-maintained pace for 10m each.	Hires Expert Vision System to measure 3D kinematic data.
Healey et al. (2008)	11 chronic LBP and 11 asymptomatic participants.	20 minutes of loaded walking tasks by wearing weighted vest (20% BM) each in the morning and in the afternoon at self-selected pace. 20 minutes of unloaded recovery position (side lying) was performed before and after the activity.	Stadiometer with linear variable high-resolution (LVDT) displacement transducer to measure changes in stature.
Fowler et al. (2006)	6 healthy males.	Walking at self-selected pace for 8500 meters with and without carrying a standard Royal Mail bag containing 17.5% BM positioned on shoulder. The bag load was reduced gradually (10% reduction from the initial load).	ELITE BTS optoelectric system to measure 3D kinematics of the spine. Stadiometer with LVDT displacement transducer to measure changes in stature.
Healey et al. (2005a)	11 chronic LBP and 11 asymptomatic participants.	Four 20-minutes sessions of loaded walking tasks by wearing weighted vest (10% BM) at self-selected pace. 20 minutes of 4 unloaded recovery positions (side lying, 50^{o} gravity inversion, spinal hyperextension, 11^{o} supported) were performed before and after the activity.	Stadiometer with LVDT displacement transducer to measure changes in stature. Surface EMG to measure changes in muscle activity.

Author (year)	**Participant**	**Carrying characteristics**	**Measurements and outcome parameters**
Healey et al. (2005b)	20 chronic LBP and 20 asymptomatic participants.	20 minutes of loaded walking tasks by wearing weighted vest (10% BM) at self-selected pace, and then followed by 40 minutes of unloaded recovery position (side lying).	Stadiometer with LVDT displacement transducer to measure changes in stature. Surface EMG to measure changes in muscle activity.
Rodacki et al. (2005)	10 obese and 10 non-obese participants.	30 minutes of walking task while carrying hand-load bilaterally (10% BM, 5% at each hand) at self-selected pace and 30 minutes of standing recovery period.	Stadiometer with LVDT displacement transducer to measure changes in stature.
LaFiandra and Harman (2004)	11 male soldiers.	Carrying a backpack with three mass conditions (13.6 kg, 27.2 kg, and 40.8 kg) on treadmill for three minutes each (speed = 1.34 m/s).	Force transducers to measure forces exerted at the lower back, upper back and shoulders, and backpack centre of mass (COM). Qualisys Motion Capture system to measure position data to calculate the forces.

2.2.4 Study Design and Statistical Analysis

Only one study was designed as experimental study (i.e. 9 x 9 Latin square design) (Muslim and Nussbaum 2016), while the rest of the studies were cross-sectional. For all studies, convenience sampling was utilized, with the exception of Majumdar et al. (2010) who had used random sampling. All studies were carried out with a small sample size (< 30) and none of these studies had reported a *priori* power calculation to determine sample size. Thus, type-II error may be presented. Nevertheless, some studies did report on the appropriate effect size to indicate the magnitude of the observed effects. For instance, Seay et al. (2011a) used Cohen's *d* as the measure of effect size to estimate the magnitude of difference after performing the multiple pairwise comparison for a two-way repeated measures ANOVA. For that, they had adopted the Cohen's *d* conventional effect size (i.e. $d > 0.5$ represents clinically meaningful difference, while d > 0.8 represents large practical difference) (Cohen 1988b). Healey et al. (2008) on the other hand, reported both correlation coefficient (r) and coefficient of determination (r^2) as the effect size measures for correlation tests. In the case of Majumdar and Pal (2010), although they had reported the changes in the mean to elaborate the magnitude of difference after performing a *post-hoc* analysis, the true magnitude of changes between the groups might still be influenced by its pooled standard deviation (with regards to the Cohen's *d* formula for independent groups). In another study, Muslim and Nussbaum (2016) claimed that the magnitude of an interaction effect between load mass and abdominal muscle activity was small without any indication of appropriate measures of effect size for ANOVA test family (e.g. partial eta squared or Cohen's f) in their results. Furthermore, their explanation regarding a significant interaction between load mass and load size on the activity of paraspinal muscle at L1 level was inconsistent with the actual *p* value that was reported to be above 5%.

2.2.5 Possible Bias and Confounding

Across the selected studies, age, body mass index (BMI), gender and level of physical activity were identified as the potential confounding variables that can interfere with the primary biomechanical outcomes. Across all studies, the mean age and BMI ranged from 21.5 to 35.1 (young adulthood) and 20.1 kgm^{-2} to 36.6 kgm^{-2} (normal weight to obese) respectively. Only Rodacki et al. (2005) recruited obese respondents in order to investigate the impact of BMI on stature variations during and after a carrying activity. Most of the studies recruited single gender respondents to eliminate any possible gender effect. Although there were two studies had been reported using mixed gender respondents, the number of male to female respondents were or almost equal (Healey et al. 2005b; Healey et al. 2008). Still, there was no baseline comparison to confirm the effect of gender on the measured parameters. While aiming to determine the differences in the alteration of paraspinal muscle activity according to different unloading positions among chronic LBP and asymptomatic respondents, of all studies, Healey et al. (2005a) was the only study that did not mention anything on gender.

Some studies attempted to evaluate or control physical activity level as one of the possible confounding factors (Healey et al. 2005a; Healey et al. 2005b; Rodacki et al. 2005; Healey et al. 2008; Smallman et al. 2013; Kim and Chai 2015). In these studies, the indication of physical activity can generally be divided into three types; habitual, short-term and immediate. As for the habitual type, the Baecke's Physical Activity Questionnaire and the NASA/Johnson Space Centre's Physical Activity Rating (PA-R) Scale were used as the measurements. The short-term physical activity was methodologically controlled by implementing strategies such as instructing the respondents to sleep for approximately 8 hours and/or preventing them from any stressful physical activity for 24 hours prior to the study session. Furthermore, to eliminate the effects of physical activities prior to arrival in the laboratory (immediate control), the participants were instructed to maintain a specific unloading position before the main experiment began. One of the methods was to maintain a left-side lying on a comfortable surface with the hip and knees flexed for 20 minutes (Healey et al. 2005a; Healey et al. 2005b; Healey et al. 2008). Other than that, Rodacki et al. (2005) instructed their respondents to lie in a Fowler's position for 30 minutes to allow spinal unloading. Other than that, warm up activities for a certain period of time prior to the experimental session was also applied possibly in order to standardize the immediate physical activity among the respondents (Smallman et al. 2013; Kim and Chai 2015; Muslim and Nussbaum 2016).

Gender difference may also affect some gait parameters because female usually walk with more anterior pelvic tilt and up-and-down oblique motion, more flexed, adducted and internally rotated hip joints and more valgus angles of the knee joint (Cho et al. 2004). However, although Krupenevich et al. (2015) reported that female exhibited more increase in forward trunk position whilst carrying a 22kg load compared to unloaded walking, the hip, knee and ankle torques showed no differences. Whilst males were reported to have a greater frequency of osteophytes, the narrowing of intervertebral disc was more frequent in females (de Schepper et al. 2010). Furthermore, as the age increased, both development of osteophytes and narrowing of intervertebral disc were also increased (de Schepper et al. 2010). The level of physical activity on the other hand, had been suggested to have a unique relationship with LBP (Heneweer et al. 2009). This relationship can be illustrated as a continuum that explains a dynamic interaction between risk of LBP and activity intensity. The risk of LBP was suggested to be at the highest level when the activity intensity is at both most minimum (i.e. total inactivity) and maximum (i.e. heaviest activity). For BMI, a meta-analytic evidence had indicated that overweight and obesity could increase the risk of LBP (Shiri et al. 2010). Spinal shrinkage was found to have a positive correlation with body mass. This could possibly be due to the impact of cumulative load from the body mass onto the spine (Yar 2008). However, the accuracy of BMI to indicate obesity is still controversial due to the fact that it cannot distinguish between fat-mass and fat free mass (Romero-Corral et al. 2008).

2.2.6 Kinetics

The kinetics of load carriage was indicated in three studies (Krupenevich et al. 2015; Dahl et al. 2016; Muslim and Nussbaum 2016). Two studies determined the carrying load based on the percentage of body mass (i.e. ranged from 10% to 50%), while only one study used a standardized load (i.e. 22kg). In general, the posterior load carriage caused an increase in the vertical ground reaction force (Krupenevich et al. 2015; Dahl et al. 2016). Muslim and Nussbaum (2016) investigated slip risk during posterior load carriage (i.e. 20%, 35% and 50% of BM) based on the calculation of required coefficient of friction (RCOF). The RCOF was determined as the ratio between the horizontal and vertical ground reaction forces (GRF) at 50-200ms just after a heel strike. In general, if the RCOF is higher than the available coefficient of friction of the floor, the risk slip will be increased. In their study, they found that the RCOF and load size were both decreased whilst carrying the heaviest load. However, a ‘U’ shaped relationship between the load mass and the shape was found whilst carrying lighter loads. For the significant main effect of load size, they explained that the difference may occur due to torso angular acceleration (Nott et al. 2010). For the non-significant main effect of load mass (RCOF increased with load mass), they claimed that there was an inconclusive findings from the previous studies on anterior load carriage (Myung and Smith 1997; Sukwon and Lockhart 2008). Nevertheless, there were several limitations in Muslim and Nussbaum (2016) study. As the participants were required to hold the upper part of the frame, the arm position might induce some discomfort in the hands and the elbows, which can possibly influence gait kinetics. Furthermore, as this study only examined the short-term effects of the current intervention, the effects may differ in a long-term period due to biomechanical adaptation over time.

LaFiandra et al. (2003) studied the force distribution on the lower back, upper back and backpack centre of mass while carrying a backpack with three mass conditions among soldiers were investigated. The backpack had an external frame, allowing both upper attachment (i.e. shoulder straps) and lower attachment (i.e. hip belt) to be the only points of contact between the backpack and the carrier. The results had shown that the vertical and anterior/posterior forces exerted on the lower back, upper back and back pack centre of mass increased as the backpack mass increased. For instance, at the lower back, mean ± standard deviation of the anterior forces resulted from the backpack with 13.6kg, 27.2kg and 40.8kg were 27.86±9.14, 58.78±14.22 and 182.27±21.63 respectively. Furthermore, regardless of the backpack mass, approximately 30% of the vertical forces generated by the backpack were transferred to the lower back by using the external frame and the hip belt. This study concluded that while the use of external frame and hip belt can possibly reduce the risk of having shoulder injury such as rucksack palsy, the consistent anterior force on the lower back as exerted by the backpack might contribute to LBP. However, the findings can only be generalized towards male population, as there were no female participants in their study. As females naturally have a broader sacrum (Tortora and Derrickson 2008), the structure may influence the degree by which a person is able to endure the backpack load onto their lower back.

Backpack carrying had been reported as one of the most prevalent carrying methods throughout the selected studies. A systematic review by Golriz and Walker (2012) had summarized that low backpack should be avoided if the load was more than 15% of the body weight. They also reported that although double pack can move the centre of gravity closer to the body and help to distribute the load between the front and back of the body, respiratory ventilation, upper limb movement and front visual field may become restricted. Other strategies such as proper positioning of the backpack on the spine (i.e. upper back, middle back or lower back) and the use of front pack and double pack (i.e. both front and back) had also being studied throughout the literature. To reduce the forces on the shoulder and the upper back while carrying a loaded backpack (i.e. vertical and anterior/posterior), one of the common reported strategy was to incorporate a frame and a hip belt to the backpack. Outside this review, other study had shown that without the frame and the hip belt, the maximal pressure of shoulder straps of a 10.2 kg backpack can reach up to 203 mmHg (Holewijn 1990), which was double than the skin threshold for irritation and redness (i.e. 105 mmHg) (Husain 1953). Although this strategy can potentially reduce the risk of getting brachial plexus lesion such as backpack palsy, the additional forces on the lower back may increase the compressive loading of the lumbar spine, which eventually may cause other problems such as vertebral body damage and further degenerative changes.

2.2.7 Kinematics

Simpson et al. (2011) had reported that carrying a backpack weighted as low as 20% of body weight (BW) can increase the trunk flexion, as they reported that there was a significant decrease in the peak trunk flexion (sagittal plane relative to horizontal planes) angle from 84±3.0 (0% of BW) to 78±3.0 (20% of BW) (i.e. a smaller angle indicates increased trunk flexion). Moreover, trunk flexion was found to be greater while carrying the backpack in a longer distance. However, as this study was conducted on experienced hikers, the participants might already accustomed with carrying heavy loads in a long distance. Therefore, they may have developed strategies to cope with fatigue due to backpack carriage compared to occasional hikers. On the other hand, Majumdar and Pal (2010) found that only during mid-stance, significant changes in the percentage of gait cycle can be observed between no-load condition (22.7±2.2) and other carrying conditions, namely a carrying rifle (24.1±1.4), carrying a light machine gun (24.4±1.2), carrying a haversack (24.5±1.0), carrying a haversack and a light machine gun (24.7±1.2) and carrying a backpack and a light machine gun (24.7±1.3). Although there had been a general increase in all spatiotemporal parameters during load carriage, most of the changes were non-significant, except for the midstance. During mid-stance, they reported that the trunk flexion can reach a maximum flexion of 9.5° while carrying the maximum load (i.e. 17.5kg). The findings on the general increase in spatiotemporal parameters were inconsistent with some previous studies across the literature that reported either no changes (Hong and Cheung 2003) or decreased stride length and increased stride frequency (Pascoe et al. 1997). However, Majumdar and Pal (2010) claimed that the changes were due to increased walking speed, which in agreement with Attwells et al. (2006) findings that reported increased stride length, cadence and speed whilst carrying a loaded military webbing.

In Fowler et al. (2006) study, the participants were instructed to carry a loaded standard Royal Mail bag (17% of body weight), and the loads decreased gradually throughout the activity. However, the number of participants were very small (N=6), and this can influence statistical power. The result showed that at the beginning of the task (i.e. heaviest loads) increased trunk lateral flexion at lumbar region in the opposite direction to the side where the bag was held (up to 12º) and increased forward flexion within the thoracic region (up to 6º). Displacing the participant's centre of mass in both planes (i.e. trunk was displaced in one plane, but coupled with movement in another different plane), had been suggested to increase the risk of LBP (Noone et al. 1993). Thus, Fowler et al. (2006) had concluded that the use of mailbag designs, which does not allow side-to-side alternation (e.g. mailbag with waist-belt that fixes carrying position only to one side), were not recommended because it may cause long-term effect of postural deviation. However, as the design of this study was a cross-sectional, the suggestion regarding the mailbags designs were not supported by causal effect. In other words, the effect of the mailbags design on the risk of low back pain cannot be confirmed as both risk and incidence were measured at the same time. Therefore, an experimental study or a cohort study was preferable to support the recommendation.

Seay et al. (2011a) carried out a study among healthy soldiers to investigate the upper body kinematics (i.e. ROM in sagittal, frontal and transverse planes), pelvis-trunk coordination (i.e. continuous relative phase) and the coordination variability while carrying a rifle (2.4 kg) in two different gait modes (i.e. walking and running). The results had shown that carrying a rifle with both hands produced a greater trunk transverse ROM (i.e. axial rotation) in running, but lower trunk sagittal ROM for both speed. In transverse plane, regardless of the gait mode, the pelvis-trunk coordination was more in-phase while carrying the weapon. These findings were later supported by Smallman et al. (2013). Moreover, decreased coordination variability can also be observed in transverse plane as a result of carrying the weapon. The decrease in coordination variability while carrying the weapon may contribute to LBP due to decreased pelvis-trunk system adaptability. Further study on the pelvis-trunk coordination was conducted by Kim and Chai (2015) to compare the changes in the coordination whilst carrying 10% BM anteriorly between chronic low back pain (CLBP) patients and healthy subjects. The study had reported that the CLBP patients exhibited more anti-phase coordination compared to the healthy subjects. However, in their discussion, while it was indicated that a more in-phase coordination in the CLBP patients across different walking speed, their results revealed a higher continuous relative phase in the CLBP compared to the healthy individuals.

2.2.8 Other Related Musculoskeletal Parameters

2.2.8.1 Stature changes

Five studies were conducted to measure changes in stature as a result of carrying activity (Healey et al. 2005a; Healey et al. 2005b; Rodacki et al. 2005; Fowler et al. 2006; Healey et al. 2008). Fowler et al. (2006) had reported that among healthy respondents, stature loss (i.e. reduction in height) was doubled after the loaded carrying activity. Furthermore, although there was no difference in stature loss as compared to the control group, the chronic LBP group was reported to have a significantly lower stature recovery during the unloading period (Healey et al. 2005a; Healey et al. 2005b; Healey et al. 2008). However, there was no stature recovery observed in the obese group in comparison with the control group (Rodacki et al. 2005). This phenomenon might be related to the fact that the obese respondents had already sustained a ‘chronic’ loading condition, in which can affect the intervertebral disc and other spinal structures, which in the future may lead to LBP.

Hung-Kay Chow et al. (2011) had studied the carry-over effects of carrying activity on trunk posture and also the repositioning ability of the spine after carrying a loaded backpack (i.e. 10% body weight). The repositioning ability was determined according to repositioning error, which can be described as the difference between trunk forward lean and spinal curvatures with regard to preload conditions. The repositioning ability was measured between preload and loaded backpack conditions. However, the effect of an unloaded backpack was unknown in this study. The results indicated that immediately after the load was removed, there were significant differences in repositioning errors of cervical lordosis (66%), thoracic kyphosis, lumbar lordosis (57%), pelvic tilt (44%) and trunk forward lean (54%). Even 30 minutes after that, the repositioning errors cannot be fully restored to the level of preload conditions. The main limitation of this study was the minimal and unequal number of participants in both groups (i.e. 5 male and 8 female), affecting the statistical power. Although there was no statistical comparison between both genders was conducted, the repositioning ability may vary due to differences in gait pattern during the load carriage (Cho et al. 2004).

2.2.8.2 Muscle activity

Three studies incorporated electromyography to analyse muscle activity during a traditional posterior load carriage (i.e. hand-held a loaded carriage bag on the back) (Muslim and Nussbaum 2016) or during a recovery period walking with a loaded vest (i.e. unloading period) (Healey et al. 2005a; Healey et al. 2005b). According to Muslim and Nussbaum (2016), two patterns of significant interaction were found between the paraspinal muscle activity at L1 and L3 and load mass depending on the size of the load. Both paraspinal muscle activity and the load size were increased whilst carrying the heaviest load. This can possibly occur to enable a sustained flexed poster in order to maintain spinal stability. However, the paraspinal muscle activity reduced as the load size was bigger whilst carrying the lightest load. From observation, they claimed that a relatively light load was commonly placed more on the upper back and neck, the participants may use the arm muscles rather than the paraspinal muscles to carry the load. Furthermore, the activity of paraspinal muscle activity (i.e. erector spinae at LI-L2 and L4-L5 interspaces) was reported to be higher in chronic LBP groups both before the carrying activity and during the unloading period (Healey et al. 2005a; Healey et al. 2005b). This higher level of activity was suggested to increase compressive load on the spine, thus, preventing the intervertebral disk to recover at its initial height (Healey et al. 2005a; Healey et al. 2005b).

In general, prolonged muscle activation could further lead to muscle fatigue. In general, fatigue was known to have a unique association with musculoskeletal pain. Outside this review, fatigue was reported to be one of the most common presentations of LBP across the literature (Demoulin et al. 2007). In general, fatigue can be described as '*the progressive decline in performance which can largely be recovered after a period of rest (reversible)'* (Allen et al. 2008). Furthermore, when the body mechanics is failing due to fatigue or pain, the body may initiate compensatory mechanisms to accommodate with the symptoms. Known as 'guarded movement' (Main and Watson 1996), this phenomenon may reduce the activity of any primary muscle and activating accessory muscles, which then could result in movement alteration from normal. In time, this phenomenon could further lead to the disuse of primary muscles by preventing any movements which are believed may trigger the pain (generally known as 'fear-avoidance' phenomenon) (Vlaeyen and Linton 2000). This may explain why the multifudus and paraspinal muscles were found smaller in chronic low back pain patients compared to healthy controls (Fortin and Macedo 2013b).

2.2.9 Review recommendation

Although studies related to carrying activity were available from the online databases, some studies were epidemiological rather than biomechanical. The main purpose of epidemiological study was to examine the relationship between the biomechanical exposure and the occurrence of low back pain rather than to explore the biomechanical mechanism behind it. This review also found that most of the studies on carrying activity had been carried out on children and adolescence rather than adult population, which was probably due to backpack carriage. Furthermore, compared to carrying, researchers tend to focus more on lifting activity in the working population. Therefore, this review recommended future research to examine the impact biomechanics of carrying activity on working-age population by focusing on the nature of the activity performed at their workplace. Finally, this review also suggested further investigation on biomechanical parameters involved in a standardized activity as commonly performed in a functional capacity evaluation. By exploring the core biomechanical aspects that were needed to be addressed during the evaluation, an emphasized, detailed, and systematic description of functional capacity of LBP can be produced to guide a safe and timely return-to-work process.

This review found that carrying characteristics among the studies varied according to the intended target population. Knowledge about these variations is beneficial in order to understand the impact of various load positions while performing the carrying activity across different types of work. Clinically, one of the major assessments to determine the physical readiness for return-to-work is Functional Capacity Evaluation (FCE). Being regarded as the gold standard of vocational assessment (McFadden et al. 2010), the major role of the assessment is to analyse the consistency between a patient's performance in work-related physical activities and the relevant job demands. Although each activity has been used for different protocols in the assessment, a number of protocols can be grouped to represent primary job demand of a profession. For instance, a heavy manual worker may undergo a set of the functional capacity evaluation protocols differently than a professional driver or a teacher due to the different work demands. In other words, each protocol should still be carried out as a standardized activity to ensure good measurement quality (i.e. validity and reliability) across various professions. Therefore, instead of carrying a static load, this review suggested the use of progressive load increment (e.g. per kilogram/pound) in conjunction with the common method of carrying in the FCE. This can enable the clinician to suggest a safe maximum carrying load limit for RTW based on a reliable method that can be used for the patients from all professions.

2.3 PROBLEM STATEMENT FOR PHD

Most studies that had attempted to explore the associations between MMH and LBP were epidemiological rather than direct examination of biomechanical mechanisms. For instance, Eriksen et al. (2004) had reported that the frequency of lifting, carrying, and pushing heavy objects statistically predicted LBP-related sick leave of longer than eight weeks. Additional force over lumbosacral region for a prolonged period of time may increase the risk of LBP. However, according to a systematic review by Wai et al. (2010), with the exception of one study (Eriksen et al. (2004), a causal relationship between occupational carrying and LBP could not be confirmed in other high quality epidemiological studies. While the 'severity' of the exposure to LBP was indicated, none of the biomechanical parameters (i.e. kinetics or kinematics) was reported in the systematic review. Therefore, an in-depth biomechanical investigation is warranted to complement these epidemiological findings in order to understand the mechanism that may lead to the development of LBP over time.

To the researcher’s knowledge, most of the studies emphasized on posterior load (e.g. backpack) rather than anterior load. The position of the load may differently influence the body posture, which may affect gait kinematics. It was assumed that the anterior load carriage is more common in the industrial setting. This assumption was supported by the fact that most of the carrying protocol in FCE was based on anterior load carriage. To fill the knowledge gap, this study attempted to explore the biomechanical impact of anterior load carriage on the body movements. Specific changes in inter-segmental coordination during carrying activity may also associate with LBP. In general, the inter-segmental coordination can be defined as the temporal-spatial coupling between adjacent body segments, while the consistency of the coordination pattern over time can be regarded as coordination variability (Yen et al. 2012). People with LBP may limit their inter-segmental coordination variability as a strategy to reduce pain (Heiderscheit et al. 2002). In LBP patients, a reduced transverse plane coordination variability between pelvis and trunk had been reported (Seay et al. 2011c). Furthermore, LBP patients tend to produce more variable coordination in frontal plane (Lamoth et al. 2002a). Therefore, this study attempted to explore the changes in pelvis-trunk coordination throughout carrying activity.

Other than that, back muscle fatigue is one of the most common presentations in LBP patients (Roy et al. 1989; Kankaanpää et al. 1998). In general, muscle fatigue can be described as the failure of muscle to maintain the required or expected force, which can be observed as a progressive decline in muscle performance which can mostly be recovered after a period of rest (Allen et al. 2008; Enoka and Duchateau 2008). The activity of paraspinal muscles during carrying activity was found to be higher in LBP patients (Healey et al. 2005a; Healey et al. 2005b). Subsequently, a prolonged muscle activation can also lead to muscle fatigue, which is a common presentation of LBP (Demoulin et al. 2007). This may enlighten why the multifudus and paraspinal muscles were found smaller in LBP patients compared to healthy controls (Fortin and Macedo 2013a). During activities of daily living, an agonist or primary muscle does not work alone in order to delegate the mechanical forces throughout the body. Thus, it is important to explore muscle fatigue among multiple muscles bilaterally during carrying activity in order to understand the biomechanical mechanism that could lead to the development of LBP. Specific compensatory mechanisms may be initiated when the body mechanics fails due to fatigue. This phenomenon may reduce the activity of primary muscles, but activates the accessory muscles, which then alter the normal movement (also known as guarded movement) (Main and Watson 1996). In time, this phenomenon may contribute to the disuse of primary muscles by avoiding any movements which is believed to cause the pain (Vlaeyen and Linton 2000). Therefore, this study will also investigate what are the changes in the body movements at the point where the body is fatigued during carrying activity.

2.4 GENERAL OBJECTIVE

To study the biomechanical impacts of carrying activity on healthy individuals.

2.5 SPECIFIC OBJECTIVES

i. To investigate the difference in muscle fatigue during Ito test between a manual and sedentary groups.
ii. To investigate the changes in muscle fatigue across carrying activity in a manual and sedentary groups.
iii. To investigate the difference in maximum carrying load between a manual and sedentary groups.
iv. To investigate the changes in spatiotemporal parameters across carrying activity in a manual and sedentary groups.
v. To investigate the changes in 3D kinematics across carrying activity in a manual and sedentary groups.
vi. To investigate the changes in pelvis-trunk coordination across carrying activity in a manual and sedentary groups.

2.6 RESEARCH HYPOTHESIS

i. Sedentary group has significantly higher muscle fatigue during Ito test compared to manual group.
ii. Manual group has significantly higher maximum carrying load compared to sedentary group.
iii. There are significant changes in spatiotemporal parameters across carrying activity for both manual and sedentary groups.
iv. There are significant changes in lower limb kinematics across carrying activity for both manual and sedentary groups.

v. There are significant changes in muscle fatigue across carrying activity for both manual and sedentary groups.

vi. There are significant changes in pelvis-trunk coordination across carrying activity for both manual and sedentary groups.

Chapter 3: GENERAL METHODOLOGY

3.1 STUDY DESIGN

The design of this study was cross-sectional with between-group and within-group comparison recruiting healthy participants (n=37). The data were collected from May 2014 to April 2015 (11 months) at the Biomechanics Laboratory, Faculty of Health Sciences, University of Southampton. This study was approved by the Faculty of Health Sciences Ethics Committee (ethics numbers: 12460). The main parameters of this study were lower limb kinematics, spatiotemporal parameters of gait, back endurance, muscle fatigue and pelvis-trunk coordination. The participants were divided into sedentary individuals (n=20) and manual workers (n=17) (see 3.4 for details) for between-group comparison. During the study, the participants were asked to perform two types of gait: standard gait (i.e. self-preferred gait) and carrying activity with progressive loads (i.e. 1 kg increment). Within-group comparison was made between the standard gait and the carrying activity with a maximum load (max-load).

There were three main components of this study: 1) the feasibility study, 2) the reliability study (within-session and between-session) and 3) the main study (Figure 3.1). Prior to the main study, a feasibility study (N=3) was conducted to identify potential issues regarding the practicality of the study and make improvements to address the issues. These had included the clarity of instruction to the participants, establishing an appropriate laboratory setting, the appropriate placement of motion analysis markers and surface electromyography (EMG) electrodes, and data acquisition and analysis from Vicon Nexus into MATLAB software. The participants and their data from the feasibility study were not included into the main study. Following the review of the feasibility study procedures, the decision was made not to change any major methodological procedures and the study was advertised to recruit participants for the main study. Concurrent with the main study, a reliability study involving within-session and between-session reliability testing was also conducted in order to investigate the measurements' reliability (see chapter 4). For the between-session reliability, only nine participants agreed to attend a second session (two-week interval). The within-session reliability was tested using all participant based on three trials for each gait activity. Therefore, the between (as noted) within session-reliability was conducted for the standard gait and max-kg gait.

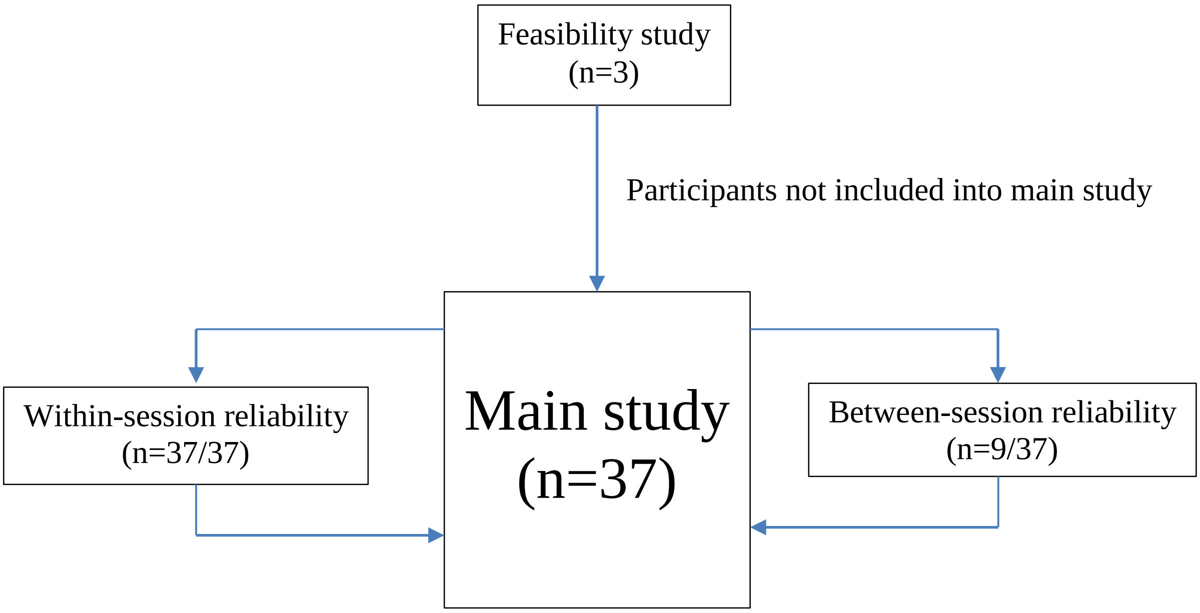


Figure 3.1: Components of study

3.2 INCLUSION AND EXCLUSION CRITERIA

Participants were included into the study if they were healthy male individuals aged between 16 to 64 years old. According to the literature, gender differences were reported to affect some gait parameters because females usually walk with more anterior pelvic tilt and up-and-down oblique pelvic motion, more flexed, adducted and internally rotated hip joints, and more valgus angles of the knee joint (Cho et al. 2004). The inclusion criterion attempted to control the influence of gender differences on gait kinematics by focusing only on males. The age group was chosen to include those who were eligible to work (UK Office for National Statistics 2013). The exclusion criteria for this study were individuals with any history of low back pain within the previous 12 months, cardiopulmonary problem and/or movement disorders or any condition that may influence motor control. This exclusion criteria were used to control any factors that can possibly influence physical performance related to the study.

3.3 DETERMINATION OF SAMPLE SIZE

Determination of sample size (DSS) for this study was conducted based on power analysis method. The power analysis method was conducted based on four main parameters: sample size, effect size, power and alpha (Type I error). If three of the parameters were known, the fourth parameter can be predicted. Therefore, the effect size, alpha, and power were determined in order to determine the sample size. According to Sparto et al. (1997), an effect size of 0.742 (Cohen's *d*) was large enough to determine the influence of muscle fatigue on movement coordination. By setting the alpha level at 5% and the statistical power at 80%, the required sample size is 17. As this study will include two groups of participants according to their types of work (i.e. sedentary and manual), the minimum required sample size for this study is 34 (i.e. 17 for each group).

3.4 ETHICAL CONSIDERATIONS

Prior to the commencement of the study, an ethical clearance was gained from the Faculty of Health Sciences Ethical Committee, University of Southampton (ethics number: 12406). This study was advertised using posters around the university campus as the sampling frame covers both the university students and staff. During the first meeting with all potential participants, a participant information sheet (Appendix B) was given to describe the purpose and the benefit of the study, participant's level of involvement, possible risk and also confidentiality. At this period, screening for the exclusion criteria had also been carried out. Then, a consent form (Appendix C) was given to formal record informed consent to participate in the study. The participant was allowed to make the decision after having had time to consider their participation, up to seven days if needed. If the individual agrees to participate by signing the consent form, an appointment will be made for a data collection session. During the session, the participant was asked to wear shorts to allow the placement of EMG electrodes on bare skin. The placement of motion analysis markers also required the participant to wear at least tight clothing to fix the markers' position on the body. At the end of the session, simple refreshment (i.e. coffee/tea and cake) was given to the participant. As for data management, the data were kept strictly confidential from anyone except the researchers. During data analysis, each participant's name was replaced by a certain code to ensure anonymity. Any written report to indicate the result of the study was presented as a whole without specifying any particular participant.

3.5 PARTICIPANT RECRUITMENT

Posters (Appendix A) were used to advertise the study. The posters were put on notice boards within approved university buildings and also the University of Southampton online website (i.e. SUSSED). The participants were recruited using a convenience sampling method and were further divided into sedentary and manual groups. In order to allocate the participants to the groups, a brief structured-interview on the physical activity level was conducted according to the International Physical Activity Questionnaire (IPAQ) (Appendix D). The IPAQ was developed to examine the level of physical activity among respondents across different countries (Ainsworth et al. 2000). In 12 countries, IPAQ was reported to have good test-retest reliability (i.e. the coefficients were clustered around 0.8) (Craig et al. 2003). The original version of the IPAQ addresses four domains that consisted of leisure time activity, domestic and gardening activity, work-related activity and transport-related physical activity. This study used a short-version of IPAQ (i.e. past-7-day version). This version was preferable because it can minimize the time taken to complete the whole research session. In this short-version, the level of physical activity for the last seven days in three specific level of activities were determined. The activities are walking, moderate-intensity activities and vigorous-intensity activities. For each activity, the duration (i.e. minutes) and frequency (i.e. days) were measured. The short-version IPAQ had an acceptable reliability, with most of the reliability coefficients were reported >0.67 across different countries (Craig et al. 2003). It was reported that the short-version IPAQ (last-7-d version; as in this study) could be used in national, regional and international prevalence studies due to its feasibility to administer. Furthermore, there was no difference in validity and reliability with the long-version IPAQ has been reported (Craig et al. 2003). Those who spent most of their working time in sitting position (i.e. > 2/3 of working time) and only stand and walk occasionally (i.e. < 1/3 of working time) were grouped into sedentary (or non-manual) group, while those who perform manual material activities (e.g. lifting, carrying, pushing, pulling) for most of the time (i.e. > 2/3 of working time) were grouped into manual workers (U.S. Department of Labour 1991).

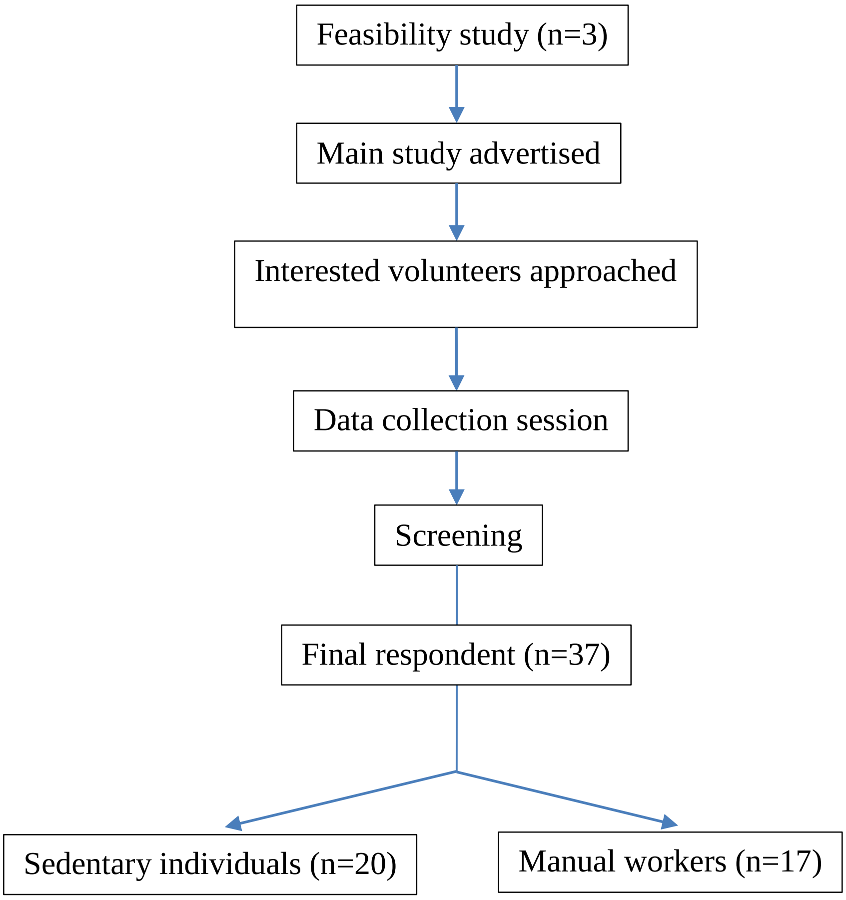


Figure 3.2. Flowchart of participant recruitment

3.6 MEASUREMENT & INSTRUMENTATION

3.6.1 Isometric Back Endurance Test

The Ito test was conducted to examine isometric back extension endurance. According to Ito et al. (1996), this test had been reported to have high test-retest reliability (i.e. 0.94 to 0.97 in healthy individuals, 0.93 to 0.95 in chronic low back pain patients). As one of the variants of Biering-Sorensen test (Biering-Sorensen 1984), it had been suggested that the Ito test was preferred over the original version because it involved lesser spinal loading, thus, minimizing the risk of injury along the lumbar area (Demoulin et al. 2006). To perform this test, participants were asked to place themselves in a prone lying position on a table with a small pillow under the lower abdomen to decrease lumbar lordosis. They were then asked to lift and maintain the sternum off the table to maximum extension. The test was terminated after five minutes or if the patient could not maintain approximately 2/3 of original test position (e.g. due to fatigue or pain) (Müller et al. 2010). During the test, the time the participants were able to keep their chest off the table to maintain a maximum extension was recorded (i.e. holding time) to indicate their level of isometric back endurance (Figure 3.3).

Figure 3.3. Ito test

3.6.2 Motion Analysis

To examine the 3D kinematics of gait activity, a camera-based motion analysis was carried out using the Vicon MX Motion Capture System (Vicon, Oxford UK). The system consisted of twelve optical cameras. Each camera had infrared light-emitting diodes (LED) that release pulses of light. During the carrying activity, the infrared light was reflected by retro-reflective markers back into the camera lens. This action enabled the system's software (Nexus) to reproduce the movement in a digital 3-dimensional environment. Therefore, it was essential to place the reflective markers at the correct locations to represent an accurate movement of the underlying body segments around any specific joint (see 3.8.1 for detail). Version 1.8 of Nexus was used for data capture with a combination of Nexus 1.8 and 2.2 used for data processing. In total, twelve Vicon MX T-series cameras were used to record the kinematics data. The cameras consisted of six Vicon's T40 and six T160 that have the resolution of 4 megapixels and 16 megapixels respectively. Both types of camera can capture between 30 to 2,000 frames per second (fps). The sampling frequency for the cameras was set to 100Hz. Each camera was positioned around the laboratory to allow a maximum coverage of the 3D kinematics movements that occur whilst walking along the platform. Once the raw 3D kinematics data were captured, these data were processed using Vicon Nexus 2.0 and exported into the Matrix Laboratory (MATLAB) software for further analysis.

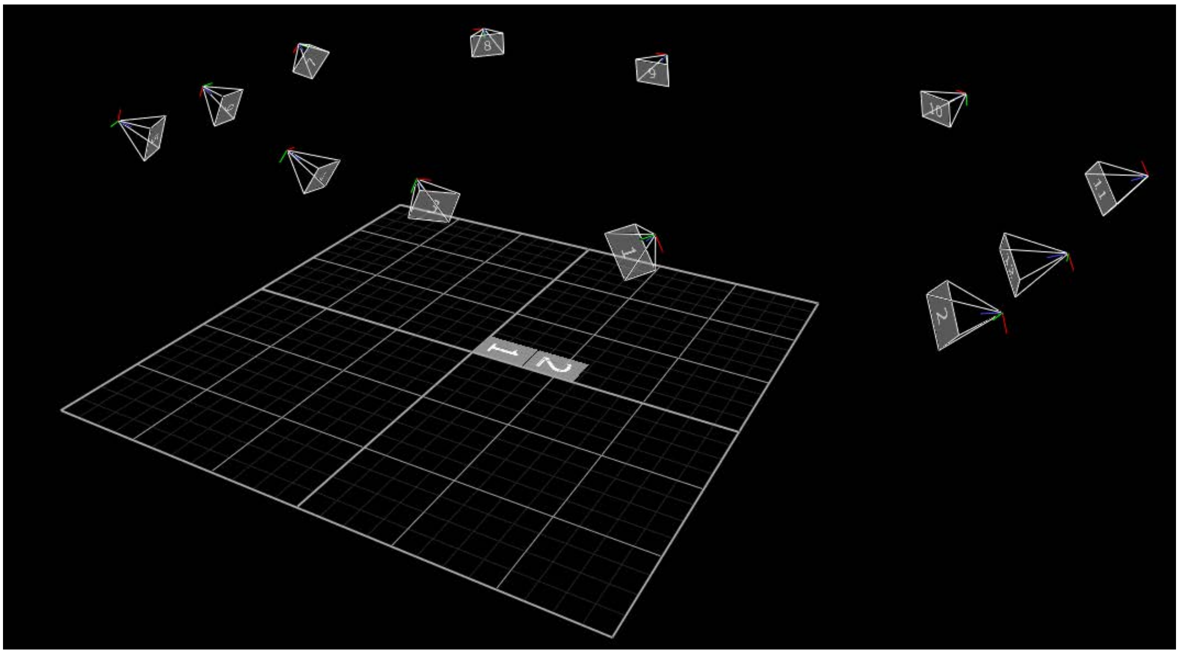


Figure 3.4. Position of motion in analysis cameras

3.6.3 Electromyography

A ZeroWire (Aurion) wireless EMG system was used to examine the muscle activity throughout the study. The system was linked to the Vicon Motion Capture System via an analogue to a digital capture board and captured concurrently with marker positions in NEXUS 2.2 enabling synchronization in the time domain for both EMG and 3D kinematic data. Equipped with 16 channels of wireless electrodes, this EMG system permitted the measurement of up to eight muscles bilaterally. In this study, all channels were used in order to record the muscle activity of illiocostalis, multifudus, gluteus maximus, biceps femoris, biceps brachii, latissimus dorsi, vastus lateralis and gastrocnemius (see 3.8.2 for more detail). The raw EMG data were recorded at 1000 Hz sampling rate. For muscle activity, the EMG measurement method was based on the recommendations from the European concerted action on Surface EMG for a non-invasive assessment of muscles (SENIAM) (Hermens and Freriks 1997; Hermens et al. 1999; Merletti and Hermens 2000; Stegeman and Hermens 2007). The latissimus dorsi was not included in the SENIAM recommendations. For this study, the electrodes were placed at approximately 4 cm below the inferior tip of the scapula, half the distance between the spine and the lateral edge of the torso (Criswell 2010). As the largest muscle of the back with the origin at the vertebrae T7 to L5, thoracolumbar fascia, iliac crest, inferior 3 or 4 ribs and inferior angle of scapula and the insertion at the floor of intertubercular groove of the humerus, the location of the electrode placement was therefore over the muscle bulk and was considered appropriate for recording the EMG signals.

As a motor control strategy to adapt with fatigue, the muscle may undergo within-muscle and between-muscles redistribution. The within-muscle redistribution can be observed as there may be a change in the onset of muscle activation. Besides, changes in the pattern of muscle activity between adjacent muscles that are connected via specific fascial webbing (i.e. myofascial meridian) can explain the between-muscles redistribution (Myers 2009). In general, Myers's classification describes seven myofascial meridians, which are the functional lines, the superficial front line, the superficial back line, the lateral line, the spiral line, the deep front lines and the arm lines. The muscles within each myofascial meridian may function together in a unique pattern as the muscles are connected via the aforementioned fascial webbings. According to the myofascial connectivity, only the muscles at the superficial back line (i.e. iliocostalis lumborum, biceps femoris and gastrocnemius) and the back functional line (latissimus dorsi, gluteus maximus and vastus lateralis) were examined in this study. The activity of these muscles were examined whilst carrying an anterior load without affecting EMG sensors, as the muscles were arranged either laterally or posteriorly. Furthermore, the activity of multifidus muscle was also examined. The multifidus muscle was one of the most commonly reported muscles in the literatures as having a strong association with low back pain (Freeman et al. 2010). Other than that, the activity of biceps brachii was also recorded. It was assumed that as the anterior load was held bilaterally with 90º of elbow flexion in front of the abdominal region (i.e. lower torso), the muscle fatigue of biceps brachii may become the main reason to terminate the carrying activity.

3.7 METHODOLOGICAL PROCEDURE

Prior to data collection session, an informed consent was obtained from each participant. During the session, the participants' level of physical activity, isometric back extension endurance, standard gait and carrying gait were determined. Figure 3.5 illustrates a detailed methodological description for this study.

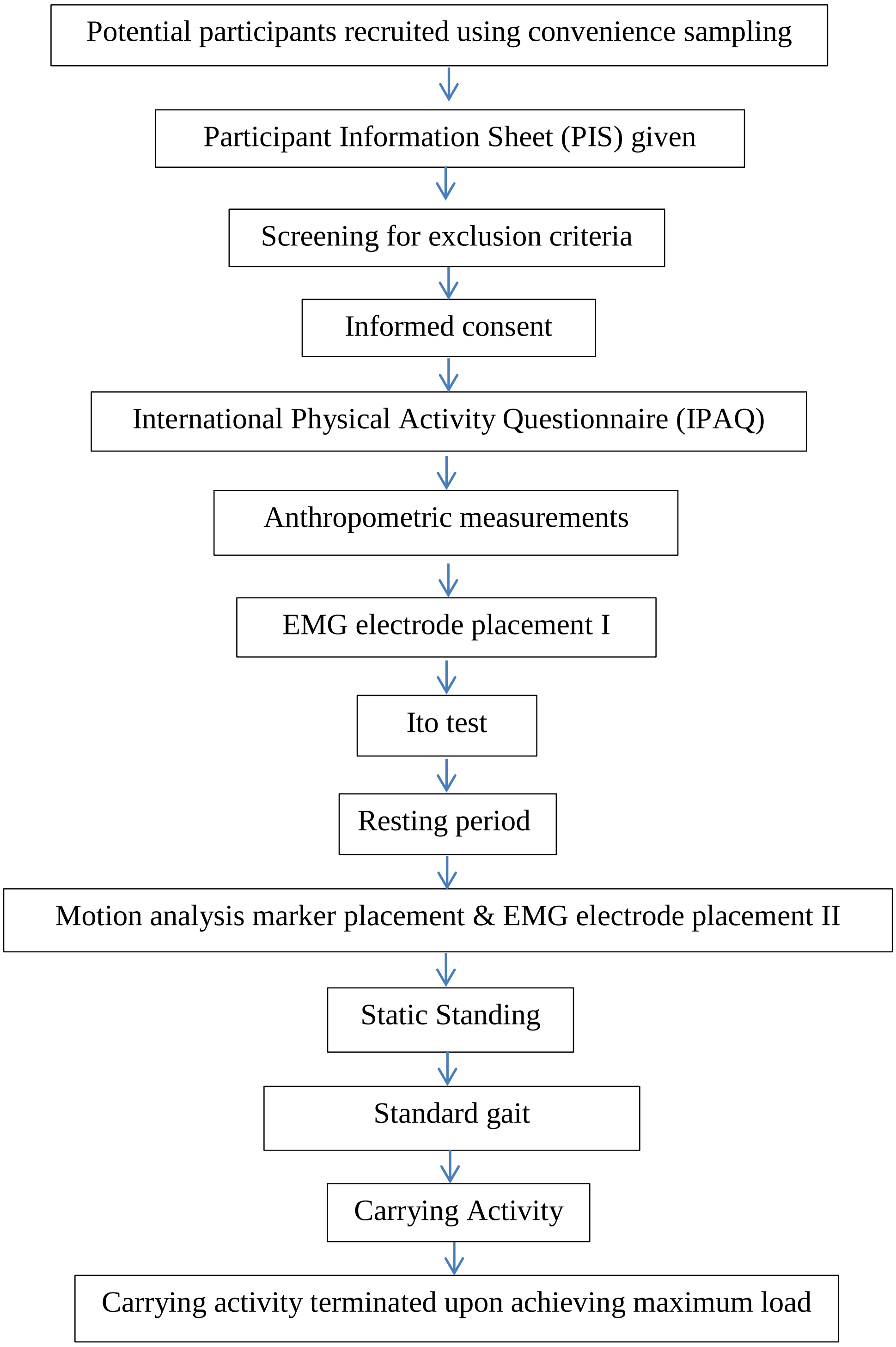


Figure 3.5. Flowchart of Study Procedure

3.7.1 Anthropometric measurements

The flowchart of study procedure is explained in Figure 3.5. After informed written consent was obtained, participants were requested to change their clothing and wear shorts only. Physical examination which consisted of taking measurement of weight (kg), height (cm), leg length (cm), knee width (cm) and ankle width (cm) were then taken. These measurements were important in order to process kinematic data based on the Plug-in-Gait model. The weight and height were measured using a weighing scale with a built-in height rod. The height was measured from top of the head to the floor while the participant stands flat on both feet. The leg length is defined as the distance between the anterior superior iliac spine (ASIS) and the medial malleolus via the knee joint and was measured using tape measurement. Knee width is defined as the distance between the lateral and medial femoral epicondyles. Ankle width is defined as the distance between the lateral and medial malleolus. Both knee width and ankle width were measured using a calliper while standing.

3.7.2 Ito test

Subsequently, the participants performed the Ito test. The main purpose of the Ito test was to measure the level of isometric back endurance in order to investigate its association with the performance in carrying activity. During the test, the time the participants were able to keep their chest off the table to maintain a maximum back extension was recorded (i.e. holding time) to indicate their level of isometric muscle endurance. At the same time, the muscle activity of iliocostalis, multifidus, gluteus maximus and biceps femoris (EMG placement I) were recorded to measure their level of muscle fatigue based on the slope of EMG median frequency (MFslope). Repeated five-second recordings were made throughout the Ito test to avoid large data files. Further details on EMG processing are explained in Section 3.8.2. After the Ito test, the participants were allowed to rest up to a maximum of 15 minutes before the next procedure.

3.7.3 Static standing

After a rest, motion analysis markers and the EMG electrodes for biceps femoris, latissimus dorsi, vastus lateralis and gastrocnemius (EMG placement II) were attached in preparation for carrying activity. Prior to any gait activities (i.e. standard gait and carrying activity) the participants were asked to statically stand on one force platform for ten seconds whilst a recording of kinematic and EMG data was made.

3.7.4 Standard gait

The participants then completed walking trails along the 10-meter platform using their self-preferred walking speed (referred as standard gait). During the standard gait, both kinematic and EMG data were recorded to obtain a baseline measure of the participants' gait. The standard gait was completed once three sets of good foot placement (i.e. both left and right foot of the same stride) on a force platform were successfully recorded.

3.7.5 Carrying activity

The carrying activity was explained to the participants and a demonstration on how to carry a plastic container whilst walking (based on the health and safety advice on the correct way to carry manual loads in a plastic container (Industrial Accident Prevention Association 2008)) was provided by the researcher. The participants were asked to perform a carrying activity for 60 meters by holding the carrying container by flexing the arm at 90º of elbow flexion in order to prevent the container from restricting the hip movement during the activity. According to the feasibility study, most participants complained about pain at both hands while holding the container because the inner edge of the container's handles were hard. The pain at the hand can probably lead to a 'premature' termination of the carrying activity. Thus, foam grips were fitted to both handles of the container and according to the participants, this had successfully reduced the hand pain throughout the carrying activity (Figure 3.6).

In this study, one set of carrying activity was carried out in the sequence of walk I, II and III (Figure 3.7). Walk I started from the middle until the end of the platform (5 meters). Then, walk II started when the participants turned around and walked along the platform until the opposite end of the platform (10 meters). Finally, walk III started when the participants turned around and walked along the platform until they reached the middle of the platform (5 meters). Therefore, the total distance for the three sets of carrying was 60 meters. At the end of each three sets of carrying activities, the participants were instructed to stand for five seconds to allow static body recording of EMG and kinematic data (Figure 3.7). The next load was then put into the container, and the subsequent carrying activity then commenced until the activity was terminated. Each participant was required to perform three sets of carrying activities for each carrying load, starting from

0kg of load and then followed by 1kg of increment until the activity was terminated. The list of criteria for terminating a carrying activity were shown in Table 3.1. The maximum load allowed whilst carrying was 20kg (Health and Safety Executive 2012).

Figure 3.6. Foam-fitted plastic container with 1kg load (sand bag)

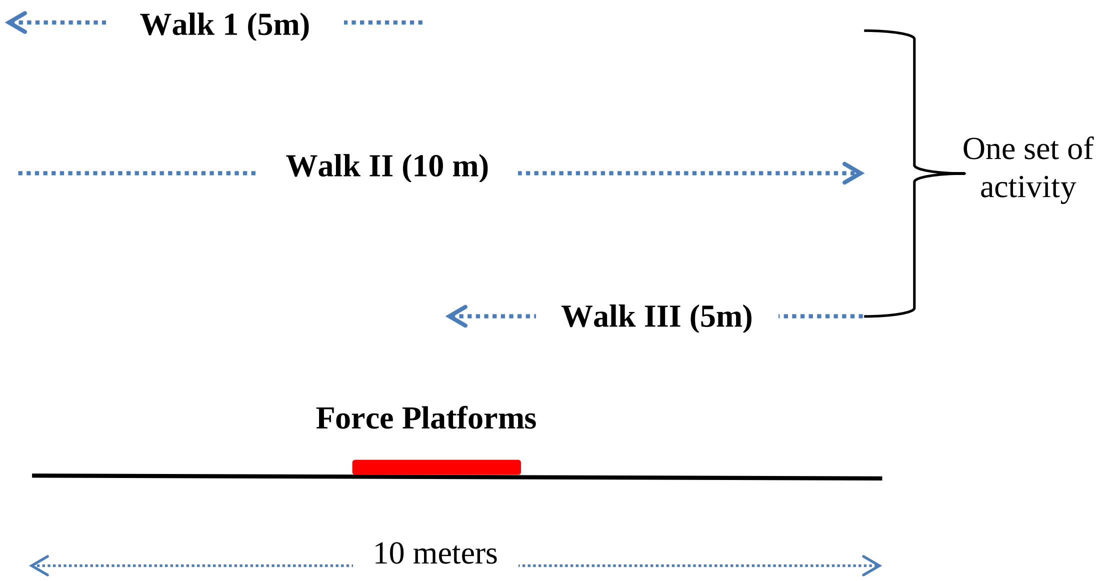


Figure 3.7. Sequence of Carrying Activity

Table 3.1. Determinants for maximum safe carrying load limit

Approaches	Determinants
Psychophysical	• Participants' verbal expression of fatigue
Kinesiophysical	• Muscle bulging of prime movers • Involuntary use of accessory muscles • Altered body mechanics, including counterbalancing or use of momentum • Loss of equilibrium • Increased base of support • Decreased efficiency and smoothness of movement • Heavy breathing patterns

Source: Gross and Battié (2005)

3.8 DATA ACQUISITION & PROCESSING

3.8.1 Kinematics

3.8.1.1 System calibration

A calibration wand was used to calibrate the motion analysis system in order to determine a global coordinate system (GCS). There were two stages of calibration: dynamic and static. The dynamic calibration was conducted by waving the calibration wand within the capture volume until all cameras had captured 2400 frames of the calibration wand (Figure 3.8). Subsequently, the static calibration was conducted by recording the calibration wand at the centre of the platform (i.e. middle of capture volume), which was aligned with the corner of a force platform.

Figure 3.8. Calibration wand

3.8.1.2 Rigid body segments

Kinematic data were analysed according to Plug-in-Gait Model (Kadaba et al. 1990; Davis et al. 1991) using Vicon Nexus 2.2 to determine the 3D kinematics of standard gait and carrying activity. In total, there were eight rigid segments in this study: trunk, pelvis, right thigh, left thigh, left thigh, right shank, left foot and right foot (Figure 3.9). In this study, only specific angles were chosen for analysis. In sagittal plane, the movements at all joint/body segments (i.e. trunk, pelvis, hip, knee and ankle) were analysed. In frontal plane, only the movements at the trunk, pelvis and hip were analysed. In transverse plane, only the movement at the trunk and pelvis were analysed. The joint angles were selected based on three principles. First of all, as the anterior load carriage can directly influence changes in kinematics within sagittal plane, all flexion-extension movements were chosen. The second principle was developed based on the Compass-Gait model (Saunders et al. 1953;

Lin et al. 2014). According to the model, there are six determinants of gait. These determinants are pelvic rotation, pelvic obliquity, knee flexion in the stance phase, foot and knee mechanism and the lateral displacement of the pelvis. Thus, all angles that were responsible for the determinants were chosen. Finally, the third principle was developed in order to determine the pelvis-trunk coordination during the carrying activity. For this purpose, the trunk and pelvis angles in all cardinal planes were included in the study.

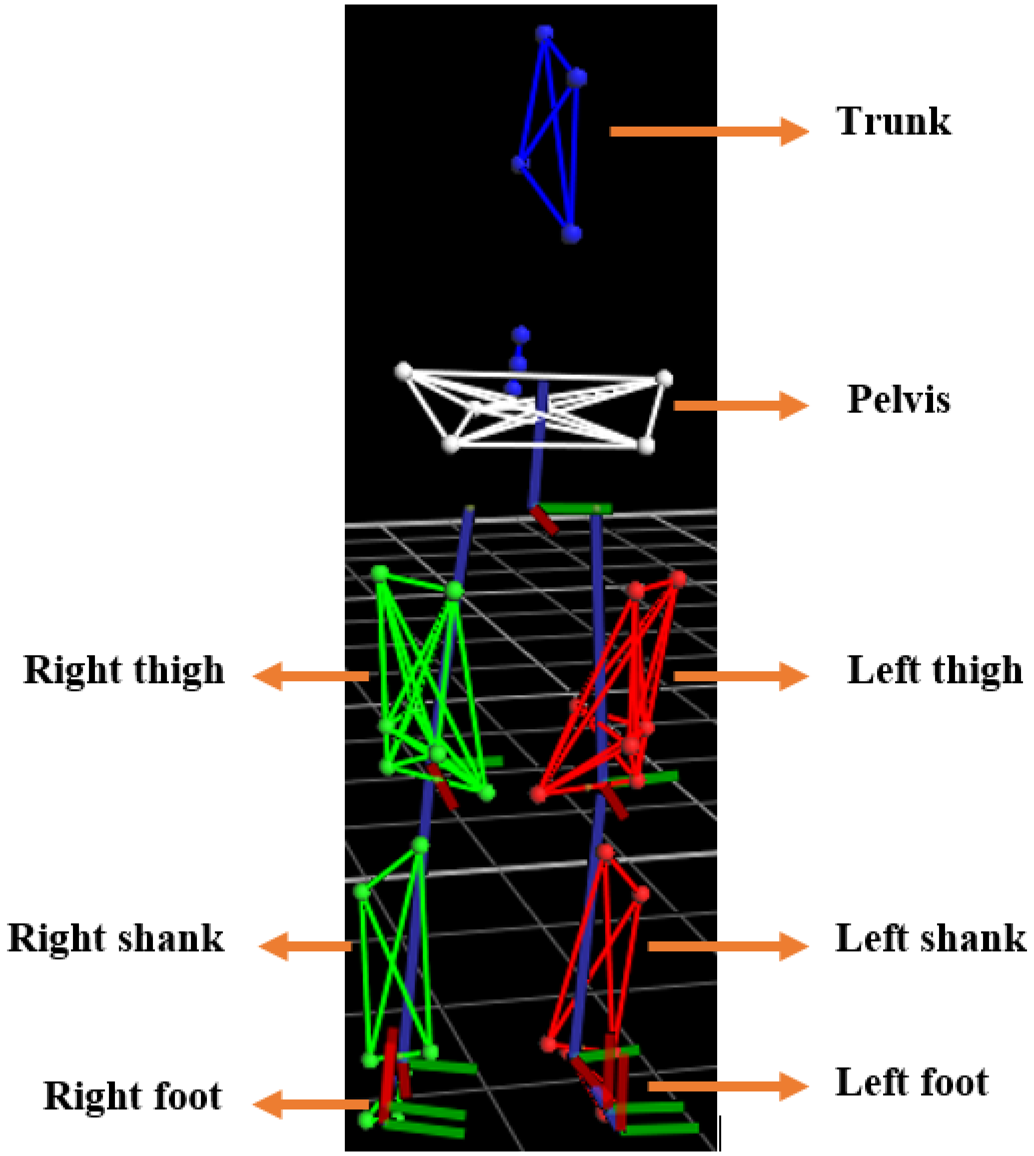


Figure 3.9. Rigid body segments

3.8.1.3 Marker location

The rigid bodies were defined based on series of retro-reflective markers that were attached on the skin either directly on specific anatomical landmarks (anatomical markers) or without any specific anatomical landmarks (technical markers) (Table 3.2). These anatomical markers can be divided into primary and accessory markers. The primary markers were used to determine the local coordinate system or for estimating the location of joint centres for each rigid body segment (Table 3.3), while the accessory markers were used to assist in determining the location of primary markers particularly when there was any presence of gaps in the primary markers' trajectory along the recorded trial. The gap filling process will be further explained in Section 3.8.1.4. For the trunk, all markers were primary markers, namely C7 vertebrae, T8 vertebrae and suprasternal notch (IJ) and xiphoid process (PX). For the pelvis, the primary markers were right and left anterior suprailiac spine (RASI & LASI) and posterior suprailiac spine (RPSI & LPSI). From the feasibility study, it was found that the anterior suprailiac spine markers were absent most of the time due to being occluded by the carrying container. Therefore, the decision was made to add markers to the most lateral part of the right and left suprailiac spine (RSIS & LSIS) as accessory markers in order to determine the absent primary markers of the pelvis.

For the thigh, the primary markers were right and left lower lateral 1/3 of the thigh (RTHI & LTHI), lateral femur epicondyle (RKNE & LKNE) and medial femur epicondyle (RMKNE & LMKNE). The original accessory markers were right and left lower medial 1/3 of the thigh (RMTHI & LMTHI). However, it was found in the feasibility study that the medial thigh markers were not effective in determining the missing gaps in the RTHI and LTHI trajectories due to limited view caused by the carrying container. Therefore, six other additional accessory markers were assigned at lower 2/3 of the thigh, namely right and left anterior thigh (RATHI & LATHI), lateral thigh (RLTHI & LLTHI) and posterior thigh (RPTHI & LPTHI) to improve the gap-filling process during data processing. For the knee, the primary markers were the right and left tibial tuberosity (RTUB & LTUB), tibia at lower lateral 1/3 of the shank (RTIB & LTIB), lateral malleolus (RANK & LANK) and medial malleolus (RMAK/LMANK). For the foot, the primary markers were the head of 2nd metatarsal (RTOE & LTOE), calcaneus (RHEE & LHEE) and the base of 5th metatarsal (R5thMET/L5thMET) (Figure 3.10 & Figure 3.11).

Table 3.2. Rigid body segment and markers

Segments	Markers	Description
Trunk	C7	The spinous process of 7th cervical vertebrae
	T8	The spinous process of 8th thoracic vertebrae
	IJ	Incisura jugularis (suprasternal notch)
	PX	Processes xiphoidens (xiphoid process)
Pelvis	RASI, LASI	Right/left anterior suprailiac spine
	RILC/LILC	Right/left iliac spine (most lateral aspect)
	RPSI/LPSI	Right/left posterior suprailiac spine
Thigh	RTHI/LTHI	Right/left thigh (lower lateral 1/3 of the thigh)
	RMTHI/LMTHI	Right/left medial thigh (lower medial 1/3 of the thigh)
	RLTHI/LLTHI	Right/left lateral thigh (lower lateral 2/3 of the thigh)
	RATHI/LATHI	Right/left anterior thigh (lower, anterior 1/3 of the thigh)
	RPTHI/LPTHI	Right/left posterior thigh (lower, posterior 1/3 of the thigh)
Shank	RKNE/LKNE	Right/left knee (lateral femur epicondyle)
	RMKNE/LMKNE	Right/left medial knee (medial femur epicondyle)
	RTUB/LTUB	Right/left tibia tuberosity
	RTIB/LTIB	Right/left tibia (lower lateral 1/3 of the shank)
	RANK/LANK	Right/left ankle (lateral malleolus)
	RMANK/LMANK	Right/left medial ankle (medial malleolus)
Foot	RTOE/LTOE	Right/left toe (over 2nd metatarsal head)
	RHEE/LHEE	Right/left heel (calcaneous)
	R5thMET/L5thMET	Right/left 5th metatarsal (lateral, base of 5th metatarsal)

Table 3.3. Local coordinate system for each rigid body segment

Body segments	**Origin**	**Axes directions**		
		X	**Y**	**Z**
Trunk	IJ	Direction of ½(C7 to T8) to ½(IJ to PX)	Cross product between X & Z unit vectors.	Direction of ½(IJ to C7) to ½(PX to T8)
Pelvis	½(RHJC to LHJC)	Cross product of Y & Z unit vectors	RASI to LASI	Perpendicular to the plane defined by RASI, LASI, RPSI, LPSI
Thigh	KJC	Perpendicular to the plane defined by HJC, KNE MKNE & THI	Cross product between Z & X unit vectors (left)	KJC to HJC
Shank	AJC = ½(ANK to MANK)	Perpendicular to the plane formed by TIB, AJC and KJC (anterior)	Cross product between Z & X unit vectors (left)	AJC to KJC
Foot	TOE	Cross product between Y & Z unit vectors	Perpendicular to the plane formed by TOE, AJC & KJC	TOE to AJC

IJ = suprasternal notch, C7 = 7th cervical vertebral process, T8 = 8th thoraxic vertebral process, PX = xiphoid process, RHJC & LHJC = right and left hip joint centres, RASI & LASI = right and left anterior suprailiac spine, RPSI & LPSI = right and left posterior iliac spine, KJC = knee joint centre, KNE = lateral femur epicondyle, MKNE = medial femur epicondyle, THI = lower lateral 1/3 of the thigh, AJC = ankle joint centre, ANK = lateral malleolus, MANK = medial malleolus, TIB = lower lateral 1/3 of the shank, TOE = over 2nd metatarsal head

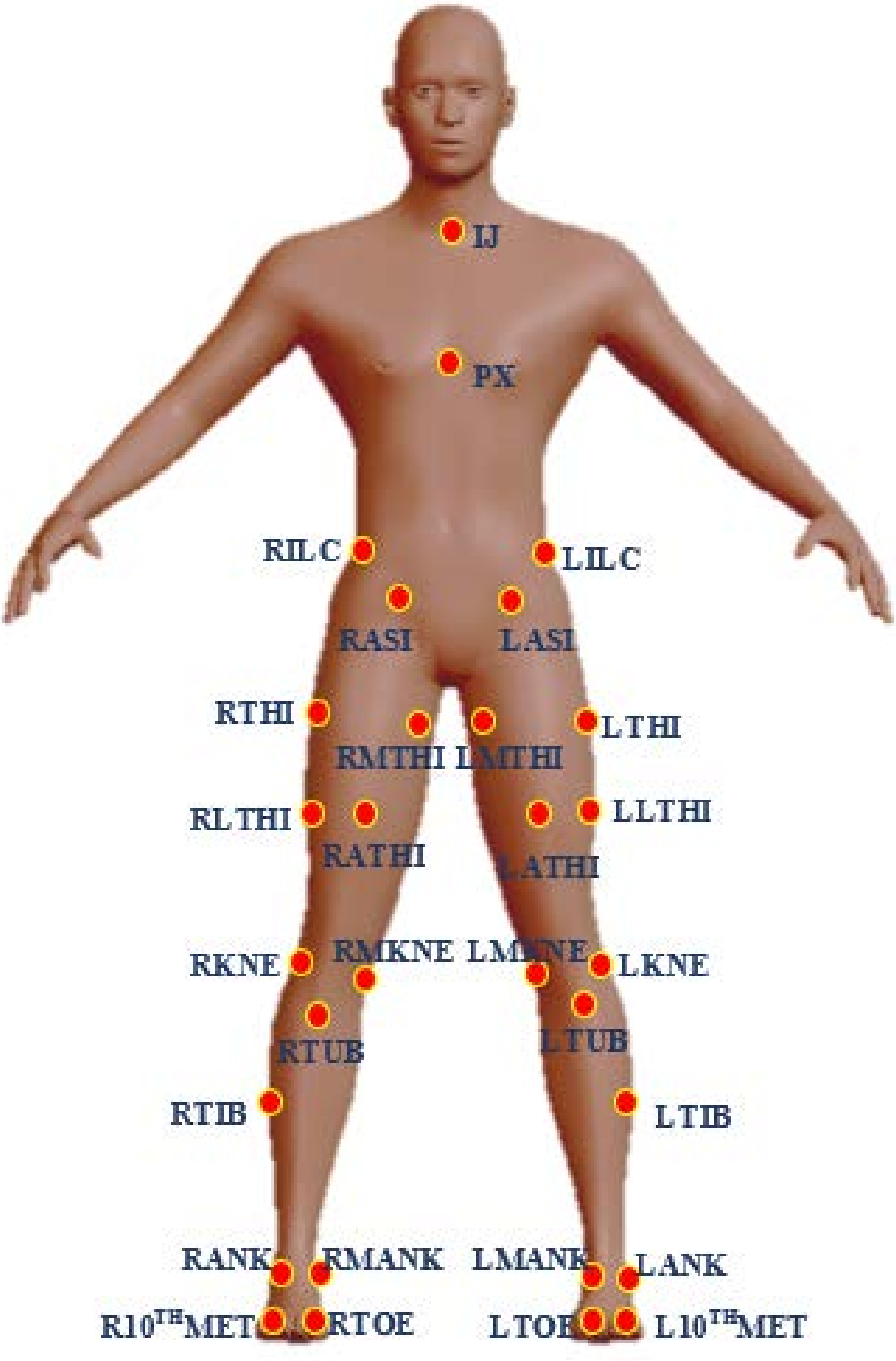


Figure 3.10. Position of reflective markers (anterior view)

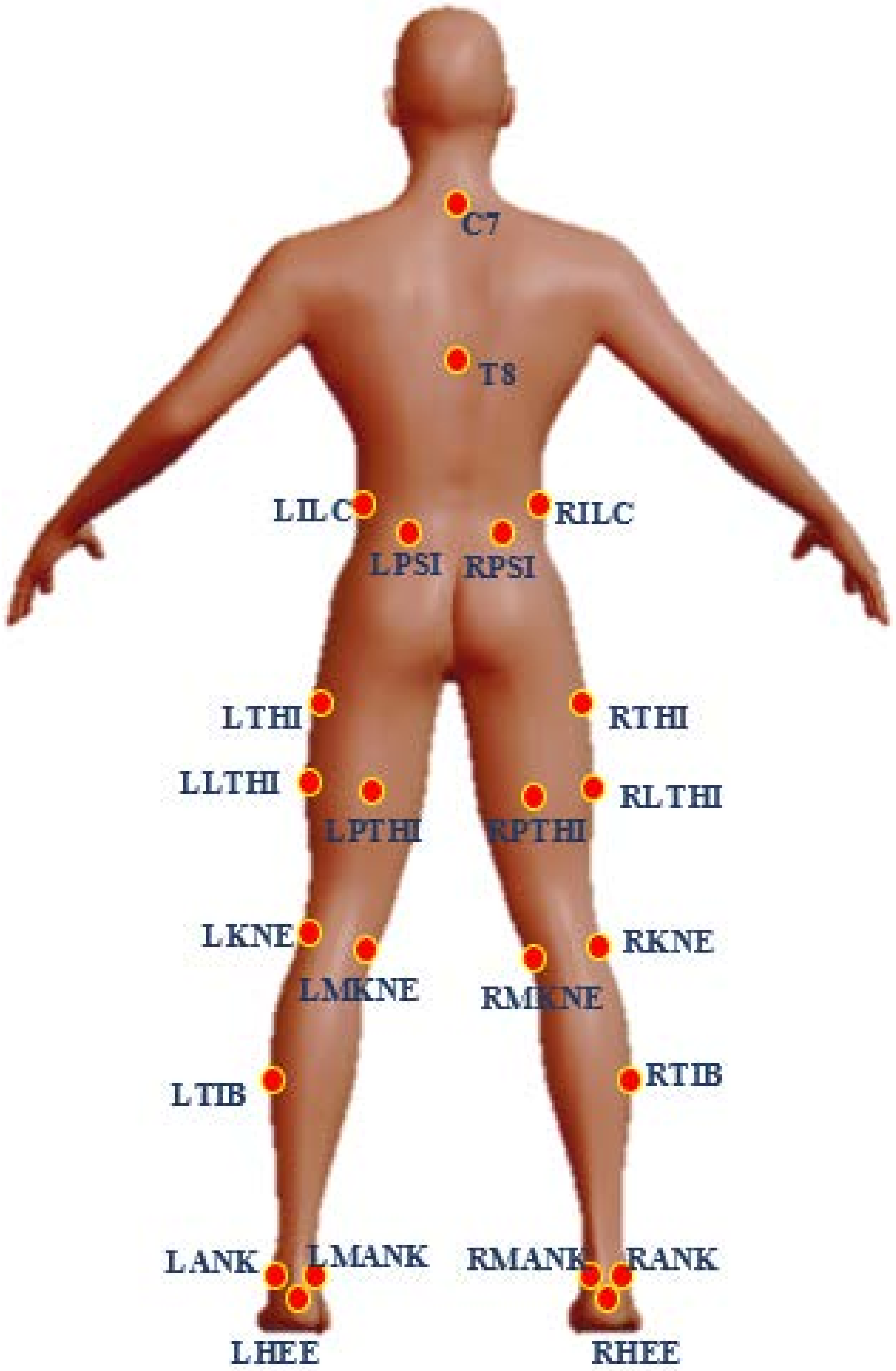


Figure 3.11. Position of reflective markers (posterior view)

3.8.1.4 Labelling and gap filling

After the kinematic trial was loaded and reconstructed, the first step was to label all the markers appropriately. Any gaps in marker trajectories (i.e. missing markers during a trial) were removed. In this study, the gap filling methods that were used to extrapolate the missing marker trajectory were spline fill, pattern fill and rigid body fill.

a. Spline fill

Spline fill was an automatic method that extrapolates the missing marker trajectory according to the last known and first recurring coordinates. However, the main limitation of this method was the larger the gap, the more likely this method will produce an unreliable result. The algorithm for the spline fill extrapolated the last trajectory based on where it was directed before the gap began. Usually, the gap occurred when the trajectory moved in a drastically irregular pattern, leading to an awkward discontinuation of the trajectory. In this case, the spline fill most likely will extrapolate the erratic direction based on the last known frame of data for the trajectory. Therefore, the spline fill was considered the weakest method of gap-filling, and was used only when the other two methods were failed.

b. Pattern fill

Unlike the spline fill, the pattern fill can be performed either automatically or by manually selecting a donor marker, which can be described as the marker which will most likely move in a similar motion with the missing marker. This method applied the displacement of the donor marker to the missing marker. To perform this method, a marker that shared the same rigid body segment with the missing markers will be chosen. For instance, when the anterior suprailiac spine marker (e.g. RASI) was missing, the other pelvis markers such as the posterior suprailiac spine (e.g. RPSI) or suprailiac spine (RSIS) can be used to determine the missing marker using the spline fill method.

c. Rigid body fill

Rigid body fill can also be performed either automatically or manually. This method determined the location of the missing marker with respect to a local coordinate system which was determined by the three donor markers. Based on this known location within the local coordinate system, the missing marker was reconstructed when a gap appeared. Three donor markers will be chosen from the same rigid body segment to extrapolate the movement of the missing markers. This method was preferable than the pattern fill because it was based on the trajectories of three donors instead of a single donor, thus, making a more accurate extrapolation. However, the rigid body fill cannot be performed to determine any missing trajectory in the foot segment, as the foot segment only had three markers (i.e. toe, 5th metatarsal and calcaneus).

3.8.1.5 3D angles and motions

There were two types of 3D movement: relative and absolute. The relative movement was based on the movement that occurs between two adjacent rigid body segments. The fixed body segment was called the parent segment, while the mobile body segment was called child segment. For instance, the movement around the knee joint was actually the movement of the shank relative to the thigh. To determine the movement of each body segments, a reference system had to be determined for each rigid body segment. This reference system was called local coordinate system (LCS). The absolute movement was the movement of a rigid body segment relative to the coordinate system of the laboratory or global coordinate system (GCS). For instance, the movement of pelvis was based on the GCS, as there was no involvement of parent segment. The movement 3D can be described based on one three axes of rotation, namely X (directs anteriorly), Y (directs laterally towards the left), and Z (directs cranially). The direction of each axis was based on an upright standing posture. In order to determine a local coordinate system, four parameters had to be determined which were the origin of movement X axis, Y axis and also Z axis (Figure 3.12).

For the trunk, the origin of movement was at IJ. Then, the midpoint of C7 to IJ and PX to T8 were determined. The direction between the two midpoints defined the Z axis. The X axis was defined based on the direction between the midpoint of C7 to T8 and IJ to PX towards the anterior. Finally, the Y axis direction was determined as the cross product between X and Z unit vectors, pointing to the left. For the pelvis, the initial origin was at the midpoint of RASI to LASI. The Y axis was determined from RASI to LASI towards the left. The Z axis was perpendicular to the plane defined by RASI, LASI, RPSI and LASI towards the cranial. The X axis was the cross product between Y and Z unit vectors towards the anterior. After hip joint centres (HJC) was determined based on Newington-Gage model, the origin of the pelvis LCS was then shifted to the halfway between left and right HJCs, while the axis orientation remains the same (Table 3.4). Finally, the kinematics was calculated based on the range between the highest and the lowest degree of rotation (peak-to-peak) across the gait cycle, which was referred as range of motion (ROM) throughout the thesis.

For the thigh, the origin of movement was at the knee joint centre (KJC). The KJC was determined as the halfway between the lateral and medial epicondyle of femur (e.g. RKNE to RMKNE). The Z axis was defined as the direction from the KJC towards HJC. The X axis was perpendicular to the plane defined by HJC, THI, RKNE and RMKNE. Finally, the Y axis was the cross product between Z and X unit vectors. For the shank, the origin of movement was at the ankle joint centre (AJC). The AJC was determined as the halfway between lateral (ANK) and medial malleolus (MANK). The Z axis was defined from AJC to KJC pointing superiorly. The X axis was perpendicular to the plane defined by lateral malleolus, medial malleolus, tibia (TIB) and KJC. Finally, Y axis was the cross product between Z and Y unit vectors towards the left side. For the foot, the origin of movement was at the toe markers (TOE). The Z axis direction was determined from TOE to AJC pointing posteriorly. The Y axis was perpendicular to the plane defined by TOE, AJC and KJC. Finally, the X axis was the cross product between Y and Z unit vectors (Table 3.4).

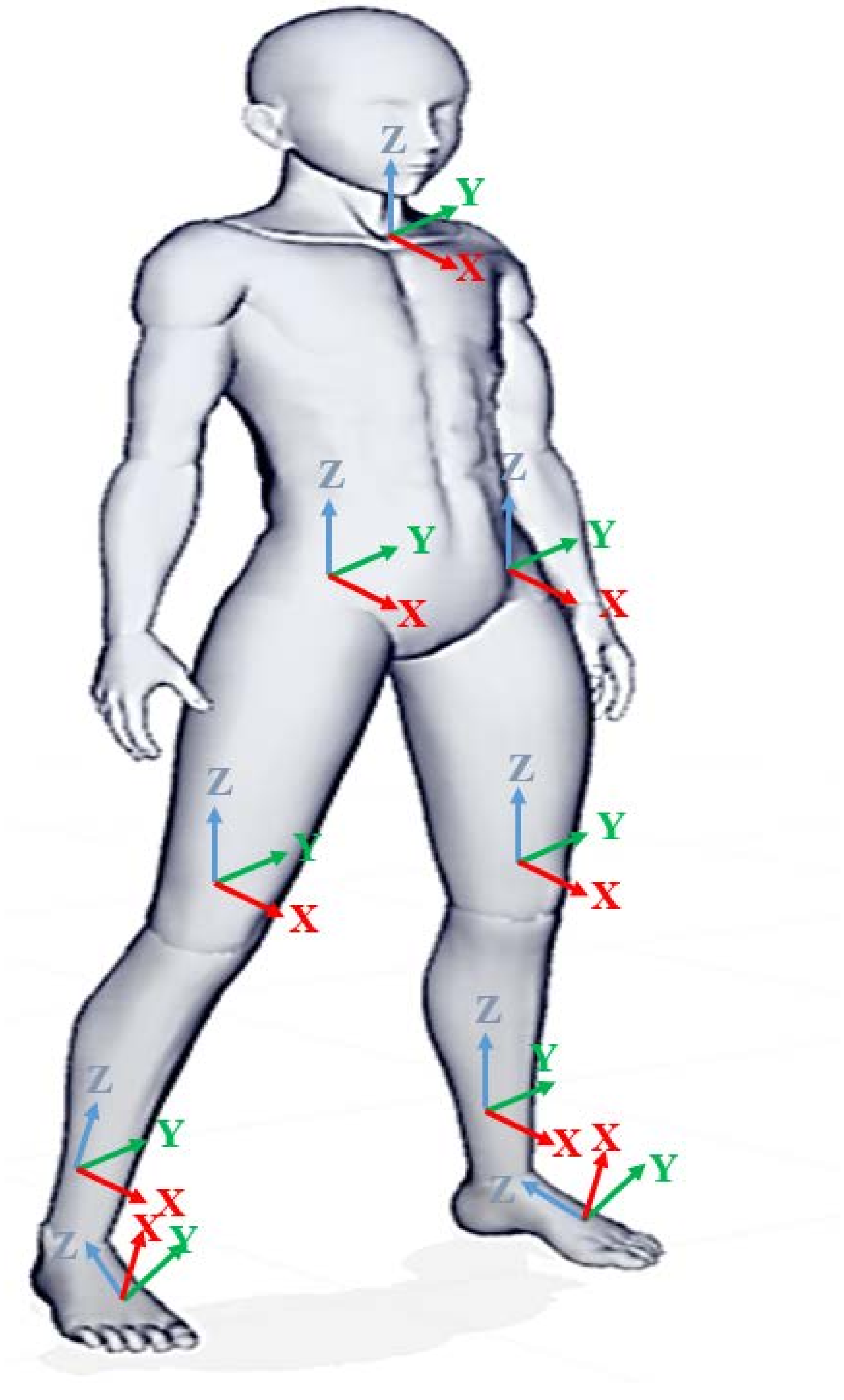


Figure 3.12. Local coordinate system for each rigid body segment

The pelvis, hip and knee angles were calculated in rotation order YXZ, whilst the ankle angle was calculated in rotation order YZX.

Table 3.4. Angles and direction of movement according to Vicon's Plug-in-Gait model

Angles	Sides	Planes	Positive Direction	Direction
Trunk (trunk relative to pelvis) & Trunk global (trunk relative to global coordinate)	Left	Sagittal	Extension	Clockwise
		Frontal	Lateral flexion	Anti-clockwise
		Transverse	Right axial rotation	Clockwise
	Right	Sagittal	Extension	Clockwise
		Frontal	Lateral flexion	Clockwise
		Transverse	Left axial rotation	Anti-clockwise
	Left	Sagittal	Anterior tilt	Anti-clockwise
		Frontal	Upward obliquity	Anti-clockwise
		Transverse	Internal rotation	Clockwise
	Right	Sagittal	Anterior tilt	Anti-clockwise
		Frontal	Upward obliquity	Clockwise
		Transverse	Internal rotation	Anti-clockwise
Hip (pelvis relative to thigh)	Left	Sagittal	Flexion	Clockwise
		Frontal	Adduction	Clockwise
		Transverse	Internal rotation	Clockwise
	Right	Sagittal	Flexion	Clockwise
		Frontal	Adduction	Anti-clockwise
		Transverse	Internal rotation	Anti-clockwise
Knee (thigh relative to shank)	Left	Sagittal	Flexion	Anti-clockwise
		Frontal	Varus	Clockwise
		Transverse	Internal rotation	Clockwise
	Right	Sagittal	Flexion	Anti-clockwise
		Frontal	Varus	Anti-clockwise
		Transverse	Internal rotation	Anti-clockwise
Ankle (shank relative to foot)	Left	Sagittal	Dorsiflexion	Clockwise
		Frontal	Inversion	Clockwise
		Transverse	Internal rotation	Clockwise
	Right	Sagittal	Dorsiflexion	Clockwise
		Frontal	Inversion	Anti-clockwise
		Transverse	Internal rotation	Anti-clockwise

Source: Vicon Motion Systems Ltd. UK (Accessed date: 1/11/16)

3.8.1.6 Spatiotemporal parameters of gait

In order to determine the spatiotemporal parameters, foot strike and foot off were determined in the Vicon Nexus for each gait trial. The spatiotemporal parameters were analysed based on one stride of the gait cycle (i.e. heel strike to heel strike of the contralateral leg). The gait cycle consisted of stance phase and swing phase. The stance phase was determined from foot strike until foot off, the swing phase was determined from foot off until foot strike. A stride was defined as heel strike to ipsilateral heel strike. A step was defined as heel strike to contralateral heel strike. For this study, the spatiotemporal parameters consisted of cadence, step length, step time, stride length, stride time and walking speed. Cadence was defined as the number of strides per minute. The left and right cadence were calculated separately based on a single stride. Step length was defined as the distance (meters) between the foot strike of one foot and the subsequent foot strike of the opposite foot, and the time taken (seconds) to complete the step length was called step time. The stride length was calculated as the distance (meters) between two successive placements of the same foot, which consists of two step lengths of left and right. The side of the step length was determined based on the foot from the side that moves forward in front of the contralateral foot. The time taken (seconds) to complete a stride length was called stride time. Finally, the walking speed was determined as the stride length divided by stride time.

3.8.2 Electromyography

3.8.2.1 Equipment

A ZeroWire (Aurion) wireless EMG system with 16 channels was used to record the muscle activity. Each transmission unit consists of a bipolar electrode and a probe with a built-in pre-amplifier and transmitter. The transmission unit was equipped with li-on rechargeable battery that can be used up to eight hours of continuous operation. The electrodes were combined with silver/silver chloride (Ag/AgCl) electrode gels (oval-shape). The standard range of operation was up to 20 meters.

Figure 3.13. ZeroWire (Aurion) wireless EMG

3.8.2.2 Electrode placement

Before the electrode was placed onto skin, the skin surface was cleaned using abrasive liquid and tissue. If the skin surface was covered with hair, the hair will be removed. The inter-electrode distance was fixed at 20mm. To determine the appropriate site for the electrode placement, a pen and a tape measurement were are used in conjunction with specific navigation technique for each muscle (mostly based on SENIAM) to locate the muscle (Table 3.5. Electrode Placement for Surface EMG electrode). Eight muscles were investigated in this study, which were illiocostalis, multifudus, gluteus maximus, biceps femoris, biceps brachii, latissimus dorsi, vastus lateralis and gastrocnemius.

Table 3.5. Electrode Placement for Surface EMG electrode

EMG channel Right	Left	Muscles	Electrode placement
1	9	Iliocostalis	One-finger width medial from the line from the PSIS to the lowest point of the lower rib, at the level of L2 (SENIAM 1999).
2	10	Multifidus	On and aligned with a line from caudal tip PSIS to the interspace between L1 and L2 interspace at the level of L5 spinous process (i.e. about 2 - 3 cm from the midline) (SENIAM 1999).
3	11	Gluteus maximus	At the middle of the line between sacral vertebrae and the greater trochanter. This position corresponds with the greatest prominence of the middle of the buttocks well above the visible bulge of the greater trochanter (SENIAM 1999).
4	12	Biceps femoris	At the middle of the line between ischial tuberosity and the lateral epicondyle of the femur (SENIAM 1999).
5	13	Biceps Brachii	On the line between the medial acromion and the fossa cubit at 1/3 from the fossa cubit (SENIAM 1999).
6	14	Latissimus dorsi	Approximately 4 cm below the inferior tip of the scapula, half the distance between the spine and the lateral edge of the torso (Criswell 2010)
7	15	Vastus lateralis	2/3 on the line from the anterior iliac spine superior to the lateral side of the patella (SENIAM 1999).
8	16	Gastrocnemius	1/3 of the line between the head of the fibula and the heel (SENIAM 1999).

3.8.2.3 Signal amplification and filtration

During muscle contraction, EMG electrode captures the electrical signal produced by the muscle fibres. The signals were pre-amplified by EMG sensors and transmitted to the EMG receiver unit (i.e. encoder) for further amplification. The analogue signals were sent from the EMG receiver unit to an analogue to digital convertor card within the Vicon Giganet interface unit and sampled at 1000Hz.

3.8.2.4 Signal analysis

Post data collection, the raw EMG data were filtered using a band pass Butterworth 4th order at 20Hz (low pass) and 500Hz (high pass) with common mode rejection ration (CMRR) of 90db. The digital signals were then sent to be displayed and recorded by the Vicon Nexus software. Once the raw EMG data were captured, these data were exported into Matrix Laboratory (MATLAB) software for further signal analysis. The EMG signals were comprised of electrical firings that occurred at different rates (i.e. frequencies). In the time domain (i.e. raw EMG), the overall signal was a composite of these multiple frequencies. The EMG frequency was measured in Hertz (Hz), and it can be described as the number of electrical firings from muscle fibres per second. There were two ways of characterising the frequency content of the signal; mean frequency and median frequency. The mean frequency was equal to the average frequency throughout the complete spectrum, whereas the median frequency divided the power density spectrum into two sections with an equal amount of power. According to De Luca (1997), the median frequency was considered a better indicator because it was less sensitive to noise and signal aliasing. When a muscle started to fatigue, the power density spectrum shifted to the left side of the frequency scale and consequently, median frequency decreased. This shift cannot be observed in raw EMG domain because although the frequency of firings decreased, the total amplitude in the time-domain can remain constant. Therefore, the median frequency was chosen as an indicator for muscle fatigue.

The median frequency, defined as the frequency that represents 50% of the power spectrum, was determined through a cumulative sum of the power distribution and subsequently determining the frequency that represent the 50th percentile of the power distribution. To construct the fatigue slope, the median frequency was calculated at each second of the test period. A 2nd order polynomial regression method was used to determine the line of best fit through the median frequency at each one second window. The slope of the regression line (MFslope) was normalised to the initial median frequency (i.e. subtracting the constant) and then used to determine the level of muscle fatigue. The more negative the slope, the more fatigue the muscle. The formula for the median frequency is presented below, where *j* is the frequency bin, *Pj* is the EMG power spectrum at the frequency bin *j*, and *M* is the length of frequency bin.

$$\sum_{j=1}^{MDF} Pj = \sum_{j=MDF}^{M} Pj = \frac{1}{2}\sum_{j=1}^{M} Pj$$

The raw EMG data were converted into the frequency domain using Discrete Fourier transformation method using MATLAB. The purpose of this transformation was to isolate each of the frequency bands into the frequency domain. In the frequency domain, the X axis displayed the frequency in Hz, while the Y axis displayed the power of the frequency. The level of muscle fatigue was determined using the slope of EMG median frequency (MFslope). To construct the slope, the median frequency was calculated at each second of the test period. Using a polynomial regression method lines of best fit, a regression line was fitted along the median frequencies. The slope of the regression line (MFslope) was used to determine the level of muscle fatigue. The more negative the slope would indicate the higher rate of muscle fatigue.

3.9 STATISTICAL ANALYSIS

The statistical analyses were conducted using the Statistical Package for Service Solution (SPSS) version 22, Matrix Laboratory (MATLAB) version R2015B and Microsoft Excel 2013. Descriptive and inferential statistics, hypothesis significance testing (NHST) and effect size method were used to interpret the statistical results.

3.9.1 Measures of central tendency and variability

In order to report the measures of central tendency and variability, an understanding of how well the data were distributed was crucial in order to choose the appropriate statistical test. The normality of the distribution was determined using Shapiro-Wilk (SW test), skewness, kurtosis and histogram. Normality was assumed if the SW test was not significant, as well as if the skewness and kurtosis values lie within two standard errors of these parameters respectively. If the data were normally distributed, mean and standard deviation were presented to indicate both measures respectively. However, if the data were not normally distributed, median and interquartile range (IQR) were presented respectively instead. A skewed distribution will affect the mean (as well as its standard deviation), as the mean will move towards the longer tail of the skewed distribution relative to the median. However, in a skewed distribution, the median was more robust compared to the mean because the value was calculated as an orderly ranked value rather than using the actual values. Thus, the IQR is the appropriate measure of variability for non-normally distributed data, as the IQR was the range within $\pm$ 25th percentile around the median (i.e. between 25th and 75th percentile).

3.9.2 Between-group, within-group and mixed-group comparisons

There were three types of statistical comparison; between-group, within-group and mixed-group comparison (Field 2013). The between-group comparison (denoted as main effect 1) can be defined as the comparison between two or more different groups on the same time (e.g. difference in gait kinematics between sedentary individuals and manual workers) (Figure 3.14). An Independent *t* test (or Mann-Whitney test if non-parametric) was used to compare between two different groups, while a One-Way Independent ANOVA (or Kruskal-Wallis test if non-parametric) was used to compare more than two groups. The within-group (denoted as main effect 2) can be described as the comparison between different conditions (e.g. difference in gait kinematics with and without carrying a load) that occur in the same group (Figure 3.15). A Paired *t* test (or Wilcoxon Signed-Rank test if non-parametric) was used to compare between two different conditions, while a One-Way Repeated Measure ANOVA (or Friedman test if non-parametric) was used to compare between more than two-conditions. Both comparisons can also be expressed in general linear model expression, which was described as the effect of independent variable (either between-group or within-group) on the dependent variable (e.g. gait kinematics).

When both between-group and within group were included into the comparison (mixed), the Split-Plot ANOVA (SPANOVA) was used to determine the effects of both between-group and within-group variables on the dependent variable. Although individual tests for between-group (e.g. independent *t* test for each standard gait and max-kg gait) and within-group (e.g. paired t test for each sedentary and manual group) can be performed, the tests were most probably will result in inflated type-I error (also known as family-wise or experiment-wise error). The familywise error can be calculated as 1- $(0.95)^n$, where n was the number of tests (e.g. four times of *t* tests). In other words, a single test (if applicable) was better than several separated tests. The interaction effect produced by the SPANOVA can be interpreted almost as similar as the individual tests, but with a corrected type-I error.

To check the normality assumption in SPANOVA, the univariate normality of the dependent variable concerning both types of independent variables had to be tested separately in the SPSS (e.g. manual-standard gait, manual-max-kg, sedentary-standard gait, and sedentary-max kg gait). In small-sampled studies, this could lead to non-normal distribution because smaller sample generally tend to be more sensitive to presence of outliers. Therefore, rather than depending on univariate normality alone, the overall influence of a case on the model was investigated using Cook’s distance. Any case that had Cook’s distance value more than one indicates the presence of influential case (also called multivariate outlier) (Cook and Weisberg 1982). Nevertheless, the SPANOVA was a robust test that can allow minimum to moderate violation of the assumption (Field 2013). However, if the normality was seriously violated (e.g. all graphical and statistical normality tests strongly suggest non-normality), the nonparametric test will be applied. Because there was no equivalent nonparametric test to the SPANOVA, the nonparametric test had to be conducted individually. To minimize the familywise error, Bonferroni correction was used to correct the inflated type-I error in the non-parametric tests. The Bonferroni correction to the type-I error was calculated as 5/n, where n was the number of test.

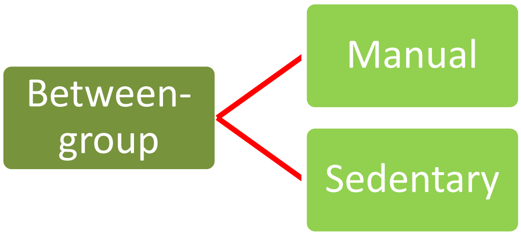


Figure 3.14. Example of between-group comparison (main effect 1)

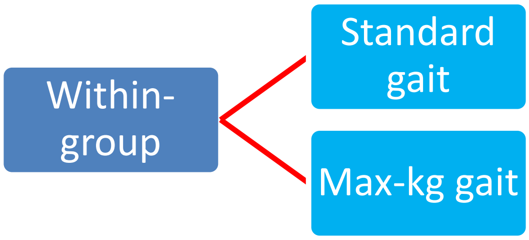


Figure 3.15. Example of within-group comparison (main effect 2)

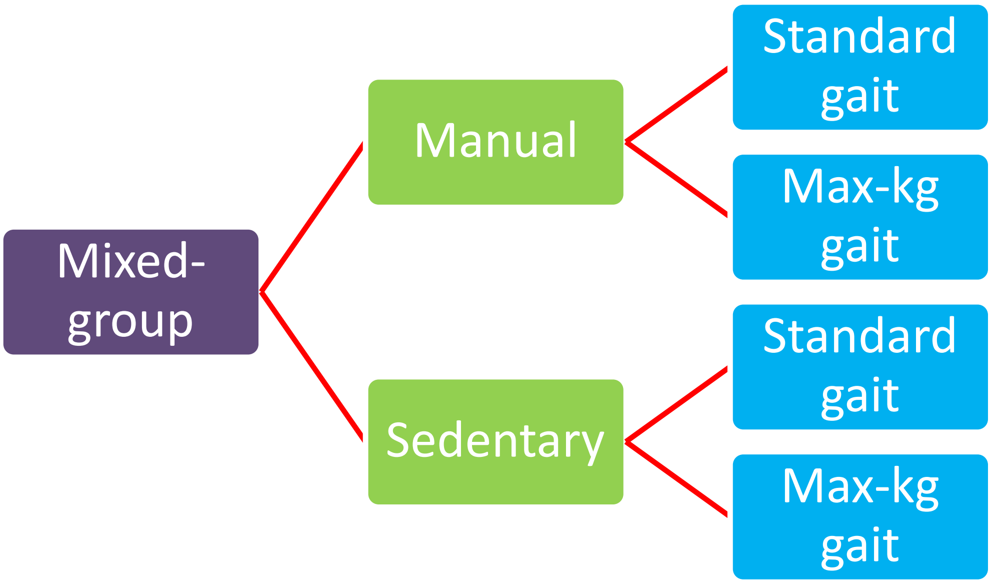


Figure 3.16. Example of mixed-group comparisons (main effect 1, main effect 2 & interaction effect)

3.9.3 Null Hypothesis Significant Testing

Null Hypothesis Significant Testing (NHST) is a conservative method of drawing conclusion from a statistical test. A null hypothesis (denoted as Ho) can be described as a type of hypothesis that contains nil-effect (e.g. no difference, association or prediction) which one most likely seeks to reject. Conversely, an alternative hypothesis (denoted as H_A) can be described as a type of hypothesis that contains the effect which one seeks to prove as true. In the NHST, type-I is the main criteria to determine whether to reject or not to reject the Ho. The type-I (false positive) error can be defined as the error of rejecting the Ho when the Ho is true. Conversely, type-II (false negative) error can be defined as the error of not rejecting the Ho when the Ho is false. In general all statistical tests have the tendency to make the errors of false positive and false negative. However, only the false positive is determined in the NHST as the error is directly related to the aim of the NHST in research, which is to indicate that there is a change in the conventional way of thinking about a certain phenomenon by rejecting the Ho. Although the false negative may still occur, the error is considered less serious because 'not rejecting' the Ho (whether Ho is true or false) would implicate similar treatment to the phenomenon (i.e. no change from traditional way of thinking). In this study, type-I error was set at 5%. The method of performing the NHST were described as below:

I. State the H_0 (e.g. trunk ROM in sedentary individuals is similar with manual workers)
II. State the H_A (e.g. trunk ROM is higher in sedentary individuals compared to manual workers)
III. Determine the acceptable type-1 error (e.g. 5%)
IV. Perform the statistical test (e.g. independent *t* test)
V. Determine the type-I error from the statistical test
VI. Draw conclusion: reject or do not reject the Ho

However, there were some limitations in using the NHST in making a statistical conclusion, as the type-I error was sensitive to the sample size (Sullivan and Feinn 2012). For instance, in an over-sampled study, the type-I error will almost always be significant, even with the presence of large effect size. Contrariwise, the type-I error will almost always be non-significant in under-sampled studies, even though the effect sizes were large. Hence, reporting both type-I error and effect size were crucial to provide a comprehensive perspective of the results.

3.9.4 Effect size

A significant test from an NHST (i.e. $p<0.05$) can only prove that an effect is present not just by chance. However, the *p* value cannot tell whether the effect has large, medium or high magnitude. Therefore, the effect size was calculated for each inferential statistics to indicate the magnitude of an effect. According to American Psychological Association (APA), it was important to include both *p* value (from NHST) and the effect size when reporting the result of an inferential statistics (Vacha-Haase et al. 2000). The common method of interpreting the effect size was by comparing the value with its conventional range (Cohen 1988a). For *t* tests (e.g. one-sample, independent & paired), the effect size was the Cohen's *d* (conventional range: 0.2 = small, 0.5 = moderate & 0.8 = large). For *F* tests (e.g. ANOVAs), the effect size was the partial eta squared (conventional range: 0.01 = small, 0.06 = medium & 0.14 = large) or Cohen's *f* (conventional range: 0.1 = weak, 0.25 = medium, 0.40 = strong) (Cohen 1988a). For correlations, the correlation coefficient (*r*) itself already indicates the strength of correlation (strength: 0.3 = small, 0.5 = moderate, 0.7 = strong) (Rumsey and Unger 2015). For linear regression, the effect size was the coefficient of determination (R^2), which can be described as the percentage of variability in dependent variable that can be explained by respective independent variable/s. For most of the nonparametric tests, the z score from the test result can be converted into *r* from $z/\sqrt{n}$ (conventional range: 0.1 = small, 0.3 = medium & 0.5 = large) (Cohen 1988a). However, Allen and Bennett (2012) suggested the conversions were only suitable for two-group non-parametric comparisons (e.g. Mann-Whitney & Wilcoxon Signed-Rank test).

Chapter 4: RELIABILITY STUDY OF 3D GAIT ANALYSIS, ISOMETRIC BACK ENDURANCE AND MUSCLE FATIGUE

4.1 INTRODUCTION

It is important to establish measurement reliability in any empirical study so that researcher or clinician can ensure the credibility of the outcome measure. One of the main purpose of reliability testing is to measure the stability of the measurement across different sessions, and this can be determined using test-retest reliability. Most of the content from this chapter were mainly used during the MPhil-to-PHD upgrade viva prior to confirming PhD candidature. The main parameters for this PhD study were 3D kinematics during gait, the isometric back endurance and muscle fatigue. The 3D gait analysis was determined based on the Plug-in-Gait model, while the isometric back endurance and muscle fatigue were determined based on holding and the slope of electromyography (EMG) median frequency (MFslope) time during the Ito test respectively. Across the literature, there were various studies reported on the reliability of 3D gait analysis. For instance, one of the earliest studies was conducted by Kadaba et al. (1989) on healthy participants. From the study, the test-retest reliability of movements in sagittal planes for hip, knee and ankle were excellent (ICC: 0.93 to 0.99). A more recent study by Yavuzer et al. (2008) investigated the between-session and within-session reliability of 3D kinematics of gait in stroke patients and found that the 3D kinematics was high for the paretic limb (CMC: 0.85 to 0.95). On the other hand, the Ito test was first introduced by Ito et al. (1996) as a reliable, safe and simple method of determining back endurance between the healthy individuals and the low back pain patients. In a study by Arab et al. (2007), they had concluded that the Ito test displayed good sensitivity, specificity and predictive value for back endurance tests. Although the measurement reliability were already reported in the literature, different studies may be exposed to internal and external errors in the measurement process that were unique to each study design (Steinwender et al. 2000).

In general, there are many factors that may influence the variability of measurement, which can be divided into two broad categories; intrinsic and extrinsic (McGinley et al. 2009). Intrinsic factors consist of individual variability among normal individuals or those with pathology and it cannot be changed (Schwartz et al. 2004). For instance, walking speed was reported to influence almost all angles in gait cycle. In this case, walking speed can be considered as an intrinsic factor because it was a natural variability of any individual that cannot be avoided (Stansfield et al. 2001). Extrinsic variability derives from technical factors such as experimental setting. Reliability studies that use the data from different sessions are susceptible to this type of error. However, proper methodological setting can help to minimize the error. For instance, marker placement between sessions can also contribute to extrinsic variability. Maynard et al. (2003) reported that the reason why the knee joint had the highest reliability while the hip joint had the lowest reliability was that the anatomical landmark for the knee can be easily identified compared to the hip. Therefore, the aim of this chapter was to determine the within-session and between-session reliability for the main measurements used in the PhD study. The within-session session reliability was determined based on repeated trials within a same experimental session, while the between-session reliability was determined based on two separate experimental sessions.

4.2 MATERIALS AND METHODS

4.2.1 Study Design

As a part of main PhD study that investigate the lower limb kinematics, spatiotemporal parameters of gait, muscle fatigue and pelvis-trunk coordination during carrying activity, a reliability study was conducted to investigate within-session and between-session reliability of the measurement. The data were collected from May 2014 to April 2015 at the Biomechanics Laboratory, Faculty of Health Sciences, University of Southampton. The within-session reliability was tested on three trials for standard gait and carrying activity whilst carrying a maximum load (i.e. max-kg gait). The measurements involved were spatiotemporal parameters of gait and 3D kinematics of a standard gait and carrying activity. For the between-session reliability, the measurements involved were the Ito test (holding time), back muscles fatigue during the Ito test and the spatiotemporal parameters and 3D kinematics of a self-preferred gait (i.e. standard gait).

4.2.2 Participants

The participants' inclusion and exclusion criteria were described as 3.2 and the recruitment strategies were described as 3.5.

4.2.3 Procedure

4.2.3.1 Measurements

Participant anthropometrics which consisted of weight (kg), height (cm), leg length (cm), knee width (cm) and ankle width (cm) were taken (see section 3.7.1 for detail). Subsequently, the participants were instructed to perform the Ito test (see section 3.7.2 for detail). During the test, the time the participants were able to keep their chest off the table to maintain a maximum back extension was recorded (i.e. holding time) to indicate their level of isometric muscle endurance. At the same time, the surface electromyography (EMG) of iliocostalis, multifidus, gluteus maximus and biceps femoris were also recorded to measure their level of muscle fatigue based on the slope of EMG median frequency (MFslope). After a rest, motion analysis markers and the remaining EMG electrode (i.e. biceps femoris, latissimus dorsi, vastus lateralis and gastrocnemius) were put onto specific sites of the body in preparation for gait activities, which were the standard gait and carrying activity. Prior to any gait activities the participants were asked to statically stand on one force platform for ten seconds whilst a recording of kinematic and EMG data were made. The participants then performed at least three trials of standard gait along a 10m walking platform. During the standard gait, both kinematic and EMG data were recorded to obtain a baseline measure of the participants' gait. The researcher explained and demonstrated to participants how to carry a plastic container whilst walking. A plastic container was carried by holding the container's handle, flexing the arm at 90° of elbow flexion and keeping the container as close as possible to the body. A set of carrying activity was performed by walking back and forth along the 10m walking platform. A minimum number of three good trials were captured for each increment of carrying activity (see section 3.7.5 for detail).

4.2.3.2 Within-session reliability

Within session was assessed on spatiotemporal and kinematic parameters on the three repeated trials of standard gait and maximum load carriage. The Ito test was determined based on a single trial, while the muscle fatigue was determined based on a 5-second period static standing the end of each gait condition. Because there were no repeated trials within-session reliability cannot be conducted for the Ito test (both holding time and MFslope during Ito test). Unlike the Ito test, which had been tested for between session-reliability, there was no reliability measure for the MFslope across the gait conditions (i.e. standard gait and max-kg gait). This is because the MFslope across the gait conditions was determined based on the median frequency from no loads towards the maximum load, and this whole process of load carriage was not repeated.

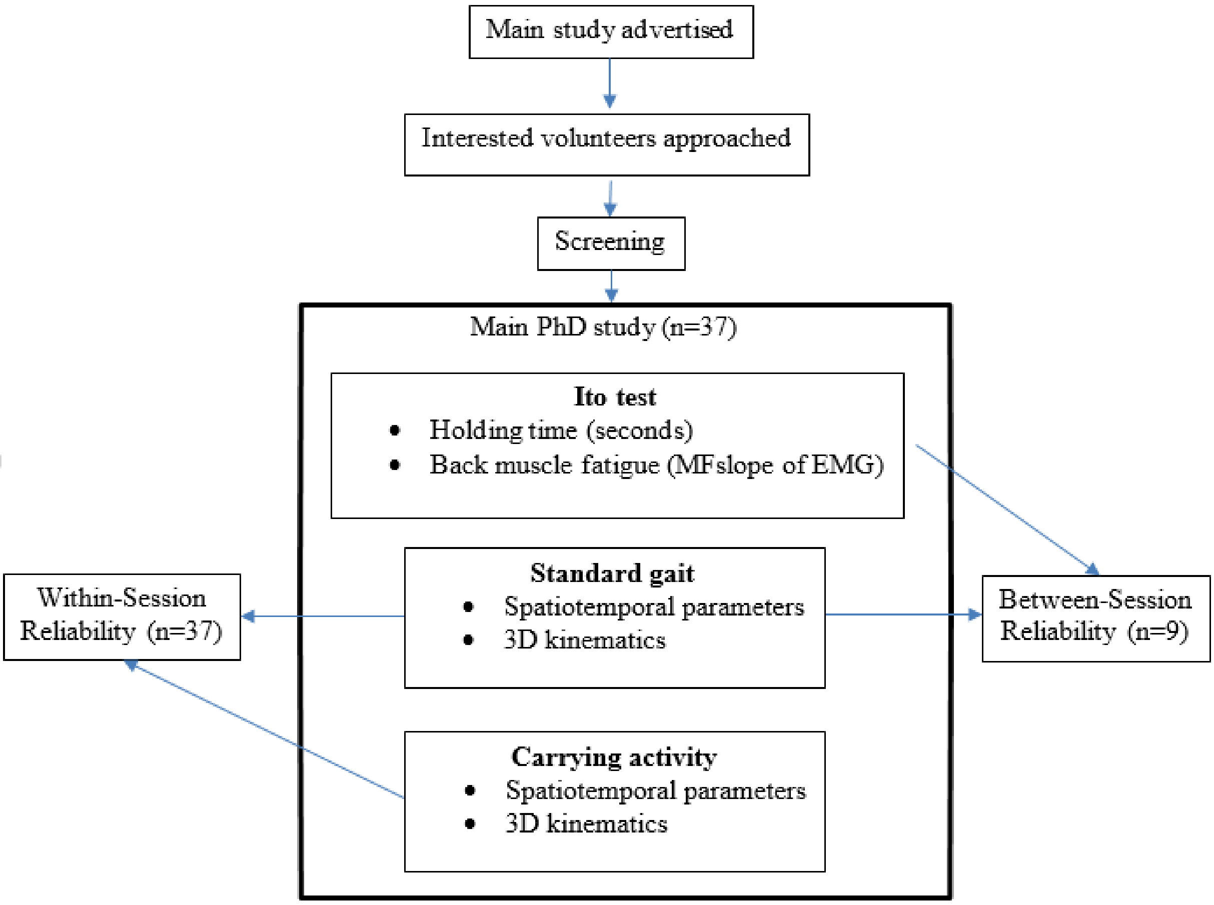


Figure 4.1. Measurements in within-session and between-session reliability

4.2.3.3 Between-session reliability

Between-session reliability was determined based on two-week interval for spatiotemporal parameters, the Ito test (holding time), the level of muscle fatigue during the Ito test as determined according to the slope of EMG median frequency (MFslope) and 3D kinematics of during gait. The spatiotemporal parameters consisted of cadence, step length, step time, stride length, stride time and walking speed. The level of muscle fatigue was determined for ilocostalis, multifidus, gluteus maximus and biceps femoris muscles. The 3D kinematics was based on the movements around ankle, knee, hip, pelvis and trunk (relative to pelvis). The between-session reliability test was conducted using nine participants. However, for the MFslope, only eight participants were analysed as raw EMG data for the second session was corrupted in one of the participants.

4.2.4 Data Processing

To be able to perform statistical analysis, the raw data from motion analysis and electromyography was processed in order to produce 3D kinematics and MFslope respectively (see section 3.8 for detail). For the motion analysis, the data processing was divided into two main aspects; spatiotemporal parameters analysis and 3D kinematics. The first stage was carried out by analysing the spatiotemporal parameters i.e. cadence, step length, step time, stride length, stride time and walking speed. The spatiotemporal parameters for the left and right sides were determined for each participant. The second stage was carried out by analysing the 3D kinematics of movements during standard gait and max-kg gait. For the ankle and knee, only flexion-extension (sagittal plane) movement was processed. For the hip, both flexion-extension (sagittal plane) and adduction-abduction (frontal plane) movements was processed. For the pelvis and trunk, the movements of flexion-extension/pelvic tilt (sagittal plane), lateral flexion/pelvic obliquity (frontal plane) and horizontal rotation (transverse plane) were determined (please see section 3.8.1.4 for detail). The trunk kinematics were analysed according to reference side (i.e. left trunk and right trunk). For instance, the right and left trunk kinematics can be defined as the 3D movements of the trunk during right and left side gait cycle respectively. The movements at each joint/body segment were measured as a degree of rotation and were plotted as waveforms.

For the Ito test, the muscle fatigue was determined using the slope of EMG median frequency (MFslope) (see section 3.8.2 for detail). In order to determine muscle fatigue, the raw EMG data were converted into the frequency domain using Fourier transformation method within MATLAB and the power of each frequency in Hz was determined. The Ito test trial was split into one-second windows where the median frequency was calculated. This median frequency, defined as the frequency that represent 50% of the power spectrum, was determined through a cumulative sum of the power distribution and subsequently determining the frequency that represents the 50^{th} percentile of the power distribution. To construct the fatigue slope, the median frequency was calculated at each second of the test period. A 2^{nd} order polynomial regression method was used to determine the line of best fit through the median frequency at each one second window. The slope of the regression line (MFslope) was normalised to the initial median frequency (i.e. subtracting the constant) and then was used to determine the level of muscle fatigue. The more negative the slope, the more fatigue the muscle.

4.2.5 Statistical Analysis

4.2.5.1 Within-session reliability

The level of agreement of the Ito test and spatiotemporal parameters of gait were calculated based on 2-way-mixed, single measure, intra-class correlation coefficient (ICC). Intra-class correlation coefficient (ICC) was a measure of agreement between continuous measurements. The ICC was graded as good (>0.75), moderate (0.5 to 0.75) and poor (<0.5) (Portney and Watkins 2000). In general, the ICC was calculated based on the ratio of the variance of interest to the sum of variance and measurement error. According to Shrout and Fleiss (1979), there were six models of ICC. The first integer of the model (i.e. 1, 2 or 3) represented types of study design, while the second integer represented (i.e. 1 or k) the unit of analysis (i.e. single measure or average measure). For ICC (1,1) model (one-way random, average measure), each participant was assessed by a set of different randomly-chosen raters (average measure). For ICC (2,1) each participant was measured by each rater, and raters were considered representative of a larger population of similar raters (average measure). For ICC (3,1), each subject was assessed by each rater of interest (average measure). For the standard gait and max-kg gait, the ICC (3,3) model was chosen because each participant was measured using only specific assessment of interest and the reliability was measured based on an average of three trials. The formula for the ICC (3,3) model was described below (Equation 1), where BMS = between-subject mean squares, WMS = within-subject mean squares and RMS = between-rater mean squares based on the ANOVA model.

$$\mathrm{ICC(3,3)} = \frac{\mathrm{BMS} - \mathrm{WMS}}{\mathrm{BMS}}$$

Equation 1. Intra-class correlation coefficient (3,3) model

The standard errors of measurements (SEM) for continuous measurements (i.e. holding time, MFslope and spatiotemporal parameters) were determined using within-subject standard deviation (WSSD). The WSSD was calculated as below (Equation 2), where the WMS was the within-subjects mean squares based on the ANOVA model.

$$WSSD = \sqrt{WMS}$$

Equation 2. Within-subject standard deviation (WSSD)

For 3D kinematics, the use of ICC to determine the level of agreement was difficult because it needs to be carried on 101 points along the movement waveforms across the gait cycle (discrete data). Therefore, the coefficient of multiple correlation (CMC) was applied due its ability to determine the degree of consistency of between the waveforms (Kadaba et al. 1989; Piotter et al. 1999). The CMC was graded as good (>0.75), moderate (0.5 to 0.75) and poor (<0.5) (Portney and Watkins 2000). The formula for the CMC (intra-tester) was shown below (Equation 3), where D was day (session) and T was the number of compared time points. In this formula, $\bar{Y}_{dt}$ represented the average of all trials within a session, $\bar{Y}_t$ represented the average of all trials and all session, and $\bar{Y}$ represented the overall average of all sessions, trials and time points.

$$\mathrm{CMC\ (intrarater)} = 1 - \sqrt{\frac{\sum_{d=1}^{D}\sum_{t=1}^{T}(\bar{Y}_{dt} - \bar{Y}_t)^2/T(D-1)}{\sum_{j=1}^{N}\sum_{i=1}^{T}(\bar{Y}_{dt} - \bar{Y}\ \ )^2/(DT-1)}}$$

Equation 3. Correlation of Multiple Correlation (CMC)

For the waveforms, the standard error of measurement (SEM) was determined according to waveform measurement error (WE) (Schwartz et al. 2004). The WE was calculated using the formula below (Equation 4), where *p* represented the trial counting index ranging from on to the total number of trials ($N_{total} = N_{respondent} \times N_{rater} \times N_{session} \times N_{trial}$) and Ø represents gait variable (e.g. hip flexion-extension).

$$\text{WE} = 1 - \sqrt{\frac{1}{N_{total} - 1}\sum_{p=1}^{N}(\Delta\emptyset^{source})^2}$$

Equation 4. Waveform measurement error (WE)

4.2.5.2 Between-session reliability

For the between-session reliability, the level of agreement (ICC) was determined for the Ito test (holding), muscle fatigue during the Ito test (i.e. MFslope of multifidus, iliocostalis, gluteus maximus and biceps femoris) and spatiotemporal parameters of gait. The ICC were calculated based on 2-way-mixed model based on single measure. The standard error of measurement was also based on WSSD. The formula for the ICC (3,1) was shown below (Equation 5), where BMS represented between-subject mean squares, WMS represented within-subject mean squares and RMS =represented between-rater mean squares.

$$\text{ICC}(3,1) = \frac{\text{BMS} - \text{WMS}}{\text{BMS} + (\text{K} + 1)\text{WMS}}$$

Equation 5. Intra-class correlation coefficient (3,1) model

For the 3D kinematics, the level of measurement was determined using the CMC (repeated trials). The formula for the CMC (repeated trials) was shown below (Equation 6), where M represented number of session, N represented number of trials and T represented the number of time points. The waveform measurement error was determined using equation 3.

$$\text{CMC (intrarater)} = 1 - \sqrt{\frac{\sum_{m=1}^{M}\sum_{n=1}^{N}\sum_{t=1}^{T} \quad (Y_{mnt} - \bar{Y}_{mt})^2/MT(N-1)}{\sum_{m=1}^{M}\sum_{n=1}^{N}\sum_{t=1}^{T} \quad (Y_{mnt} - \bar{Y}_{t})^2/M(NT-1)}}$$

Equation 6. Coefficient of multiple correlation (repeated trials)

The limits of agreement based on the Bland-Altman plot was also determined for the Ito test (holding), muscle fatigue during the Ito test and spatiotemporal parameters of gait. To determine the limits of agreement, Bland-Altman plot was constructed by plotting the differences between sessions against the mean for both sessions. The 95% limits of agreement (LOA) was then determined based on the formula below (Equation 7), where the mean differences was calculated based on the mean of differences between of two sessions.

Upper limit of 95% LOA = mean difference + 2SD of mean difference

Lower limit of 95% LOA = mean difference - 2SD of mean difference

Equation 7. 95% limits of agreement (LOA)

4.3 RESULTS

4.3.1 Within-Session Reliability

4.3.1.1 Participant characteristics (N=37)

A total of 37 participants were recruited for the study (Table 4.1). All participants were right-handed. Based on the results, the participants' mean age was 31.54±8.48, ranging from 18 to 39 years old). The mean of the body mass index (BMI) slightly fell into overweight category (26.20±3.62). For physical activity, the participants spent most of the time sitting (360.00±194.94 minutes/day), followed by vigorous activity (100.54±82.36 minutes/day), walking (85.19±74.56 minutes/day) and moderate activity (67.70±57.19 minutes/day).

Table 4.1. Participant characteristics for within-session reliability study

Variables	**Mean**	**Standard deviation**
Age	31.54	8.48
Height	168.74	5.90
Weight	74.57	10.79
Body mass index (BMI)	26.20	3.62
Physical activity*		
Vigorous activity: days/week	3.03	2.40
Vigorous activity: minutes/day	100.54	82.36
Moderate activity: days/week	2.84	2.22
Moderate activity: minutes/day	67.70	57.19
Walking: days/week	6.03	1.36
Walking: minutes/days	85.19	74.56
Sitting on weekday: minutes/day	360.00	194.94

*Measured based on The International Physical Activity Questionnaire (IPAQ)

4.3.1.2 Standard gait

a. Spatiotemporal parameters

Intra-class correlation coefficient (ICC) and within-subject standard deviation (WSSD) were used to examine the level of agreement (within-session) and variability respectively for spatiotemporal parameters (i.e. cadence, step length, step time, stride length, stride time and walking speed) for standard gait (Table 4.2). According to the results, all spatiotemporal parameters had good level of agreements for both left and right gait cycles in standard gait (ICC: 0.933 to 0.980). According to WSSD, the variability for all spatiotemporal parameters were relatively small (WSSD: 0.017 to 2.728).

Table 4.2. Level of agreement of spatiotemporal parameters for standard gait

Side	Parameters	ICC (standard gait)	ICC (max-kg gait)	WSSD (standard gait)	WSSD (max-kg gait)
Right	Cadence	0.961	0.972	2.726	3.177
	Step length	0.970	0.972	0.017	0.017
	Step time	0.933	0.942	0.019	0.023
	Stride length	0.939	0.980	0.049	0.029
	Stride time	0.949	0.971	0.033	0.030
	Walking speed	0.977	0.976	0.040	0.044
Left	Cadence	0.974	0.963	2.181	3.623
	Step length	0.964	0.968	0.019	0.018
	Step time	0.934	0.913	0.018	0.025
	Stride length	0.976	0.981	0.030	0.028
	Stride time	0.972	0.961	0.024	0.034
	Walking speed	0.980	0.973	0.038	0.045

*Units of measurement: cadence = steps/minutes, step length & stride length = meter, step time & stride time = seconds, walking speed = meter/seconds

b. 3D kinematics

Coefficient of multiple correlations (CMC) was used to determine the level of agreement for the 3D movements (i.e. sagittal, frontal and transverse planes) around ankle, knee, hip, pelvis, trunk (relative to pelvis) and trunk global (relative to global coordinate system) for standard gait (Table 4.3). In the sagittal plane (flexion-extension), the level of agreement was good for ankle, knee and hip (MC: 0.938 to 0.960) but moderate for pelvis, trunk and trunk global (CMC: 0.535 to 0.640). In the frontal plane, the level of agreement was good for hip, pelvis and trunk (CMC: 0.882 to 0.937), but moderate for the trunk global. All movements in the transverse plane had good level of agreement (CMC: 0.870 to 0.944). According to waveform measurement error (WE), the variability for all movements were relatively small (WE: 0.608 to 0.3990)

Table 4.3. Within-session level of agreement of movements during standard gait

Joint/segment	**Flexion-extension**		**Adduction-abduction**		**Horizontal rotation**	
	CMC	**WE**	**CMC**	**WE**	**CMC**	**WE**
Rt. ankle	0.920	2.700				
Rt. knee	0.938	3.262				
Rt. hip	0.954	2.313	0.934	1.125		
Rt. pelvis	0.535	1.051	0.882	0.762	0.872	1.353
Rt. trunk	0.561	1.156	0.917	0.901	0.944	1.246
Rt. trunk (g)	0.615	3.774	0.610	1.966	0.898	0.899
Lt. ankle	0.916	2.677				
Lt. knee	0.957	2.805				
Lt. hip	0.960	2.180	0.937	1.048		
Lt. pelvis	0.543	1.041	0.897	0.745	0.870	1.325
Lt. trunk	0.579	1.190	0.924	0.917	0.947	1.234
Lt. trunk (g)	0.640	3.990	0.608	1.982	0.897	0.890

Trunk (g) = trunk relative to global coordinate system (trunk global)
CMC = Coefficient of multiple correlations
WE = waveform (measurement) error

4.3.1.3 Max-kg gait

a. Spatiotemporal parameters

Intra-class correlation coefficient (ICC) and within-subject standard deviation (WSSD) were used to examine the level of agreement (within-session) and variability respectively for spatiotemporal parameters (i.e. cadence, step length, step time, stride length, stride time and walking speed) for max-kg gait (Table 4.4). According to the results, all spatiotemporal parameters had good level of agreements for both left and right gait cycles in max-kg gait (ICC: 0.913 to 0.981). According to the WSSD, the variability for all spatiotemporal parameters were relatively small (WE: 0.017 to 3.623).

Table 4.4. Level of agreement of spatiotemporal parameters for max-kg gait

Side	Parameters*	ICC	95% Confidence interval		WSSD
			Lower limit	Upper limit	
Right	Cadence	0.972	0.952	0.985	3.177
	Step length	0.972	0.952	0.985	0.017
	Step time	0.942	0.900	0.968	0.023
	Stride length	0.980	0.966	0.989	0.029
	Stride time	0.971	0.950	0.984	0.030
	Walking speed	0.976	0.958	0.987	0.044
Left	Cadence	0.963	0.936	0.980	3.623
	Step length	0.968	0.945	0.983	0.018
	Step time	0.913	0.850	0.952	0.025
	Stride length	0.981	0.967	0.990	0.028
	Stride time	0.961	0.933	0.979	0.034
	Walking speed	0.973	0.953	0.985	0.045

*Units of measurement: cadence = steps/minutes, step length & stride length = meter, step time & stride time = seconds, walking speed = meter/seconds

b. 3D kinematics

Coefficient of multiple correlation (CMC) was used to determine the level of agreement for the 3D movements (i.e. sagittal, frontal and transverse planes) around ankle, knee, hip, pelvis, trunk (relative to pelvis) and trunk global (relative to global coordinate system) for the max-kg gait (Table 4.5). In the sagittal plane (flexion-extension), the level of agreement was good for the ankle, knee and hip (CMC: 0.829 to 0.852), moderate for the pelvis and trunk (CMC: 0.649 to 0.668) and poor for the trunk global (CMC: 0.479 to 0.495). In the frontal plane (adduction-abduction), the level of agreement was good for the hip, pelvis and trunk (CMC: 0.752 to 0.863) and moderate for the trunk global (0.694 to 0.708). In the transverse plane, the level of agreement was good for the trunk global (CMC: 0.821 to 0.825) and moderate for the pelvis and trunk (CMC: 0.681 to 0.717). According to the waveform measurement error (WE), the reliability was relatively small for all movements (WE: 0.745 to 5.447).

Table 4.5. Within-session level of agreement of movements during max-kg gait

Joint/segment	**Flexion-extension**		**Adduction-abduction**		**Horizontal rotation**	
	CMC	**WE**	**CMC**	**WE**	**CMC**	**WE**
Rt. ankle	0.843	2.773				
Rt. knee	0.841	5.316				
Rt. hip	0.849	5.447	0.839	3.255		
Rt. pelvis	0.668	3.360	0.808	1.264	0.681	3.039
Rt. trunk	0.649	4.055	0.752	1.461	0.717	2.700
Rt. trunk (g)	0.495	1.587	0.708	0.879	0.825	2.331
Lt. ankle	0.845	3.984				
Lt. knee	0.852	5.158				
Lt. hip	0.829	4.510	0.863	2.027		
Lt. pelvis	0.686	2.637	0.817	1.176	0.667	1.653
Lt. trunk	0.675	3.540	0.799	1.401	0.691	1.912
Lt. trunk (g)	0.479	1.720	0.694	0.908	0.821	3.214

Trunk (g) = trunk relative to global coordinate system (trunk global)
CMC = Coefficient of multiple correlations
WE = waveform (measurement) error

4.3.2 Between-Session Reliability

4.3.2.1 Participant characteristics (N=9)

A total of nine participants were recruited for the study. Based on the results, the participants’ mean age was 27.00±7.28, ranging from 18 to 39 years old) (Table 4.6). The mean of the body mass index falls into normal body mass index (BMI) category (25.16±4.66). For physical activity, the participants spent most of the time on sitting (366.67±145.26 minutes/day), followed by walking (114.44±787.76 minutes/day), vigorous activity (73.33±21.85 minutes/day) and moderate activity (66.67±46.90 minutes/day).

Table 4.6. Participant characteristics for between-session reliability study (n=7)

Variables	Mean	Standard deviation
Age	27.00	7.28
Height	168.26	7.73
Weight	71.47	15.54
Body mass index (BMI)	25.16	4.66
Physical activity*		
Vigorous activity: days/week	1.44	1.59
Vigorous activity: minutes/day	73.33	21.85
Moderate activity: days/week	3.56	2.19
Moderate activity: minutes/day	66.67	46.90
Walking: days/week	5.89	1.45
Walking: minutes/days	114.44	78.76
Sitting on weekday: minutes/day	366.67	145.26

*based on the International Physical Activity Questionnaire (IPAQ)

4.3.2.2 Between-session reliability of spatiotemporal parameters

Intra-class correlation coefficient (ICC) and within-subject standard deviation (WSSD) were used to examine the level of agreement (between-session) and variability respectively for spatiotemporal parameters (Table 4.7). According to the ICC, all parameters indicated good level of agreements (ICC: 0.820 to 0.956). The highest level of agreement can be observed in the stride length. According to the WSSD, the variability of all spatiotemporal parameters were relatively small (WSSD: 0.014 to 3.729).

Table 4.7. Level of agreement and variability of spatiotemporal parameters between sessions

Side	Parameters	ICC	95% Confidence interval		Mean	SD	WSSD
			Lower limit	Upper limit			
Right	Cadence	0.892	0.519	0.976	109.478	8.451	3.717
	Step length	0.955	0.800	0.990	0.613	0.053	0.014
	Step time	0.820	0.202	0.959	0.551	0.044	0.024
	Stride length	0.956	0.805	0.990	1.235	0.103	0.029
	Stride time	0.857	0.364	0.968	1.104	0.085	0.042
	Walking speed	0.921	0.651	0.982	1.127	0.133	0.050
Left	Cadence	0.919	0.642	0.982	108.474	9.393	3.729
	Step length	0.937	0.720	0.986	0.623	0.052	0.017
	Step time	0.843	0.302	0.964	0.553	0.043	0.022
	Stride length	0.938	0.723	0.986	1.238	0.104	0.032
	Stride time	0.886	0.496	0.974	1.115	0.097	0.045
	Walking speed	0.914	0.617	0.981	1.12	0.143	0.055

*Units of measurement: cadence = steps/minutes, step length & stride length = meter, step time & stride time = seconds, walking speed = meter/seconds

For the spatiotemporal parameters, limits of agreement were determined based on Bland-Altman plot. According to the results, the differences between sessions of each participant are within 95% limits of agreement (Table 4.8). To examine how close the mean difference (i.e. middle horizontal line) to zero, Cohen *d* values were calculated. For all spatiotemporal parameters of gait, the distances were either negligible (Cohen *d*: 0.076 to 0.193) or small (Cohen *d*: 0.201 to 0.327). However, proportional bias can be observed in the left stride length (β =-0.676, p<0.05) (Figure 4.2).

Table 4.8. Limit of agreement for spatiotemporal parameters of normal gait

Sides	Variables	Mean diff. ($\bar{d}$)	SD	95% Limits of agreement		Cohen *d*	Linear Regression	
				Lower limit	Upper limit		β	p
Right	Cadence	-0.328	5.564	-11.233	10.579	0.059	-0.041	0.917
	Step length	-0.007	0.023	-0.05	0.037	-0.291	-0.602	0.087
	Step time	-0.004	0.037	-0.076	0.069	-0.091	-0.080	0.838
	Stride length	-0.009	0.044	-0.093	0.077	-0.194	-0.583	0.100
	Stride time	0.003	0.064	-0.123	0.129	0.046	-0.063	0.872
	Walking speed	-0.012	0.075	-0.159	0.135	-0.162	-0.297	0.438
Left	Cadence	-1.576	5.339	-12.039	8.888	-0.296	-0.010	0.980
	Step length	-0.003	0.026	-0.054	0.048	-0.109	-0.486	0.185
	Step time	0.007	0.034	-0.06	0.073	0.201	-0.076	0.845
	Stride length	-0.002	0.052	-0.104	0.100	-0.035	-0.676	0.044
	Stride time	0.016	0.065	-0.112	0.144	0.243	-0.068	0.863
	Walking speed	-0.019	0.084	-0.183	0.145	-0.227	-0.339	0.372

*Units of measurement: cadence = steps/minutes, step length & stride length = meter, step time & stride time = seconds, walking speed = meter/seconds

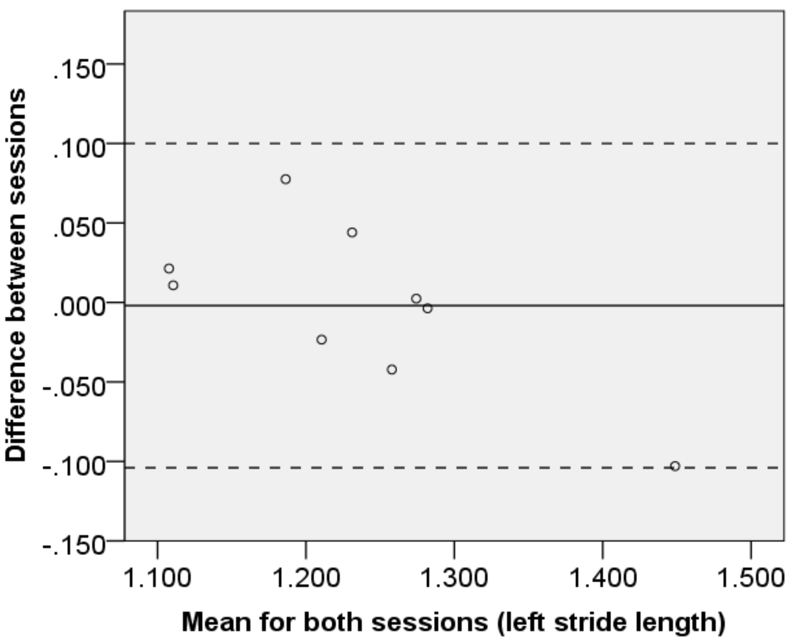


Figure 4.2. Bland-Altman plot for left stride length (meters)

To examine the systematic errors between two sessions for the spatiotemporal parameters, paired *t* test was conducted (Table 4.9). The results indicated that there were no significant mean differences between the sessions, indicating no systematic errors between the sessions.

Table 4.9. Between-session systematic errors for spatiotemporal parameters of gait

Sides	**Parameters***	**Session**	**Mean**	**SD**	***P***	**95% Confidence interval**	
						Lower limit	**Upper limit**
Right	Cadence	Session 1	109.291	9.625	0.864	3.95	-4.605
		Session 2	109.332	9.343			
	Step length	Session 1	0.608	0.050	0.405	0.011	-0.024
		Session 2	0.613	0.063			
	Step time	Session 1	0.551	0.049	0.793	0.025	-0.032
		Session 2	0.555	0.052			
	Stride length	Session 1	1.230	0.099	0.580	0.025	-0.042
		Session 2	1.234	0.124			
	Stride time	Session 1	1.106	0.095	0.893	0.053	-0.047
		Session 2	1.106	0.098			
	Walking speed	Session 1	1.121	0.136	0.641	0.046	-0.07
		Session 2	1.126	0.157			
Left	Cadence	Session 1	107.582	10.406	0.402	2.528	-5.679
		Session 2	108.784	10.354			
	Step length	Session 1	0.623	0.050	0.758	0.018	-0.023
		Session 2	0.623	0.063			
	Step time	Session 1	0.560	0.048	0.559	0.033	-0.02
		Session 2	0.552	0.050			
	Stride length	Session 1	1.238	0.095	0.920	0.038	-0.042
		Session 2	1.235	0.131			
	Stride time	Session 1	1.125	0.107	0.487	0.066	-0.035
		Session 2	1.113	0.110			
	Walking speed	Session 1	1.111	0.144	0.516	0.046	-0.084
		Session 2	1.121	0.170			

*Units of measurement: cadence = steps/minutes, step length & stride length = meter, step time & stride time = seconds, walking speed = meter/seconds

4.3.2.3 Between-session reliability of 3D kinematics

For joint rotations, the results of coefficient of multiple correlations (CMC) and waveform measurement errors (WE) were presented (Table 4.10). Examples of waveform (Figure 4.3) and root mean square errors (RMSE) (Figure 4.4) for the joint across gait cycle were presented. Except for the trunk (CMC: right=0.245, left=0.246) and pelvis (CMC: right=0.311, left=0.326) that had poor level of agreement between the sessions in sagittal plane, all other movements for ankle, knee and hip had good level of agreement in all selected planes (ICC: 0.797 to 0.960).

Table 4.10. Level of agreement and waveform errors of 3D kinematics of normal gait

Joint/segment	**Sagittal plane**		**Frontal plane**		**Transverse**	
	CMC	**WE**	**CMC**	**WE**	**CMC**	**WE**
Right ankle	0.930	2.820				
Right knee	0.831	6.060				
Right hip	0.939	3.882	0.831	2.542		
Right pelvis	0.311	1.721	0.896	0.974	0.858	2.005
Right trunk	0.245	2.309	0.844	1.426	0.848	2.427
Left ankle	0.802	5.101				
Left knee	0.941	4.449				
Left hip	0.960	3.043	0.904	1.702		
Left pelvis	0.326	1.597	0.898	0.994	0.869	2.008
Left trunk	0.246	2.480	0.837	1.417	0.797	6.757

CMC: coefficient of multiple correlation, WE: waveform errors

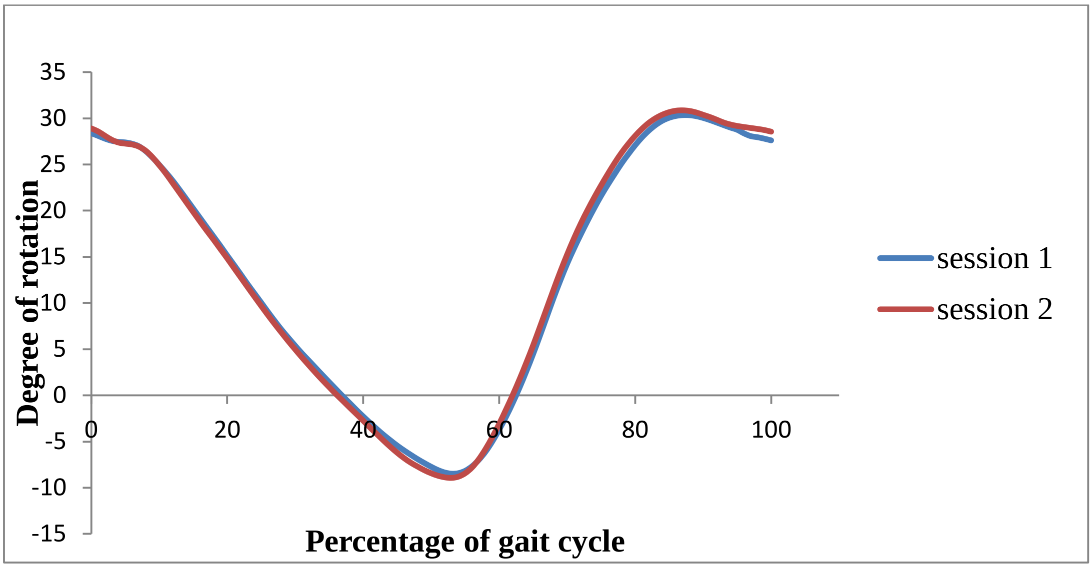


Figure 4.3. Flexion-extension movement around hip joint between sessions for standard gait (both groups)

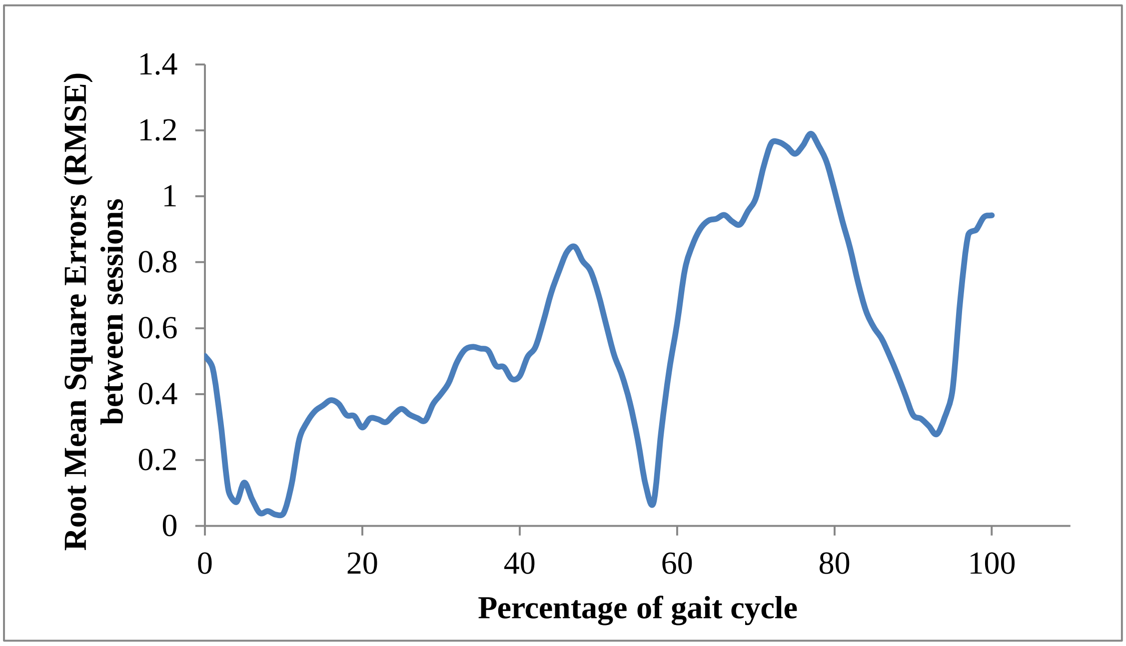


Figure 4.4. Root mean square errors (RMSE) for flexion-extension movement around hip joint between sessions for standard gait (both groups)

4.3.2.4 Ito test

Intra-class correlation coefficient (ICC) and within-subject standard deviation (WSSD) were used to examine the level of agreement (between-session) and variability respectively for the Ito test (i.e. holding time) and muscle fatigue for iliocostalis, multifidus, gluteus maximus and biceps femoris during the Ito test (i.e. MFslope of EMG). According to the results, the Ito test had a good level of agreement between the sessions (ICC=0.759) (Table 4.11). For the MFslope, only right iliocostalis (ICC=0.521) and left biceps femoris (ICC=0.684) had moderate level of agreements between the sessions, while the rest of the muscles had poor level of agreements (ICC: -0.109 to 0.441).

Table 4.11. Level of agreement for holding time and muscle fatigue during Ito test

Side	**Variables**	**ICC**	**95% Confidence interval**		**Mean**	**SD**	**WSSD**
			Lower limit	**Upper limit**			
	Ito test (holding time)[1]	0.759	0.244	0.940	161.611	50.765	26.59
Right	Iliocostalis[2]	0.521	-0.222	0.881	0.033	0.156	0.155
	Multifidus[2]	0.317	-0.443	0.812	-0.148	0.065	0.063
	Glut. maximus[2]	-0.125	-0.731	0.591	-0.020	0.105	0.158
	Biceps femoris[2]	0.441	-0.319	0.856	-0.111	0.108	0.095
Left	Iliocostalis[2]	-0.109	-0.723	0.601	0.000	0.087	0.152
	Multifidus[2]	0.291	-0.465	0.802	-0.121	0.076	0.077
	Glut. maximus[2]	0.063	-0.630	0.700	-0.017	0.137	0.170
	Biceps femoris[2]	0.684	0.032	0.928	-0.066	0.085	0.055

Unit of measurements: [1] = seconds, [2] = EMG MFslope

Limits of agreement were determined based on Bland-Altman plot (Table 4.12). For the holding time, the differences between sessions of each participant were within 95% limits of agreement. To examine how close the mean difference (i.e. middle horizontal line) to zero, Cohen *d* values were calculated. The distance was small (Cohen $d = 0.27$) for the Ito test. For the right iliocostalis (Figure 4.5), the distance from zero was large, indicating a presence of fixed bias (Cohen d=0.84). The plot also indicated that there was no proportional bias found in the Ito test ($\beta = -0.048$, p=0.903). For the left biceps femoris (Figure 4.6), a significant linear regression indicated presence of proportional bias (β=0.62, $p<0.05$).

Table 4.12. Limit of agreement of Ito test and MF slope in Ito test

Sides	**Variables**	**Mean difference ($\bar{d}$)**	**SD**	**95% Limits of agreement**		**Cohen *d***	**Linear Regression**	
				Upper limit	**Lower limit**		***B***	***p***
	Ito test[1]	8.667	32.357	72.087	-54.754	0.27	-0.048	0.903
Right	Iliocostalis[2]	-0.146	0.175	0.196	-0.489	0.84	0.172	0.717
	Multifidus[2]	-0.037	0.093	0.146	-0.220	0.40	-0.181	0.767
	Glut. maximus[2]	-0.019	0.239	0.450	-0.488	0.08	-0.631	0.504
	Biceps femoris[2]	-0.045	0.134	0.219	-0.308	0.33	0.002	0.998
Left	Iliocostalis[2]	-0.118	0.194	0.262	-0.498	0.61	-0.127	0.894
	Multifidus[2]	-0.043	0.112	0.177	-0.264	0.39	-0.194	0.758
	Glut. maximus[2]	0.012	0.257	0.517	-0.493	0.05	0.938	0.208
	Biceps femoris[2]	0.026	0.073	0.170	-0.118	0.36	0.622	0.045

Unit of measurements: [1] = seconds, [2] = slope of EMG median frequency

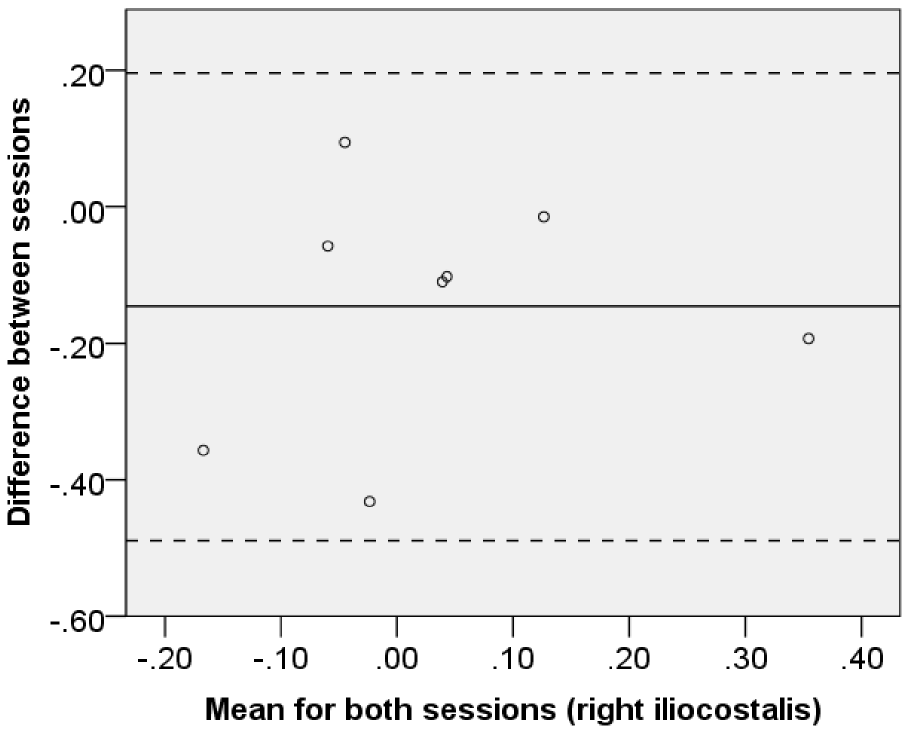


Figure 4.5. Bland-Altman plot for right iliocostalis

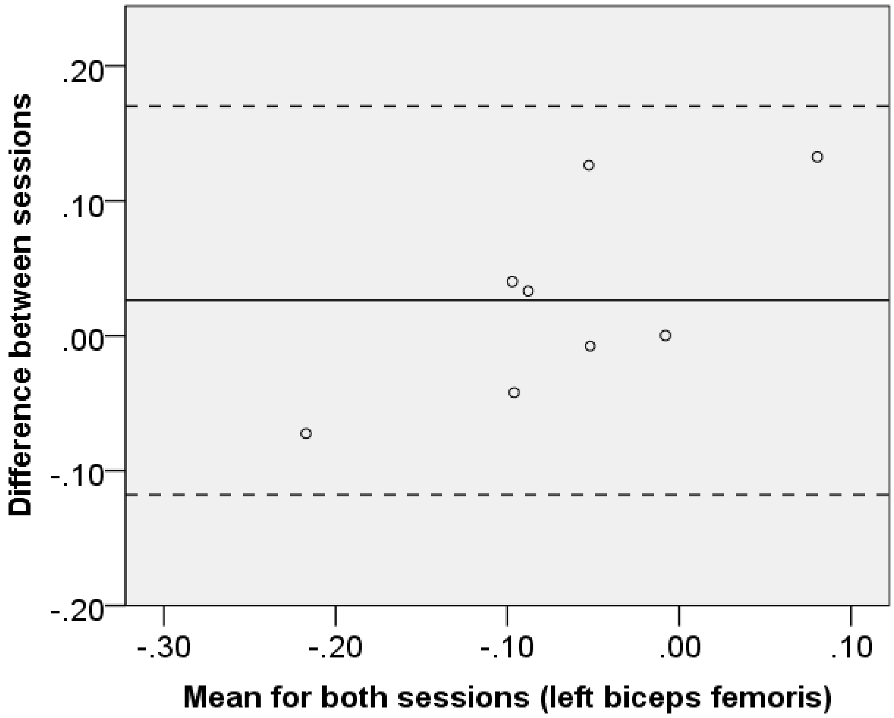


Figure 4.6. Bland-Altman plot for left biceps femoris

4.4 DISCUSSION

The level of agreement for all spatiotemporal parameters were good for both within-session-session and between-session reliability. The Bland-Altman plot cannot be produced for within-session reliability because there were more than two observations (i.e. three trials). For between-session reliability, a presence of proportional bias was indicated in the left stride. The proportional bias was tested based on simple linear regression by regressing the differences between both sessions. Although no outlier was found outside 95% limits of agreement for the stride, there was one participant who had a very low difference between the sessions, and the difference was on the lower limit agreement (-0.1 meter). If the participant was to be excluded from the Bland-Altman plot, the proportional bias would not exist. However, the main purpose of this reliability study was not to generalize to reliability towards a bigger population, but rather, to give an idea on possible issues that may affect the reproducibility of the measurements. Due to small sample size (N=9), any pattern can be easily misinterpreted as bias, thus, can mislead the interpretation of Bland-Altman plot (Altman 1990). For instance, it was very difficult to decide for any presence of tunnelling bias (heteroscedasticity) because the pattern was observable only due to a small number of participants. If the participant was going to be excluded, there was still a possibility that the remaining participants had homoscedastic differences. Therefore, it can be concluded that the proportional bias for the left stride only happened due to the small sample size, and the left stride was still reliable to be used in the main study.

The within-subject standard deviation (WSSD) for the spatiotemporal parameters were relatively low for both within-session and between-session reliability (WSSD: 0.014 to 3.623), indicating low variability between and within the sessions. Although the findings from this study are supported by the literature, the coefficient of variation (CV) value was used to indicate variability from the previous studies. Kadaba et al. (1989) reported that the CV for all spatiotemporal parameters were below 10% of the respective mean values. Tsushima et al. (2003) reported that the coefficient of variation (CV) of all spatiotemporal variables was less than 5%, suggesting low variability and thus, good repeatability. In general, CV is expressed as the percentage of the standard deviation of sample divided by mean. One of the main limitations of CV is that the highest score will greatly differ from the lowest score because due to expression in the percentage. Thus, the use of CV is no longer advisable in determining reliability (Atkinson and Nevill 1998; Bland 2015). Therefore, the WSSD is more preferable as the measure of variability because it is based on the difference between the sessions that are likely to occur within 95% of probability (Bruton et al. 2000).The findings also indicated that the walking speed was the most reliable spatiotemporal parameter (within-session: ICC = 0.977, WSSD = 0.004 & between-session: ICC = 0.921, WSSD: 0.005), and was supported by the findings from Yavuzer et al. (2008) (within-session: ICC = 0.99, CV =3.9 & between session: ICC = 0.98, CV = 6.1).

For within-session reliability, the movement of trunk global in sagittal plane was found to have a poor level of agreement during max-kg gait. When a participant carried a container with progressive loads, the load can cause the trunk to bend following the antero-posterior force from the load (Seay et al. 2011a). As a part of motor control system to maintain a balanced posture, back muscles will counter-act the force from the load. When the trunk was extended, a specific coupling mechanism between adjacent body segments (i.e. trunk and pelvis) may be activated as a part of the motor control system. According to the result, the variability of the trunk global was higher compared to the trunk (relative to pelvis) in the sagittal plane. The trunk (global) was determined as the movement of the local coordinate system (LCS) of the trunk relative to global coordinate system (GCS). Because GCS was fixed vertically, the variability may become more obvious for the trunk global. However, for the trunk (relative to pelvis), the variability may be lower due to possible occurrence of similar direction of movement in the pelvis. This pattern was commonly referred as in-phase coordination (Seay et al. 2011a; Seay et al. 2011c). The pelvis may be continuously adapting with the change in the trunk movement, the variability in the trunk (relative to pelvis) was lower than the trunk global. A good intersegmental coordination is important in maintaining dynamic balance throughout gait cycle, which will be discussed later in Chapter 7.

The Ito test was measured based on the time-to-failure method (i.e. holding time). Across the literature, to the researcher’s knowledge, none of Ito test studies reported the limit of agreement whilst performing reliability testing. For instance, although it had been reported that the Ito test had good test-retest correlation in both healthy individuals and low back pain patients, the limit of agreement for the correlations were unknown (Ito et al. 1996; McIntosh et al. 1998). Even though the ICC value for the holding of Ito test was good (ICC = 0.759), the width of limits of agreement was relatively wide. Most of the participants with major differences in the Ito test performed their second test session in the afternoon (2pm to 5pm). It was reported in the literature that the diurnal change in central body temperature from morning to evening can affect the muscle activity (i.e. tend to be higher in the evening) (Martin et al. 1999; Chtourou et al. 2011). Furthermore, prior to test session, the participants were instructed not to do any heavy physical activity. However, their actual physical activity prior to the test were beyond researcher's control. The carry-over effect of physical activity prior to the test session can be one of the possible factors that may influence their performance in the Ito test. In this study, the Ito test was chosen because of its simplicity (i.e. cost-effective without any specific equipment). It was assumed that this test can minimize the pressure on the lower back compared to Biering-Sorensen test. However, the issue that may limit the use of Ito test among the clinician and lack of standardization of test procedure (Demoulin et al. 2006). During the Ito test, a pillow was placed under the lower abdomen to decrease the lumbar lordosis. This was to prevent the possibility of bulging intervertebral discs and also buckling of the ligamentum flavum, which can cause narrowing of the intervertebral foramen (Ito et al. 1996). Yet, there was no precise documentation for the type of pad throughout the literature. This variability may affect the degree of back extension during the test because it may influence the lordosis of the lumbar vertebrae. In the current study, the researcher used a folded rectangular pillow. The use of the pad was standardized in this study to minimize within-subject variability. However, some participants complained that the reason why the Ito test was terminated was not so much of the fatigue at the back, but rather, caused by an uncomfortable feeling under the abdomen due to the pad. In this case, there was a possibility that the termination of the Ito test was premature, thus, may alter the accuracy of the result.

During the Ito test, right iliocostalis and left biceps femoris had moderate level of agreement between the sessions, while the rest of the muscles had poor level of agreements. However, as the concept of reliability is multidimensional, the results from the level of agreement alone cannot solely determine the reliability of the muscle fatigue. According to Bland-Altman Plot, the distance from zero was significantly large for the right iliocostalis, indicating a presence of fixed (or constant) bias. The fixed bias exists when the scores from one trial (or method) is consistently higher or lower compared to the second trial and is indicated when there is a significant departure of the mean difference from zero (Pottel 2015). In this study, the distance from zero was calculated based on magnitude of effect (i.e. Cohen *d*) rather than statistical significance (i.e. one sample *t* test). Small sample size can influence type-I error (i.e. a false positive result where the null hypothesis is rejected erroneously), but the effect size is generally not affected by the sample size (Olejnik and Algina 2000; Nakagawa and Cuthill 2007). For the right iliocostalis, majority of the scores were below zero, indicating that the MFslope was increased (i.e. less fatigued) in most of the participants during the second session. According to the effect size, the distance from zero for the right iliocostalis was large.

There is also a possibility that internal factors such as motivation may have influenced the effort to perform the test. Other than pain tolerance and competitiveness, motivation is among the personal factors that can influence the variability in an isometric back extension test (Mannion et al. 1996; Moffroid 1997; Demoulin et al. 2006). It is assumed that the participants may have higher anxiety about the strength of their first session, so they performed better compared to the second session. However due to a small sample size, the interpretation of the Bland-Altman plot cannot be generalized to a bigger population of study. It has been suggested that a minimum of 50 participants is required to enable the limits of agreement to be estimated well (Altman 1990; Rankin and Stokes 1998). The suggested number for minimum sample can still be argued because the Bland-Altman is generally not an inferential statistic, thus, the power calculation is not applicable for sample size estimation. However, to interpret the pattern in Bland-Altman plot (e.g. proportional bias, tunnelling), the use of bigger sample is generally recommended.

For the between-session reliability, except trunk and pelvis between the sessions in sagittal plane, all other waveforms had good level of agreement. The findings for pelvis, hip, knee and ankle in this study were similar to the findings from Tsushima et al. (2003), Steinwender et al. (2000), Growney et al. (1997) and Kadaba et al. (1989). The highest CMC value was found at the hip, followed by knee and ankle. The values of CMC have to be interpreted with care because the calculation method of CMC was highly influenced by the magnitude of joint angle. Because of the mathematical formula of CMC, joints with large range of motion will commonly produce high CMC, and joints with small range of motion will commonly produce low CMC (Steinwender et al. 2000; Yavuzer et al. 2008). In general, the movement at hip, knee and ankle joints in the sagittal plane had the largest ROM compared to the other planes of movement because these movements allow the body to project itself forward. According to Neumann (2013), 10º of dorsiflexion and 20º of plantar flexion are required to walk on an average walking speed. Furthermore, in stance phase (i.e. from 0% to 60% of gait cycle), greater dorsiflexion was needed compared to swing phase. For the knee joint, a minimum of nearly full extension to approximately 60º of flexion (from normal position) were required during the gait cycle. For the hip joint, approximately 10º of extension and 30º of flexion (from normal position) were required during the gait cycle. The number of studies on the reliability of 3D trunk kinematics during walking was limited. Most of these studies focused on lower limb kinematics (i.e. pelvis, hip, knee and ankle) without including the trunk. Because the trunk was reported to function as an active unit rather than just a passive unit during gait (Armand et al. 2009; Kiernan et al. 2014), the present study had included the trunk as a body segment to describe a comprehensive body movement during gait analysis. As well as the pelvis, the trunk movement also consists of a two-full-cycles sinewave, and a relatively small range of motion occurred throughout the gait cycle (Neumann 2013). The standard errors of measurement for the trunk movement in the sagittal plane were relatively small and thus, were acceptable. Therefore, it was concluded that from this study, the small CMC values for the trunk movement in sagittal plane may also be affected by the small range of motion.

Chapter 5: COMPARING MUSCLE FATIGUE DURING ITO TEST AND ANTERIOR LOAD CARRIAGE

5.1 INTRODUCTION

In rehabilitation, isometric back endurance test is an important clinical measure of low back pain. Controversially, some studies even claimed that the test was able to differentiate between those with and without low back pain (Moreau et al. 2001; Arab et al. 2007). In general, the isometric back endurance test is manually tested based on a time-to-failure method, and can be classified into extensor endurance tests such as Biering-Sorensen test (Biering-Sorensen 1984), prone isometric chest raise test/Ito test (Ito et al. 1996) and the prone double straight leg raise test (McIntosh et al. 1998). Both types are used to measure the isometric endurance of the trunk extensors. The flexor endurance tests such as supine isometric chest raise test (McIntosh et al. 1998) and supine double straight-leg test (McIntosh et al. 1998) are used to indicate the isometric endurance of the trunk flexors. Amongst the tests, the Biering-Sorensen test is the most common measure of isometric back endurance, and the standardization of procedures is well reported across the literature (Moreau et al. 2001; Demoulin et al. 2006; Arab et al. 2007; Demoulin et al. 2007; Beneck et al. 2013; Álvarez-Álvarez et al. 2014). However, it was argued that the Biering-Sorensen is rather a test of hip extensor endurance than trunk extensor endurance due to major involvement of biceps femoris over the erector spinae (Moreau et al. 2001). Therefore, the use of the Ito test as an optional measure of the isometric test is rising. The Ito test was considered to be more cost-effective, requires minimal equipment and assumed to inflict a relatively lower pressure to the spine compared to the Biering-Sorensen test (Ito et al. 1996; Demoulin et al. 2006).

Although the isometric back endurance was reported to associate with work disability related to low back pain (Rissanen et al. 2002), there are many possible factors that can mediate or moderate the association. For instance, biomechanical factors such as activity frequency, moment of load, lateral and axial velocity of the trunk, and trunk rotation in sagittal plane were reported to have associations with low back pain (Burton 1997). Such data regarding the biomechanical factors can only be obtained via an assessment of dynamic movement whilst performing a functional activity such as manual material handling. In clinical setting, Functional Capacity Evaluation (FCE) is an assessment that measures work-related capabilities of an individual in performing manual material handling activities related to his/her job. It was reported that a successful return to work can be predicted by the improvement in the FCE (Fore et al. 2015). Amongst the manual material handling activities, carrying can be considered as an activity that involves various dynamic movements. In contrast to the isometric endurance test, the performance in FCE can be used to indicate the functional endurance of a person, which is indeed an important indicator for return-to-work. As the carrying activity involves dynamic body movements, a prolonged carrying activity can potentially lead to increased muscle activity that is responsible for the movements. Therefore, the aim of this chapter was to compare the muscle fatigue during the Ito test and anterior load carriage with progressive anterior load carriage between sedentary individuals and manual workers.

5.2 METHODOLOGY

5.2.1 Study Design

The design of this study was cross-sectional with mixed-group comparisons recruiting healthy participants (n=37). The data were collected from May 2014 to April 2015 (11 months) at the Biomechanics Laboratory, Faculty of Health Sciences, University of Southampton. The participants were divided into sedentary individuals (n=20) and manual workers (n=17) for between-group comparison. During the study, the participants were asked to perform an isometric back endurance test (i.e. Ito test) and two types of gait: standard gait (i.e. self-preferred gait) and carrying activity with progressive loads (i.e. one kg increment).

5.2.2 Participants

The participants' inclusion and exclusion criteria were described as 3.2 and the recruitment strategies were described as 3.5.

5.2.3 Procedures

After written informed consent was obtained, research participants were asked to indicate their level of physical activity for the past seven days based on International Physical Activity Questionnaires (IPAQ) (Ainsworth et al. 2000). After that, the participants were asked to position themselves in prone-lying on a table to prepare for the Ito test (Ito et al.

1996). In order to perform the test, the participants were asked to lift and maintain the sternum off the table to maximum extension (please see 3.6.1 for more detail). Then, the participants were asked to perform a carrying activity for 60 meters by holding the carrying container by flexing the arm at 90º of elbow flexion. The carrying activity was performed by walking back and forth along the 10m walking platform (see section 3.7.5 for detail).

5.2.4 Data Processing

5.2.4.1 Muscle fatigue during Ito test

The rate of muscle fatigue was determined based on the EMG median frequency (MF) during the Ito test. The MF for each load increment during carrying activity was determined, and a slope (MFslope) was calculated based on the MF at each second of holding time (Figure 5.1). The MF of four bilateral muscles were analysed. The muscles that involved were iliocostalis, multifidus, gluteus maximus and biceps femoris. The MFslope of all participants (N=37) were then determined (please see section 3.8.2 for more detail on electrode placement and signal amplification, filtration and signal analysis).

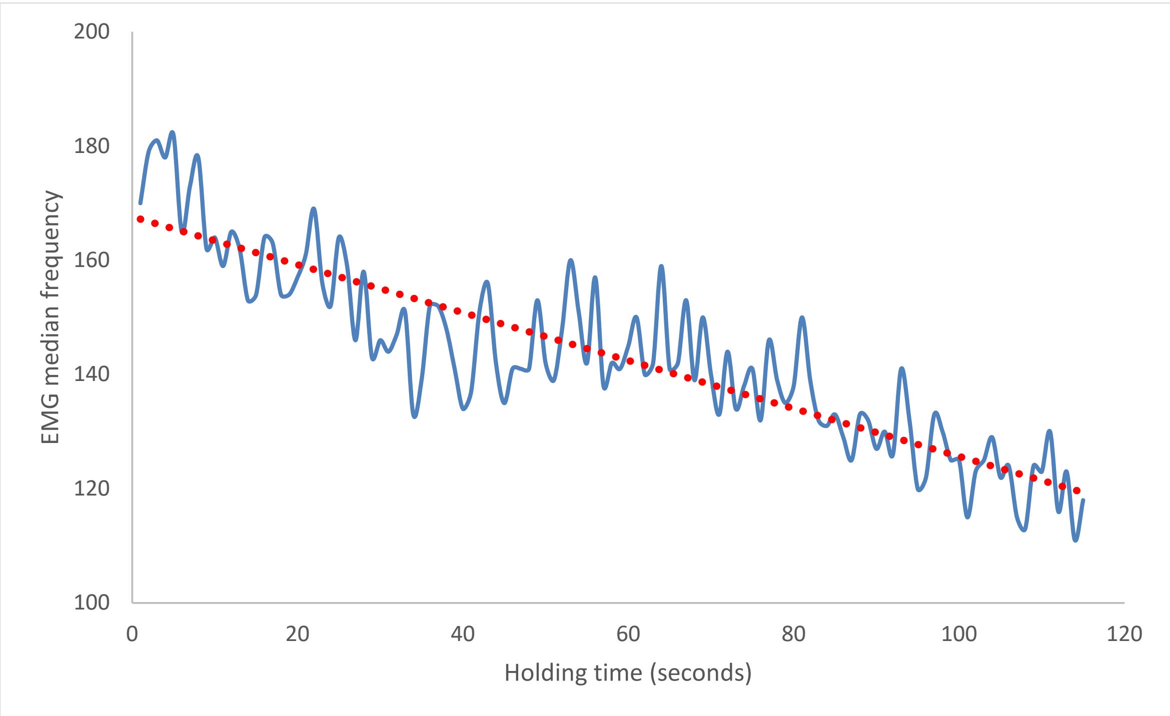


Figure 5.1. Exemplar slope of median frequency (MFslope) for biceps femoris during Ito test (maximum holding time = 115 seconds)

5.2.4.2 Muscle fatigue during carrying activity

The muscle fatigue during the carrying activity was determined based on the EMG median frequency (MF) during static standing (i.e. five seconds) immediately following one set of carrying activity, prior to the increase in load. The MF for each load increment during carrying activity was determined, and a slope (MFslope) was calculated based on the average of MF at each second of the static standing period. The MF of eight bilateral muscles were analysed which were; iliocostalis, multifidus, gluteus maximus, biceps femoris, biceps brachii, latissimus dorsi, vastus lateralis and gastrocnemius. However, the right latissimus dorsi was excluded from the analysis due to a non-responsive EMG wireless receiver for that particular muscle. Furthermore, three participants were excluded from the analysis because the EMG sensors detached from the original location on the skin, which can possibly mislead the analysis. The possible causes for this detachment can be due to excessive sweating, stretched skin or excessive movement. Therefore, only 34 participants were analysed for this section (i.e. sedentary=19, manual=15).

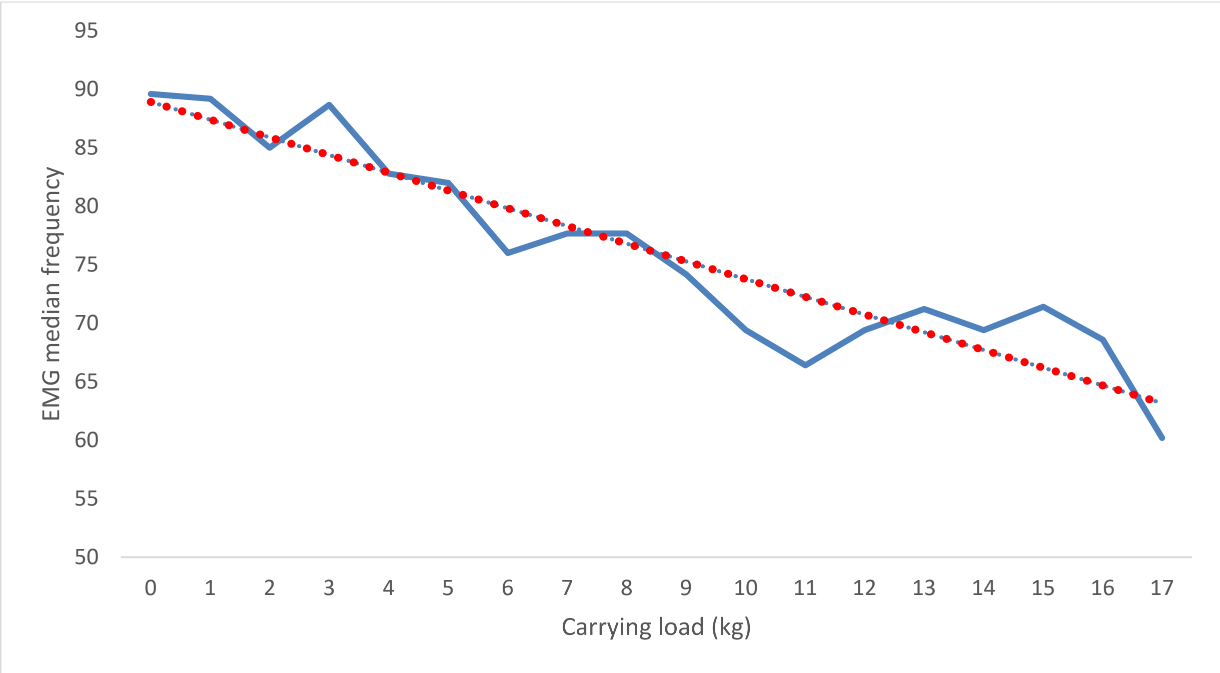


Figure 5.2. Exemplar slope of median frequency (MFslope) for biceps femoris (maximum load = 17kg).

5.2.5 Statistical Analysis

To compare the MFslope of the muscles between the sedentary and the manual group, independent *t* test was conducted for normally distributed data whilst Mann-Whitney test was conducted for not normally distributed data. For the independent *t* test, the effect size (i.e. magnitude of difference) was based on Cohen's *d:* 0.2 (small), 0.5 (medium) and 0.8 (large) (Cohen 1988a). For the Mann Whitney test, the effect size was based on correlation coefficient (r): 0.1 (small), 0.3 (moderate) and 0.5 (large) (Cohen 1988a). To investigate the relationship between holding time and maximum load carriage, Pearson product moment correlation and simple linear regression was conducted. The effect size for the Pearson product-moment correlation (i.e. strength of association) was based on correlation coefficient (*r*): 0.3 (weak), 0.5 (moderate) and 0.7 (strong) (Rumsey and Unger 2015). The effect size for the simple linear regression was based on coefficient of determination (R^2): 0.01 (small), 0.059 (moderate) and 0.138 (large) (Cohen 1988a).

5.3 FINDINGS

5.3.1 Participant Characteristics

A total of 37 participants were recruited for the study that consisted of 20 sedentary individuals and 17 manual workers. For manual workers, there were no workers who performed heavy or very heavy manual works in this study, as all the workers performed only light to medium manual work for more than 2/3 working hours per day. For each characteristic, comparison between the manual and the sedentary groups were carried out. For normally distributed data (i.e. height, weight and body mass index (BMI), independent *t* test was conducted. For non-normally distributed data (i.e. age and all physical activities), Mann-Whitney test was conducted. According to the results, there were no statistical differences in age, height, weight, BMI and the duration spent on moderate activities and walking. However, there were significant differences in the duration spent on vigorous activity (days/week: $p<0.001$ & minutes/day: $p<0.001$) and sitting ($p<0.01$). The manual workers spent more time in performing vigorous activities compared to the sedentary individuals. On the other hand, the sedentary individuals spent more time in sitting compared to the manual workers ($p<0.01$) (Table 5.1).

Table 5.1. Anthropometric and physical activity characteristics

Characteristics	Sedentary (N=20)		Manual (N=17)		*P*
	Mean/Mdn	**SD/IqR**	**Mean/Mdn**	**SD/IqR**	
Age (years)[2]	27.5	21.8-34.0	31.0	29.0-38.5	0.102
Height (cm)[1]	167.7	5.43	167.0	6.3	0.247
Weight (kg)[1]	73.1	12.6	76.4	8.2	0.253
Body mass index[1]	26.0	4.3	26.5	2.8	0.692
Physical activities (duration spent)					
- Vigorous: days/week[2]	1.0	0.0-2.8	5.0	3.5-6.0	<0.001*
- Vigorous: minutes/day[2]	35.0	0.0-60.0	180.0	120.0-240.0	<0.001*
- Moderate- days/week[2]	3.0	1.0-4.0	2.0	0.5-5.0	0.916
- Moderate: minutes/day[2]	30.0	16.25-120.00	90.0	120.0	0.442
- Walking: days/week[2]	7.0	5.0-7.0	7.0	5.0-7.0	0.617
- Walking: min/day[2]	30.0	16.3-120.0	60.0	45.0-210.0	0.061
- Sitting: min/weekday[2]	480.0	300.0-600.0	240.0	120.0-360.0	0.003*

1=Normally distributed (statistics: mean±SD & independent *t* test)
2=Not normally distributed (statistics: median, IqR & Mann Whitney test)
* significant at $p<0.05$

5.3.2 Isometric Back Endurance

5.3.2.1 Ito test holding time

Isometric back endurance among all participants was examined using Ito test. The isometric back endurance during Ito test was measured based on time-to-failure method (i.e. holding time during Ito test). Independent *t* test was conducted to compare the level of isometric back endurance during Ito test between sedentary and manual groups (Table 5.2). The isometric back endurance was measured based on the holding time in performing the Ito test. In general, the manual workers can hold their back 7 seconds longer than the sedentary individuals during the Ito test. However, the result of independent *t* test indicated that between the sedentary and manual group, there was no significant difference in the level of isometric back endurance.

Table 5.2. Difference in the level of isometric back endurance (holding time in seconds) during Ito test (N=37)

Groups	Mean	SD	*p*	95% Confidence Interval		Cohen *d*
				Lower limit	Upper limit	
Sedentary	146.65	43.00	0.38[1]	-40.00	15.18	0.03
Manual	159.10	38.95				

5.3.2.2 Muscle fatigue during Ito Test

In general, the multifidus muscles had the most negative MFslope for both sides, indicating the most fatigued muscles during Ito test. Comparisons of MFslope were made between sedentary individuals and manual workers for the right (Table 5.3) and left (Table 5.4) muscles. Independent *t* test was conducted to compare the level of muscle fatigue during the Ito test between the groups for variables with normal distribution (i.e. right multifidus, right biceps femoris, left iliocostalis, and left multifidus), while Mann Whitney test was conducted for variables with non-normal distribution (i.e. right iliocostalis, right gluteus maximum, left gluteus maximus and left biceps femoris). The results indicated that no significant difference in MFslope was found between the groups in all muscles with low to medium effect size.

Table 5.3. Differences in level of muscle fatigue (MFslope) during Ito test for right muscles (N=37)

Variables	**Groups**	**Mean**	**SD**	**Median**	**IqR**	***p***	**ES**
Right iliocostalis	Sedentary			-0.01	-0.16 to 0.07	0.30[2]	0.17[b]
	Manual			<-0.01	-0.07 to 0.20		
Right multifidus	Sedentary	-0.12	0.10			0.49[1]	0.22[a]
	Manual	-0.14	0.08				
Right gluteus maximus	Sedentary			-0.01	-0.08 to 0.10	0.76[2]	0.05[b]
	Manual			<-0.01	-0.04 to 0.04		
Right biceps femoris	Sedentary	-0.11	0.12			0.08[1]	0.51[a]
	Manual	-0.06	0.06				

1=Independent *t* test, 2=Mann-Whitney test,
ES= effect size; a = Cohen *d*, b = correlation coefficient (*r*)

Table 5.4. Differences in level of muscle fatigue (MFslope) during Ito test for left muscles (N=37)

Variables	**Groups**	**Mean**	**SD**	**Median**	**IqR**	***p***	**ES**
Left iliocostalis	Sedentary	-0.02	0.14			0.76[1]	0.11[a]
	Manual	-0.04	0.23				
Left multifidus	Sedentary	-0.12	0.09			0.64[1]	0.23[a]
	Manual	-0.14	0.08				
Left gluteus maximus	Sedentary			-0.02	-0.14 to 0.14	0.46[2]	0.13[b]
	Manual			-0.04	-0.15 to 0.00		
Left biceps femoris	Sedentary			-0.08	-0.19 to -0.04	0.58[2]	0.10[b]
	Manual			-0.05	-0.11 to -0.03		

1=Independent *t* test, 2=Mann-Whitney test,
ES= effect size; a = Cohen *d*, b = correlation coefficient (*r*)

5.3.3 Carrying Activity

5.3.3.1 Maximum load

Independent *t* test was conducted to compare maximum carrying load between the sedentary and the manual groups. The maximum carrying load was measured based on the last safe maximum carrying load that can be carried by the participants. The result indicated that there was a significant difference in the maximum carrying load between the groups. The manual workers were able to carry 4kg more load compared to the sedentary group (Table 5.5).

Table 5.5. Differences in maximum carrying load (kg) between sedentary and manual group (N=37)

Groups	Mean	SD	*p*	95% Confidence Interval		Cohen *d*
				Lower limit	Upper limit	
Sedentary	7.65	2.21	<0.001	-5.94	-2.29	1.49
Manual	11.76	3.23				

5.3.3.2 Muscle fatigue during carrying activity

The result had shown that there was no significant between-group difference in the MFslope of bilateral iliocostalis, multifidus, gluteus maximus, biceps femoris, biceps brachii, left latissimus dorsi, right vastus lateralis and left gastrocnemius with low effect size. For the left vastus lateralis, the sedentary individuals were found to have a significantly higher rate of muscle fatigue compared to the manual workers. For the right gastrocnemius, the manual workers were found to have a significantly higher rate of muscle fatigue compared to the sedentary individuals. Both significant differences had moderate effect size (r = 0.3 to 0.5).

Table 5.6. Difference in level of muscle fatigue (MFslope) during carrying for right muscles (N=34)

Variables	Groups	Mean	SD	Median	IqR	*p*	ES
Right iliocostalis	Sedentary			-0.53	-0.78 to -0.06	0.34[2]	0.16[b]
	Manual			-0.78	-1.97 to -0.25		
Right multifidus	Sedentary			-0.77	-1.08 to -0.09	0.57[2]	0.10[b]
	Manual			-0.77	-1.08 to -0.09		
Right gluteus maximus	Sedentary	-0.96	2.33			0.90[1]	0.05[a]
	Manual	-1.06	2.02				
Right biceps femoris	Sedentary			-0.14	-1.54 to 0.09	0.13[2]	0.26[b]
	Manual			-1.54	-2.41 to -2.00		
Right biceps brachii	Sedentary			-0.98	-4.11 to -0.20	0.62[2]	0.09[b]
	Manual			-1.63	-2.06 to 0.14		
Right latissimus dorsi	Sedentary	NA	NA	NA	NA	NA	NA
	Manual	NA	NA	NA	NA	NA	NA
Right vastus lateralis	Sedentary			-1.36	-4.02 to -0.28	0.36[2]	0.16[b]
	Manual			-0.98	-1.56 to -0.64		
Right gastrocnemius	Sedentary			0.82	-0.14 to 1.81	0.01[2]	0.47[b]
	Manual			-0.36	-1.14 to -0.37		

1=Independent *t* test, 2=Mann-Whitney test,
ES= effect size; a = Cohen *d*, b = correlation coefficient (*r)*
NA = not available

Table 5.7. Difference in level of muscle fatigue (MFslope) during carrying for right muscles (N=34)

Variables	Groups	Mean	SD	Median	IqR	*p*	ES
Left iliocostalis	Sedentary			-0.72	-2.73 to 0.07	0.99[2]	<0.01[b]
	Manual			-0.77	-2.26 to -0.98		
Left multifidus	Sedentary			-0.70	-2.03 to -0.18	0.23[2]	0.21[b]
	Manual			-1.14	-2.20 to -0.50		
Left gluteus maximus	Sedentary	-1.06	2.02			0.35[1]	0.33[a]
	Manual	-1.48	1.51				
Left biceps femoris	Sedentary			-0.68	-2.05 to -0.18	0.72[2]	0.06[b]
	Manual			-1.01	-2.00 to -0.51		
Left biceps brachii	Sedentary			-1.58	-2.34 to 0.46	0.22[2]	0.21[b]
	Manual			-0.87	-1.04 to -0.75		
Left latissimus dorsi	Sedentary	NA	NA	NA	NA	NA	NA
	Manual	NA	NA	NA	NA	NA	NA
Left vastus lateralis	Sedentary			-1.20	-2.79 to -0.10	0.02[2]	0.40[b]
	Manual			0.05	-2.79 to 1.82		
Left gastrocnemius	Sedentary			-0.10	-0.88 to 1.85	0.93[2]	0.01[b]
	Manual			1.66	-0.45 to 0.30		

1=Independent *t* test, 2=Mann-Whitney test,
ES= effect size; a = Cohen *d*, b = correlation coefficient (*r*)
NA = not available

5.3.4 Association between Ito Test and Anterior Load Carriage

5.3.4.1 Association between Ito test and maximum load during carrying activity

A simple linear regression was conducted to examine the influence of isometric back endurance on carrying performance (Table 5.8). The results indicated that there was no significant association between isometric back endurance and maximum carrying load in both groups (r=0.266, p<0.071) (Figure 5.3). Furthermore, the isometric back endurance during the Ito test had no significant influence on maximum carrying load. However, the manual group had a higher R^2 value compared to the sedentary group. According to the R^2 value, in the manual group, 12% of variability in the maximum carrying load can be explained by the level of isometric back endurance, whilst the sedentary group only constituted 1%.

Table 5.8. Association between isometric back endurance and maximum carrying load

Groups	***R***	**Unstandardized Coefficient**		***p***	$\mathbf{R^2}$
		Constant	**Beta**		
Sedentary	0.088	133.545	1.713	0.712	0.008
Manual	0.351	109.254	2.914	0.167	0.123
Both	0.266	121.68	3.215	0.112	0.071

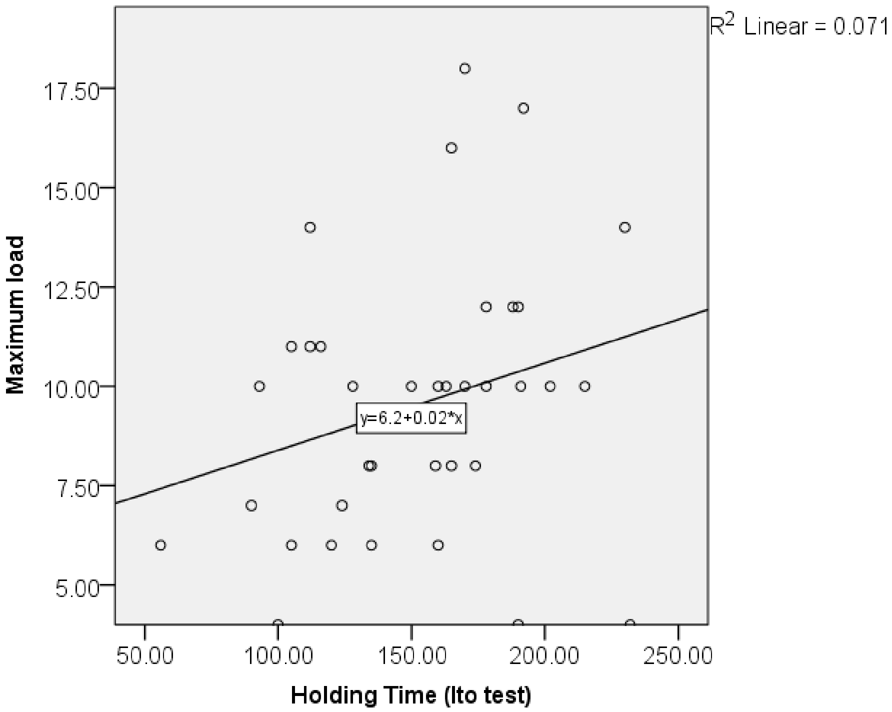


Figure 5.3. Association between Ito test and maximum load (both groups)

5.3.4.2 Association between muscle fatigue (MFslope) during Ito test and anterior load carriage

i. Iliocostalis (N=34)

A simple linear regression was conducted to investigate the association between the slope of EMG median frequency (MFslope) of iliocostalis during Ito test and anterior load carriage. The results indicated that in both groups, there was no significant association between the MFslope during the Ito test and anterior load carriage for both right (r=0.007, p<0.967) and left sides (r=0.030, p<0.861) (Table 5.9).

Table 5.9. Association between iliocostalis muscle fatigue during Ito test and carrying activity according to slope of EMG median frequency (MFslope)

Side	**Groups**	***r***	**Unstandardized coefficient**		***p***	**R^2**
			Constant	**Beta**		
Right	Sedentary	0.086	0.060	1.377	0.717	0.007
	Manual	0.011	-0.769	0.060	0.965	<0.001
	Both	0.007	-0.339	0.079	0.967	<0.001
Left	Sedentary	-0.275	-0.585	-4.727	0.241	0.076
	Manual	0.247	-0.746	1.494	0.339	0.061
	Both	0.030	-0.636	-0.320	0.861	0.001

ii. Multifidus (N=34)

A simple linear regression was conducted to investigate the association between the MFslope of multifidus during Ito test and anterior load carriage. The results indicated that in both groups, there was no significant association between the MFslope during the Ito test and anterior load carriage for both right (r=-0.231, p<0.170) and left sides (r=0.005, p<0.978) (Table 5.10).

Table 5.10. Association between multifidus muscle fatigue during Ito test and carrying activity according to slope of EMG median frequency (MFslope)

Side	**Groups**	***r***	**Unstandardized coefficient**		***p***	**R^2**
			Constant	**Beta**		
Right	Sedentary	-0.121	-0.762	-1.863	0.613	0.015
	Manual	-0.419	-1.207	-6.321	0.094	0.176
	Both	-0.231	-0.893	-3.524	0.170	0.053
Left	Sedentary	-0.026	-0.343	-1.028	0.915	0.001
	Manual	0.007	-1.048	0.099	0.979	<0.001
	Both	-0.005	-0.624	-0.148	0.978	<0.001

iii. Gluteus maximus (N=34)

A simple linear regression was conducted to investigate the association between the MFslope of gluteus during Ito test and anterior load carriage. The results indicated that in both groups, there was no significant association between the MFslope during the Ito test and anterior load carriage for both right (*r*=-0.331, *p*<0.055) and left sides (*r*=-0.010, *p*<0.951) (Table 5.11).

Table 5.11. Association between gluteus maximus muscle fatigue during Ito test and carrying activity according to slope of EMG median frequency (MFslope)

Side	**Groups**	***r***	**Unstandardized coefficient**		***p***	**R^2**
			Constant	**Beta**		
Right	Sedentary	-0.402	-0.666	-4.487	0.079	0.162
	Manual	-0.243	-0.750	-2.008	0.347	0.059
	Both	-0.331	-0.681	-3.228	0.055	0.110
Left	Sedentary	-0.153	-0.789	-1.166	0.519	0.024
	Manual	0.225	-1.266	1.902	0.384	0.051
	Both	-0.010	-1.032	-0.083	0.951	<0.001

iv. Biceps femoris (N=34)

A simple linear regression was conducted to investigate the association between the MFslope of biceps femoris during Ito test and anterior load carriage. The results indicated that in both groups, there was no significant association between the MFslope during the Ito test and anterior load carriage for both right (*r*=0.073, *p*<0.666) and left sides (*r*=0.039, *p*<0.821) (Table 5.12).

Table 5.12. Association between biceps femoris muscle fatigue during Ito test and carrying activity according to slope of EMG median frequency (MFslope)

Side	**Groups**	***r***	**Unstandardized coefficient**		***p***	**R^2**
			Constant	**Beta**		
Right	Sedentary	0.155	-0.586	1.921	0.513	0.024
	Manual	0.083	-1.127	2.133	0.753	0.007
	Both	0.073	-0.911	1.117	0.666	0.005
Left	Sedentary	-0.233	-1.683	-5.638	0.324	0.002
	Manual	0.470	-0.883	0.456	0.057	0.221
	Both	0.039	-1.190	0.710	0.821	0.001

5.4 DISCUSSION

There was no significant difference in the level of isometric back endurance during the Ito test between the manual and the sedentary groups. This finding was also supported by no significant differences in the level of muscle fatigue for iliocostalis, multifidus, gluteus maximus and biceps femoris between the groups. However, there was a significant difference in the maximum carrying load between the groups. In this study, two back muscles were investigated which includes iliocostalis and multifidus. The iliocostalis, the bilateral action of the muscles extends and maintains vertebral column at lumbar region in erect posture, while the unilateral action of the muscle laterally flexes the vertebral column respectively according to the side of activation (Tortora and Derrickson 2008). During the Ito test, this study suggested that the primary action of iliocostalis was to maintain isometric back extension where the muscle was bilaterally activated. In order to adapt with the fatiguing position while keeping the back in extension there was a possibility that the unilateral function of the muscle was also activated.

This study found no significant difference in the muscle fatigue between the groups during the Ito test. However, it can be observed descriptively that the multifidus had the highest rate of muscle fatigue (as indicated by the most negative slope of median frequency) compared to the other muscles in both groups. The multifidus was known to have a strong association with low back pain (Freeman et al. 2010). A major role in spinal stability was assumed by the multifidus because the muscle was responsible for counteracting the forces in the sagittal plane, while the iliocostalis was responsible for the frontal plane (Ng et al. 1997). On the structural design, the physiologic cross-sectional area of the multifidus muscle (i.e. area of the cross section of a muscle perpendicular to its fibers) were relatively greater than other lumbar spine muscles (Ward et al. 2009). Furthermore, with the high physiologic cross-sectional area and relatively short fibres, this design allows the multifidus to produce a large force in order to stabilize the spine, but less contribution to the spine movement. It was previously reported that the multifidus was the most affected muscle in low back pain patients (Freeman et al. 2010), where 80% of low back pain patients had multifidus muscle atrophy, with positive correlation with leg pain (Kader et al. 2000). Therefore, any dysfunction of multifidus can greatly affect spinal stability, particularly during load carriage.

The back muscles also played an important role during an anterior load carriage. In the current study, as the load was placed anteriorly, the body may generate forces to extend the trunk in order to counteract with the gravitational forces that acted on the load during the anterior load carriage. Thus, the activity of back muscles was increased in order to maintain the spinal stability, which consequently lead to muscle fatigue. Furthermore, it can be observed during experimental session that the orientation of the trunk was more extended whilst carrying the maximum load compared to the standard gait (see section 6.3.2.5 for detail). Hence, this finding suggested that the anterior load carriage may lead to a greater muscle fatigue of back muscles than the posterior load carriage, such as backpack carriage (i.e. twice EMG amplitude). To counteract the anterior gravitational pull onto the load, the body had to produce a significant amount of forces to prevent from an exaggerated forward lean. Due to a higher compressive forces onto the spine (Motmans et al. 2006), the anterior load carriage can possibly contribute to a higher risk of low back pain.

To stabilize the position of the trunk so that the extension position can be maintained during the Ito test, biceps femoris may also be activated consistently as a stabilizer. In general, the role of biceps femoris is to enable hip extension and knee flexion. From laboratory observation, a slight intermittent leg movement can be seen throughout the Ito test whilst maintaining the back extension. This phenomenon can possibly be a form of motor control mechanism to adapt to the fatiguing position, which involved a redistribution of activity between the muscles (Hodges and Tucker 2011). By the use of EMG, the redistribution can be determined according to specific group of muscles that are functionally interrelated (Janda et al. 1996). In order to perform a dynamic movement, the muscles tend to work together in a unique pattern, but with variable degree of activation. Unlike the Biering-Sorensen test, the lower body is fixed to the examination table at the pelvis, knee and ankle by a strap. Therefore, it is very unlikely that the motor adaptation phenomenon as seen in the Ito test will happen in the Biering-Sorensen test. As well, the biceps femoris also became fatigued in both groups during the carrying activity. According to Ogden (2002), the muscle was first activated at approximately 80% of gait cycle, and the activation was continued passing the foot strike of a new gait cycle until 20% to 30% of gait cycle. The period of muscle activation may suggest that the role biceps femoris is to prepare the leg for the push-off sequence. As the biceps femoris is responsible to bring the body forward, it has to endure the body weight during the process. Therefore, as the load increases, the muscle may eventually become fatigued. Furthermore, this study also suggested that the biceps femoris muscle fatigue can also be a possible contributor to the reduction in the step/stride length, which will be further discussed in the next chapter (see 6.3.1 for detail). Although the onset of muscle activation was not directly measured for both Ito test and carrying activity in the current study, the findings from this study may offer an insight for future investigations.

Across the carrying activity, the manual workers were found to have a significantly lower rate of muscle fatigue for the left vastus lateralis but a significantly higher rate of muscle fatigue for the right gastrocnemius compared to the sedentary individuals. According to the MFslope, it can be observed that the differences were due to a positive MFslope of a group against a negative MFslope of another group. In other words, between the right and left gastrocnemius and vastus lateralis, the muscles became fatigued on one side, but less fatigued on another side. A positive MFslope would indicate that as the load increases, the median frequency will also be increased, indicating a reduction in the rate of muscle fatigue. As the largest part of quadriceps femoris muscle group, the roles of vastus lateralis are to extend and stabilize the knee (Saladin 2007). During gait, the vastus lateralis is activated mostly during late-swing, loading and initial mid stance stages (Childs et al. 2004; Huang and Ferris 2012). For the left vastus lateralis, the significant difference between the groups can probably due to positive MFslope among the manual workers. This may indicate that the manual workers tend to use the left leg for stability rather than mobility. Furthermore, it can also be observed that the range of motion (ROM) of the left knee was significantly reduced from standard gait to max-kg gait (see section 6.3.2.2 for detail).

Contrariwise, the right gastrocnemius was found to have a significantly higher rate of muscle fatigue in the manual workers compared to the sedentary individuals. The gastrocnemius belongs to triceps surae muscle group along with soleus that regulate plantar flexion around the ankle joint (Saladin 2007). During gait, the gastrocnemius responsible to maintain the ankle stability, to generate the propulsive forces and to accelerate the leg into swing phase (Kepple et al. 1997; Meinders et al. 1998; Neptune et al. 2001; Perry and Burnfield 2010; Silder et al. 2013). Both gastrocnemius and soleus are inserted at calcaneus via Achilles' tendon, which is the strongest tendon of the body (Saladin 2007; Doral et al. 2010) but also associated commonly with sports injury due to sudden stress (Järvinen et al. 2005). To explain the unilateral increased in muscle fatigue (i.e. left side only), previous studies suggested that the right leg was generally responsible for mobility and manipulation, whilst the left leg was generally responsible for stability and postural control (Spry et al. 1993; Velotta et al. 2011). As the right leg tend to focus on mobility, a stronger muscle contraction during the push-off may be needed to enable forward progression, particularly whilst carrying a heavier maximum. This may explain why the manual workers had a lower MFslope for gastrocnemius compared to the sedentary individuals, indicating a relatively higher rate of muscle fatigue. However, these findings had to be interpreted carefully since the manual workers took a relatively longer period of carrying activity as they were able to carry a significantly heavier weight. In other words, the rate of muscle fatigue should be determined with regards to the point where the participant reached their safe maximum carrying load limit.

For the Ito test, the reason of activity termination was due to the participants no longer being able to maintain their isometric back contraction, which was indicated when they were unable to maintain 2/3 of original test position. However, the holding time from the current study (i.e. sedentary: 146.7s±43.0, manual 159.1s±40.0) was lower than the previous studies. The earliest report on the Ito test claimed that the holding time were 208.2s±66.2 and 85.1s±55.6 respectively for the healthy controls and the chronic low back pain patients (Ito et al. 1996). A more recent study by Müller et al. (2010) reported that the holding time for healthy individuals was even higher (i.e. 249.8s±64.1). The variability of the holding time across the studies can probably be due to unavailable precise standardization for the type of pad across the literature (Demoulin et al. 2007). For instance, Müller et al. (2010) used foam pad, while there was no precise description of the pad except ‘small pillow’. This can possibly affect the degree of back extension during the test because it may influence the lordosis of the lumbar vertebrae, which can also influence the holding time. It was assumed that a harder pad can promote a longer holding time due to a higher degree of extension as a consequence of a lesser reduction in lumbar lordosis. During the Ito test, a folded rectangular pillow was placed under the lower abdomen to decrease the lumbar lordosis. Nevertheless, the use of the pillow was standardized in this study to minimize variability. Moreover, the variability (i.e. standard deviation) within the current study was relatively low compared to both previous studies. Therefore, the Ito test in this study was still considered a reliable test according to the holding time. To improve the quality of the test, further study is needed to obtain a consensus on the type of foam used.

Chapter 6: SPATIOTEMPORAL PARAMETERS OF GAIT AND 3D KINEMATICS DURING ANTERIOR LOAD CARRIAGE

6.1 INTRODUCTION

Studies have shown that carrying activity can possibly lead to low back pain due to changes in biomechanical factors such as consistent anterior force on the lower back as exerted by a back pack (LaFiandra et al. 2003), modification in spinal proprioception after load carriage (Hung-Kay Chow et al. 2011), decreased coordination variability in load carriage (Seay et al. 2011a) and increased paraspinal muscle activity (Healey et al. 2005b). Anterior load carriage is one of the most frequent method of carrying in the industrial setting. During the carrying activity, the load is positioned within the visual field that can allow maximum control of the load. The anterior load carriage is suggested to influence the gait evolution from quadripedalism to bipedalism in apes (Hewes 1961; Watson et al. 2009; Carvalho et al. 2012). For instance, bipedalism in chimpanzee can allow full utilization of both hands for tool to obtain and carry food, enhanced visibility, and gives more effective security from potential predators (Tanner and Zihlman 1976). However, to date, studies on the anterior load carriage are very limited as the focus on the literature was on posterior load carriage, particularly on the effects of backpack carriage in schoolchildren (Sharan et al. 2012; Kistner et al. 2013; Dockrell et al. 2015) and soldiers (LaFiandra et al. 2003; Majumdar et al. 2010; Seay et al. 2011a).

To evaluate the carrying activity, a measurement of gait analysis is performed. The gait analysis is a method of breaking down the sequence of body movement that occurs during gait and parameterizing the space (e.g. stride length), time (e.g. cadence) and mixed (e.g. speed) components of the gait (i.e. spatiotemporal parameters) (Rose and Gamble 2006; Perry and Davids 2010). In clinical setting, gait analysis is commonly used to identify any deviation from a normal gait. In functional capacity evaluation (FCE), progressive weights over time are used to simulate fatigue development in the actual carrying activity. To the researcher's knowledge, most of the previous carrying-related study emphasis on the use a fixed load such as 10% to 40% of body weight rather than simulating the FCE by using progressive loads (Neumann and Cook 1985; Cook and Neumann 1987; Hong et al. 2008; Simpson et al. 2012). It is hypothesized that the carrying activity with progressive anterior loads can lead to specific biomechanical changes in the healthy individuals that can be used to assist clinical judgement of FCE. Therefore, the aim of this chapter is to investigate the changes in the spatiotemporal parameters of gait and 3D kinematics of carrying activity between the sedentary individuals and the manual workers. This chapter hopes to establish a thorough understanding on the impact of fatigue on gait whilst carrying a progressive anterior load, as well as to explore any possible health and safety related issues from the carrying activity.

6.2 METHODOLOGY

6.2.1 Study Design

The design of this study was cross-sectional with mixed-group comparisons recruiting healthy participants (n=37). The data were collected from May 2014 to April 2015 (11 months) at the Biomechanics Laboratory, Faculty of Health Sciences, University of Southampton. The main parameters of this study were spatiotemporal parameters of gait and 3D kinematics of ankle, knee, hip, pelvis and trunk. The participants were divided into sedentary individuals (n=20) and manual workers (n=17) for between-group comparison. During the study, the participants were asked to perform two types of gait: standard gait (i.e. self-preferred gait) and carrying activity with progressive loads with 1 kg increment. Within-group comparison was made between the standard gait and the carrying activity with a maximum load (max-kg).

6.2.2 Participants

The participants' inclusion and exclusion criteria were described as 3.2 and the recruitment strategies were described as 3.5.

6.2.3 Procedures

Physical examination that consisted of weight (kg), height (cm), leg length (cm), knee width (cm) and ankle width (cm) were taken (see section 3.7.1 for detail). A series of motion analysis markers were put onto specific sites of the body in preparation for gait activities, which were the standard gait and carrying activity. Prior to any gait activities, the participants were asked to stand static on one force platform for 10 seconds whilst a recording of kinematic was made. The participants then performed standard gait along a 10m walking platform. During the standard gait, the kinematic data were recorded to obtain a baseline measure of the participants' gait. The participants were explained and demonstrated on how to carry a plastic container whilst walking. A plastic container was carried by holding the container's handle, flexing the arm at 90º of elbow flexion and keeping the container as close as possible to the body. A set of carrying activity was performed by walking back and forth along the 10m walking platform (see section 3.7.5 for detail).

6.2.4 Data Processing

To be able to perform statistical analysis, the raw data from motion analysis had to be processed in order to produce 3D kinematics (see section 3.8.1 for detail). The first stage was carried out by analysing the spatiotemporal parameters, which were the duration of stance and swing phase, cadence, step length, step time, stride length, stride time and walking speed. The spatiotemporal parameters for the left and right sides were determined for each participant. The second stage was carried out by analysing the 3D kinematics of movements during standard gait and max-kg gait (Table 3.3). For the ankle and knee, only flexion-extension (sagittal plane) movement was processed. For the hip, both flexion-extension (sagittal plane) and adduction-abduction (frontal plane) movements was processed. For the pelvis and trunk, the movements of flexion-extension (sagittal plane), lateral flexion (frontal plane) and axial rotation (transverse plane) were determined (please see section 3.8.1.4 for detail). The trunk kinematics were analysed according to reference side during gait (i.e. left trunk and right trunk). For instance, the right and left trunk kinematics can be defined as the 3D movements of the trunk during right and left side gait cycle respectively. These movements were measured as degree of rotation and were plotted as waveforms.

6.2.5 Statistical Analysis

Split-plot analysis of variance (SPANOVA) was conducted to compare the changes in spatiotemporal parameters and 3D kinematics across standard and max-kg gait in the manual and the sedentary groups. For spatiotemporal parameters, the duration of stance phase and swing phase, cadence, step length, step time, stride length, stride time and walking speed were determined. For 3D kinematics, the movements were determined for ankle, knee, hip, pelvis and trunk. In this study, only specific angles were chosen for analysis. In sagittal plane, the movements at all joint/body segments (i.e. trunk, pelvis, hip, knee and ankle) were analysed. In frontal plane, only the movements at the trunk, pelvis and hip were analysed. In transverse plane, only the movement at the trunk and pelvis were analysed. All parameters were analysed for both left and right gait cycles. The comparisons for 3D kinematics were conducted based on range of motion (ROM) for two types of gait cycle: full gait cycle and stance phase only. The effect size for the SPANOVA (i.e. magnitude of effect) was based on partial eta squared (ηp^2): 0.1 (good), 0.059 (moderate) and 0.138 (large) (Cohen 1988a).

6.3 FINDINGS

6.3.1 Spatiotemporal Parameters

6.3.1.1 Stance Phase

The duration of stance phase was determined based on the percentage of the duration between foot strike and the subsequent foot off event of the same foot during a full gait cycle. In order to determine the effect of group and gait condition on the duration of stance phase, a SPANOVA test was conducted. The results indicated that there was no significant main effects of group and gait condition on the duration of stance phase (Table 6.1). The interaction effects between group and gait condition for both sides also indicate non-significant results. Therefore, it can be concluded that in both groups there were no significant changes in the duration of stance phase from standard gait to max-kg gait. The average walking speed was 61% for both groups and gait conditions.

Table 6.1. Effect of group and gait condition on stance phase (percentage of gait cycle)

Side	Group	Gait conditions		Within-group		Between-group		Interaction	
		Standard	Max-kg	*p*	ηp2	*p*	ηp2	*p*	ηp2
Right	Sedentary	61.15±1.98	61.23±1.55	0.512	0.012	0.629	0.007	0.343	0.026
	Manual	61.15±1.90	60.72±1.68						
	Both	61.15±1.92	61.00±1.61						
Left	Sedentary	60.86±1.79	61.20±1.78	0.577	0.009	0.816	0.002	0.631	0.007
	Manual	61.12±1.48	61.14±1.16						
	Both	60.98±1.64	61.17±1.51						

6.3.1.2 Swing Phase

The duration of swing phase was determined based on the percentage of the duration between foot off and the subsequent foot strike of the same foot during a full gait cycle. In order to determine the effect of group and gait condition on the duration of swing phase, a SPANOVA test was conducted. The results indicated that there was no significant main effects of group and gait condition on the duration of swing phase (Table 6.2). The interaction effects between group and gait condition for both sides also indicating non-significant results. Therefore, it can be concluded that there were no significant changes in the duration of swing phase from standard gait (right: 38.85±1.92, left: 39.02±1.64) to max-kg gait (right: 39.00±1.61, left: 38.83±1.51) in both the sedentary individuals and the manual workers (i.e. 39% for both groups and gait conditions).

Table 6.2. Effect of group and gait condition on swing phase (percentage of gait cycle)

Side	Group	Gait conditions		Within-group		Between-group		Interaction	
		Standard	Max-kg	*p*	ηp2	*p*	ηp2	*p*	ηp2
Right	Sedentary	38.84±1.98	38.77±1.55	0.512	0.012	0.629	0.007	0.343	0.026
	Manual	38.85±1.90	39.28±1.68						
	Both	38.85±1.92	39.00±1.61						
Left	Sedentary	39.14±1.79	38.80±1.78	0.557	0.009	0.816	0.002	0.631	0.007
	Manual	38.88±1.48	38.86±1.16						
	Both	39.02±1.64	38.83±1.51						

6.3.1.3 Cadence

Cadence was determined based on the number of steps per minute. To determine the effect of group and gait condition on cadence, a SPANOVA test was conducted. The results indicated that there was a significant effect of gait condition on cadence for both sides with large effect size (Table 6.3). However, there was no significant effect of group on cadence for both sides. The interaction effects between group and gait condition for both sides also indicating non-significant results. (Figure 6.1). Therefore, it can be concluded that in both groups, there was a significantly large increase in cadence (i.e. about 10 steps/minutes) from standard gait to max-kg gait.

Table 6.3. Effect of group and gait condition on cadence (steps/minutes)

Side	Group	Gait conditions		Within-group		Between-group		Interaction	
		Standard	Max-kg	*p*	ηp2	*p*	ηp2	*p*	ηp2
Right	Sedentary	107.46±9.52	116.39±10.93	<0.001	0.664	0.619	0.007	0.549	0.010
	Manual	105.26±5.95	115.60±11.18						
	Both	106.45±9.05	116.03±10.90						
Left	Sedentary	107.46±9.09	116.64±10.81	<0.001	0.692	0.672	0.005	0.455	0.016
	Manual	105.35±6.36	116.24±11.07						
	Both	106.49±7.92	116.46±10.78						

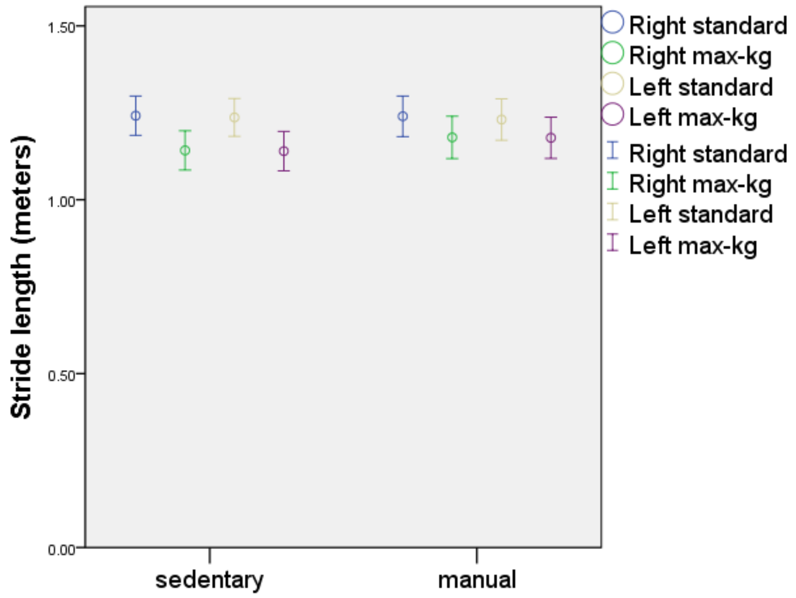


Figure 6.1. Differences in cadence between groups across gait conditions (error bars represent 95% confidence interval)

6.3.1.4 Step length

Step length was determined based on the distance (meters) between the foot strike of one foot and the subsequent foot strike of the opposite foot. To determine the effect of group and gait condition on the step length, a SPANOVA was conducted. The results indicated that there was a significant effect of gait condition on step length for both sides with large effect size (Table 6.4). However, there was no significant effect of group on the step length for both sides. The interaction effects between group and gait condition for both sides also indicating non-significant results (Figure 6.2). Therefore, it can be concluded that in both groups, there was a significantly large decrease in step length (i.e. about 4 cm) from standard gait to max-kg gait.

Table 6.4. Effect of group and gait condition on step length (meters)

Side	Group	Gait conditions		Within-group		Between-group		Interaction	
		Standard	Max-kg	*p*	ηp2	*p*	ηp2	*p*	ηp2
Right	Sedentary	0.62±0.06	0.57±0.06	<0.001	0.475	0.533	0.011	0.351	0.025
	Manual	0.62±0.06	0.59±0.06						
	Both	0.62±0.06	0.58±0.06						
Left	Sedentary	0.62±0.06	0.57±0.06	<0.001	0.375	0.804	0.002	0.132	0.064
	Manual	0.61±0.06	0.59±0.06						
	Both	0.62±0.96	0.58±0.06						

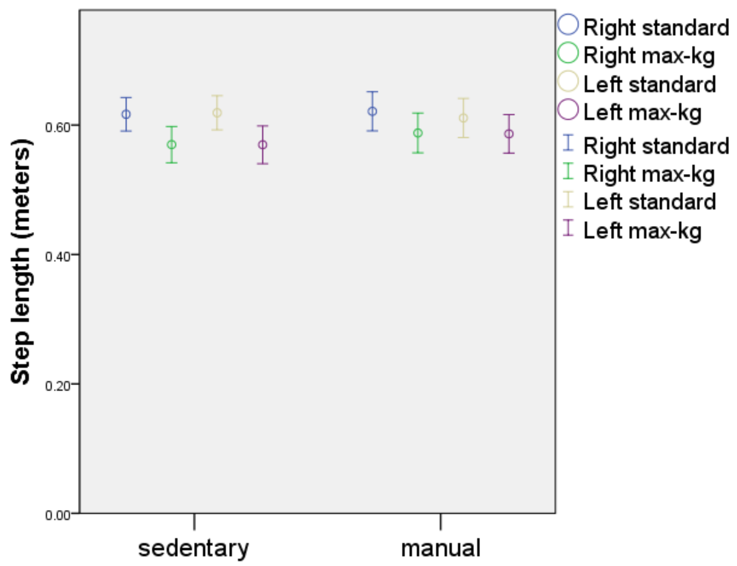


Figure 6.2. Differences in step length between groups across gait conditions (error bars represent 95% confidence intervals)

6.3.1.5 Step time

Step time was determined based on the time taken to complete a step length. To determine the effect of groups and gait conditions on the step time, a SPANOVA was conducted. The results indicated that there was a significant effect of gait condition on step time for both sides with large effect size (Table 6.3). However, there was no significant effect of groups on the step time for both sides. The interaction effects between group and gait condition for both sides also indicating non-significant results (Figure 6.1). Therefore, it can be concluded that in both groups, there was a significant large decrease in step time (i.e. about 0.05 seconds) from standard gait to max-kg gait.

Table 6.5. Effect of group and gait condition on step time (seconds)

Side	Group	Gait conditions		Within-group		Between-group		Interaction	
		Standard	Max-kg	*P*	ηp2	*p*	ηp2	*p*	ηp2
Right	Sedentary	0.56±0.05	0.52±0.06	<0.001	0.619	0.636	0.006	0.540	0.011
	Manual	0.57±0.03	0.52±0.06						
	Both	0.56±0.04	0.52±0.05						
Left	Sedentary	0.56±0.04	0.52±0.05	<0.001	0.699	0.740	0.003	0.773	0.002
	Manual	0.57±0.03	0.52±0.05						
	Both	0.57±0.04	0.52±0.05						

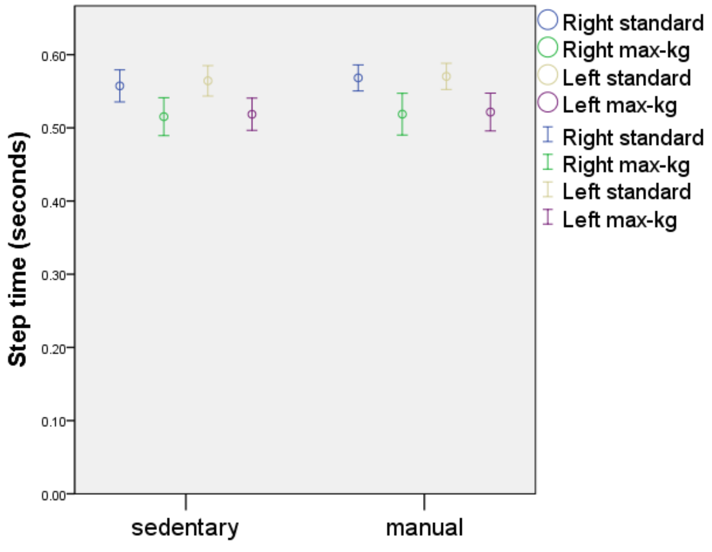


Figure 6.3. Differences in step time between groups across gait conditions (error bars represent 95% confidence intervals)

6.3.1.6 Stride length

Stride length was determined based on the distance (meters) between the foot strike of one foot and the subsequent foot strike of the same foot. To determine the effect of group and gait condition on the stride length, a SPANOVA was conducted. The results indicated that there was a significant effect of gait condition on stride length for both sides with large effect size (Table 6.6). However, there was no significant effect of group on the step length for both sides. The interaction effects between group and gait condition for both sides also indicating non-significant results (Figure 6.4). Therefore, it can be concluded that in both groups, there was a significantly large decrease in step length (i.e. about 8 cm) from standard gait to max-kg gait.

Table 6.6. Effect of group and gait condition on stride length (meters)

Stride length	Group	Gait conditions		Within-group		Between-group		Interaction	
		Standard	Max-kg	*p*	ηp2	*p*	ηp2	*p*	ηp2
Right	Sedentary	1.24±0.12	1.14±0.12	<0.001	0.453	0.625	0.007	0.195	0.047
	Manual	1.24±0.11	1.18±0.12						
	Both	1.24±0.12	1.16±0.12						
Left	Sedentary	1.24±0.12	1.14±0.12	<0.001	0.402	0.652	0.006	0.160	0.056
	Manual	1.23±0.11	1.16±0.12						
	Both	1.23±0.11	1.16±0.12						

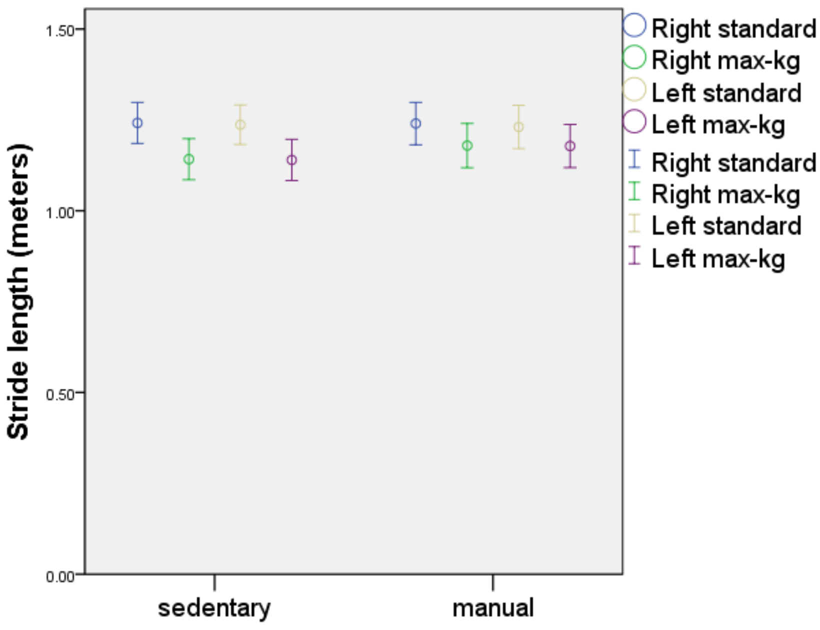


Figure 6.4. Differences in stride length between groups across gait conditions (error bars represent 95% confidence interval)

6.3.1.7 Stride time

Stride time was determined based on the time taken to complete a stride length. To determine the effect of groups and gait conditions on the stride time, a SPANOVA was conducted. The results indicated that there was a significant effect of gait condition on stride time for both sides with large effect size (Table 6.7). However, there was no significant effect of groups on the stride time for both sides. The interaction effects between group and gait condition for both sides also indicating non-significant results (Figure 6.5). Therefore, it can be concluded that in both groups there was a significant decrease in stride time (i.e. about 0.10 seconds) from standard gait to max-kg gait.

Table 6.7. Effect of group and gait condition on stride time (seconds)

Side	Group	Gait conditions		Within-group		Between-group		Interaction	
		Standard	Max-kg	*p*	ηp2	*p*	ηp2	*p*	ηp2
Right	Sedentary	1.13±0.10	1.04±0.10	<0.001	0.646	0.664	0.005	0.644	0.006
	Manual	1.14±0.07	1.05±0.10						
	Both	1.12±0.08	1.04±0.10						
Left	Sedentary	1.12±0.09	1.14±0.12	<0.001	0.702	0.697	0.004	0.473	0.015
	Manual	1.14±0.07	1.18±0.12						
	Both	1.13±0.08	1.04±0.10						

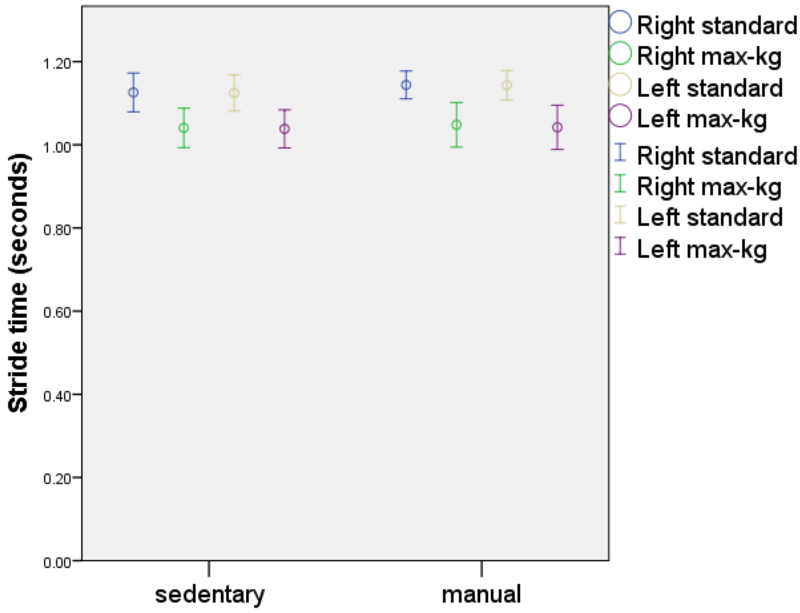


Figure 6.5. Differences in stride time between groups across gait conditions (error bars represent 95% confidence intervals)

6.3.1.8 Walking speed

Walking speed was determined as the stride length divided by stride time. In order to determine the effect of group and gait condition on the walking speed, a SPANOVA test was conducted. The results indicated that there was no significant main effects of group and gait condition on the walking speed (Table 6.8). The interaction effects between group and gait condition for both sides also indicating non-significant results. Therefore, it can be concluded that in both groups, there were no significant changes in the walking speed from standard gait to max-kg gait in both sedentary individuals and manual workers. The average walking speed was about 1.11 meters/second for both groups and gait conditions.

Table 6.8. Effect of group and gait condition on walking speed (meter/second)

Side	Group	Gait conditions		Within-group		Between-group		Interaction	
		Standard	Max-kg	*p*	ηp2	*p*	ηp2	*p*	ηp2
Right	Sedentary	1.11±0.17	1.11±0.16	0.248	0.038	0.985	<0.001	0.146	0.059
	Manual	1.09±0.13	1.14±0.15						
	Both	1.10±0.15	1.12±0.16						
Left	Sedentary	1.11±0.18	1.11±0.17	0.139	0.061	0.993	<0.001	0.105	0.073
	Manual	1.08±0.13	1.14±0.15						
	Both	1.10±0.16	1.12±0.16						

6.3.1.9 Gait-Stability Ratio (GSR)

Gait stability ratio (GSR) was calculated as the ratio between cadence and walking speed. A higher the GSR indicates more stability. To determine the effect of group and gait condition on the GSR, a SPANOVA was conducted. The results indicated that there was a significant effect of gait condition on the GSR for both sides with large effect size (Table 6.9). However, there was no significant effect of group on the cadence for both sides. The interaction effects between group and gait condition for both sides also indicating non-significant results (Figure 6.6). Therefore, it can be concluded that in both groups, there was a significantly large increase in gait stability (i.e. about 7 steps/meter) from standard gait to max-kg gait.

Table 6.9. Effect of group and gait condition on gait-stability ratio (step/meter)

Side	Group	Gait conditions		Within-group		Between-group		Interaction	
		Standard	Max-kg	*p*	ηp2	*p*	ηp2	*p*	ηp2
Right	Sedentary	97.52±8.98	106.17±11.02	<0.001	0.455	0.587	0.009	0.183	0.050
	Manual	97.58±9.31	102.76±10.72						
	Both	97.54±9.01	104.61±10.87						
Left	Sedentary	97.79±8.59	106.38±11.05	<0.001	0.407	0.619	0.007	0.128	0.065
	Manual	98.36±9.66	102.81±10.34						
	Both	98.05±8.97	104.74±10.73						

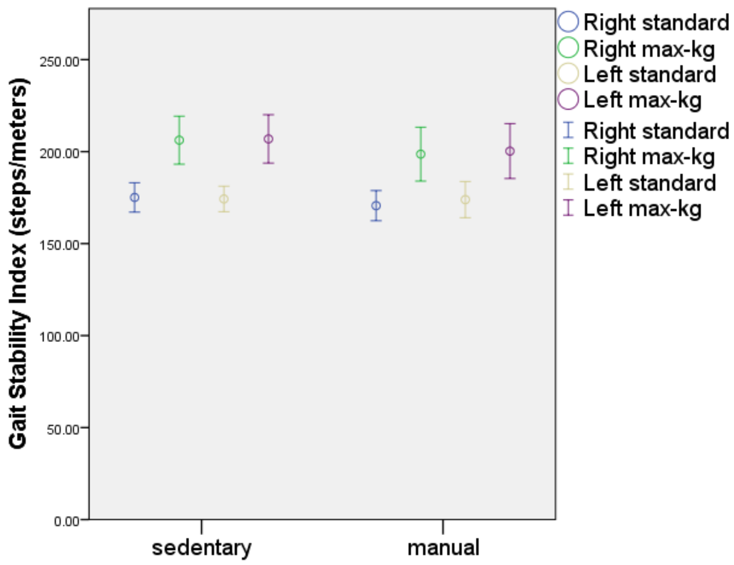


Figure 6.6. Differences in Gait Stability Index (GSR) between groups across gait conditions (error bars represent 95% confidence interval)

6.3.2 3D Kinematics

6.3.2.1 Ankle

To determine the effect of group and gait condition on the ankle dorsiflexion-plantarflexion range of motion (ROM), a split-plot analysis of variance (SPANOVA) was conducted (Table 6.10). According to the result, there was a significant effect of gait condition on ankle dorsiflexion-plantarflexion ROM for both full gait cycle and stance phase only. In both groups, the ankle ROM was decreased from standard gait to max-kg gait (about 3º reduction in ROM). However, there was no significant effect of group on the ROM in both types of gait cycle. The interaction effect between group and gait condition for both sides also indicating non-significant results. According to the ankle ROM waveform, the dorsiflexion was peaked at the end of right single limb support (about 10º), while the plantarflexion was peaked during early swing phase (about -13º). The highest standard deviation for both types of gait were approximately at 60% of gait cycle, which was during foot off event (Figure 6.7).

Table 6.10. Effect of gait condition and group on ankle range of motion (dorsiflexion-plantarflexion)

Joints	Group	Standard	Max-kg	Within-group		Between-group		Interaction	
				p	ηp^2	*p*	ηp^2	*p*	ηp^2
RA[1]	Sedentary	26.80±5.86	22.72±6.18	0.002*	0.240	0.980	<0.001	0.430	0.018
	Manual	25.97±4.40	23.47±5.69						
	Both	26.41±5.18	23.06±5.89						
LA[1]	Sedentary	34.35±12.81	29.59±14.08	0.031*	0.126	0.102	0.075	0.278	0.034
	Manual	26.98±7.28	25.35±8.81						
	Both	30.97±11.14	27.64±11.99						
RA[2]	Sedentary	23.63±6.10	19.90±5.75	0.002*	0.241	0.918	<0.001	0.419	0.019
	Manual	23.06±4.21	20.80±5.38						
	Both	23.37±5.25	20.32±5.52						
LA[2]	Sedentary	30.07±11.38	25.16±9.81	0.023*	0.138	0.146	0.059	0.131	0.064
	Manual	23.93±7.24	22.89±8.02						
	Both	27.25±10.07	24.12±8.98						

RA=Right ankle, LA=Left ankle, 1=Full gait cycle, 2=Stance phase only
*significant at $p<0.05$

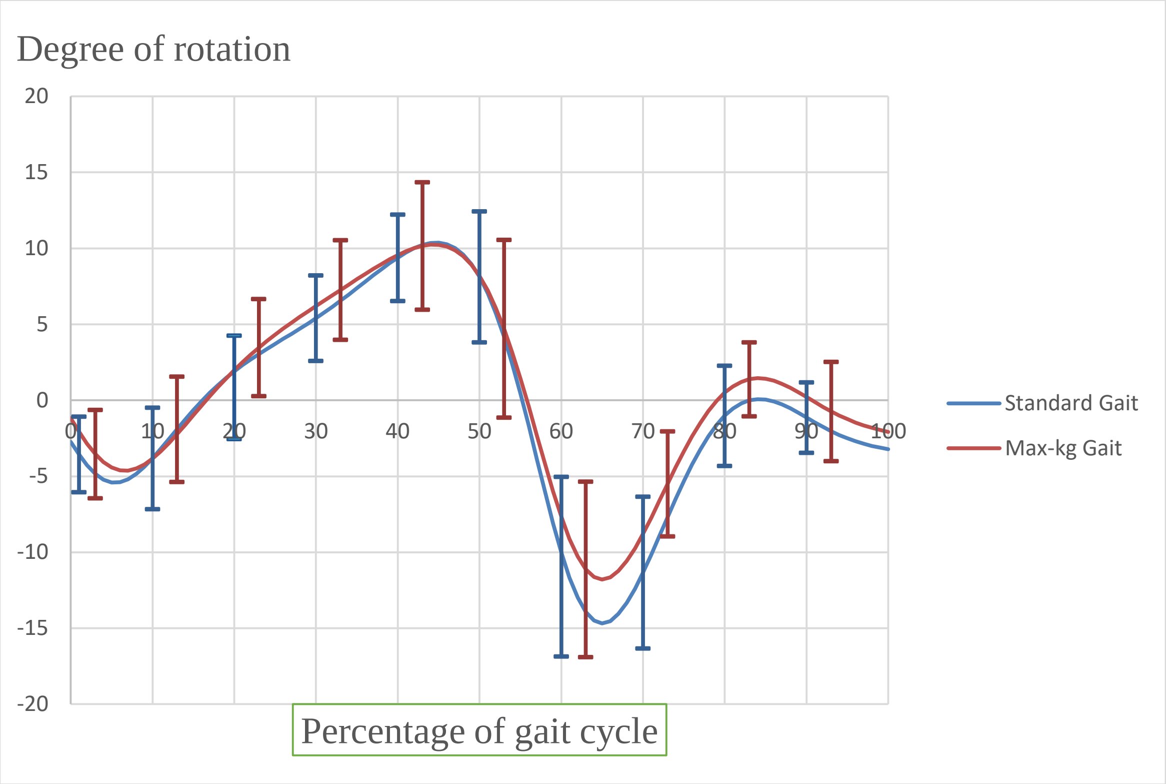


Figure 6.7. Mean flexion-extension waveform of ankle during standard gait and max-kg gait (error bars represent 2 standard deviations).

6.3.2.2 Knee

To determine the effect of group and gait condition on the knee flexion-extension ROM, a SPANOVA was conducted (Table 6.11). During full gait cycle, a significant effect of gait condition on flexion-extension ROM was observed only for the left knee. The ROM for the left knee was decreased from standard gait to max-kg gait in about 5º of ROM. However, during stance phase, the effect of gait condition on flexion-extension ROM was not significant for both sides. According to the waveform, the variability for the standard gait was relatively lower at 30% (terminal stance) until 60% (pre-swing) of the gait cycle compared to the max-kg gait. A relatively lower degree of rotation can be observed in the max-kg gait ROM compared to the standard gait ROM (Figure 6.8). The inconsistent findings of the ROM between full gait cycle and stance phase only was due to the highest knee flexion ROM that occurs during the swing phase, thus, resulted in a relatively lower ROM in the stance phase compared to the full gait cycle. Other than the left knee ROM during the full gait cycle, there was no significant effect of group on the knee ROM, as well as no significant interaction effect in both types of gait cycle.

Table 6.11. Effect of gait condition and group on knee joint range of motion (flexion-extension)

Joints	Group	Standard	Max-kg	Within-group		Between-group		Interaction	
				p	ηp^2	*p*	ηp^2	*p*	ηp^2
RK[1]	Sedentary	55.49±7.78	52.28±9.46	0.064	0.094	0.222	0.042	0.763	0.003
	Manual	57.76±5.12	55.43±8.42						
	Both	56.53±6.70	53.73±9.02						
LK[1]	Sedentary	55.47±9.32	49.91±11.72	0.024*	0.137	0.171	0.053	0.738	0.003
	Manual	57.76±4.88	53.60±8.00						
	Both	56.52±7.60	51.61±10.22						
RK[2]	Sedentary	34.99±7.81	35.46±8.22	0.309	0.030	0.510	0.012	0.555	0.010
	Manual	35.86±6.48	37.61±7.75						
	Both	35.39±7.15	36.45±7.97						
LK[2]	Sedentary	33.94±7.29	32.79±7.33	0.814	0.002	0.126	0.066	0.338	0.026
	Manual	34.93±5.97	36.83±6.37						
	Both	34.40±6.64	34.65±7.11						

RK=Right knee, LK=Left knee, 1=Full gait cycle, 2=Stance phase only
*significant at $p<0.05$

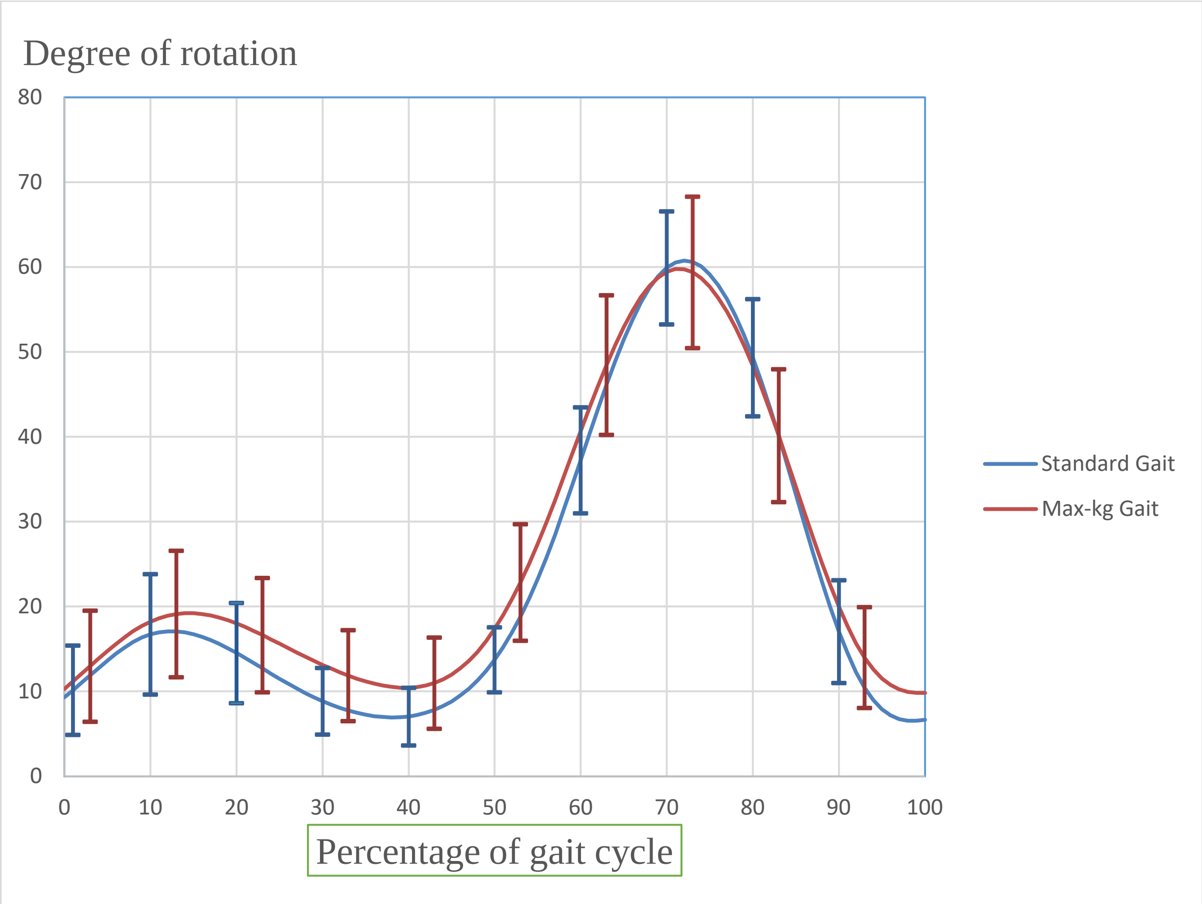


Figure 6.8. Knee flexion-extension waveform during standard gait and max-kg gait (error bars represent 2 standard deviations).

6.3.2.3 Hip

a. Flexion-extension

To determine the effect of group and gait condition on hip flexion-extension ROM, a SPANOVA was conducted (Table 6.12). For both full gait cycle and stance phase only, there was a significant effect of gait condition on the left hip ROM with a significant interaction effect between group and gait condition. There was a significant decrease in the left hip flexion-extension ROM from standard gait to max-kg gait, which was higher in the sedentary group (about 12º decrease in ROM) compared to the manual group (about 4º decrease in ROM). However, there was no significant effect of group, gait condition and interaction between group and gait condition on the right hip ROM for both full gait cycle and stance phase only. The 95% confidence interval for the max-kg gait was wider, indicating more variability compared to the standard gait (Figure 6.9). This finding was also supported by a relatively lower standard deviation for the max-kg gait waveform along the gait cycle (Figure 6.10). The pattern of hip flexion-extension ROM throughout full gait cycle was generally sinusoidal. The highest hip flexion occurred twice during heel strike and towards foot off, while the highest hip extension occurred during the opposite foot contact.

Table 6.12. Effect of gait condition and group on hip joint range of motion (flexion-extension)

Joints	Group	Standard	Max-kg	Within-group		Between-group		Interaction	
				p	ηp^2	p	ηp^2	p	ηp^2
RH[1]	Sedentary	39.12±5.70	41.37±12.15	0.938	<0.001	0.300	0.031	0.180	0.051
	Manual	39.26±5.24	36.73±8.09						
	Both	39.18±5.42	39.24±10.61						
LH[1]	Sedentary	38.62±5.22	26.77±6.76	<0.001*	0.441	0.016*	0.154	0.013*	0.162
	Manual	40.09±5.33	36.09±12.67						
	Both	39.30±5.25	31.05±10.85						
RH[2]	Sedentary	37.84±5.88	38.22±14.61	0.138	0.062	0.268	0.035	0.098	0.076
	Manual	38.54±5.16	32.01±9.30						
	Both	38.17±5.50	35.37±12.69						
LH[2]	Sedentary	36.70±5.66	23.68±9.28	<0.001*	0.440	0.013	0.163	0.026	0.134
	Manual	39.02±5.62	34.02±13.56						
	Both	37.77±5.69	28.43±12.43						

RH=Right hip, LH=Left hip, 1=Full gait cycle, 2=Stance phase only
*significant at p<0.05

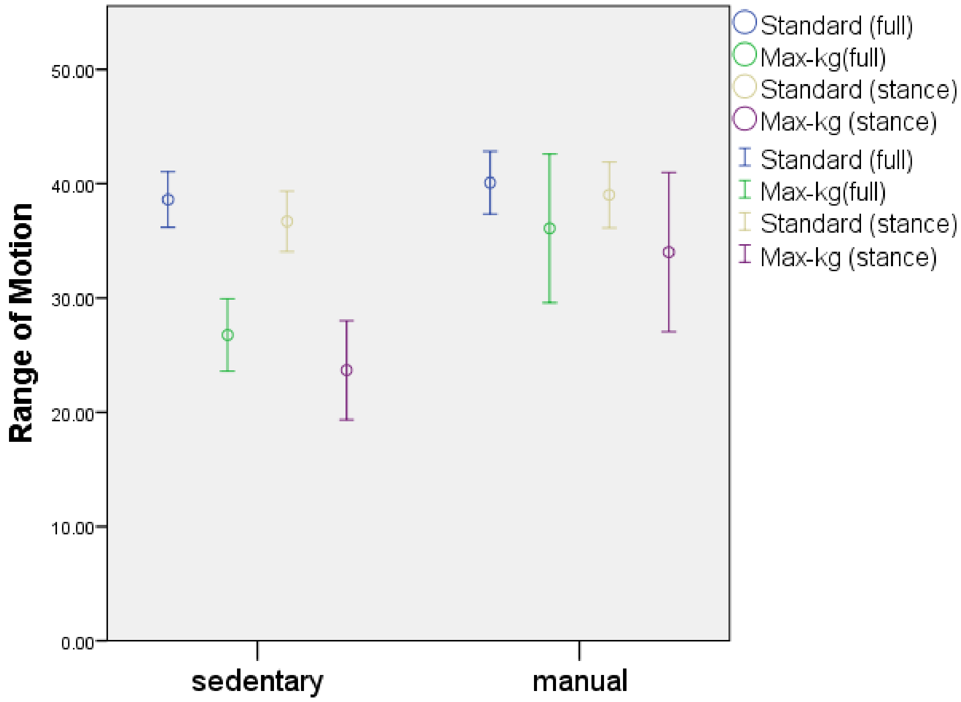


Figure 6.9. Differences in left hip ROM between groups across gait conditions (error bars represent 95% confidence interval).

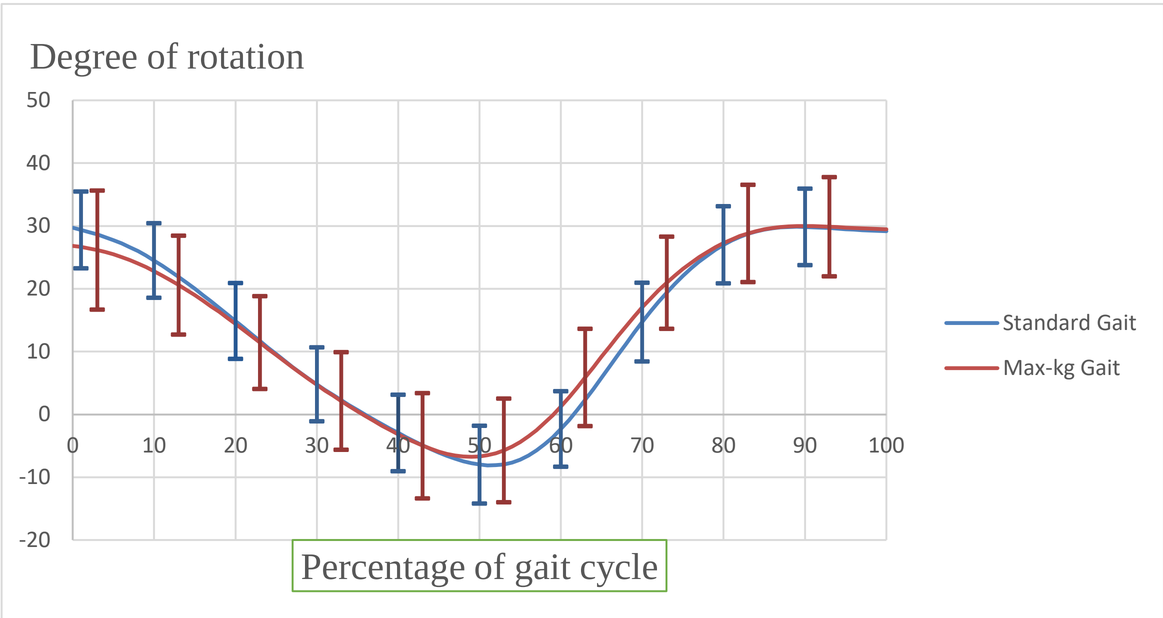


Figure 6.10. Hip flexion-extension ROM waveform during standard gait and max-kg gait (error bars represent 2 standard deviations)

b. Adduction-abduction

To determine the effect of group and gait condition on hip adduction-abduction ROM, a SPANOVA was conducted (Table 6.13). According to the result, a significant effect of gait condition on the adduction-abduction ROM was observed in both right and left hip joints only during full gait cycle. In general, the ROM was increased in about 3º from standard gait to max-kg gait. However, during stance phase only, the effect of gait condition on the ROM was not significant for both sides. The inconsistent finding for between the full gait cycle and the stance phase only was due to the highest abduction that occurred during swing phase, thus, resulted in a lower ROM during the stance phase only. According to the waveform, the adduction and abduction of the hip in the max-kg gait had a relatively wider variability during the beginning of pre-swing event (i.e. about 50° of gait cycle) and towards the end of the gait cycle (Figure 6.11). There was no significant effect of group on the left hip adduction-abduction ROM, as well as no significant interaction effect in both types of gait cycle.

Table 6.13. Effect of gait condition and group on hip joint range of motion (adduction-abduction)

Joints	Group	Standard	Max-kg	Within-group		Between-group		Interaction	
				p	ηp^2	*p*	ηp^2	*p*	ηp^2
RH[1]	Sedentary	12.41±2.72	16.00±11.07	0.019*	0.148	0.822	0.001	0.974	<0.001
	Manual	12.82±2.92	16.32±5.04						
	Both	12.60±2.79	16.15±8.72						
LH[1]	Sedentary	12.52±2.98	13.57±4.48	0.023*	0.140	0.228	0.041	0.275	0.034
	Manual	12.93±3.24	15.81±5.42						
	Both	12.71±3.07	14.60±4.99						
RH[2]	Sedentary	11.02±2.74	13.67±11.66	0.136	0.062	0.906	<0.001	0.800	0.002
	Manual	11.60±2.87	12.49±4.79						
	Both	11.29±2.77	13.59±9.05						
LH[2]	Sedentary	11.31±3.16	10.72±4.26	0.731	0.003	0.161	0.055	0.301	0.031
	Manual	11.80±3.13	12.96±4.60						
	Both	11.53±3.11	11.75±4.50						

RH=Right hip, LH=Left hip, 1=Full gait cycle, 2=Stance phase only
*significant at $p<0.05$

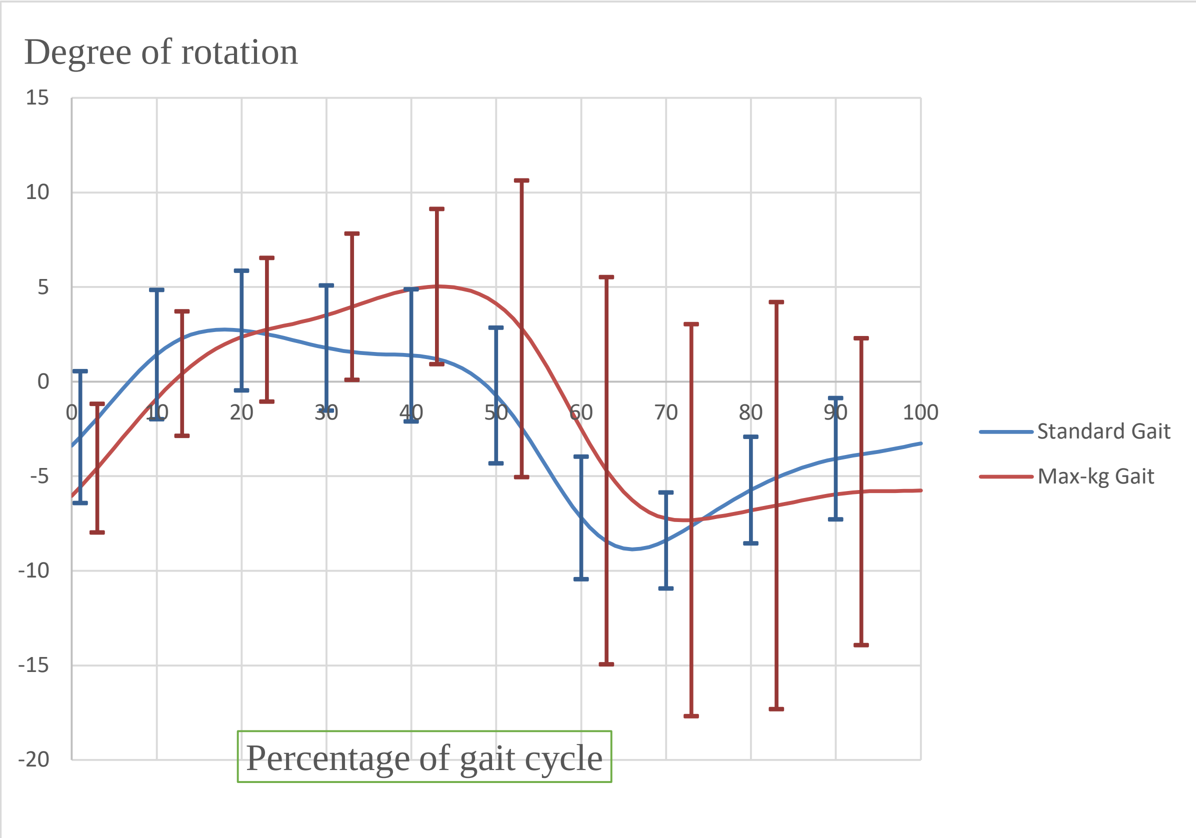


Figure 6.11. Hip lateral flexion waveform during standard gait and max-kg gait (error bars represent 2 standard deviations)

6.3.2.4 Pelvis

a. Pelvic tilt

To determine the effect of group and gait condition on pelvic tilt ROM, a SPANOVA was conducted. For both sides, there was a significant effect of gait condition on pelvic tilt ROM in both full gait cycle and stance phase only. In general, the variability of pelvic tilt motion for both types of gait were relatively high among the participants. Furthermore, the variability during the max-kg gait was relatively higher during the terminal stance (i.e. about 20% to 50% of gait cycle. According to the mean, the pelvic tilt ROM was increased about 7º from standard gait to max-kg gait in a full gait cycle. However, there was no significant effect of group on the pelvic tilt ROM, as well as no significant interaction effect in both types of gait cycle (Table 6.14). According to the waveforms, the pattern pelvic tilt ROM was double sinusoidal, particularly for the standard gait. For the max-kg gait, the pelvic tilt was lower at the beginning and the end of gait cycle (Figure 6.12).

Table 6.14. Effect of gait condition and group on pelvis range of motion (pelvic tilt)

Joints	Group	Standard	Max-kg	Within-group		Between-group		Interaction	
				p	ηp^2	*p*	ηp^2	*p*	ηp^2
RP[1]	Sedentary	3.21±1.79	11.43±10.36	<0.001*	0.403	0.099	0.076	0.210	0.044
	Manual	2.69±0.80	7.48±3.17						
	Both	2.97±1.43	9.62±8.07						
LP[1]	Sedentary	3.10±1.94	9.20±4.29	<0.001*	0.631	0.258	0.036	0.487	0.014
	Manual	2.80±0.66	7.89±3.73						
	Both	2.97±1.48	8.60±4.04						
RP[2]	Sedentary	2.67±1.79	10.26±9.68	<0.001*	0.374	0.106	0.073	0.151	0.058
	Manual	2.43±0.81	6.34±3.30						
	Both	2.56±1.41	8.46±7.63						
LP[2]	Sedentary	2.80±2.00	8.25±4.56	<0.001*	0.564	0.314	0.029	0.540	0.011
	Manual	2.51±0.83	7.04±3.69						
	Both	2.67±1.56	7.69±4.17						

RP=Right pelvis, LP=Left pelvis, 1=Full gait cycle, 2=Stance phase only
*significant at p<0.05

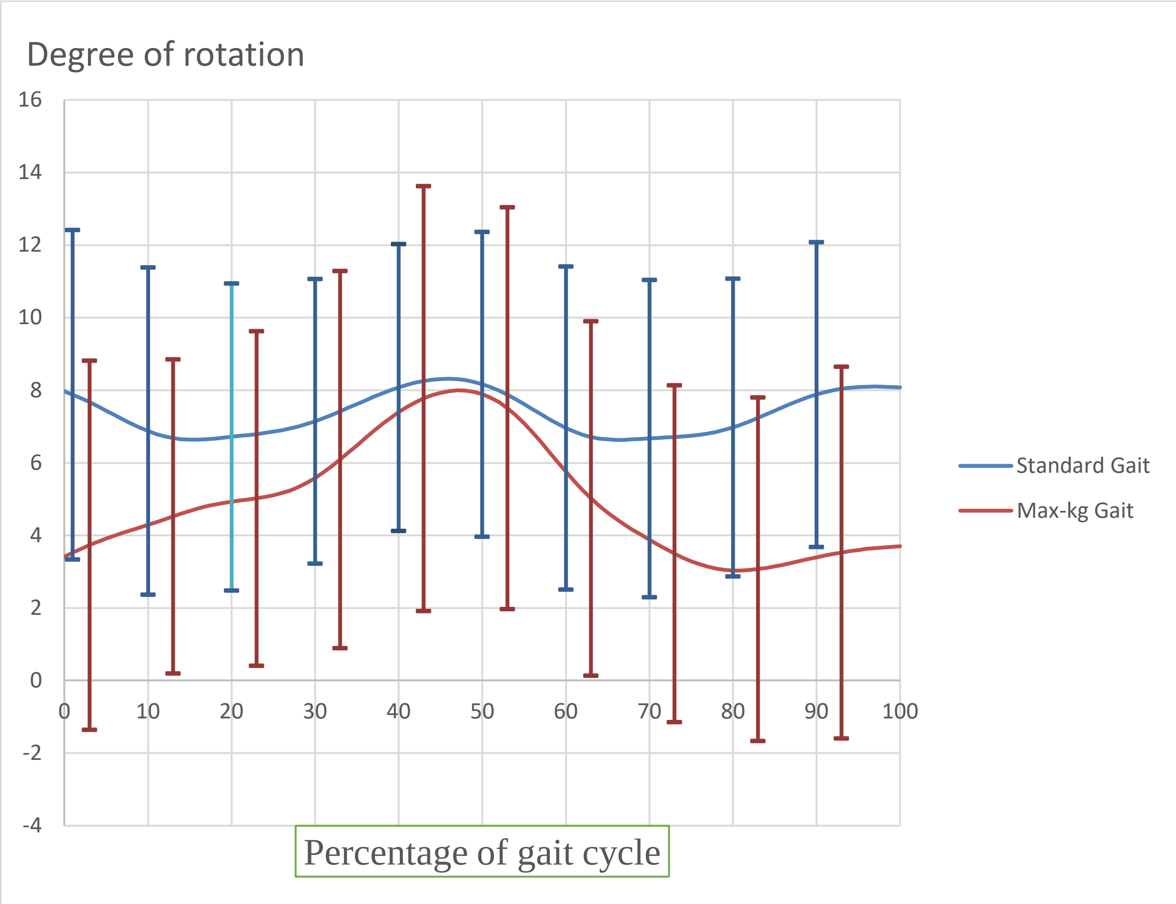


Figure 6.12. Pelvic tilt waveform during standard gait and max-kg gait (error bars represent 2 standard deviations).

b. Pelvic obliquity

To determine the effect of group and gait condition on pelvic obliquity ROM, a SPANOVA was conducted (Table 6.15). For both sides, the effect of gait condition, group and the interaction between gait condition and group on pelvic obliquity ROM were not significant for both full gait cycle and stance phase only. Across the standard gait and the max-kg gait, the pelvic obliquity remained stable at around 7º of ROM.

Table 6.15. Effect of gait condition and group on pelvis range of motion (pelvic obliquity)

Joints	Group	Standard	Max-kg	Within-group		Between-group		Interaction	
				p	ηp²	*p*	ηp²	*p*	ηp²
RP[1]	Sedentary	6.80±2.15	6.84±2.89	0.286	0.033	0.143	0.060	0.316	0.029
	Manual	6.87±2.24	8.36±3.32						
	Both	6.83±2.16	7.54±3.14						
LP[1]	Sedentary	7.03±2.15	6.92±3.04	0.372	0.023	0.213	0.044	0.297	0.031
	Manual	6.94±2.24	8.38±3.46						
	Both	6.88±2.16	7.59±3.28						
RP[2]	Sedentary	6.51±2.28	6.44±3.09	0.615	0.007	0.238	0.040	0.558	0.010
	Manual	6.73±2.36	7.54±3.32						
	Both	6.61±2.28	6.95±3.20						
LP[2]	Sedentary	6.77±2.32	6.30±3.11	0.781	0.002	0.218	0.043	0.369	0.023
	Manual	6.78±2.33	7.68±3.40						
	Both	6.78±2.29	6.93±3.27						

RP=Right pelvis, LP=Left pelvis, 1=Full gait cycle, 2=Stance phase only

c. Axial rotation

To determine the effect of group and gait condition on pelvic axial rotation ROM, a SPANOVA was conducted. During both full gait cycle and stance phase only, the effect of gait condition on pelvis axial rotation ROM were significant for the left pelvis. There was a significant decrease in the both pelvis axial rotation in about 6º ROM from standard gait to max-kg gait (Table 6.16). The interaction effect between group and gait condition for both sides also indicating non-significant results. According to the waveform, a relatively low range ROM was found in the max-kg gait compared to the standard gait. The variability of the motion was relatively consistent throughout the gait cycle for both types of gait (Figure 6.13).

Table 6.16. Effect of gait condition and group on pelvis range of motion (axial rotation)

Joints	Group	Standard	Max-kg	Within-group		Between-group		Interaction	
				p	ηp^2	*p*	ηp^2	*p*	ηp^2
RP[1]	Sedentary	10.51±3.30	4.73±1.72	<0.001*	0.666	0.060	0.097	0.898	<0.001
	Manual	11.72±3.68	6.12±2.59						
	Both	11.07±3.49	5.37±2.24						
LP[1]	Sedentary	10.94±3.31	4.93±1.68	<0.001*	0.687	0.363	0.024	0.664	0.005
	Manual	11.27±3.73	5.84±2.25						
	Both	11.09±3.46	5.35±1.99						
RP[2]	Sedentary	10.11±3.40	3.79±1.79	<0.001*	0.716	0.081	0.084	0.937	<0.001
	Manual	11.41±1.79	4.98±2.25						
	Both	10.71±3.63	4.34±2.07						
LP[2]	Sedentary	10.58±3.33	4.22±1.74	<0.001*	0.720	0.593	0.008	0.994	<0.001
	Manual	10.91±3.72	4.56±1.72						
	Both	10.73±3.47	4.37±1.72						

RP=Right pelvis, LP=Left pelvis, 1=Full gait cycle, 2=Stance phase only
*significant at $p<0.05$

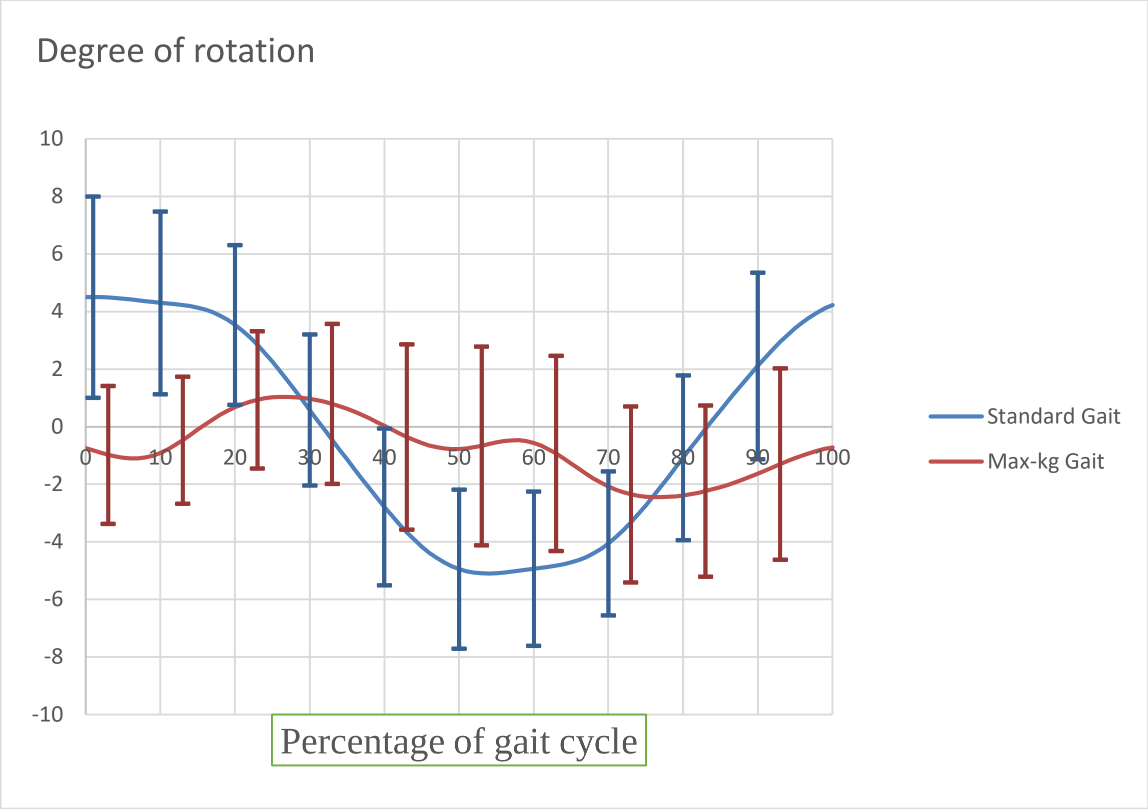


Figure 6.13. Pelvis axial rotation waveform during standard gait and max-kg gait (error bars represent 2 standard deviations)

6.3.2.5 Trunk

a. Flexion-extension

To determine the effect of group and gait condition on the trunk flexion-extension ROM, a SPANOVA was conducted. For both sides, there was significant effect of gait condition on trunk flexion-extension ROM in both full gait cycle and stance phase only (Table 6.17). In general, the variability of the flexion-extension motion of the trunk for both types of gait were relatively high among the participants. For the max-kg gait, the variability became higher starting at the middle of the terminal stance (40% of gait cycle) and towards the end of the gait cycle. According to the waveform, the mean was increased in about 7º from standard gait to max-kg gait in full gait cycle. However, there was no significant effect of group on the ROM. There was also no significant interaction effect between group and gait condition in both types of gait cycle. According to the waveform, the trunk was relatively more extended during the max-kg gait compared to the standard gait (Figure 6.14).

Table 6.17. Effect of gait condition and group on trunk range of motion (flexion-extension)

Joints	Group	Standard	Max-kg	Within-group		Between-group		Interaction	
				p	ηp²	*p*	ηp²	*p*	ηp²
RT¹	Sedentary	3.28±1.81	11.75±10.59	<0.001*	0.415	0.161	0.055	0.340	0.026
	Manual	2.67±0.80	8.39±4.58						
	Both	3.00±1.45	10.21±8.45						
LT¹	Sedentary	3.31±1.83	12.32±11.12	<0.001*	0.498	0.154	0.057	0.164	0.055
	Manual	2.3±0.87	8.24±3.48						
	Both	3.04±1.48	10.45±8.66						
RT²	Sedentary	2.87±1.81	10.86±10.76	<0.001*	0.357	0.115	0.069	0.239	0.039
	Manual	2.33±0.80	6.90±4.20						
	Both	2.62±1.45	9.04±8.54						
LT²	Sedentary	2.82±1.88	9.21±5.71	<0.001*	0.579	0.140	0.061	0.248	0.038
	Manual	2.35±0.87	6.89±3.96						
	Both	2.61±1.50	8.15±5.05						

RT=Right trunk, LT=Left trunk, 1=Full gait cycle, 2=Stance phase only
*significant at $p<0.05$

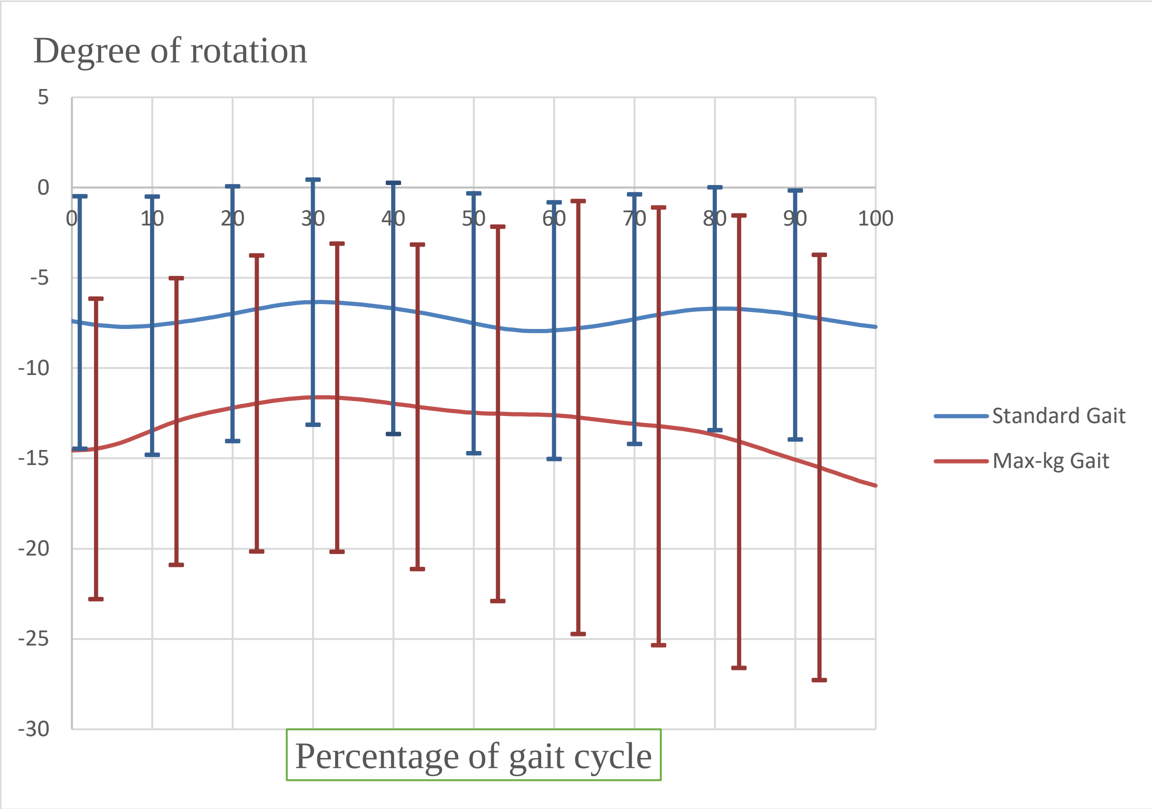


Figure 6.14. Trunk flexion-extension waveform during standard gait and max-kg gait (error bars represent 2 standard deviations)

b. Lateral flexion

To determine the effect of group and gait condition on the trunk lateral flexion ROM, a SPANOVA was conducted. For both sides, there was a significant effect of gait condition on the trunk lateral flexion ROM in both full gait cycle and stance phase only (Table 6.18). In general, the lateral flexion was decreased in about 2º from standard gait to max-kg gait in a full gait cycle (Figure 6.15). However, there was no significant effects of group on the ROM. There was also no significant interaction effect between group and gait condition in both types of gait cycle. The variability of the motion was relatively consistent throughout the gait cycle for both types of gait.

Table 6.18. Effect of gait condition and group on trunk range of motion (lateral flexion)

Joints	Group	Standard	Max-kg	Within-group		Between-group		Interaction	
				p	ηp^2	*p*	ηp^2	*p*	ηp^2
RT[1]	Sedentary	9.68±2.51	7.22±2.87	0.001*	0.253	0.174	0.052	0.861	0.001
	Manual	10.35±2.68	8.13±2.68						
	Both	9.99±2.57	7.63±2.78						
LT[1]	Sedentary	9.83±2.95	7.28±2.77	0.001*	0.278	0.230	0.041	0.870	0.001
	Manual	10.51±2.55	8.18±2.90						
	Both	10.14±2.76	7.70±2.83						
RT[2]	Sedentary	9.07±2.77	6.45±3.32	0.001*	0.294	0.142	0.060	0.901	<0.001
	Manual	10.12±2.77	7.32±2.65						
	Both	9.55±2.78	6.85±3.02						
LT[2]	Sedentary	9.48±3.10	6.77±2.79	<0.001*	0.313	0.244	0.039	0.957	<0.001
	Manual	10.27±2.69	7.48±2.71						
	Both	9.85±2.91	7.09±2.74						

RT=Right trunk, LT=Left trunk, 1=Full gait cycle, 2=Stance phase only
*significant at $p<0.05$

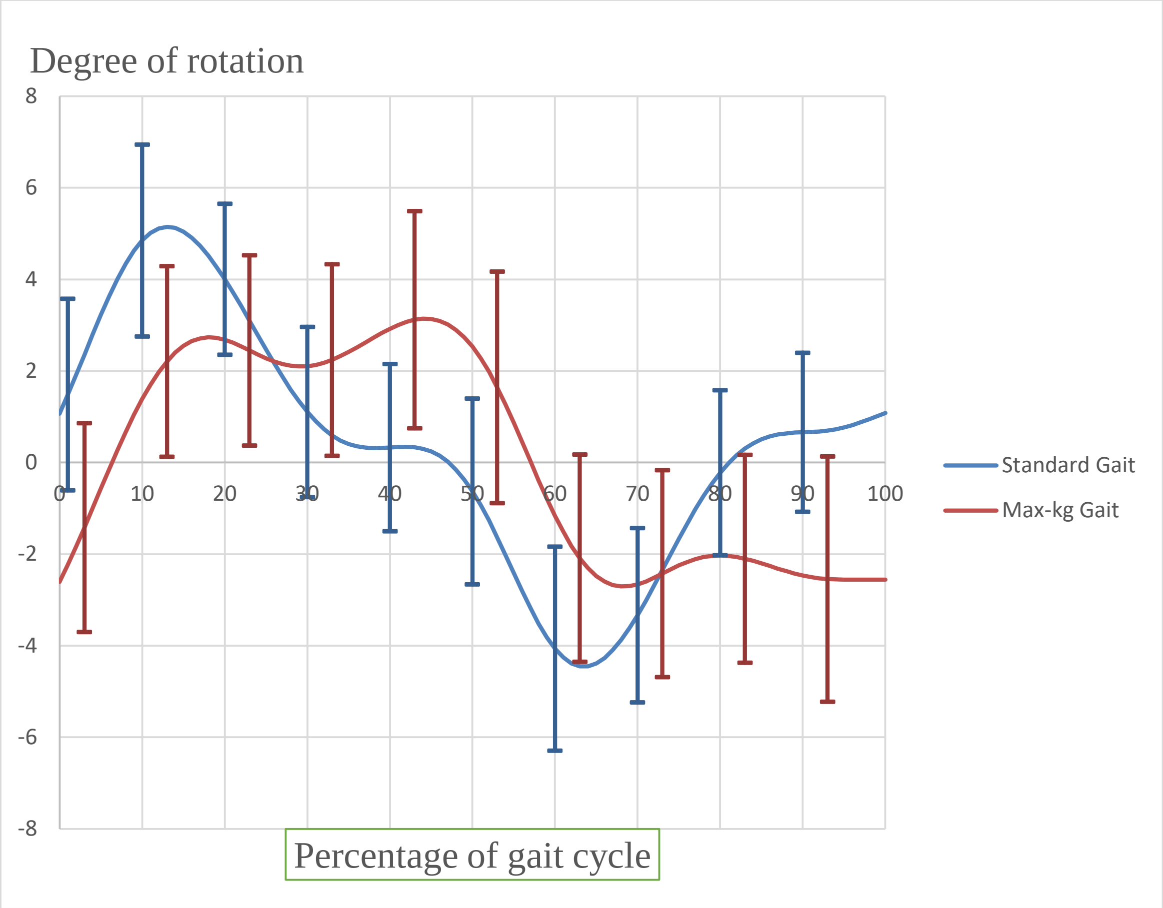


Figure 6.15. Trunk lateral flexion waveforms during standard gait and max-kg gait (error bars represent 2 standard deviations)

c. Axial rotation

To determine the effect of group and gait condition on pelvic axial rotation ROM, a SPANOVA was conducted (Table 6.19). For both sides, there was a significant effect of gait condition on trunk axial rotation ROM in both full gait cycle and stance phase only. In general, the ROM was decreased in about 7º of ROM from standard gait to max-kg gait in a full gait cycle (Figure 6.16). However, there was no significant effects of group on the ROM, as well as no significant interaction effect in both types of gait cycle. For the max-kg gait, the variability became higher during heel off (40% of gait cycle) until the end of the initial swing (80% of gait cycle).

Table 6.19. Effect of gait condition and group on trunk range of motion (axial rotation)

Joints	Group	Standard	Max-kg	Within-group		Between-group		Interaction	
				p	ηp^2	*p*	ηp^2	*p*	ηp^2
RT[1]	Sedentary	14.50±4.03	8.74±13.31	<0.001*	0.348	0.890	0.001	0.329	0.027
	Manual	16.47±3.95	7.28±3.24						
	Both	15.40±4.06	8.07±9.93						
LT[1]	Sedentary	14.70±4.40	8.48+11.52	<0.001*	0.418	0.949	<0.001	0.346	0.025
	Manual	16.06±3.84	6.91±3.07						
	Both	15.32±4.15	7.76±8.65						
RT[2]	Sedentary	14.22±4.11	8.33±13.13	<0.001*	0.372	0.951	<0.001	0.268	0.035
	Manual	16.27±3.76	6.50±2.81						
	Both	15.16±4.03	7.49±9.76						
LT[2]	Sedentary	14.37±4.42	4.78±2.30	<0.001*	0.827	0.096	0.077	0.905	<0.001
	Manual	15.92±3.87	6.14±2.73						
	Both	15.08±4.20	5.41±2.57						

RT=Right trunk, LT=Left trunk, 1=Overall gait cycle, 2=Stance phase only
*significant at $p<0.05$

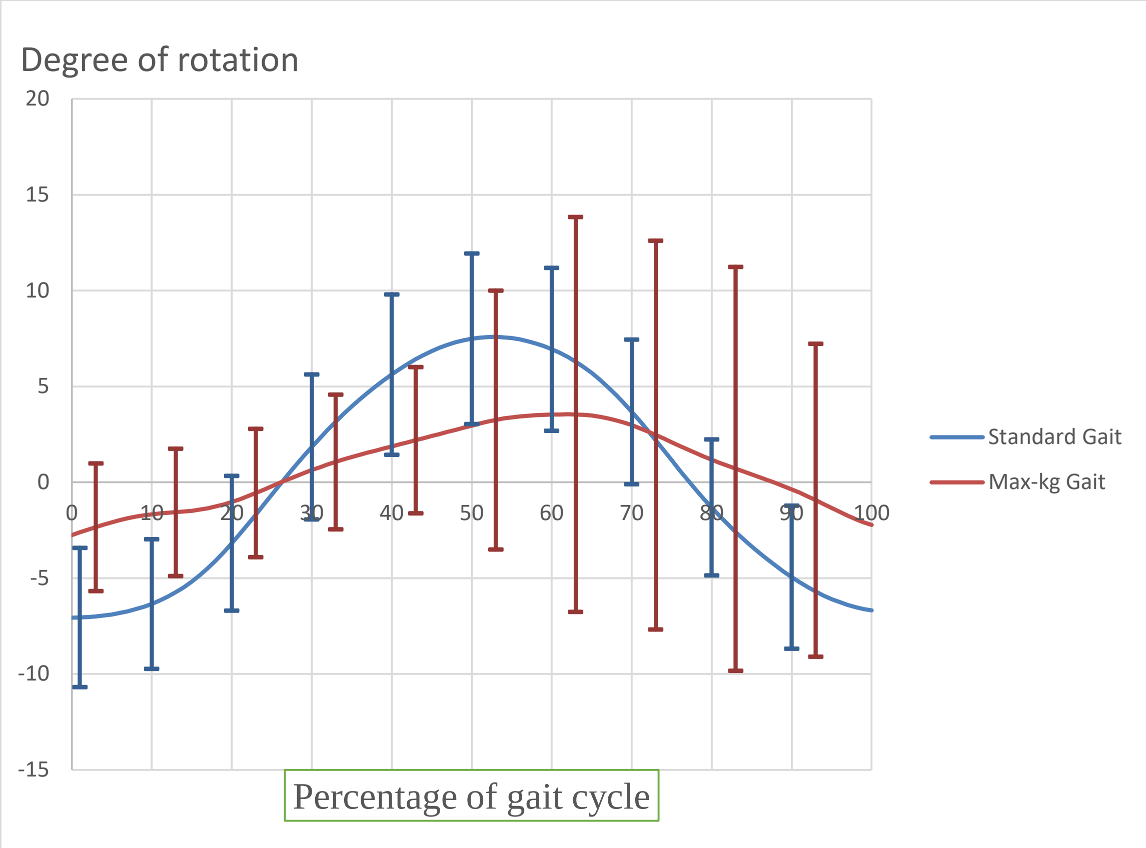


Figure 6.16. Trunk axial rotation waveform during standard gait and max-kg gait (error bars represent 2 standard deviations)

6.3.3 Summary of Changes

The changes in spatiotemporal parameters (Table 6.20) and 3D kinematics (Table 6.21) from standard gait to max-kg gait were investigated in this study. For the spatiotemporal parameters, there was no significant changes in the duration of stance and swing phase and walking speed. There was a significant increase in the cadence and gait stability ratio. However, the step length, step time, stride length and stride time were significantly decreased. The changes in 3D kinematics was organized based on three planes of movement. For flexion-extension movements (sagittal plane), a significant decrease in range of motion from standard gait to max-kg gait was found at the ankle, the knee (left only) and the hip (left only). However, pelvis and trunk flexion-extension ROM were significantly decreased. For adduction-abduction movements (frontal plane), a significant increase was found at the hip (left only). However, the trunk lateral flexion ROM was significantly decreased. For axial rotation movements (transverse plane), a significant increase was found at the pelvis. However, the trunk axial rotation ROM was significantly increased.

Table 6.20. Summary of changes for spatiotemporal parameters from standard gait to max-kg gait

Spatiotemporal Parameters	**Sedentary**	**Manual**
Stance & Swing Phase	NS	NS
Cadence	Increased	Increased
Step length	Reduced	Reduced
Step time	Reduced	Reduced
Stride length	Reduced	Reduced
Stride time	Reduced	Reduced
Walking speed	NS	NS
Gait stability ratio	Increased	Increased

Significant at $p<0.05$, NS = Non-significant

Table 6.21. Summary of significant changes for 3D kinematics during full gait cycle from standard gait to max-kg gait

Joint/segment	**Movement (ROM)**	**Sedentary**	**Manual**
Ankle	Flexion-extension	Reduced	Reduced
Knee	Flexion-extension	Reduced (left only)	Reduced (left only)
Hip	Flexion-extension	Reduced (left only)	Reduced (left only)
	Adduction-Abduction	Increased (right only)	Increased (right only)
Pelvis	Pelvis tilt	Increased	Increased
	Pelvis obliquity	NS	NS
	Axial rotation	Decreased	Decreased
Trunk	Flexion-extension	Increased	Increased
	Lateral Flexion	Decreased	Decreased
	Axial rotation	Decreased	Decreased

Significant at $p<0.05$, NS = Non-significant, ROM = Range of motion

6.4 DISCUSSION

There was a significant change on cadence (increased), step/stride length (reduced) and step/stride time (reduced). However, there was no significant change in walking speed between the gait conditions. These findings indicated an increase in gait frequency when carrying a maximum load without affecting the walking speed. According to the literature, the reduced step/stride length and higher gait frequency can be as a result of a decreased pelvic rotation (Pascoe et al. 1997; LaFiandra et al. 2003). A gait-stability ratio (GSR) was determined to provide a measure of dynamic stability by calculating the ratio between cadence and walking speed. Expressed as steps per meter, the GSR indicated how long the gait cycle was spent in double-limb support. According to the findings, there was a significant increase of dynamic balance from standard gait to max-kg gait. As the dynamic component of gait became reduced, the gait should ideally become more stable (Cromwell and Newton 2004). Other than that, a slower walking speed length and longer steps were reported to associate with decreased stability, hence, contributing to a higher risk of fall (You et al. 2001; Cham and Redfern 2002; Cromwell and Newton 2004; Espy et al. 2010).

There was a significant decrease in range of motion (ROM) from standard gait to max-kg gait for both left and right ankle joints. The reduced ankle ROM may occur due to muscle fatigue around the ankle joint particularly from the plantar flexors of the ankle joint (i.e. triceps surae and perimalleor muscles). Among the muscles, only the activity of gastrocnemius (i.e. plantar flexor) was recorded using the surface electromyography (EMG) (please see Chapter 5 for detail). In general, as a part of triceps surae (i.e. calf muscles), the gastrocnemius contributed to propulsive forces during the gait (Silder et al. 2013). According to the literature, there were three main theories that described the role of plantar flexors during the gait: 1) controlled roll-off, 2) active push-off, and 3) accelerate leg into swing (Neptune et al. 2001). The controlled rolled-off theory stated that as the body rotates over the stance leg (i.e. during single support), the plantar flexors provide ankle stability that allows the foot and tibia to roll forward over forefoot rocker (i.e. controlled ‘fall’) for forward progression (Perry and Burnfield 2010). According to the active push-off theory, rather than a ‘passive’ roll off, active plantar flexors produce an energy that enables forward progression (assisted by knee extensor), which then transferred to the upper body to provide support (particularly during single support) (Kepple et al. 1997). Final theory suggested that as the leg was accelerated into swing by a push off, the energy that derived from the swing leg will be transferred to the trunk (Meinders et al. 1998). From these theories, it can be assumed that the plantar flexors may fatigue greatly during the carrying activity due to the aforementioned actions, thus, affecting the active ROM of the ankle. Nevertheless, in order to prevent the foot from ‘slapping’ the floor during the gait, concentric contraction of the pretibial muscles (i.e. tibialis anterior, extensor digitorum longus, extensor halucis longus) played a major role as the dorsiflexors of the ankle joint. Concentric contractions that occurred as the tibia was pulled the dorsiflexors in the second half of loading response (approximately 8% to 12% of gait cycle) contributed to the ankle motion (Perry and Burnfield 2010). Other than that, it can also be observed that the reduction in the left hip ROM was relatively smaller in the manual group compared to the sedentary group. Furthermore, the left ankle, the knee and the hip ROM (i.e. left leg ROM) were significantly decreased from standard gait to max-kg gait during a full gait cycle. These findings may indicate that the sedentary group required a more adjustment in the left leg ROM to maintain stability control whilst carrying the maximum load (see 8.3 for details).

A significant increase of pelvic tilt ROM can also be observed from standard gait to max-kg gait. This finding was supported by previous studies (Smith et al. 2006; Birrell and Haslam 2009). However, these previous studies were based on posterior load carriage (i.e. backpack), which explained the increase in the pelvic tilt ROM in order to keep an upright posture when the COM was displaced posteriorly. Whilst carrying a maximum anterior load, the anterior pelvic tilt was reduced particularly at the beginning and at the end of the gait cycle. However, there was no apparent changes in the pelvis tilt from standard gait to max-kg gait at approximately 50% of the gait cycle where the opposite foot strike may take place. It was assumed that the pelvic tilt ROM was limited due to the extended position of trunk corresponding to the anterior load carriage, particularly at the beginning of the gait cycle. For pelvic obliquity, there was no significant ROM change from standard gait to max-kg gait, which was similar to the findings from (Birrell and Haslam 2009). The importance of pelvis motion in carrying activity can be further explained according to the compass gait model (Saunders et al. 1953; Rose and Gamble 2006; Charalambous 2014). As one of the classical model gait, the model had proposed six fundamental determinants of human locomotion, namely pelvic rotation as the first determinant of gait, followed by pelvic obliquity, knee flexion in the stance phase, foot and knee mechanism and the lateral displacement of the pelvis. According to the model, the role of pelvis rotation was to improve energy conservation by flattening the arcs of the COM during gait. In the current study, about 6º decrease in the left pelvic axial rotation ROM from standard gait to max-kg gait can be observed in this study. As the left hip, left knee and left ankle flexion-extension

ROM were also significantly reduced, it was suggested that the reduction in the left leg ROM were necessary to enable greater support from the left leg whilst carrying a maximum load.

A significant increase in trunk ROM can also be observed in both groups. Based on the flexion-extension movement waveform, the max-kg waveform occurred in a more extended orientation. The trunk extension manifested the opposing force produced by the body against the anterior load that decreases the moment of inertia of the load in order to maintain an erected trunk posture during the gait. According to the Newton's law of acceleration, the acceleration (A) of an object is dependent upon both mass (M) and force (A). As the anterior load (M) increased alongside with a continuous gravitational force (F), the body may experience an increased acceleration of forward leaning of the trunk during the carrying activity (i.e. F=MA). To counteract the anterior loading, the body had to produce a significant amount of forces to prevent from an exaggerated forward lean. Contrariwise, forward leaning of the trunk was indicated in posterior load carriage (e.g. backpack carriage) in order to counteract the posterior gravitational pull onto the load (Chansirinukor et al. 2001; Li et al. 2003; Chow et al. 2005). Other than that, this study also found significant decrease in the trunk lateral flexion for both groups. As the displacement of COM in the frontal plane can also be influenced by the mediolateral moment of inertia, gait stability may be compromised. Thus, the body tends to restrict the lateral flexion in about 2º of ROM to produce a more rigid trunk movement in the frontal plane. A more decrease in ROM can be observed in the axial rotation of the trunk in about 7º of ROM. The extended trunk movement with a reduced lateral flexion and a reduced external rotation during max-kg gait can possibly be due to an increased back muscle activity. It was assumed that this can possibly relate the guarding mechanism against fatigue. Main and Watson (1996) was the first to introduce the guarding mechanism to explain the changes in movement pattern among chronic low back pain (LBP) patients, which includes the increase in muscle activity. Van der Hulst et al. (2010) also reported an increase in the activity of erector spinae muscles in chronic LBP participants compared to asymptomatic controls during gait. Furthermore, the current findings also suggested that the cumulative forces generated from the muscle contractions were directed towards lumbar spine, which can possibly occurs in order to improve spinal stability of that region (van Dieën et al. 2003). Increased tension around the lumbar spine may increase the pressure onto the vertebral structures and nearby soft tissues, which may lead to the development of LBP.

The findings on the trunk kinematics were based on the trunk movement relative to the pelvis. According to literature, another common method to determine the trunk kinematics was based on global coordinate system (GCS) (Seay et al. 2011a; Seay et al. 2011c, b). In other words, instead of using the pelvis local coordinate system (LCS) as the reference, the movement of the trunk was based on a predetermined laboratory coordinate system (GCS). This method is commonly used for determining pelvis-trunk coordination. Otherwise, an accurate measure of independent kinematics of the trunk and the pelvis cannot be established. In order to compare the findings on the trunk kinematics across the literature, a careful interpretation has to be made to differentiate between the trunk movements relative to the pelvis or laboratory. Furthermore, it was reported in the literature that the changes in the pelvis-trunk coordination are associated with low back pain (LBP). Therefore, the next chapter will focus on the pelvis-trunk coordination to explore the changes in the coordination whilst carrying a maximum load among healthy individuals and to investigate any possible mechanism that may lead to the development of LBP.

Chapter 7: CHANGES IN PELVIS-TRUNK COORDINATION DURING ANTERIOR LOAD CARRIAGE

7.1 INTRODUCTION

In kinematics studies, the angular movement of a body segment can be described either relative to other body segment (i.e. relative motion) or to a predefined global coordinate system (i.e. absolute motion) (Robertson et al. 2013). In order to perform a functional movement such as walking, the movement of a segment is often coupled with other segment to maintain postural stability (Krasovsky and Levin 2009). A good movement coupling between the body segments is important to enable an effective and smooth human locomotion. This movement coupling can be divided into intra-limb, inter-limb coordination and intersegmental coordination (Haddad et al. 2006; Plotnik et al. 2013). Intra-limb coordination represents the coupling within the same limb such as between the knee and ankle (Reisman et al. 2005; Hu et al. 2016; Ohmura et al. 2016). The inter-limb coordination represents the coupling between two different limbs such as arms and legs (Donker and Beek 2002; Zehr and Duysens 2004) or between bilaterally same limbs (Swinnen et al. 2010). The intersegmental coordination represents the coupling between adjacent or non-adjacent different body segment such as between the pelvis and trunk (Lamoth et al. 2002a; Lamoth et al. 2002b; Lamoth et al. 2006; Seay et al. 2011a; Seay et al. 2011c, b). In general, the patterns of coordination are commonly described as either in-phase or anti-phase. The in-phase coordination is indicated when two segments synchronously move in the same direction, while the anti-phase coordination is indicated when two segments synchronously move in the different directions (Seay et al. 2011c).

One of the most commonly studied coordination during gait is the intersegmental coordination between the pelvis and trunk, or simply pelvis-trunk coordination. Studies have shown that the pelvis-trunk coordination has a unique association with low back pain (LBP). Currently, there is an increasing trend of research that observes how these segments coincide functionally during specific types of gait. It is commonly reported that the pelvis-trunk coordination was different among the LBP patients compared to the healthy controls. The LBP patients generally spent most of the time in in-phase coordination which is also known as guarded activity to prevent from further injury (Van der Hulst et al. 2010; Seay et al. 2011c). For instance, the fast rule to investigate the difference is by comparing the coordination pattern between walking and running. In a standard walking speed, individuals with no LBP would exhibit an in-phase coordination in transverse plane. As the speed increases, the coordination gradually changed from in-phase to anti-phase coordination (Lamoth et al. 2002a). However, the ability to change the coordination during running was diminished in LBP patients (Lamoth et al. 2002b; Lamoth et al. 2006; Bruijn et al. 2008). Although the literature had established the changes in the pelvis-trunk coordination is a prevalent clinical manifestations of LBP, little is known about how the changes can lead to the development of LBP (LaFiandra et al. 2003). Therefore, this chapter aimed to investigate the changes in pelvis-trunk coordination in anterior load with progressive loads in healthy individuals. This chapter hopes to provide a baseline information on how the changes can relate to the mechanism of LBP.

7.2 METHODOLOGY

7.2.1 Study Design

The design of this study was cross-sectional with mixed-group comparisons recruiting healthy participants (n=37). The data were collected from May 2014 to April 2015 (11 months) at the Biomechanics Laboratory, Faculty of Health Sciences, University of Southampton. The participants were divided into sedentary individuals (n=20) and manual workers (n=17) for between-group comparison. During the study, the participants were asked to perform two types of gait: standard gait (i.e. self-preferred gait) and carrying activity with progressive loads (i.e. one kg increment). Within-group comparison was made between the standard gait and the carrying activity with a maximum load (max-load).

7.2.2 Participants

The participants' inclusion and exclusion criteria were described as 3.2 and the recruitment strategies were described as 3.5.

7.2.3 Procedures

Anthropometric measurements which consisted of weight (kg), height (cm), leg length (cm), knee width (cm) and ankle width (cm) were then taken (see section 3.7.1. for detail). A series of motion analysis markers were put onto specific sites of the body in preparation for gait activities, which were the standard gait and carrying activity. For the trunk, the markers were C7 vertebrae, T8 vertebrae and suprasternal notch (IJ) and xiphoid process (PX). For the pelvis, markers were right and left anterior suprailiac spine (RASI & LASI), posterior suprailiac spine (RPSI & LPSI) and the most lateral part of the right and left suprailiac spine (RSIS & LSIS). Prior to any gait activities, the participants were asked to stand static on one force platform for ten seconds whilst a recording of kinematic was made. The participants then performed standard gait along a 10m walking platform. During the standard gait, the kinematic data were recorded to obtain a baseline measure of the participants' gait. The participants were explained and demonstrated on how to carry a plastic container whilst walking. A plastic container was carried by holding the container's handle, flexing the arm at 90º of elbow flexion and keeping the container as close as possible to the body. A set of carrying activity was performed by walking back and forth along the 10m walking platform (see section 3.7.5 for detail).

7.2.4 Data Processing

To be able to perform statistical analysis, raw data from motion analysis had to be processed in order to produce 3D kinematics (see section 3.8 for detail). The data processing was carried out by analysing the 3D kinematics of movements during standard gait and max-kg gait (Table 3.3). The 3D kinematics of trunk relative to pelvis (trunk), trunk relative to global coordinate system (trunk global) and pelvis was calculated based on the plug-in-gait model (see section 3.8.1 for detail). The movement of trunk, trunk global and pelvis were determined for flexion-extension (sagittal plane), lateral flexion (frontal plane) and axial rotation (transverse plane). The trunk kinematics were analysed according to reference side (i.e. left trunk and right trunk). For instance, the right and left trunk kinematics can be defined as the 3D movements of the trunk during right and left side gait cycle respectively.

Pelvis-trunk coordination can be described as the intersegmental coupling of between pelvis and trunk movements. In this study, the coordination was determined using vector coding (Chang et al. 2008; Seay et al. 2011b). In general, vector coding is a quantification of angle-angle diagram, i.e. movement plot of one body segment against another, to produce a continuous joint coordination angle (i.e. coupling angle). For pelvis-trunk coordination, the relative motion plot was constructed based on the movements of pelvis and trunk relative to the laboratory coordinate system (Figure 7.1). The coupling angle was calculated at each percentage of gait cycle for the left and right stride. There are four coordination patterns that can be interpreted based on the coupling angles (Figure 7.2), which are trunk only (only trunk is moving), in-phase (both pelvis and trunk move simultaneously in same direction), pelvis only (only pelvis is moving) and anti-phase (both pelvis and trunk move simultaneously in opposite direction) (Figure 7.3). For statistical analysis, the percentage of each coordination pattern during gait cycle was calculated to compare the coupling angles between sedentary and manual groups.

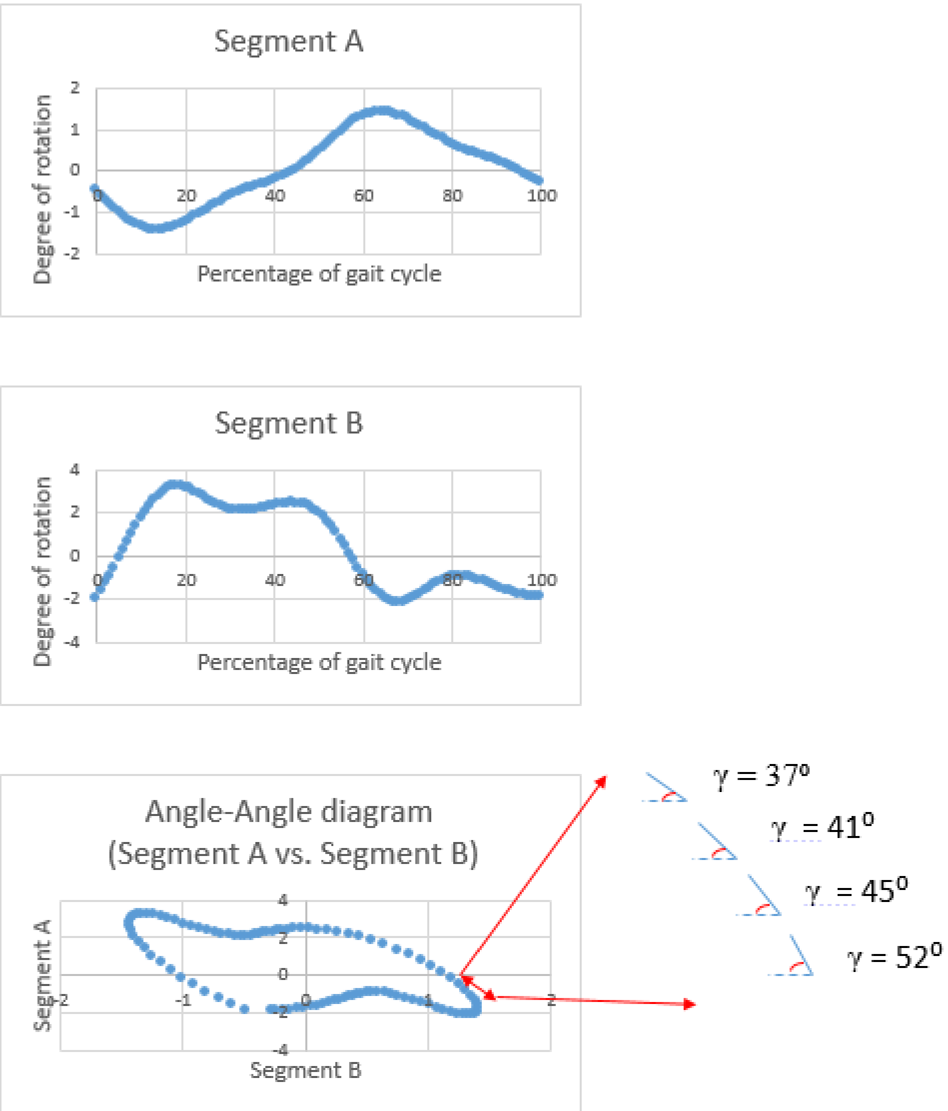
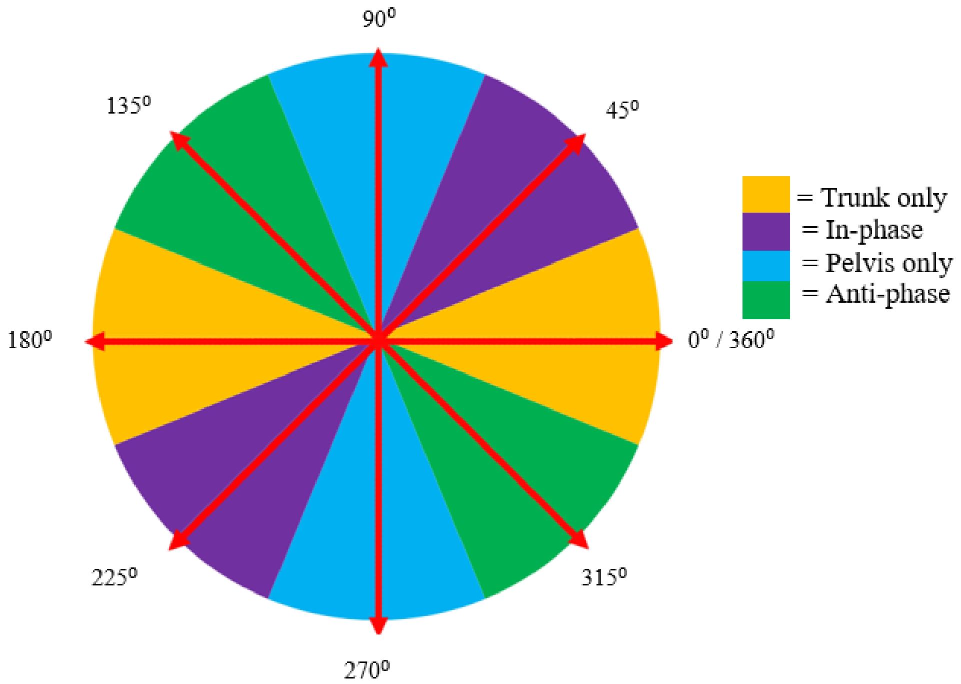


Figure 7.1. Example of angle-angle diagram (γ = coupling angles)



From	22.5°	202.5°	67.5°	247.5°	112.5°	292.5°	0°	157.5°	337.5°
To	67.49°	247.49°	112.49°	292.49°	157.49°	337.49°	22.49°	202.49°	360°
Coordination pattern	In-phase		Pelvis only		Anti-phase		Trunk only		

Figure 7.2. Categorization of coupling angles into four types of coordination (Chang et al. 2008; Seay et al. 2011b)

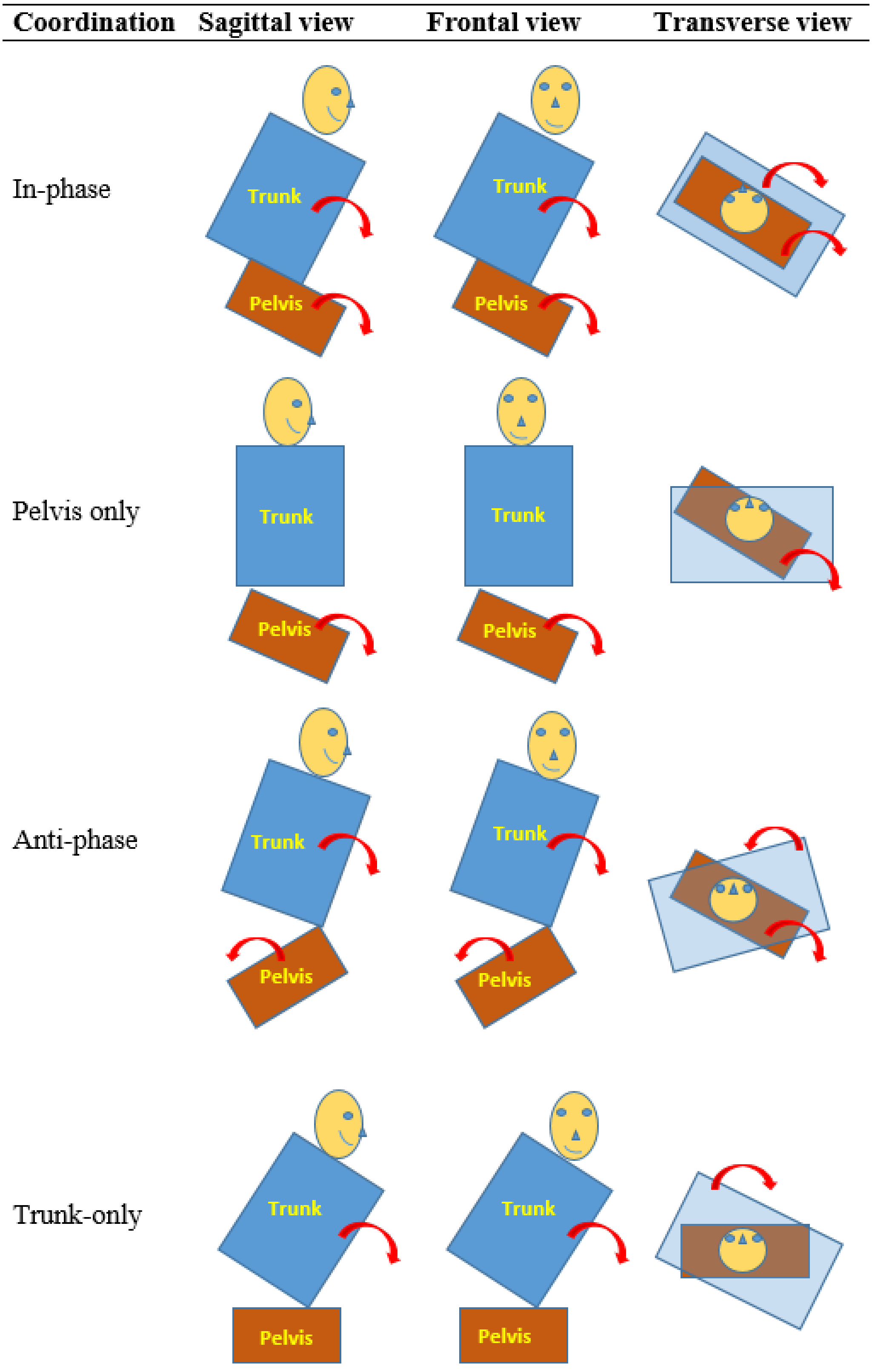


Figure 7.3. Visual representation of pelvis-trunk coordination

7.2.5 Statistical Analysis

Split-plot analysis of variance (SPANOVA) was conducted to compare the changes in 3D kinematics of trunk relative to pelvis (trunk), trunk relative to global coordinate system (trunk global) and pelvis across standard and max-kg gait in the manual and the sedentary groups. All angles were analysed for both left and right gait cycles. The comparisons for 3D kinematics were conducted based on range of motion (ROM), which was calculated as the difference between the minimum and maximum degree of rotation. The effect size for the SPANOVA (i.e. magnitude of effect) was based on partial eta squared (ηp^2): 0.1 (good), 0.059 (moderate) and 0.138 (large) (Cohen 1988a). The pelvis-trunk coordination was determined based on a specific range of coupling angles (Figure 7.2). The coupling angles across the gait cycle were calculated for each participant. To plot an average of coupling angles, circular statistics was used due to the discontinuity problem for circular variable (Hamill et al. 2012). The percentage of coordination that occurred during standard gait and max-kg gait cycles (bilateral) was analysed for each coordination, namely the in-phase (IP), pelvis only (PO), anti-phase (AP) and trunk-only (TO). Due to non-normal distribution of the percentage of coordination data, non-parametric tests were conducted with Bonferroni correction for type-I error. Mann-Whitney test was carried out to compare the percentage of coordination between sedentary and manual groups (i.e. between-group comparison). The type-I error for this test was adjusted to 0.025 for this test (p=0.05/2), as there were two possible comparisons which were during sedentary gait and max-kg gait. Wilcoxon signed-rank test was carried out to compare the percentage of coordination between standard gait and max-kg gait (i.e. within-group comparison). The type-I error for the test was also adjusted to 0.025 (p=0.05/2), as there were two possible comparisons, which were in the manual and the sedentary groups. Descriptive statistics were indicated as median (Mdn) and inter-quartile range (IqR). For both Mann Whitney test and Wilcoxon signed-rank test, the effect size were based on correlation coefficient (r): 0.1 (weak), 0.3 (moderate) and 0.5 (strong) (Rumsey and Unger 2015).

7.3 FINDINGS

7.3.1 Flexion-Extension Coordination

7.3.1.1 3D kinematics

To determine the effect of group and gait condition on the trunk (relative to pelvis), trunk global (relative to laboratory) and pelvis ROM in sagittal plane, split-plot analyses of variance (SPANOVA) were conducted (Table 7.1). According to the result, there was a significant effect of gait condition on the ROM of all segments. Regardless of groups, there was a significant increase in the trunk (right: 7.21º, left: 7.41º), trunk global (right: 0.61º, left: 0.65º) and pelvis (right: 6.65º, left: 5.63º) ROM. However, there was no significant effect of group on the ROM in both sides. The interaction effect between group and gait condition for both sides also indicating non-significant results.

Table 7.1. Effect of group and gait condition on the trunk relative to pelvis, trunk global and pelvis range of motion (ROM) in sagittal plane

Segments	**Group**	**Standard**	**Max-kg**	**Within-group**		**Between-group**		**Interaction**	
				p	**ηp^2**	***p***	**ηp^2**	***p***	**ηp^2**
Right trunk (relative to pelvis)	Sed. Man. Both	3.28±1.81 2.67±0.80 3.00±1.45	11.75±10.59 8.39±4.58 10.21±8.45	<0.001*	0.415	0.161	0.055	0.340	0.026
Right trunk global	Sed. Man. Both	1.63±0.64 1.23±0.48 1.44±0.60	2.07±1.45 2.02±1.10 2.05±1.28	0.018*	0.150	0.309	0.029	0.478	0.014
Right pelvis	Sed. Man. Both	3.21±1.79 2.69±0.80 2.97±1.43	11.43±10.36 7.48±3.17 9.62±8.07	<0.001*	0.403	0.099	0.076	0.210	0.044
Left trunk (relative to pelvis)	Sed. Man. Both	3.31±1.83 2.30±0.87 3.04±1.48	12.32±11.12 8.24±3.48 10.45±8.66	<0.001*	0.498	0.154	0.057	0.164	0.055
Left trunk global	Sed. Man. Both	1.53±0.59 1.22±0.40 1.39±0.53	2.05±1.24 2.04±1.06 2.04±1.14	0.002*	0.250	0.472	0.015	0.449	0.016
Left pelvis	Sed. Man. Both	3.10±1.94 2.80±0.66 2.97±1.48	9.20±4.29 7.89±3.73 8.60±4.04	<0.001*	0.631	0.258	0.036	0.487	0.014

7.3.1.2 Coordination

The mean of coupling angle (circular statistics) during standard gait and max-kg gait between sedentary (Figure 7.4) and manual (Figure 7.5) groups were plotted to generally visualize pelvis-trunk flexion-extension coordination. Further analysis of within-group and between-group comparisons were then conducted for each coordination.

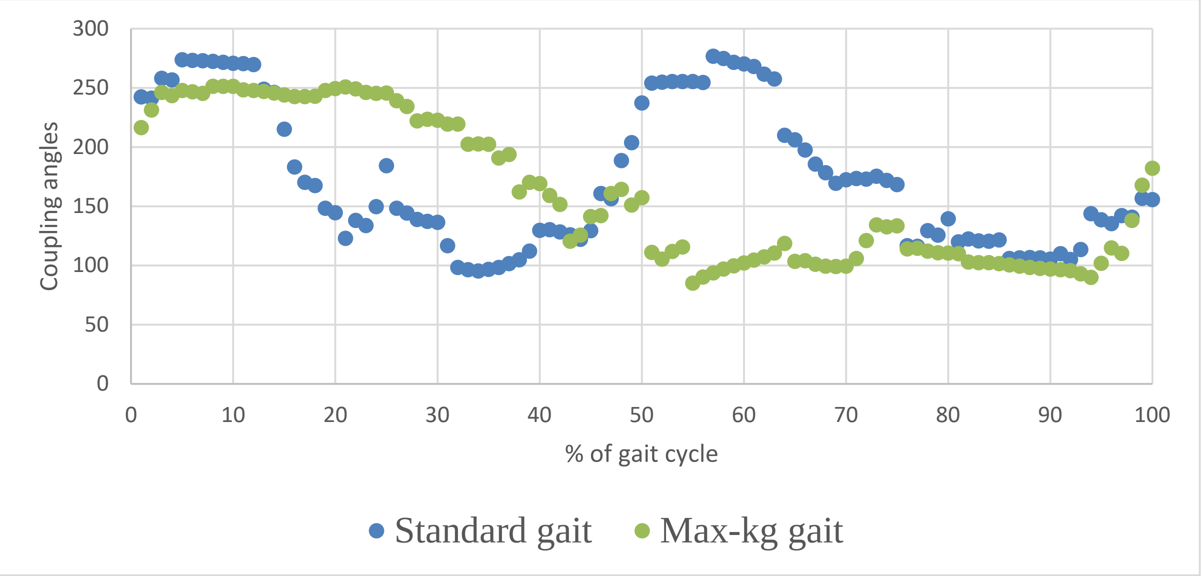


Figure 7.4. Pelvis-trunk flexion-extension coordination in sedentary group (right gait cycle)

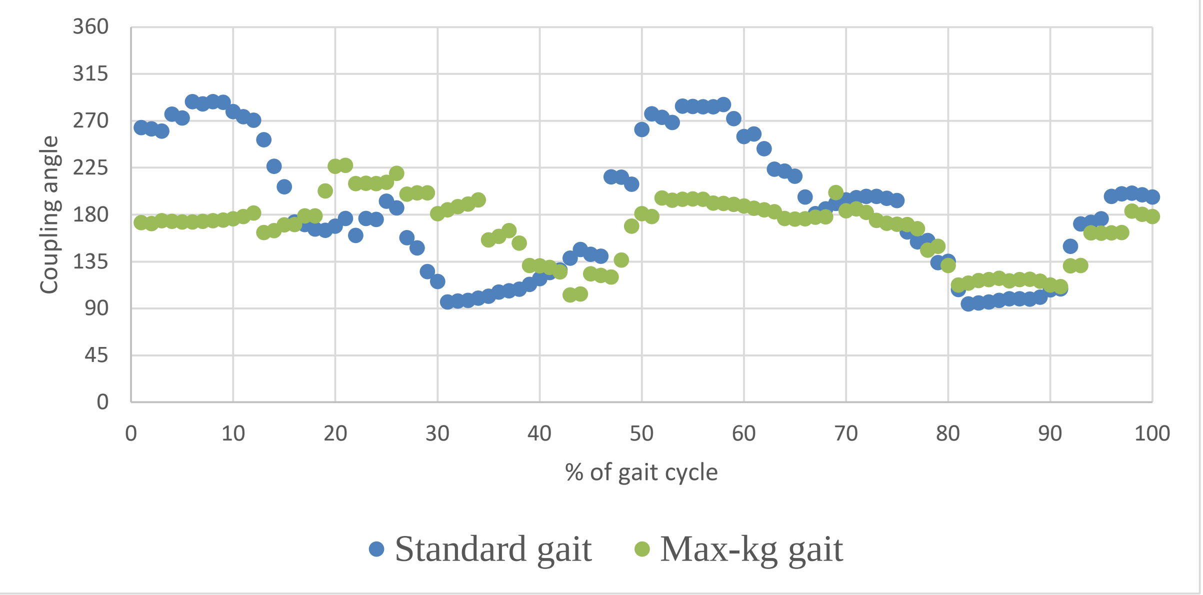


Figure 7.5. Pelvis-trunk flexion-extension coordination in manual group (right gait cycle)

a. In-phase

Significant increase in in-phase coordination (flexion-extension) from standard gait to max-kg gait can be observed only in the manual group for both sides (Table 7.2). There was no significant change in coordination from standard gait to max-kg gait in the sedentary group for both sides. Furthermore, there were no significant differences in coordination found in all between-group comparisons for both standard gait and max-kg gait.

Table 7.2. Comparison of in-phase coordination for flexion-extension based on the percentage of occurrence in a full gait cycle

Side	**Group**	**Standard**	**Max-kg**	**Within-group**		**Between-group**			
						Standard gait		**Max-kg gait**	
				p^1	r^1	p^2	r^2	p^2	r^2
Right	Sedentary	17.00 (10.50-21.75)	15.00 (10.25-21.75)	0.727	0.057	0.038	0.341	0.170	0.226
	Manual	7.00 (4.00-18.50)	19.00 (12.50-28.50)	0.014*	0.405				
Left	Sedentary	16.00 (7.00-23.00)	17.50 (10.25-25.50)	0.654	0.074	0.200	0.211	0.891	0.023
	Manual	12.00 (7.50-14.00)	22.00 (10.00-28.00)	0.014*	0.405				

1=Wilcoxon signed-rank test, 2=Mann Whitney test, *significant at $p<0.025$

b. Pelvis only

There was no significant difference in pelvis only coordination (flexion-extension) in all within-group and between group comparisons for both standard gait and max-kg gait (Table 7.3).

Table 7.3. Comparison of pelvis-only coordination for flexion-extension movement based on the percentage of occurrence in a full gait cycle

Side	Group	Standard	Max-kg	Within-group		Between-group			
						Standard gait		Max-kg gait	
				***p*[1]**	***r*[1]**	***p*[2]**	***r*[2]**	***p*[2]**	***r*[2]**
Right	Sedentary	48.00 (38.25-62.75)	50.00 (35.00-76.00)	0.751	0.052	0.048	0.326	0.161	0.231
	Manual	57.00 (50.50-66.50)	43.00 (25.50-54.50)	0.026	0.366				
Left	Sedentary	45.00 (26.75-56.00)	56.00 (38.50-71.75)	0.198	0.212	0.057	0.313	0.502	0.110
	Manual	55.00 (45.50-67.50)	46.00 (37.00-56.50)	0.224	0.200				

1=Wilcoxon signed-rank test, 2=Mann Whitney test

c. Anti-phase

Significant decrease in anti-phase coordination (flexion extension) from standard gait to max-kg gait can only be observed in the left side, while there was no significant changes in the right anti-phase coordination (Table 7.4). Furthermore, there was no significant difference in the right anti-phase coordination. Other than that, there was no significant difference in coordination found in all between-group comparisons for both standard gait and max-kg gait.

Table 7.4. Comparison of anti-phase coordination for flexion-extension based on the percentage of occurrence in a full gait cycle

Side	Group	Standard	Max-kg	Within-group		Between-group			
						Standard gait		Max-kg gait	
				p^1	r^1	p^2	r^2	p^2	r^2
Right	Sedentary	20.50 (30.50-14.00)	20.50 (9.50-30.50)	0.433	0.129	0.669	0.070	0.217	0.203
	Manual	28.00 (11.50-31.00)	24.00 (19.00-34.00)	0.586	0.090				
Left	Sedentary	25.50 (12.50-38.75)	17.50 (6.25-22.75)	0.016*	0.396	0.483	0.115	0.410	0.135
	Manual	25.00 (15.00-28.00)	17.00 (13.00-28.00)	0.587	0.089				

1=Wilcoxon signed-rank test, 2=Mann Whitney test, * significant at $p<0.025$

d. Trunk only

There was no significant difference in trunk only coordination (flexion extension) in all within-group and between-group comparisons for both standard gait and max-kg gait (Table 7.5).

Table 7.5. Comparison of trunk-only coordination for flexion-extension based on the percentage of occurrence in a full gait cycle

Side	**Group**	**Standard**	**Max-kg**	**Within-group**		**Between-group**			
						Standard gait		**Max-kg gait**	
				p[1]	***r***[1]	***p***[2]	***r***[2]	***p***[2]	***r***[2]
Right	Sedentary	10.50 (5.00-17.50)	7.50 (3.25-15.50)	0.643	0.076	0.126	0.251	0.963	0.008
	Manual	7.00 (4.00-10.00)	7.00 (3.50-16.00)	0.569	0.094				
Left	Sedentary	10.00 (4.50-17.75)	6.00 (3.25-18.00)	0.765	0.049	0.691	0.065	0.725	0.058
	Manual	7.00 (4.50-14.00)	11.00 (3.50-14.50)	0.740	0.055				

1=Wilcoxon signed-rank test, 2=Mann Whitney test

7.3.2 Lateral Flexion Coordination

7.3.2.1 3D kinematics

To determine the effect of group and gait condition on the trunk (relative to pelvis), trunk global (relative to laboratory) and pelvis ROM in frontal plane, split-plot analyses of variance (SPANOVA) were conducted (Table 7.6). According to the result, the ROM was significantly increased from standard gait to max-kg gait for the trunk global (right: 1.38º, left: 1.25º) but significantly decreased for the trunk (right: 2.36º, left: 1.25º). Furthermore, as indicated by a significant interaction effect in the trunk global, the increase in ROM was more obvious in the manual group compared to the sedentary group (Figure 7.6). There was no significant effect of gait condition on the pelvis ROM.

Table 7.6. Effect of group and gait condition on the trunk relative to pelvis, trunk global and pelvis range of motion (ROM) in frontal plane

Segments	Group	Standard	Max-kg	Within-group		Between-group		Interaction	
				p	**ηp^2**	***p***	**ηp^2**	***p***	**ηp^2**
Right trunk (relative to pelvis)	Sed. Man. Both	9.68±2.51 10.35±2.68 9.99±2.57	7.22±2.87 8.13±2.68 7.63±2.78	0.001*	0.253	0.174	0.052	0.861	0.001
Right trunk global	Sed. Man. Both	1.74±0.68 1.87±0.69 1.80±0.68	2.73±1.11 3.71±1.48 3.18±1.35	<0.001*	0.587	0.049*	0.106	0.042*	0.112
Right pelvis	Sed. Man. Both	6.80±2.15 6.87±2.24 6.83±2.16	6.84±2.89 8.36±3.32 7.54±3.14	0.286	0.033	0.143	0.060	0.316	0.029
Left trunk (relative to pelvis)	Sed. Man. Both	9.83±2.95 10.51±2.55 10.14±2.76	7.28±2.77 8.18±2.90 7.70±2.83	0.001*	0.278	0.230	0.041	0.870	0.001
Left trunk global	Sed. Man. Both	1.87±0.75 1.82±0.54 1.85±0.66	2.87±1.09 3.36±1.43 3.10±1.26	<0.001*	0.484	0.390	0.021	0.232	0.040
Left pelvis	Sed. Man. Both	7.03±2.15 6.94±2.24 6.88±2.16	6.92±3.04 8.38±3.46 7.59±3.28	0.372	0.023	0.213	0.044	0.297	0.031

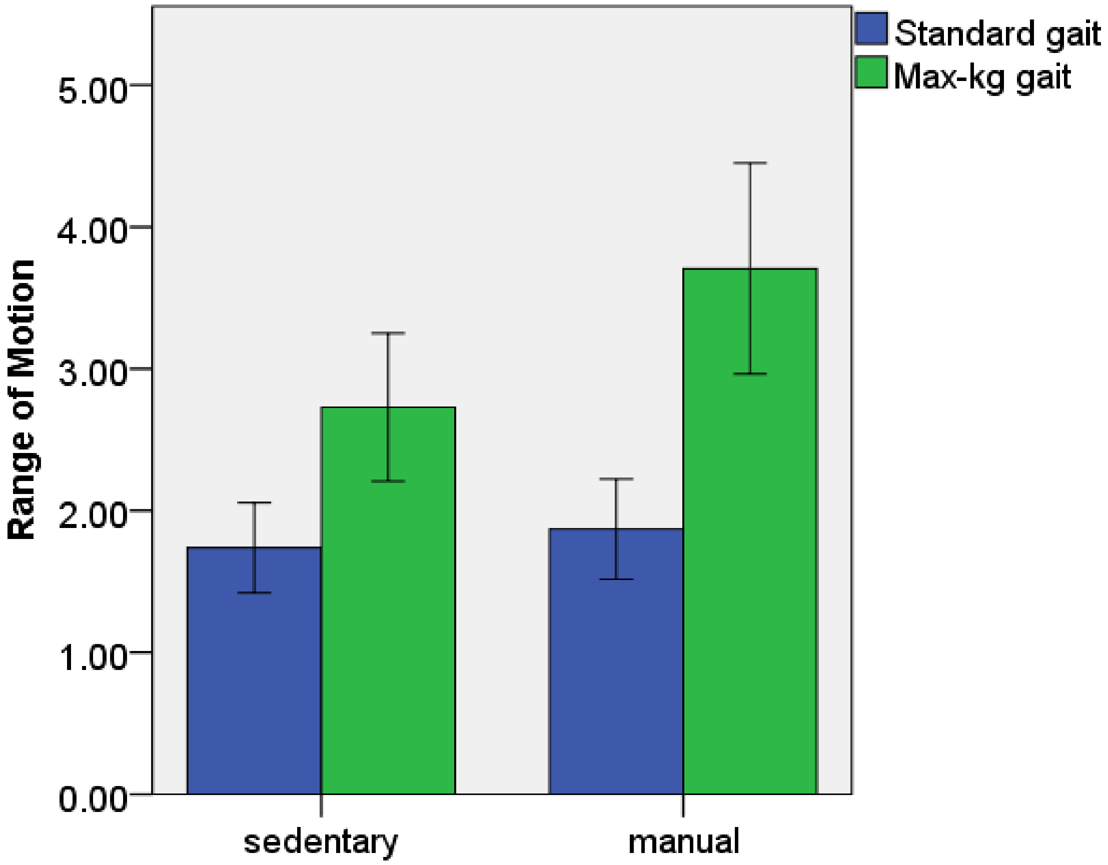


Figure 7.6. Difference in the trunk (global) range of motion between sedentary and manual groups (error bars represent 95% confidence interval)

7.3.2.2 Changes in specific coordination

The mean of coupling angle (circular statistics) during standard gait and max-kg gait between sedentary (Figure 7.7) and manual (Figure 7.8) groups were plotted to generally visualize the pelvis-trunk flexion-extension coordination. Further analysis of within-group and between-group comparisons were then conducted for each coordination.

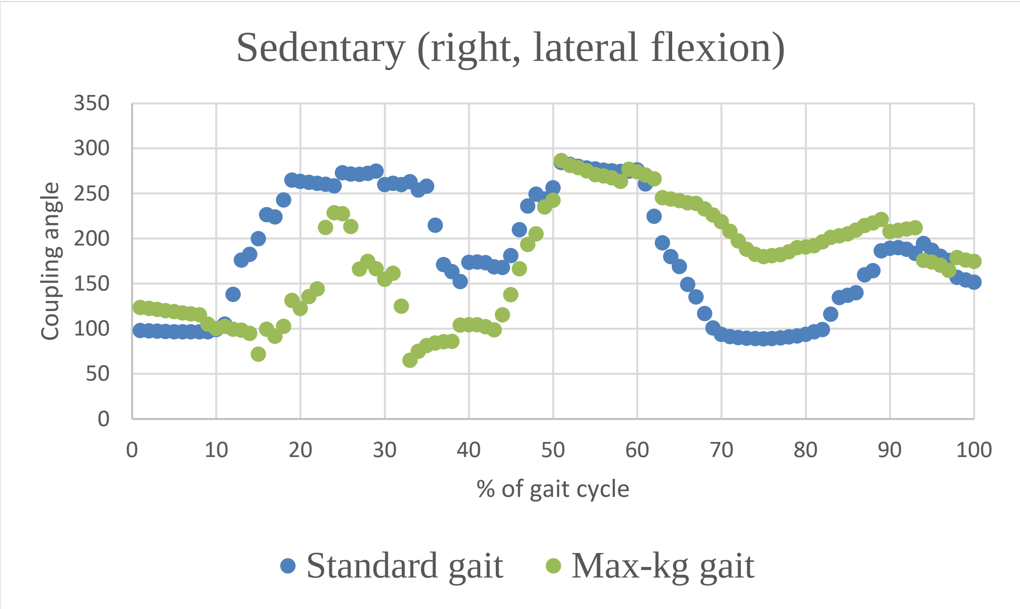


Figure 7.7. Pelvis-trunk lateral flexion coordination in sedentary group (right gait cycle)

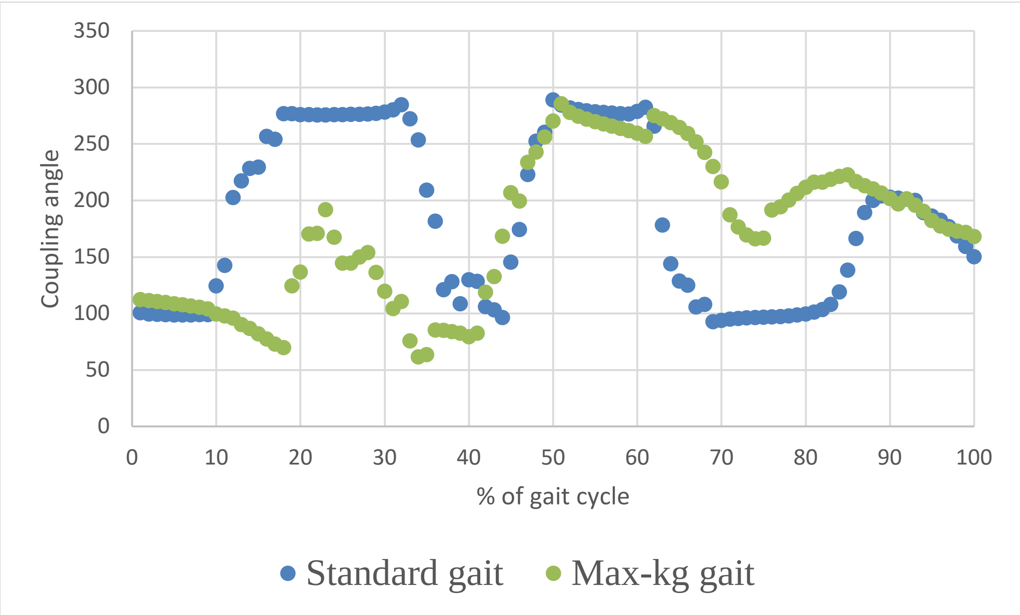


Figure 7.8. Exemplar pelvis-trunk lateral flexion coordination in manual group (right gait cycle)

a. In-phase

Significant increase in in-phase coordination from standard gait to max-kg gait can only be observed in the left side, while there was no significant difference found in the right anti-phase coordination (Table 7.7). Other than that, there was no significant difference in the coordination of all between-group comparisons for both standard gait and max-kg gait. Therefore, in general, it can be concluded that there was an obvious increase in the in-phase, lateral flexion coordination in most of the participants.

Table 7.7. Comparison of in-phase coordination for lateral flexion based on the percentage of occurrence in a full gait cycle

Side	**Group**	**Standard**	**Max-kg**	**Within-group**		**Between-group**			
						Standard gait		**Max-kg gait**	
				p^1	r^1	p^2	r^2	p^2	r^2
Right	Sedentary	11.00 (6.00-17.75)	21.00 (15.25-30.75)	0.013*	0.408	0.562	0.095	0.976	0.005
	Manual	10.00 (4.50-16.00)	21.00 (11.00-37.00)	0.002*	0.514				
Left	Sedentary	11.50 (8.00-19.00)	19.00 (10.25-32.50)	0.044	0.331	0.760	0.050	0.879	0.025
	Manual	14.00 (7.00-18.50)	19.00 (13.00-27.00)	0.024*	0.370				

1=Wilcoxon signed-rank test, 2=Mann Whitney test, * significant at $p<0.025$

b. Pelvis only

Significant decrease in pelvis only coordination (lateral flexion) from standard gait to max-kg gait can be observed in all within-group comparisons for both standard gait and max-kg gait (Table 7.8).

Table 7.8. Comparison of pelvis only coordination for lateral flexion based on the percentage of occurrence in a full gait cycle

Side	Group	Standard	Max-kg	Within-group		Between-group			
						Standard gait		Max-kg gait	
				***p*[1]**	***r*[1]**	***p*[2]**	***r*[2]**	***p*[2]**	***r*[2]**
Right	Sedentary	68.00 (60.50-78.75)	43.00 (25.00-49.50)	0.001*	0.562	0.772	0.048	0.726	0.058
	Manual	69.00 (61.50-81.00)	39.00 (24.50-49.00)	0.001*	0.595				
Left	Sedentary	67.00 (58.50-76.00)	39.50 (23.00-53.75)	0.002*	0.522	0.784	0.045	0.726	0.058
	Manual	66.00 (59.50-78.00)	47.00 (31.00-52.50)	0.001*	0.596				

1=Wilcoxon signed-rank test, 2=Mann Whitney test, * significant at $p<0.025$

c. Anti-phase

Significant increase in anti-phase coordination (lateral flexion) from standard gait to max-kg gait can be observed only in manual group for both sides (Table 7.9). There was no significant change in coordination from standard gait to max-kg gait found in the sedentary group for both sides. Furthermore, there was no significant difference in the coordination found in all between-group comparisons for both standard gait and max-kg gait.

Table 7.9. Comparison of anti-phase coordination for lateral flexion based on the percentage of occurrence in a full gait cycle

Side	**Group**	**Standard**	**Max-kg**	**Within-group**		**Between-group**			
						Standard gait		**Max-kg gait**	
				p[1]	***r***[1]	***p***[2]	***r***[2]	***p***[2]	***r***[2]
Right	Sedentary	14.50 (7.25-18.00)	17.00 (10.00-28.00)	0.100	0.270	0.783	0.045	0.562	0.095
	Manual	13.00 (7.50-17.50)	21.00 (16.00-21.00)	0.009*	0.433				
Left	Sedentary	13.00 (6.75-21.00)	23.00 (17.00-29.50)	0.036	0.344	0.352	0.153	0.988	0.003
	Manual	9.00 (5.50-19.00)	19.00 (16.50-38.50)	0.002*	0.518				

1=Wilcoxon signed-rank test, 2=Mann Whitney test, * significant at $p<0.025$

d. Trunk only

Significant decrease in trunk only coordination (lateral flexion) can be observed in all within-group and between-group comparisons in both standard gait and max-kg gait (Table 7.10).

Table 7.10. Comparison of anti-phase coordination for lateral flexion based on the percentage of occurrence in a full gait cycle

Side	**Group**	**Standard**	**Max-kg**	**Within-group**		**Between-group**			
						Standard gait		**Max-kg gait**	
				p^{1}	***r***^{1}	***p***^{2}	***r***^{2}	***p***^{2}	***r***^{2}
Right	Sedentary	5.00 (2.00-7.75)	14.00 (8.25-20.75)	0.001*	0.562	0.580	0.091	0.784	0.045
	Manual	5.00 (5.00-5.00)	16.00 (6.00-23.00)	0.003*	0.491				
Left	Sedentary	5.00 (2.00-8.75)	14.00 (10.25-18.50)	0.001*	0.525	0.914	0.018	0.951	0.010
	Manual	5.00 (2.00-6.00)	8.50 (12.00-24.00)	0.001*	0.530				

1=Wilcoxon signed-rank test, 2=Mann Whitney test, *significant at $p<0.05$

7.3.3 Axial Rotation Coordination

7.3.3.1 3D kinematics

To determine the effect of group and gait condition on the trunk (relative to pelvis), trunk global (relative to laboratory) and pelvis ROM in transverse plane, split-plot analyses of variance (SPANOVA) were conducted (Table 7.11). According to the result, there was a significant decrease of ROM in the trunk (right: 7.33º, left: 7.56º) and pelvis (right: 5.70º, left: 5.74º). There was no significant effect of gait condition on the trunk global ROM. For all segments, there was no significant effect of group on the ROM, as well as no significant interaction effect between group and gait condition.

Table 7.11. Effect of group and gait condition on the trunk relative to pelvis, trunk global and pelvis range of motion (ROM) in transverse plane

Segments	**Group**	**Standard**	**Max-kg**	**Within-group**		**Between-group**		**Interaction**	
				p	**ηp²**	***p***	**ηp²**	***p***	**ηp²**
Right trunk relative to pelvis	Sed. Man. Both	14.50±4.03 16.47±3.95 15.40±4.06	8.74±3.31 7.28±3.24 8.07±9.93	<0.001*	0.348	0.890	0.001	0.329	0.027
Right trunk global	Sed. Man. Both	6.53±2.79 7.99±3.42 7.20±3.14	8.63±7.44 8.49±3.51 8.57±5.89	0.176	0.052	0.604	0.008	0.403	0.020
Right pelvis	Sed. Man. Both	10.51±3.30 11.72±3.68 11.07±3.49	4.73±1.72 6.12±2.59 5.37±2.24	<0.001*	0.666	0.060	0.097	0.898	<0.001
Left trunk relative to pelvis	Sed. Man. Both	14.70±4.40 16.06±3.84 15.32±4.15	8.48+11.52 6.91±3.07 7.76±8.65	<0.001*	0.418	0.949	<0.001	0.346	0.025
Left trunk global	Sed. Man. Both	6.57±2.87 7.71±3.39 7.09±3.13	6.73±2.54 8.30±3.42 7.45±3.04	0.413	0.019	0.139	0.061	0.641	0.006
Left pelvis	Sed. Man. Both	10.94±3.31 11.27±3.73 11.09±3.46	4.93±1.68 5.84±2.25 5.35±1.99	<0.001*	0.687	0.363	0.024	0.664	0.005

7.3.3.2 Changes in coordination

The mean of coupling angle (circular statistics) during standard gait and max-kg gait between sedentary (Figure 7.9) and manual (Figure 7.10) groups were plotted to generally visualize pelvis-trunk axial rotation coordination. Further analysis of within-group and between-group comparisons were then conducted for each coordination.

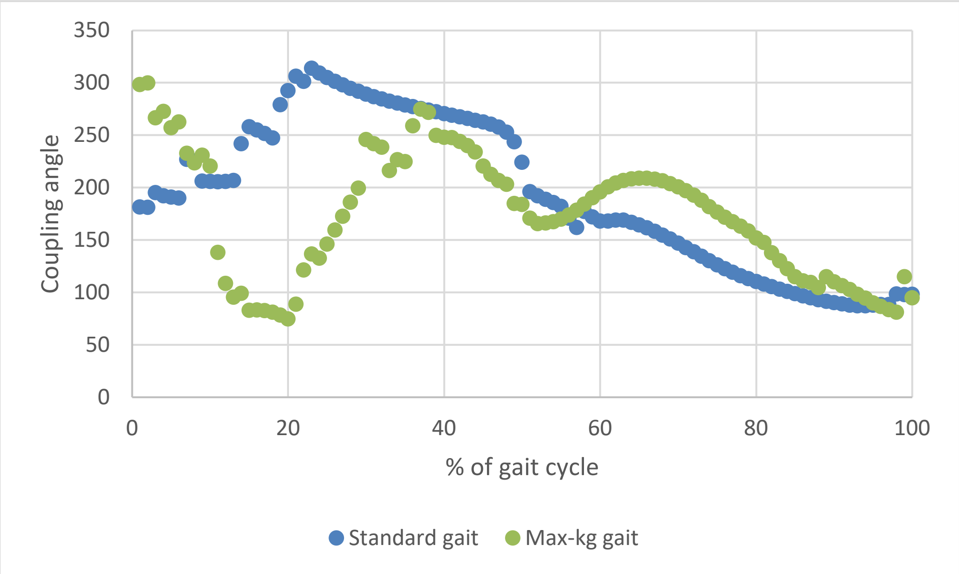


Figure 7.9. Exemplar pelvis-trunk axial rotation coordination in sedentary group (right gait cycle)

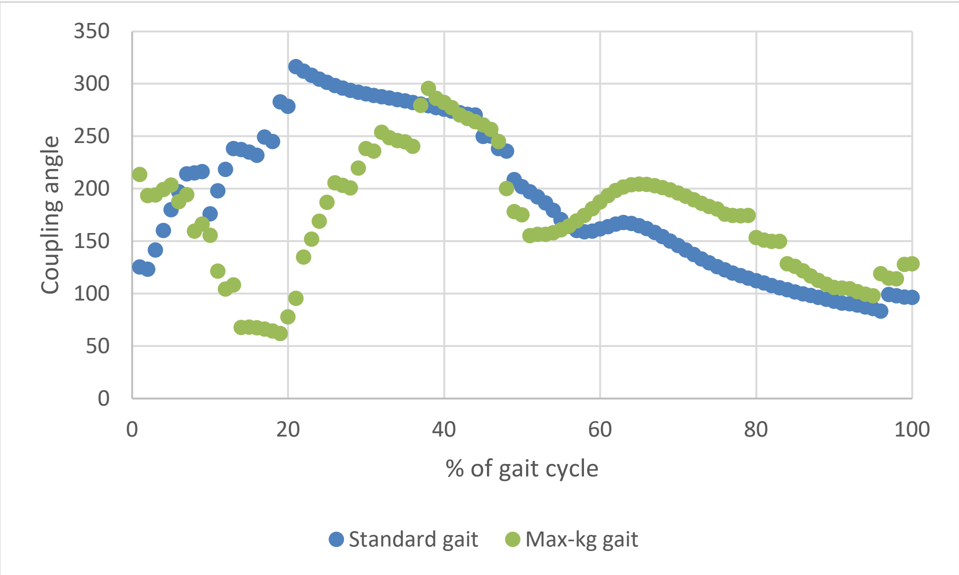


Figure 7.10. Exemplar pelvis-trunk axial rotation coordination in manual group (right gait cycle)

a. In-phase

Significant increase in in-phase coordination (axial rotation) from standard gait to max-kg gait can only be observed for the right side in both manual and sedentary groups (Table 7.12). There was no significant change from standard gait to max-kg gait for left anti-phase coordination in both groups. There was no significant difference in coordination found in all between-group comparisons.

Table 7.12. Comparison of in-phase coordination for axial rotation based on the percentage of occurrence in a full gait cycle

Side	**Group**	**Standard**	**Max-kg**	**Within-group**		**Between-group**			
						Standard gait		**Max-kg gait**	
				p^1	r^1	p^2	r^2	p^2	r^2
Right	Sedentary	7.00 (5.00-16.00)	27.50 (19.50-40.50)	0.001*	0.533	0.842	0.033	0.625	0.080
	Manual	9.00 (4.50-16.00)	35.00 (18.50-38.00)	0.001*	0.572				
Left	Sedentary	28.00 (19.25-30.75)	19.00 (13.25-25.00)	0.035	0.348	0.831	0.035	1.000	0.000
	Manual	24.00 (19.50-38.50)	20.00 (10.50-24.00)	0.177	0.222				

1=Wilcoxon signed-rank test, 2=Mann Whitney test, *significant at $p<0.025$

b. Pelvis only

Significant decrease in pelvis-only coordination (axial rotation) from standard gait to max-kg gait can be observed in all within-group comparisons for both standard gait and max-kg gait (Table 7.13). There was no significant difference in coordination in all between-group comparisons.

Table 7.13. Comparison of pelvis only coordination for axial rotation based on the percentage of occurrence in a complete gait cycle

Side	**Group**	**Standard**	**Max-kg**	**Within-group**		**Between-group**			
						Standard gait		**Max-kg gait**	
				p^1	r^1	p^2	r^2	p^2	r^2
Right	Sedentary	44.00 (32.25-54.75)	18.50 (13.00-27.25)	0.001*	0.544	0.807	0.040	0.156	0.233
	Manual	48.00 (23.00-56.00)	11.00 (7.50-23.50)	0.001*	0.588				
Left	Sedentary	43.50 (29.25-57.50)	21.00 (9.00-29.75)	0.001*	0.608	0.322	0.163	0.360	0.150
	Manual	42.00 (27.00-53.50)	14.00 (9.50-23.00)	0.001*	0.561				

1=Wilcoxon signed-rank test, 2=Mann Whitney test, *significant at p<0.025

c. Anti-phase

Significant increase in the anti-phase coordination from standard gait to max-kg gait can only be observed in the left side (axial rotation) in both manual and sedentary groups (Table 7.14). There was no significant change from standard gait to max-kg gait for the right anti-phase coordination in both groups, as well as no significant difference in coordination found in all between-group comparisons.

Table 7.14. Comparison of anti-phase coordination for axial rotation based on the percentage of occurrence in a full gait cycle

Side	**Group**	**Standard**	**Max-kg**	**Within-group**		**Between-group**			
						Standard gait		**Max-kg gait**	
				p^1	r^1	p^2	r^2	p^2	r^2
Right	Sedentary	25.50 (24.00-35.50)	18.00 (14.50-25.75)	0.030	0.357	0.409	0.136	0.542	0.100
	Manual	24.00 (19.00-41.00)	21.00 (12.50-31.50)	0.434	0.129				
Left	Sedentary	8.00 (6.00-12.75)	33.00 (25.00-44.75)	0.001*	0.605	0.582	0.090	0.573	0.093
	Manual	10.00 (5.00-16.00)	34.00 (17.50-42.50)	0.002*	0.518				

1=Wilcoxon signed-rank test, 2=Mann Whitney test, significant at $p<0.025$

d. Trunk only

Significant increase in the trunk only coordination from standard gait to max-kg gait be observed in the left trunk only coordination (Table 7.15). However, only the right trunk in the sedentary group had significant change in coordination across the gait conditions, while no significant change can be observed in manual group. There was no significant differences in coordination found in all between-group comparisons.

Table 7.15. Comparison of trunk-only coordination for axial rotation based on the percentage of occurrence in full gait cycle

Side	**Group**	**Standard**	**Max-kg**	**Within-group**		**Between-group**			
						Standard gait		**Max-kg gait**	
				p[1]	***r***[1]	***p***[2]	***r***[2]	***p***[2]	***r***[2]
Right	Sedentary	16.00 (8.00-22.50)	30.00 (19.50-30.00)	0.001*	0.599	0.393	0.140	0.915	0.018
	Manual	19.00 (8.00-30.50)	30.00 (23.00-39.00)	0.028	0.362				
Left	Sedentary	15.00 (9.00-23.50)	24.00 (16.50-34.00)	0.001*	0.559	0.502	0.110	0.246	0.191
	Manual	16.00 (9.00-30.00)	29.00 (20.50-43.50)	0.006*	0.455				

1=Wilcoxon signed-rank test, 2=Mann Whitney test, *significant at p<0.025

7.3.4 Summary of Changes in Pelvis-Trunk Coordination

The findings for pelvis-trunk coordination changes from standard gait to max-kg gait in manual workers and sedentary group were summarized (Table 7.16). In general, in-phase coordination was significantly increased in flexion extension (manual only), lateral flexion (sedentary-right only) and axial rotation (right only) planes of movement. Pelvis-only coordination was significantly decreased for the lateral flexion and axial rotation planes of movement. For anti-phase coordination, a significant decrease was found in the flexion-extension (sedentary-left only) and axial rotation planes of movement. However, a significant increase was found in the lateral flexion (manual group only) for the anti-phase coordination. For trunk-only coordination, a significant increase was found in the lateral flexion and axial rotation (except manual-right)

Table 7.16. Summary of changes pelvis-trunk coordination from standard gat to max-kg gait

Angle	**Coordination**	**Manual**	**Sedentary**
Flexion-extension	In-phase	Increased*	NS
	Pelvis-only	NS	NS
	Anti-phase	NS	Decreased (left only)*
	Trunk-Only	NS	NS
Lateral flexion	In-phase	Increased*	Increased (right only)*
	Pelvis-only	Decreased*	Decreased*
	Anti-phase	Increased*	NS
	Trunk-Only	Increased*	Increased*
Axial rotation	In-phase	Increased (right only)*	Increased (right only)*
	Pelvis-only	Decreased*	Decreased*
	Anti-phase	Decreased (left only)	Decreased (left only)*
	Trunk-Only	Increased (left only)	Increased

Significant at p<0.025 (Bonferroni correction)
NS = Non-significant

7.4 DISCUSSION

It was hypothesized that the changes in the pelvis-trunk coordination will occur across the load carriage activity to reflect a motor control strategy in adapting to a fatiguing situation. For flexion-extension movement, the range of motion (ROM) was increased for trunk (relative to pelvis), trunk global (relative to laboratory) and pelvis. A generally similar pattern of ROM changes in all segments may indicate minimal changes in the pelvis-trunk coordination. It can be observed in the sagittal plane that only in-phase coordination was significantly increased for both left and right in manual workers. In order to carry more loads than the sedentary group, an increased in-phase coordination can be observed in the manual group, indicating both trunk and pelvis were rotating synchronously in the same direction more frequently. The anterior pelvic tilt was known to have a mechanical association with increased lumbar lordosis (Levine and Whittle 1996). In this study, the anterior pelvic tilt may increase to correspond with trunk extension in order to counter-act the carrying load's moment inertia during the activity. If the anterior pelvic tilt become exaggerated, it may lead to low back pain (LBP) due to an excessive stress on the sacroiliac joints and lumbar spine (Denslow and Chace 1962; Minicozzi et al. 2016).

For lateral flexion movement, the kinematics indicated that there was a significant decrease in the trunk ROM, but significant increase in the trunk global ROM. There was no significant change in pelvic obliquity ROM across the activity. The inconsistent findings between the trunk and the trunk global could possibly be due to the influence of pelvic orientation during the carrying activity. An in-phase coordination would indicate a synchronous dropping of the pelvis of the same side as the trunk lateral flexion. If the trunk lateral flexion occurred whilst the contra-lateral pelvis was dropping (i.e. opposite pelvic obliquity), this may increase the degree of the trunk lateral flexion. Therefore, the significant decrease in the trunk lateral flexion ROM relative to pelvis can possibly indicate that the synchronous opposite pelvic obliquity was decreased in max-kg gait, leading to a lower trunk lateral flexion ROM. It was also found in the frontal plane that there was a significant increase in the trunk-only coordination, but significant decrease in the pelvis-only coordination for the lateral flexion movement. In other words, carrying a maximum load tend to restrict the movement of pelvic obliquity whilst allowing the trunk lateral flexion, which supports the notion of descriptive reduced pelvic obliquity during max-kg gait. However, it was unexpected to discover that both in-phase and anti-phase coordination were significantly increased during max-kg gait in the manual workers. Those changes may occur in two different periods along the gait cycle. For instance, during initial contact (0% to 2% of gait cycle) and loading response (2% to 12% of gait cycle) of the stance phase (Perry and Davids 2010), the anti-phase coordination was prevalent. During this period, the pelvic obliquity was increasing but the trunk lateral flexion was reducing possibly in order to maintain initial limb stability by promoting an upright body posture. However, during mid swing (75% to 87%) and terminal swing (87% to 100% of gait cycle) (Perry and Davids 2010), both pelvic obliquity and trunk lateral flexion were decreasing to allow a complete the limb advancement. The increase in the in-phase coordination in the frontal plane can also be observed in the LBP patients that may represent a guarded activity to prevent from further injury (Seay et al. 2011c).

For axial rotation, pelvis ROM was significantly reduced whilst carrying a maximum load, as well as the trunk (relative to pelvis) ROM. However, the significant reduction in the ROM cannot be observed in the trunk global ROM. A significant decrease was also found in the pelvis-only coordination during max-kg gait. According to the movement waveform (see 6.3.2.4 for detail), it can be observed that the pelvis rotation in the transverse plane was less variable whilst carrying the maximum load compared to the standard gait. This study suggested that the pelvic axial rotation ROM was significantly reduced in order to minimize the step length. As the gait frequency increased, the body can spend more time in the double stance cumulatively from each gait cycle that completed the carrying activity. Studies had reported that a minimized stride length was important in order to increase gait stability (You et al. 2001; Cham and Redfern 2002; Cromwell and Newton 2004; Espy et al. 2010). Furthermore, it was also found that the maximum load had led to a significant increase in the trunk-only coordination. From this study, it can be suggested that the trunk-only movement assisted the change into a more in-phase coordination during the max-kg gait. During the max-kg gait, the pelvis may produce a smaller amount of axial rotation as the stride length was shortened. However, the trunk axial rotation may still occur in order to compensate with the absence of arm swing. Consequently, when the pelvis started to rotate in order to allow forward progression, the trunk-only rotation will change into the in-phase coordination.

The change in coordination can also be associated with the upper limb muscle fatigue around the shoulder and elbow joints (Dedieu and Zanone 2012). In a standard gait, the pelvis rotation was concurrently balanced by the contralateral arm swing in order to optimize gait stability and minimize energy consumption (also known as the anti-phase pelvis-shoulder inter-girdle coordination) (Meyns et al. 2013). For instance, whilst the right trunk and leg move forward, the left arm will coincide to move forward. However, the arm swing was completely restricted during the carrying activity in order to hold the load. The restricted arm swing may increase metabolic consumption of gait due to the need to counterbalance the mediolateral momentum (Herr and Popovic 2008; Collins et al. 2009; Yizhar et al. 2009), as well as to counterbalance the vertical movement of the trunk (i.e. downward translation of the trunk is usually balanced by upward direction of arm swing) (Park 2008). It was also reported that a restricted arm swing can induce a more in-phase pelvis-shoulder coordination in both walking and running (Dedieu and Zanone 2012). Furthermore, the restricted arm swing was reported to accompany by the activation of latissimus dorsi, teres major and deltoid muscles (Ballesteros et al. 1965; Kuhtz-Buschbeck and Jing 2011). Therefore, it was assumed that the muscle activation of the aforementioned muscles had occurred in order to compensate with the loss of pelvis-shoulder coordination.

In this study, it was found that while the right in-phase coordination was significantly increased, the left anti-phase was significantly reduced during axial rotation. This can be due to the fatigue on the muscle that controls the movement. For instance, in an isometric fatiguing axial rotation, Ng et al. (2002) reported a significantly higher right external oblique muscle activity during left axial rotation in the back pain group compared to the healthy group. However, there was no significant increase in the left external oblique muscle activity during the right axial rotation. As the right external oblique controlled contralateral trunk rotation, it was assumed that the guarded activity of the muscle decreases the occurrences of anti-phase coordination for the left side. In low back pain patients, studies had reported an increased pattern in muscle activity in conjunction with a lesser muscle relaxation (Van der Hulst et al. 2010). Interestingly, this phenomenon could also lead to decreased ROM, hence, the development of the guarding phenomenon (Van der Hulst et al. 2010). Therefore, this may suggest that the fatigued muscle could prevent the eccentric contraction of the right external oblique, which can prevent the action of the left internal oblique from performing a smooth and optimal contralateral axial trunk rotation.

Chapter 8: GENERAL DISCUSSION

8.1 STUDY NOVELTY: THE DEVELOPMENT AND UTILITY OF CARRYING PROTOCOL AS A COMPONENT OF FUNCTIONAL CAPACITY EVALUATION

An innovative aspect from this PhD study is the application of a novel carrying protocol that involves progressive increase in load to assess carrying ability based on anterior load carriage. In general, the studies on carrying activity can be divided into biomechanics and clinical (Hanif Farhan et al. 2015). For biomechanics studies, the load that is used during the carrying activity is based either on specific work equipment (e.g. weaponry and military vest) or according to percentage of body mass. According to the literature, most of the clinical studies related to Functional Capacity Evaluation (FCE) aimed to investigate the reliability of the protocol (Gross and Battié 2002; Reneman et al. 2002; Brouwer et al. 2003; Gouttebarge et al. 2004; Reneman et al. 2004) or to determine the patients' functional capacity to return-to-work (Gross et al. 2004; Gross and Battié 2005; Wind et al. 2006). Although the current study is laboratory-based, the carrying activity protocol is tailored to suit the clinical purpose of the FCE. Furthermore, to the researcher's knowledge, this study is the only study that utilized a comprehensive biomechanical analysis of an FCE protocol. Therefore, the findings from this study can be used as the basis of scientific reasoning to justify the importance FCE in clinical setting.

There are several advantages of using a progressive load increment. First, the use of progressive load increment is important to determine the maximum load limit that can be used as a guideline to propose the extent of physical load the patient could handle after returning to work. As the loads are standardized, the test should improve the reliability of FCE as a measurement of readiness for return to work. In reality, the patients may not necessarily return to their previous job or employer, depending on their functional capacity. In other words, the patients can be either returning to their same, similar or new job, as well as either to a same or different employer. In the cases where patients are unable to fully return into his/her previous job due, an indication of safe maximum load limit is highly important to any human resource department in order to search for available jobs that suit the patient's need. Other than that, a standardized FCE report also important for any work insurance company to evaluate for reimbursement eligibility and/or to suggest other necessary actions.

As a profession that utilizes human occupation as a medium of rehabilitation, occupational therapy purports that an active engagement in occupation is essential for mood, health status and quality of life (Graff et al. 2007). For occupational therapists, knowledge of biomechanics is imperative to explain how a particular movement is executed in order to plan for appropriate targeted intervention. The profession is pursuing more objective and measurable means of describing practice (Kielhofner 2009). The reliability of the carrying protocol has also been reported in this study (see Chapter 4 for more details). The musculoskeletal biomechanics of carrying are the major parameters in this study, and the findings has provided comprehensive movement descriptions of anterior load carriage. Therefore, the use of carrying activity as a tool for therapy can be categorized as a part of biomechanical strategies of rehabilitation.

8.2 SYNTHESIS OF FINDINGS: COMPENSATORY STRATEGIES DURING ANTERIOR LOAD CARRIAGE

In general, the changes in muscle activity, spatiotemporal parameters and 3D kinematics of gait and pelvis-trunk coordination during anterior load carriage can be considered as a musculoskeletal strategy to maintain gait stability. The current study reveals a significant increase in gait stability ratio (GSR) throughout the anterior load carriage, indicating more steps were taken per unit of distance without increasing the walking speed in order to allow a more frequent stance phase to maintain balance. This finding was supported by significant increase in cadence but significant decrease in the stride length. As the walking speed was kept constant throughout the carrying activity, a general increase in the in-phase coordination was observed possibly in order to protect the lumbo-pelvic structure. Unlike in a higher walking speed (e.g. running), the in-phase coordination during standard gait may shift to an anti-phase coordination in healthy individuals, particularly in the transverse plane (Seay et al. 2011b). For instance, the trunk usually rotates towards the opposite site of the pelvic rotation during running. This action was to ensure that the angular momentum from the pelvis can be stabilized in order to maintain a dynamic balance (Herr and Popovic 2008). Therefore, that study suggested that an increase in gait frequency is generally associated with a more frequent occurrence of in-phase pelvis-trunk coordination during anterior load carriage.

This study found that the manual workers can significantly carry a heavier load compared to the sedentary group. As both groups consisted of healthy individuals, the findings from this study can be utilized to gain an empirical understanding on how the general population compensate with the fatigue due to a progressively loaded anterior load carriage. In this study, the compensatory strategies were grouped into two categories, namely functional leg preference and guarding mechanism. For the functional leg preference, there were two important findings from this study that can explain the impact of anterior load carriage on motor control, which were the unilateral localization of leg muscle fatigue and also the unilateral reduction in range of motion (ROM) of the leg. The results gave an insight on the preferable use of the left leg for stability and postural control, whilst the right leg was more preferable for mobility and manipulation (Spry et al. 1993; Velotta et al. 2011). For muscle fatigue, it can be observed that the manual workers had a significantly lower rate of muscle fatigue for the left vastus lateralis but a significantly higher rate of muscle fatigue for the right gastrocnemius compared to the sedentary individuals. The vastus lateralis is responsible for knee extension and stabilization and the gastrocnemius is responsible to generate the propulsive forces to accelerate the leg into swing phase (Kepple et al. 1997; Meinders et al. 1998; Neptune et al. 2001; Saladin 2007; Perry and Burnfield 2010; Silder et al. 2013). Therefore, the findings from this study suggested that the manual workers use of left leg to stabilize the gait, and the right to maintain forward propulsion of the gait.

The guarding mechanism was first introduced by Main and Watson (1996) to explain the changes in movement pattern in chronic low back pain (LBP) patients. In this study, in-phase coordination occurred most of the time whilst carrying a maximum load in both manual and sedentary groups. It was assumed that these changes are important in order to ‘guard’ the structure and function around the pelvis and trunk from injury that can derive from the anterior load carriage. As the body started to fatigue, in-phase coordination became more frequent in all planes of motion. The repetitive strain around the pelvic-lumbar structure due to a prolonged carrying activity may eventually lead to the occurrence of LBP. As the load increases, the in-phase coordination became more frequent along the carrying activity in order to maintain a good stability control. For instance, to maintain balance, a significant amount of force has to be generated by the body to prevent an exaggerated forward leaning of the trunk due to moment of inertia of the anterior load. If an anti-phase coordination was to be dominant in the sagittal plane during the anterior load carriage, this would potentially lead to an increased risk of fall due to loss of posterior balance, particularly just after the heel strike as the whole body weight was transferred to the stepping leg in order to maintain forward progression (Redfern et al. 2001).

To the researcher's knowledge, there was no previous studies on the pelvis-trunk coordination that compare bilateral difference (i.e. right vs. left) of the coordination. Therefore, it was challenging to compare the findings from this study against the literature. However, the findings on the increase in-phase coordination was expected and supported by the literature. Previous studies reported that during standard gait, the most frequent pelvis-trunk coordination in the transverse plane was the in-phase coordination (Lamoth et al. 2002a; Lamoth et al. 2002b). Furthermore, it was also reported in the literature that there was a shift from the in-phase to anti-phase pelvis-trunk coordination as the walking speed increases (Seay et al. 2011b). In other words, any increase in the walking speed may lead to increased anti-phase coordination in healthy individuals. However in the low back pain individuals, the ability to shift from the in-phase to anti-phase was diminished (Seay et al. 2011b). As the pelvic rotation produced moment inertia during running, the trunk may rotate towards the opposite site in order to stabilize the angular momentum (Herr and Popovic 2008). However, in the current study, there was a significant decrease in stride length but a significant increase in cadence, suggesting an increase in gait frequency without affecting the walking speed (see section 6.3.1). Therefore, the changes from in-phase to anti-phase cannot be observed as there was no change in the walking speed.

8.3 CLINICAL IMPLICATION I: STATIC VS. FUNCTIONAL ENDURANCE

This study suggested that the test of endurance can be categorized into static and functional. With a proper standardization, the Ito test is a cost-effective screening tool for static endurance as the test is easy-to-conduct, require minimal equipment and may reduce the forces acted on the spine. Across the literature, the isometric back endurance was reported to be lower in the low back pain patients compared to the healthy individuals (Moreau et al. 2001). However, to the researcher's knowledge, no previous studies had compared the endurance between sedentary individuals and manual workers, as well as the difference in carrying activity performance. Therefore, this study attempted to fill the gap, as well as to recommend the carrying activity as a measure of functional endurance among workers. The carrying activity for this study was designed to mimic a standard protocol of carrying in a common functional capacity evaluation (FCE). Therefore, rather than carrying a load with predetermined weight (e.g. according to percentage of body mass), the participants were required to carry progressive loads throughout the activity. The advantage of this method is that the clinician could simulate actual working environment where the actual carrying loads varied according to the physical demands of a job.

Furthermore, long-term goal of rehabilitation is commonly more functional and relative to the patient's need, instead of focusing on symptom reduction. Therefore, the use of carrying activity as a test of functional endurance is suggested to indicate the level of occupational performance. The test of functional endurance can provide more information that is useful for making clinical decision towards the rehabilitation goal such as determining ability return-to-work after injury.

The isometric back endurance was achieved as a function of a sustained dynamic control between back extensor and hip extensor muscles. As the activity during the Ito test was static, the test can be considered as a measure of static endurance. In order to accomplish the carrying activity, a more complex dynamic control that may involve more muscles had to be maintained effectively. Due to a possibly frequent use of manual material handling (MMH) activities among manual workers, it was initially assumed that the manual workers may develop a higher functional endurance compared to the sedentary group due to the physical nature of their job. Nevertheless, the functional endurance should not be mistaken with a static and localized measure of endurance such as the Ito test because it should be tested in a more dynamic and global condition. The current study also found no significant association between the Ito test and the maximum carrying load. This can further confirm that static and functional endurance are two distinctive and unrelated measures that assess different types of performance. Workers who performed carrying activity as a part of their job description may regard the activity as meaningful as it relates to their work role as compared to other activities that are occupationally irrelevant to the workers. The meaningfulness of an activity can be either within the scope of the activity alone (i.e. purposeful activity), or can be extended towards the life role (occupation-based activity) (Early 2013). The use of carrying activity as a work-related assessment or intervention should incorporate the relevant job description in order to maximize its therapeutic values that may further benefit the workers (Daud et al. 2015; Ikiugu and Pollard 2015)

In an industrial work environment, a carrying activity may also performed with a presence of a pre-existing fatigue from previous activities that were conducted to accomplish a routine job circuit. In order to simulate the condition, the implementation of carrying protocol in this study was carefully designed to replicate the common work situation. This can permit the respondents to experience the carry-over fatigue from the previous carrying activity, thus, serving the original purpose of an FCE (Valpar International Corporation 2007). Previous studies had reported that the EMG level varied according to muscle length (Liberson et al. 1962; Lunnen et al. 1981; Pincivero et al. 2004). In this study, after each 60 meters of carrying activity, the participants stopped and performed a static stand. In order to capture reliable data for the carrying activity, the raw EMG signal was recorded for approximately five seconds during the static standing. For each gait condition, there was only one trial for static standing to indicate each load increment. Thus, a more reliable EMG data without any possible changes in muscle length could be captured.

8.4 CLINICAL IMPLICATION II: CLINICAL GAIT ANALYSIS

This study found that more steps were taken per unit of distance as the load became heavier, indicating an increase in dynamic stability during gait. These phenomena were influenced by two main factors: position of centre of mass (COM) and the size of base of support (BOS) (Pai and Patton 1997). During a static standing, a wide BOS was necessary to maintain stability. A good stability was maintained during static standing if the projection of COM is within the BOS. However, it was believed that a wide BOS can be a signpost of unsteady gait (Nutt et al. 1993; Snijders et al. 2007). During walking, the COM is always outside the BOS, and the dynamic stability was achieved when the gait enters the stance phase (Winter 1995). Studies had found that in an event of slipping, one of the common strategies to recover the loss of balance was by producing compensatory stepping (Jensen et al. 2001; Mansfield et al. 2010). The main objective of the compensatory stepping was to stop the gait without falling. In that case, a higher walking speed can bring COM more effectively to follow the slipping BOS in order to recover the loss of stability. However, if the walking speed was to be increased during a carrying activity with an anterior load, the compensatory stepping may not be able to hold the load inertia in an event of slipping, thus, may lead to forward balance loss. Therefore, the walking speed was constantly retained during the carrying activity as a part of stability control. This finding is in conjunction with LaFiandra et al. (2003) which also reported no significant interaction effect between walking speed and load carriage and explained that a constant walking speed was sustained by a decreased pelvic rotation and increased stride frequency.

The concept of leg preference also had clinical importance. For instance, the incidence of anterior cruciate ligament (ACL) injury among females tend to involve their non-preferred leg (i.e. supporting leg) (Brophy et al. 2010). It was generally understood that due to differential functions of brain hemispheres, there should be some preferential use of one limb over another under voluntary control. Nevertheless, leg preference cannot simply be determined by measuring the strongest leg, as the leg preference was greatly influenced by the type of activity performed by individuals (Spry et al. 1993). The activities that require mobility and manipulative skills (e.g. kick a ball, take a step forward) tend to use the right leg, whilst the activities that require stability and postural controls (e.g. static standing on one or two leg/s) tend to use the left leg (Velotta et al. 2011). Although carrying activity may utilize both sets of tasks, the participants may preferably use the left leg to maintain the dynamic stability whilst carrying the maximum load. In other words, the left leg movement became more rigid during the gait. This may indicate the use of left leg to stabilize the gait by decreasing the step length.

A significant increase in the trunk flexion-extension ROM was also observed whilst carrying a maximum load. Based on the kinematic waveform, the orientation of the trunk became more extended throughout the activity. In order to counteract with the gravitational forces that acted on the load during the anterior load carriage, the back muscles generated forces to extend the trunk. Both iliocostalis and multifidus were responsible for this action. The main responsibility of the back muscles during the anterior load carriage was to produce an opposing force against the moment of inertia of the load in order to maintain stability. However, the cumulative forces generated from the back muscle contractions were directed towards lumbar spine may increase the pressure onto the vertebral structures and nearby soft tissues. As for the pelvis, a repetitive and exaggerated pelvic tilt may eventually lead cumulative stress on the sacroiliac joints and lumbar spine (Denslow and Chace 1962; Minicozzi et al. 2016). Overall, these changes were important to maintain safety, stability and conserve energy, as well as to prevent from any possible injuries. Therefore, clinician should observe any changes in body alignment, posture and the spatiotemporal parameters of gait during the carrying activity and relate how these changes could affect health. The findings from the current study hoped to assist the clinicians towards a better understanding on the biomechanics of carrying in order to strengthen the clinical reasoning that justifies their practice.

Chapter 9: CONCLUSION TO PHD THESIS

9.1 INTRODUCTION

This chapter discussed the strength and limitation of the study and recommendations for future works. The strengths and limitations of the study were discussed based on important issues related to the study design and methodology. These issues consisted of intrinsic and extrinsic errors of surface electromyography, test-retest reliability of the Ito test, soft tissue artefact and occluded markers during motion analysis, the use of range of motion as the main kinematics parameter and the qualitative experience of fatigue. For the recommendations for future works, this section suggested some future research topics that can be conducted to fulfil gaps in knowledge. These topics should consist of comparison between healthy male individuals with female, heavy workers and/or patients, 3D kinematics of other manual material handling activities (e.g. lifting, lowering, pulling, pushing), the development of a cost-effective Functional Capacity Evaluation (FCE) and the use of carrying activity as a biomechanical intervention.

9.2 STRENGTHS AND LIMITATIONS

9.2.1 Intrinsic and Extrinsic Errors of Surface Electromyography

To determine the between-session reliability, the experiment session was repeated after two weeks. Within that interval period, there was a number of possible factors that can influence the test results. For instance, according to Gamet et al. (1996), there were two factors that can influence the reproducibility of EMG signals; extrinsic and intrinsic. Extrinsic factors were mainly related to the techniques of EMG recording, such as errors from reapplication of electrodes, intrinsic factors were mainly related to the physiological events such as metabolic changes between the sessions and muscle temperature. As the EMG was recorded on two separate sessions, there were possibilities that the variability in EMG may be influenced by both extrinsic factors and intrinsic factors. Although extrinsic factors were carefully controlled by consistently following the SENIAM recommendations (e.g. sensor location and inter-electrode distance) throughout the activity, it was very difficult to control the intrinsic factors when repeated in a different session (e.g. more than one day). To mitigate the effect of differing intrinsic effects on EMG, the temperature of the laboratory was kept at a reasonable temperature according to the participants' condition. Furthermore, there was a considerable time of rest between the arrival to the lab and the first recording due to having to fulfil the International Physical Activity Questionnaire (IPAQ) and also to attach EMG sensors and motion analysis markers. These strategies can potentially stabilize the muscle temperature after coming in from outside where the environment may be hot or cold. However, controlling for motor unit recruitment and ionic and metabolic alterations was not possible because muscle physiology was different for each individual. Other possible intrinsic factors that can influence the EMG content were changes in changes in muscle temperature, motor unit recruitment, ionic and/or metabolic alterations (Petrofsky and Lind 1980; Duchêne and Goubel 1993). However, both extrinsic and intrinsic factors affecting EMG recordings might be reduced when assessing within-session reliability. For instance, errors that were derived from reapplication of electrode can be minimized because the electrodes were placed only once during the whole session. Furthermore, as there were only few seconds of interval between each trial, the change in metabolic factors was assumed minimal. Nevertheless, the within-session reliability of EMG recordings could not be assessed in all conditions. The level of agreement cannot be determined for the Ito test and the muscle fatigue across gait conditions because there was only a single trial for each measurement.

9.2.2 Test-Retest Reliability of the Ito test

Generally, there are two types of reliability, namely between-session and within-session reliability. In the current study, the test-retest reliability of the Ito test was determined based on only between-session reliability with a 2-week interval. There were two outcome measures of the Ito test, which were the holding time (in seconds) and the muscle fatigue according to the slope of EMG median frequency (MFslope). Ideally, both outcome measures should correspond to each other, since the high rate of muscle fatigue should theoretically reduce the holding time. However, the findings indicated that the between-session reliability was good for the holding time,but was very low for the MFslope. The decision to maintain the Ito test as one of the main study parameters was made based on the between-reliability of the holding time, since the findings MFslope was exposed to more errors related to the EMG data acquisition and processing. For the between-session reliability, the main limitation was a wide range of interval between the two measurements. As the participants may vary within two weeks, this may affect their physical performance depending on the extent of physical activities throughout the period. Furthermore, diurnal temperature variability of the measurement was also varied based on the participants' availability, which may also affect the muscle physiology. Previous studies suggested that that the diurnal change in central body temperature tend to be higher in the evening (Martin et al. 1999; Chtourou et al. 2011). Therefore, a minimal interval between measurements (e.g. less than a week) and standardization for the time of day can improve the test-retest reliability. Nevertheless, the between-session reliability may be influenced by errors in sensor relocation. Thus, within-session reliability was more recommended for future works because the measurement will be repeated within the same without involving detachment and re-attachment of the EMG sensors, preventing the sensor relocation errors.

9.2.3 Soft Tissue Artefacts and Occluded Markers during Motion Analysis

In this study, the 3D motion analysis was determined based on photogrammetry method that uses the trajectories of selected markers to calculate the kinematics of each selected joint/body segment. The location of the markers was generally based on the Plug-in-Gait model, which determines the local coordinate systems (LCS) of each body segment. It was ideally assumed that the kinematic calculation was reliable as the markers were fixed around the body using a double-sided tape. Specifically, the soft tissue artefacts could occur due to inertial effects, skin deformation and sliding, gravity and muscle contraction (Andriacchi and Alexander 2000; Stagni et al. 2005). In this study, the soft tissue artefacts may exist due to dynamic movements that occurred whilst carrying the anterior load. Before the carrying activity began, all participants were instructed to hold the carrying container with a 90º of elbow flexion. During the carrying activity, as the carrying load and time increased, the elbow flexors eventually became fatigued and the elbow degree became more extended (i.e. reduced elbow flexion). This phenomenon can lead to a lowered position of the container towards the pelvis, which can alter the original position of the anterior supra iliac spine (ASIS) markers. Furthermore, the lowered position of the container occluded the markers from the Vicon’s cameras on occasions, preventing the motion analysis system from capturing the trajectories. As the position of the markers respective to the pelvic bone was changed, this could lead to soft tissue artefacts and error in the calculation of pelvic rotations. As the ASIS markers were the primary markers of pelvis according to the Plug-in-Gait model, the markers cannot be excluded from the kinematic analysis. Any gaps in the pelvis markers’ trajectories were processed based on Vicon’s gap filling methods. Apart from the primary markers of the pelvis (i.e. anterior and posterior suprailiac spine, ASIS & PSIS respectively), this study added two more pelvis markers on the most lateral part of the iliac spine bilaterally, which had improved the gap filling quality. Nevertheless, as the gap filling process works based on approximation of possible marker location, the errors in defining the location of the pelvis could still occur, which may further the location of hip joint centre, as well as the hip, knee and ankle kinematics.

9.2.4 Range of Motion as Kinematics Parameter

In order to compare the 3D kinematics between the sedentary individuals and the manual workers, the range of motion (ROM) was calculated by measuring the range between the minimum and maximum degree of rotation along the gait cycle. Also known as peak-to-peak waveform analysis, the ROM is important in determining the variability of a movement. One of the main implication of the study is to provide a sufficient background for rehabilitation practitioners such as occupational therapist and physiotherapist to inform and reason their practice in work assessment. As the ROM is commonly used in the clinical setting as a part of physical assessment, the clinicians can more understand, able to visualize, and make more sense of the current findings. The use of ROM as the main kinematic parameters is commonly reported in other studies that investigated the impact of load carriage on pelvis trunk coordination. For instance, Seay et al. (2011a) reported that carrying a rifle with both hands produced a greater trunk transverse ROM (i.e. axial rotation) in running, but lower trunk sagittal ROM for both speed. They also found that in transverse plane, the pelvis-trunk coordination was more in-phase while carrying the weapon. From this instance, it can be postulated that the ROM can provide an insight on how guarded phenomenon appears during the carrying activity. However, the main limitation of using the ROM is that the variability of minimum and maximum (peak) values of angular excursion that occurs around the selected joints. Although the use of ROM is considered adequate for this study to signify the coordination variability, the peak values are also important parameters in determining how safe the activity was conducted (e.g. extreme trunk flexion or extension may result in poor body mechanics). While the kinematics waveform for each angular excursion is presented and described in the findings, there are e no further statistical analysis conducted to synthesize the findings. Therefore, it is recommended that future studies should include peak values in statistical analysis such as analysis of covariance (ANCOVA) to further explore the changes in kinematics during carrying activity. One of the possible analysis strategies using the ANCOVA is to control the minimum peak value as a covariate whilst comparing the maximum peak value during the activity. By doing this, the magnitude of varying ROM throughout the activity can be controlled.

9.2.5 Qualitative Experience of Fatigue

Fatigue was an important parameter in this study as it aimed to understand its impact on body mechanics. The participants were instructed to terminate the carrying activity when they can no longer hold the load and the safe maximum carrying load limit was then determined. Once the activity was stopped, the participants were asked about their overall experience during the activity and what made them to discontinue the activity, which includes which parts of the body segments fatigued the most. However, this information was not analysed using a proper qualitative data analysis method (e.g. thematic analysis). One of the reasons was that instead of an in-depth interview, the participants were briefly asked about their experience along the activity. The second reason was that due to the current study can take up to a maximum of 3 hours, any forms of in-depth interview (e.g. structured, semi-structured or non-structured) would be time-consuming for both participants and researcher. Nevertheless, the information regarding the qualitative experience of fatigue can be a great addition to the study. The participants may describe their feeling of fatigue along the carrying activity, as well as their concerns with safety issues. This qualitative information can be used to triangulate the main findings, which is mainly quantitative.

9.3 RECOMMENDATION FOR FUTURE WORKS

9.3.1 Comparison with Female, Heavy Workers & Patients

The current study was conducted to establish a baseline information regarding muscle fatigue, spatiotemporal parameters, 3D kinematics and pelvis-trunk coordination during carrying activity. The carrying activity in this study may benefit rehabilitation practitioners such as occupational therapists when performing activity analysis to guide treatment planning and injury prevention at worksite. The baseline information regarding the biomechanics of carrying is important to serve as a scientific reasoning to be used as a possible method of assessment or intervention. However, the findings from this study cannot be inferred beyond the study population. For instance, as the study was conducted on the male participants, the findings cannot be fully generalized for both genders due to potential differences in gait kinematics (Bruening et al. 2015). Furthermore, the manual workers that were recruited into this study only performed light to moderate manual work. Therefore, an inclusion of heavy to very heavy manual workers is suggested for future studies to explore the biomechanical changes in more detail. Finally, in order to be used in clinical settings, it is recommended that this using actual patients. Patients may include those experiencing low back pain and also other orthopaedic conditions that may affect gait such as arthritis, and those using lower limb prosthetics. The comparison between the healthy individuals and the patients can provide more clinical-related knowledge such as deviation from standard movement due to fatigue or pain, which can serve both clinicians and researchers to guide their practice.

9.3.2 3D Kinematics of other Manual Material Handling Activities

The current study examined biomechanical changes in carrying activity because it was generally assumed that the activity can potentially lead to the development of low back pain. Although the literature on the carrying activity was available, none of the studies examined the carrying activity with progressive load. In clinical practice, the use of progressive load is important in order to determine the safe maximum load limit of a patient before returning back to a full employment. This method can also be applied to other manual material handling activities such as lifting, lowering, pushing and pulling. The use of 3D kinematics to evaluate the body movement and electromyography to evaluate muscle activity whilst performing those activities are needed to provide biomechanical characteristics of the activities. This can also be added to the knowledge on how these activities can potentially contribute to musculoskeletal disorders. As mentioned previously, a comparison between healthy individuals and LBP patients can inform on the impacts of specific musculoskeletal disorders on biomechanical functioning.

9.3.3 Development of Cost-Effective Functional Capacity Evaluation

Generally, a functional capacity evaluation (FCE) system is an expensive assessment tool in a standard clinical setting. It is common to observe that only specialized rehabilitation centres that focus on vocational rehabilitation and occupational health are equipped with the FCE. To ensure validity and reliability of the FCE, specific certification is required to enable a therapist to conduct the assessment. As the cost for both equipment and training may require extensive funding, this may give an advantage for some rehabilitation centres which operate using limited financial resources. As such, a more cost-effective assessment approach needs to be developed in order to ensure sustainability of rehabilitation services. Apart from 3D motion analysis and electromyography instrumentation, the carrying protocol that was developed from this study requires only minimal equipment such as a container and sand bags. However, the performance in carrying activity alone cannot be the representative for one's functional capacity. Therefore, other work-related activities such as lifting, pulling, pushing, standing, walking, sitting and stair climbing have to be included so that the physical demands of a job can be evaluated.

9.3.4 Carrying Activity as Biomechanical Intervention

In occupational therapy, the biomechanical strategies can be described as the use of biomechanical principles to understand and utilize physical functions such as strength, endurance and range of motion to facilitate and support the engagement in occupation. In order to determine the therapeutic potential of anterior load carriage, a process known as activity analysis is implemented by breaking down the activity into several components of tasks. From there, each component of the task can be used as a medium of activity gradation or adaptation. For instance, this study indicated that the carrying protocol consisted of static standing, walking and turning. This protocol was developed to simulate the standard anterior load carriage during working. Gradation is the process of increasing or decreasing the demand of an activity. This can be carried out by gradually altering some quantifiable aspects of the activity such as time, repetition or strength. For the carrying activity, gradation can possibly be implemented by reducing the frequency of static standing (i.e. less stop), increasing the frequency of turning and also increasing the carrying load progressively. Other than that, the current study also indicated that as the carrying increases and the participants became more fatigued, the body had to adapt with the situation by changing the kinematic, spatiotemporal parameters and also pelvis-trunk coordination during the activity.

9.4 CONCLUSIONS

9.4.1 Reliability of 3D Gait Analysis, Isometric Back Endurance and Muscle Fatigue

The motion analysis is a reliable measurement of the 3D kinematics of gait, as well the holding time of the Ito test to measure the isometric back endurance. However, the use of MFslope to measure the level of muscle fatigue during Ito test needs to be interpreted carefully if comparison between two different sessions are to be made. Due to inconsistent findings among the reliability measures for the level of muscle fatigue based on the MFslope of EMG, between-session reliability cannot be confirmed for all muscles. Therefore, the findings from this study suggested any comparison of muscle fatigue based on the MFslope should be performed during the same session in order to minimize errors that may occurred between different sessions.

9.4.2 Comparing Muscle Fatigue during Ito Test and Anterior Load Carriage

There was no significant difference in the level of isometric back endurance between the sedentary and the manual group. During the Ito test, no significant difference in MFslope was found between the groups in all muscles with low to medium effect size. For the maximum carrying load, the manual group can carry heavier load compared to the sedentary group. Finally, there was no significant association between isometric back endurance and maximum carrying load in both groups. It can be concluded that the Ito test and carrying activity represented different types of physical endurance, which were static and functional endurance respectively. This study suggested that both measures of endurance were important in assessing the impact of work-related musculoskeletal disorders on occupational performance at different stages of rehabilitation. Therefore, further investigations are needed to examine the clinical utility of the newly developed carrying protocol, as well to establish a more precise standardization for the Ito test.

9.4.3 Spatiotemporal Parameters of Gait and 3D Kinematics during Anterior Load Carriage

Both sedentary individuals and manual workers generally exhibit the same pattern of spatiotemporal parameters and 3D kinematics across the gait condition. Although the manual workers can carry heavier load compared to the sedentary individuals, both groups exhibit motor control and compensatory strategies to cope with fatigue. For spatiotemporal parameters, it can be concluded that the anterior load carriage can lead to increased gait frequency without influencing walking speed. For 3D kinematics, it can be concluded that the activity can lead to specific changes in ROM of the ankle, knee, hip, pelvis and trunk. The changes in the pelvis and trunk across the carrying activity may lead to the development of low back pain (LBP). Therefore, the next chapter will investigate the changes in the intersegmental coordination between the pelvis and trunk its relation to the LBP.

9.4.4 Changes in Pelvis-Trunk Coordination during Anterior Load Carriage

The changes in pelvis-trunk coordination as the result of carrying a maximum load can be observed in all planes of movements. As the pelvis and trunk are mechanically adjacent structures and functionally inter-related, any alteration from normal pelvis-trunk coordination to adapt to a fatiguing activity may eventually contribute to a mechanism that may result or sustain in low back pain. The findings from this study have provided an insight on how a carrying activity with progressive load can lead to changes in the pelvis-trunk coordination. Further investigations are now needed in the future to observe the changes in the coordination relative to different walking speed and to compare the changes between the low back pain patients and the healthy controls.

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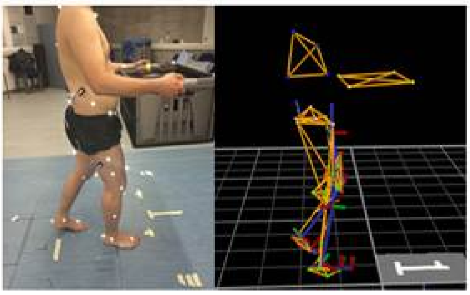
Appendix A POSTER FOR PARTICIPANT RECRUITMENT

The Rehabilitation & Health Technologies Group

UNIVERSITY OF Southampton
School of Health Sciences

How does your body move while carrying?



Carrying is known to be one of the most common activities in normal daily life and also for manual workers in industrial settings. Carrying tasks may be related to people getting low back pain and we want to look further at how the whole body moves and reacts while performing the activity.

To investigate this problem, we are looking for male individuals aged between 20 to 64 years old without any history of low back pain within the previous 12 months, and free from any chest or breathing problems (i.e. heart problem, asthma) or any medical problems that can affect your balance (e.g. dizziness, eye problem, foot problem).

What will happen during the research?

- A number of sticky patches will be put on specific sites of your body (e.g. trunk, buttocks, arms, legs).
- You will be instructed to carry a container that will have increasing loads up to the point you feel tired.
- Your body movements throughout the activity will be recorded by high-resolution cameras.
- The whole session will be completed within 3 hours.
- The test will be repeated again after 3-7 days to see whether there is any change in your carrying performance. However, the repeated session will only take a maximum 2 hours.

This research hopes to gain a new insight into the mechanism and prevention of low back pain. If you are interested, please contact:

Researcher: Mr. Hanif Farhan Mohd Rasdi (Hanif)
Email: hfmr1g12@soton.ac.uk
Supervisors: Dr. Peter White, Dr. Martin Warner, Dr. Jo Adams **Ethics number:** 8763

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Appendix B PARTICIPANT INFORMATION SHEET

Study Title: Kinetics and kinematics of a standardized bilateral carrying activity with progressive loads

Researcher: Hanif Farhan Mohd Rasdi

Please read this information carefully before deciding to take part in this research. If you are happy to participate you will be asked to sign a consent form.

What is the research about?

Carrying is known to be one of the most common activities not only for industrial work, but also in normal daily living activity. However, findings from previous research have suggested that this activity may give rise to low back pain. To investigate the problem, this research hopes to explore how and why the carrying activity can possibly contribute to low back pain. Therefore, the aim of this research is to observe the body movement during carrying activity. This research is self-funded by a PhD researcher which will be using healthy individual as the participant.

Why have I been chosen?

This research is recruiting male individuals aged between 20 to 65 years old without any history of clinical low back pain within the previous 12 months and any chest or breathing problem (e.g. heart disease, asthma).

What will happen to me if I take part?

Your participation is voluntary. If you decide to take part, you will be asked to sign a consent form, provide some basic information about yourself and complete a questionnaire to indicate your level of physical activity. Then, both you and the researcher will decide on the date for the main research, which will be divided into two sessions. Both sessions will be carried out at the Biomechanics Laboratory, Faculty of Health Sciences (Building 45), University of Southampton. In the laboratory, you will be asked to wear short pants to allow a number of sticky patches to be put on the skin at some specific locations of your body such as at the back, buttocks, arms and legs. Some measurements of your body such as knee width and ankle width will also be taken. This is to prepare you for the main sessions, in which can be divided into two sessions:

Session 1 (Part I and Part II)

Part I (Endurance of your back): While lying on a bed facing downward (supine lying), you will be instructed to lift your chest off the table as long as you can. This activity will be stopped whenever you feel tired.

Part II (How your body works while carrying): You will be instructed to carry a plastic container. During the activity, the researcher will put progressive loads eventually until you feel tired or being told to stop (1 kg load at each 60 meters; the maximum load is 20 kg). Throughout the carrying activity, your body movements will be recorded by high-resolution cameras. However, the cameras will only capture the movements of the sticky patches instead of the whole body image.

Session 2 (One week after)

In this session, only Part II will be repeated again after one week. This purpose is to detect whether there are any changes to your carrying performance after 7 days.

Are there any benefits in my taking part?

There will be no direct benefit (i.e. imbursement) you. The outcome of this research hopes to gain a new insight in the prevention and the mechanism of low back pain.

Are there any risks involved?

This research will be carried out in a very controlled environment. However, during the carrying, there is a risk of muscle soreness. This symptom is normal and likely to happen following any strenuous physical activity. Research has shown that this symptom will occurs as minimal as 24 hours after the activity, peaked within 72 hours, but slowly resolve in 5 to 7 days. To manage the muscle soreness, the use of cold pack on the sore site will be recommended. If the symptom extends beyond 7 days, you will be advised to see a doctor for further medical attention. Before the carrying activity begins, the researcher will give a clear instruction to demonstrate the safe carrying principle. Furthermore, the carrying activity will be terminated whenever you feel tired or when the researcher detects any unsafe movement. If you feel that the load is dropping from your hand, you can just let it fall onto the ground or push it away from the body.

Appendix C CONSENT FORM

Study title: Kinetics and Kinematics of a Standardized Bilateral Carrying Activity with Progressive Loads

Researcher name: Hanif Farhan Mohd Rasdi

Study reference: Carrying

Ethics reference: 8763

Please initial the box(es) if you agree with the statement(s):

I have read and understood the information sheet (28/11/13, version 1) and have had the opportunity to ask questions about the study.	☐
I agree to take part in this research project and agree for my data to be used for the purpose of this study	☐
I understand my participation is voluntary and I may withdraw at any time without my legal rights being affected	☐

Data Protection

I understand that information collected about me during my participation in this study will be stored on a password protected computer and that this information will only be used for the purpose of this study. All files containing any personal data will be made anonymous.

Name of participant …………………………………………………………………….

Signature of participant…………………………………………………………………….

Date………………………………………………………………………………………

Appendix D INTERNATIONAL PHYSICAL ACTIVITY QUESTIONNAIRE (IPAQ)

We are interested in finding out about the kinds of physical activities that people do as part of their everyday lives. The questions will ask you about the time you spent being physically active in the **last 7 days**. Please answer each question even if you do not consider yourself to be an active person. Please think about the activities you do at work, as part of your house and yard work, to get from place to place, and in your spare time for recreation, exercise or sport.

Think about all the **vigorous** activities that you did in the **last 7 days**. **Vigorous** physical activities refer to activities that take hard physical effort and make you breathe much harder than normal. Think *only* about those physical activities that you did for at least 10 minutes at a time.

1. During the **last 7 days**, on how many days did you do **vigorous** physical activities like heavy lifting, digging, aerobics, or fast bicycling?

   _____ **days per week**

   ☐ No vigorous physical activities → ***Skip to question 3***

2. How much time did you usually spend doing **vigorous** physical activities on one of those days?

   _____ **hours per day**
   _____ **minutes per day**

   ☐ Don't know/Not sure

Think about all the **moderate** activities that you did in the **last 7 days**. **Moderate** activities refer to activities that take moderate physical effort and make you breathe somewhat harder than normal. Think only about those physical activities that you did for at least 10 minutes at a time.

3. During the **last 7 days**, on how many days did you do **moderate** physical activities like carrying light loads, bicycling at a regular pace, or doubles tennis? Do not include walking.

   _____ **days per week**

   ☐ No moderate physical activities → ***Skip to question 5***

4. How much time did you usually spend doing **moderate** physical activities on one of those days?

_____ **hours per day**
_____ **minutes per day**

☐ Don't know/Not sure

Think about the time you spent **walking** in the **last 7 days**. This includes at work and at home, walking to travel from place to place, and any other walking that you have done solely for recreation, sport, exercise, or leisure.

5. During the **last 7 days**, on how many days did you **walk** for at least 10 minutes at a time?

_____ **days per week**

☐ No walking → ***Skip to question 7***

6. How much time did you usually spend **walking** on one of those days?

_____ **hours per day**
_____ **minutes per day**

☐ Don't know/Not sure

The last question is about the time you spent **sitting** on weekdays during the **last 7 days**. Include time spent at work, at home, while doing course work and during leisure time. This may include time spent sitting at a desk, visiting friends, reading, or sitting or lying down to watch television.

7. During the **last 7 days**, how much time did you spend **sitting** on a **week day**?

_____ **hours per day**
_____ **minutes per day**

☐ Don't know/Not sure

This is the end of the questionnaire, thank you for participating.

Appendix E PARTICIPANT CHARACTERISTICS

BASIC INFORMATION	
Participant ID: ________________ Age: __________________ Height: __________________ Weight: ___________________ Knee width: _______________ Ankle width: ________________ Leg length: ________________ Occupation: ________________	
Research procedure (guide for researcher)	**Remarks**
1. Calibration 2. Participant arrived, give and/or explain the PIS 3. Sign the consent form 4. Give the IPAQ 5. Antropometric measurement 6. Put 4 EMG electrodes (illiocostalis, multifudus, gluteus maximus, biceps femoris) 7. Ito test 8. Rest (up to a maximum of 15 minutes) 9. Put another 4 emg electrodes (biceps brachii,) 10. Put markers for motion analysis 11. Static standing for 10 seconds	

Appendix F CHECKLIST FOR MOTION ANALYSIS MARKERS

	Codes	Description	Remarks
1	IJ	Jugular Notch	
2	PX	Xiphoid Process	
3	C7	7th cervical vertebrae	
4	T8	8th Thoracic vertebrae	
5	L1	1st lumbar vetebrae	small
6	L3	3rd lumbar vetebrae	small
7	L5	3rd lumbar vetebrae	small
8	LASI	Left anterior sacroiliac spine	
9	RASI	Right anterior sacroiliac spine	
10	LPSI	Left posterior sacroiliac spine	
11	RPSI	Right posterior sacroiliac spine	
12	LSI	Left Sacroiliac Spine	
13	RSI	Right Sacroiliac Spine	
14	LTHI	Left thigh	
15	LMTHI	Left medial thigh	small
16	LKNE	Left knee	
17	LMKNE	Left medial knee	small
18	LTUB	Left tibial tuberosity	small
19	LTIB	Left tibial	
20	LANK	Left ankle	
21	LMANK	Left medial ankle	
22	LHEE	Left heel	
23	L5thMET	Left 5th metatarsal	small
24	LTOE	Left toe	
25	RTHI	Right thigh	
26	RMTHI	Right medial thigh	small
27	RKNE	Right knee	
28	RMKNE	Right medial knee	small
29	RTUB	Right tibial tuberosity	small
30	RTIB	Right tibial	
31	RANK	Right ankle	
32	RMANK	Right medial ankle	
33	RHEE	Right heel	
34	R5thMET	Right 5th metatarsal	small
35	RTOE	Right toe	
36	BFL	Anterior-Left edge of container	
37	BFR	Anterior-Right edge of container	
38	BRL	Posterior-Left edge of container	
39	BRR	Posterior-Right edge of container	

Appendix G CHECKLIST FOR EMG ELECTRODES

EMG chanel Right	Left	Muscles	Electrode placement	Remarks
1	9	Iliocostalis	One-finger width medial from the line from the PSIS to the lowest point of the lower rib, at the level of L2 (SENIAM 1999).	
2	10	Multifidus	On and aligned with a line from caudal tip PSIS to the interspace between L1 and L2 interspace at the level of L5 spinous process (i.e. about 2 - 3 cm from the midline) (SENIAM 1999).	
3	11	Gluteus maximus	At the middle of the line between sacral vertebrae and the greater trochanter. This position corresponds with the greatest prominence of the middle of the buttocks well above the visible bulge of the greater trochanter (SENIAM 1999).	
4	12	Biceps femoris	At the middle of the line between ischial tuberosity and the lateral epicondyle of the tibia (SENIAM 1999).	
5	13	Biceps Brachii	On the line between the medial acromion and the fossa cubit at 1/3 from the fossa cubit (SENIAM 1999).	
6	14	Latissimus dorsi	Approximately 4 cm below the inferior tip of the scapula, half the distance between the spine and the lateral edge of the torso (Criswell 2010)	
7	15	Vastus lateralis	2/3 on the line from the anterior iliac spine superior to the lateral side of the patella (SENIAM 1999).	
8	16	Gastrocnemius	1/3 of the line between the head of the fibula and the heel (SENIAM 1999)	