The Effects of Age on Gait and Functional Movement Characteristics in an Older Adult Population

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Dedication

I dedicate my thesis to my family, my true inspiration.

'The road to success is not easy, we will stumble and lose our path, but with support, hard work, determination and passion, it is possible to achieve our dreams.'

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Appendix One

Abbreviations

%GC	Percentage of the Gait Cycle
ALSA	Australian Longitudinal Study of Ageing
ASIS	Anterior Superior Iliac Spine
BLSA	Baltimore Longitudinal Study of Aging
BMI	Body Mass Index
C7	7th Cervical Vertebrae
CHAMPS	Community Health Activities Model Program for Seniors
DT	Dual Task
EAGLES	Essex Ageing and Gait Longitudinal Study
ELSA	English Longitudinal Study of Ageing
F1	First Peak Vertical Force
F2	Minimum Vertical Peak Force
F3	Second Peak Vertical Force
F4	Breaking Peak Force
F5	Propulsive Peak Force
FPC	Force Plate Contact
Fy	Anterior-Posterior Force
Fz	Vertical Force
GRF	Ground Reaction Forces
L5	Fifth Lumbar Vertebrae
LR	Loading Response
manual DT	Manual Dual Task Walking
MET	Metabolic Equivalent Intensity
MMSE	Mini-Mental State Examination
MTC	Minimum Toe-Clearance
MxT1	First Maximum Toe-Clearance

MxT2	Second Maximum Toe-Clearance
NHS	National Health Service
NW	Normal Walking
NW (with FPC)	Normal Walking with Force Plate Contact
NW (without FPC)	Normal Walking without Force Plate Contact
PiG	Plug-in Gait Marker Model
PSw	Pre-Swing
SON	Stepping Onto an Obstacle
SOV	Stepping Over an Obstacle
SPM	Statistical Parametric Mapping
T10	10th Thoracic Vertebrae
TILDA	The Irish Longitudinal Study on Ageing
TS	Terminal Stance
TUG	Timed Up and Go

Thesis Abstract

The maintenance of function in an ageing population is essential to ensure current and future health in older people. The ability to walk independently in a range of situations and environments is key to successful ageing. Age-related gait adaptations including spatial-temporal parameters, joint kinematics and kinetics have been identified to be a consequence of the ageing process. For example, reduced walking speed and increased pelvic tilt are suggestive of compensation strategies to minimise falls. The majority of research has compared young adults (20-40 yrs) to older adults (≥ 50 yrs), categorising older adults into a single group regardless of actual age. An alternative approach is to explore the effects of age on gait and functional movement characteristics within an older adult population. One-hundred and fifty-eight community-dwelling older adults, age range 55 to 86 years (65.7 ± 6.8 yrs) were recruited to create a new gait database. Three-dimensional motion analysis captured five walking tasks: normal walking (with and without force plate contact), manual dual task walking and walking with obstacle clearance (stepping onto, off and over an obstacle). Age-related adaptations to walking occurred from age 75 years by adopting a joint kinetic strategy (including reduced hip extension moment) and altering gait (including a reduced walking speed). Increasing the task complexity was associated with altered gait patterns for this older adult group including a reduction in toe-clearance during manual dual task walking (increasing the likelihood of tripping) and increased arm swing during obstacle clearance (potentially increasing stability). This work represents the creation of one of the largest databases of gait in older people including three-dimensional motion analysis for normal walking and three functional walking tasks for healthy high-functioning older adults. It has the potential to be used to identify factors that predispose older adults to falling or with previously unidentified pathological changes.

Chapter One: Literature Review

1.1. Ageing Population

Since the 1960s life span in the UK has increased by ten years, with the most recurring age at death being 86 for men and 89 for women (Office for National Statistics, 2013). In 2016, the UK population reached 65.6 million, with a predicted population of over 74 million by 2039 (Office for National Statistics, 2017). An estimated 23.6 million people are aged fifty years and older, which is a third of the total UK population (Office for Natioanl Statistics, 2016), with 18 % of the population aged 65 years and older and 2.4 % aged 85 years and above (Office for National Statistics, 2017a). Births in the UK are outnumbering deaths, resulting in a growing population. Consequently, the old age dependency ratio, which is the number of older adults (≥ 65 yrs) in relation to every 1000 people (aged 16-64 yrs) has increased (Office for National Statistics, 2017a). For example, in 2016, the UK old age dependency ratio was 285 (Office for National Statistics, 2017a).

While longevity is advantageous, an increased 'pensioned' age portion of the population does question the sustainability of public and social-care services such as the National Health Service (NHS) (Office for National Statistics, 2017a), which could impact the well-being of older adults. The UK Government since 2010, account for almost half the UK expenditure in health and social welfare spending, with the NHS predominantly spending care for older adults (≥ 65 yrs) (Cracknell, 2010). Older adults over the age of 85 have found on average to cost the NHS three times more than an older adult aged 65 (Cracknell, 2010). As such, the NHS since 2014 have executed the 'five year forward view', which is currently responding to predicted changes in the future delivery of health and social welfare care, with the emphasis on care improvement, promoter diagnosis and smart technology (assistive technology) for older adults for example (NHS England, 2014).

1.2. Consequences of Ageing

An ageing population creates both societal and individual challenges in terms of 'ensuring independence' and minimising the risk of disability among older adults. Ageing will most likely increase the expenditure of declined health and disability in the population, as older adults typically shift from acute ill health to chronic condition, morbidities, cognitive decline and impairment and increasing frailty. As well, the burden/obligation of an ageing population for families and communities to provide care services to aid quality of life for older adults (Smith, 2015). According to Age UK (2018), there is an estimated 4 million older adults living with a longstanding illness (65-74 yrs = 36 % and ≥ 75 yrs = 47 %), which will impact on health and social care services and expenditure (estimated £5 billion additional funds). It has been found approximately 7.6 million (41 %) hospital adult admissions out of 18.7 million were older adults aged 65 years and over (NHS, 2015), which increases for A&E admissions to forty-seven percent (NHS Benchmarking Network, 2017). Consequently, average length of hospital stays increases with age (65-74 yrs = 6.5 days, 75-84 yrs = 8.3 days and ≥ 85 yrs = 10.1 days) (NHS Benchmarking Network, 2017).

As such, the quality of older adults' physical function is important for the maintenance of health and well-being, as this is influenced by the ageing process (Guralnik and Simonsick, 1993). Frailty as been linked to quality of life in older adults, with this being described as progressive decline in physical, mental and social functions (Van Campen, 2015). Frailty can cause reduced recovery from acute ill health (Nicholson *et al.*, 2017), resulting in increased vulnerability to sudden health deterioration (Covinsky *et al.*, 2003, Turner and Clegg, 2014). Although, not all older adults become frail, it is more prevalent with increased age, for example ten percent become frail aged 65 years and over and sixty-five percent from the age of 90 years (Gale *et al.*, 2015). Older adults with frailty are found to have greater risk of disability, hospitalisation, care home admission and ultimately death compared to healthy older adults (Fried *et al.*, 2001, Rockwood *et al.*, 2006). For example, in 2006-2012, older adults with frailty accounted for 4000 daily hospital admissions which resulted in over one million deaths (Soong *et al.*, 2015).

Sixty percent of older adults will contend with at least one chronic condition such as arthritis, dementia or congestive heart failure (Comas-Herrera *et al.*, 2011, Charlesworth, 2013, Giuli *et al.*, 2014). Dementia is the leading cause of disability for older adults compared to cardiovascular disease and stroke for example (Department of Health, 2009) and consequently results in the most deaths for women in the UK (Alzheimer's Research UK, 2016). There are estimated 850,000 adults in the UK living with dementia and 808,000 are older adults aged 65 years and above (Alzheimer's Society, 2017). One in fourteen older adults aged 65 has dementia, however this condition increases with age, which affects one in six for older adults over 80 years and one in three over 95 years (Alzheimer's Society, 2017). In the UK, 3.2 million older adults aged 65 and over have urinary incontinence (Buckley and Lapitan, 2009). Twenty-five percent of all deaths in older adults aged 65 years and over were caused by cardiovascular disease in the UK (British Heart Foundation, 2015). It is estimated in the UK, thirty-four percent (older men) and twenty-two percent (older women) of older adults aged 65-74 years and twenty-eight percent (older men) and twenty-nine percent (older women) of older adults aged 75 years and over have cardiovascular disease (British Heart Foundation, 2015). It is estimated 152,000 strokes occur each year in the UK (The Stroke Association, 2016). Stroke risk doubles for adults aged 55 and above, with seventy-four percent of strokes in the UK occurring for older adults (≥ 65 yrs) (The Stroke Association, 2016), costing NHS and society over £8.9 billion in health and social care (Saka *et al.*, 2009).

For older adults aged 65 years and older, trips and falls are the leading cause of unintentional injury and hospitalisation (Dellinger and Stevens, 2006, Deandrea *et al.*, 2013, Stevens *et al.*, 2014). Falls in the England have resulted in 220,000 A&E hospital admissions (Public Health England, 2016). One in three of this age group will fall each year (Hausdorff *et al.*, 2001, Panel on Prevention of Falls in Older Persons and British Geriatrics, 2011, Stevens *et al.*, 2014), with 20-30% of those incurring a moderate-to-severe injury impacting on independence and increasing the risk of early mortality (Sterling *et al.*, 2001, Nachreiner *et al.*, 2007, Bleijlevens *et al.*, 2010, Panel on Prevention of Falls in Older Persons and British Geriatrics, 2011). Extrinsic risk factors which are commonly found in the home (e.g. house clutter), cost the NHS in England around £435 million (BRE, 2016), with fragility fractures costing the

UK NHS £4.4 billion and £1.1 billion for social care (Svedbom *et al.*, 2013). In 2016, 4,984 falls for older adults over the age of 65 resulted in death (Office for National Statistics, 2017b).

After hospitalisation from a fall, there is an approximately 50% risk of mortality within 12 months (Rubenstein, 2006). Older adults who fall, even those without injury, may develop a fear of falling causing individuals to limit their activities leading to reduced mobility and functionality (Delbaere *et al.*, 2004). Tinetti and Kumar (2010) systematically reviewed 33 studies which assessed risk factors of falls in community-dwelling older adults, demonstrating a strong association with previous falls, reduced muscle strength, gait and balance impairments. Causes of falls are generally multifactorial with combined intrinsic (e.g. joint stiffness) and extrinsic (e.g. loose rug) risk factors (Tinetti *et al.*, 1988, Stevens *et al.*, 2014). However, the majority occur when tripping on a step, turning or whilst walking (Winter *et al.*, 1990, Scott *et al.*, 2007, Stevens *et al.*, 2014), which commonly (around 65,000 falls) result in hip fractures for older adults (≥ 60 yrs), costing the NHS £1 billion per year (Royal College of Physicians, 2016). It is rare for an older adult to regain complete recovery post hip fracture, typically there is an increased dependency which results in walking difficulty (reliance of walking aids) and the need for long-term care (e.g. care home) (Royal College of Physicians, 2016).

The comorbidities described above, not only have been found to cause loneliness and isolation but also can be a consequence of an increased loneliness and isolation (Age UK, 2018). For example, social relationship deficiencies were associated with an increased health risk of cardiovascular disease and stroke (Department of Health, 2009). Figure 1.1. illustrates the effects of loneliness and social isolation on physical health. Older adults who experience long periods of loneliness are twice more likely to develop Alzheimer's or dementia (Wilson *et al.*, 2007). Consequently, loneliness/isolation and comorbidities typically result in an overall reduced physical activity level and time (e.g. walking) and increased sedentary time (e.g. sitting), which impacts on quality of life (Age UK, 2018). For example, sedentary lifestyle above the age of 50 is associated with an increased risk of mortality compared to physical active adults (≥ 50 yrs) (ELSA, 2016). As such, older adults who have retired have found to change from high-medium to low levels of physical activity, compared to older adults who work

(Matthew *et al.*, 2014), which results in weaker muscles (particularly for the lower body) (Bijlsma *et al.*, 2013, Sillanpaa *et al.*, 2014); leading to premature onset of ill health and frailty (McPhee *et al.*, 2016).

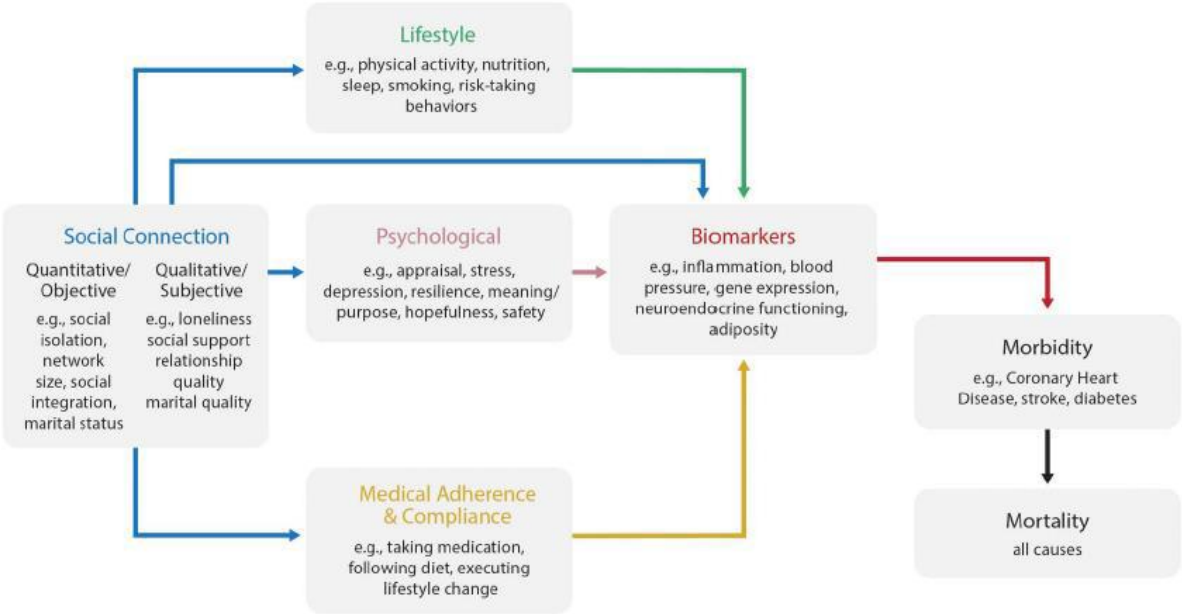


Figure 1.1. The effect of loneliness and social isolation in older adults (Age UK, 2018).

1.3. Example of a Comorbidity and the Impact on Gait

Osteoarthritis is a common, degenerative condition which progressively destroys joint cartilage and can affect several joints, especially weight-bearing joints such as the hip and knee (Arthritis Research UK, 2014). This results in joint swelling, stiffness, instability, joint pain and structural changes such as bone deformity (Broström *et al.*, 2012). In the UK, osteoarthritis treatment occurs in 33 % of middle-aged older adults (≥ 45 yrs), with 49 % of women and 42 % of men over the age of 75 years seeking treatment (Arthritis Research UK, 2014).

Gait adaptations of older adults with osteoarthritis results from soft-tissue stiffness and structural joint changes (Broström *et al.*, 2012). Older adults with hip osteoarthritis have commonly reported reductions in walking speed, stride length, flexion and extension of the hip during normal walking (NW) compared to healthy older adults (Hulet *et al.*, 2000, Kyriazis and Rigas, 2002, Mont *et al.*, 2007, Lamontagne *et* *al.*, 2009, Bijlsma *et al.*, 2011). In addition, reduced cadence, stride/step length and increased double-support time was also reported for older adults with osteoarthritis (Ouellet and Moffet, 2002, Dyrby *et al.*, 2004, Khazzam *et al.*, 2006, Weidow *et al.*, 2006, Kubota *et al.*, 2007, Valderrabano *et al.*, 2007, Houdijk *et al.*, 2008, Brodsky *et al.*, 2011, Nuesch *et al.*, 2012). Kubota *et al.* (2007) found women (59.4 ± 11.1 yrs) with hip osteoarthritis had an increased anterior tilt and ankle power generation, with decreased step width, hip extension and abduction and hip abduction moment compared to healthy women (64.3 ± 2.8 yrs).

Older adults with knee osteoarthritis have been found to reduce their knee range of motion (RoM) during weight acceptance, causing higher impact loads in the knee (Lafortune *et al.*, 1996, Cook *et al.*, 1997). However, research (Rudolph *et al.*, 2007, Boyer and Andriacchi, 2016) has also reported reduced knee flexion and greater knee adduction, with no difference in knee RoM (knee flexion/extension). Predominantly changes at the knee joint results in an increased adduction moment during normal walking (NW) (Weidow *et al.*, 2006, Rudolph *et al.*, 2007, Zeni and Higginson, 2009). In addition, patients with ankle osteoarthritis demonstrated decreased dorsiflexion RoM and no first rocker, which is known as rapid plantarflexion (Khazzam *et al.*, 2006, Valderrabano *et al.*, 2007, Brodsky *et al.*, 2011).

1.4. <u>Functionality and Mobility of the Ageing Process</u>

Physical functionality describes a person's ability to perform everyday tasks (Cooper *et al.*, 2011b), whereas mobility is broadly defined as the effects of the musculoskeletal system to locomote through more than one plane within the environment (e.g. walking at home) (Grillner *et al.*, 2008, Webber *et al.*, 2010). Walking is an important daily task which requires systematic actions of the musculoskeletal system. Previous research (Faulkner *et al.*, 2007, Snijders *et al.*, 2007) found older adults have reduced musculoskeletal function resulting from physiological and neuromuscular changes. These changes include; cross-sectional muscle mass loss (10-40 %), decrease in muscle fibres (type I and type II), sensory and motor nerve conduction velocity decreases in the central and peripheral nervous system (Lewis and Bottomley, 1994, Kauffman, 2007) and loss of elastic fibres within the articular cartilage, resulting in stiffer joints (Lewis and Bottomley, 1994, Kauffman *et al.*, 2001). As such, these changes are contributory to gait ageing effects, which affect neural control, muscle function and postural control (Harris *et al.*, 2008). The above, age-related variations contribute to altered joint ranges of motion and reduced muscle mass and strength, decreased reaction time (Kang and Dingwell, 2008b) and consequently older adults modify their gait pattern. Figure 1.2. illustrates the ageing trajectories of physical activity on the musculoskeletal system, which identifies good physiological function to around the age of 50 years, with subsequent progressive decline (>50 yrs). Although, ageing declines in the physiological and neurological system occur for healthy active older adults, the rate of this decline is slower compared to inactive-moderately active older adults (Pearson *et al.*, 2002, Michaelis *et al.*, 2008, Wilks *et al.*, 2009, Power *et al.*, 2010, Degens *et al.*, 2013, Trappe *et al.*, 2013, Ireland *et al.*, 2014). As such, walking for older adults is a key factor for healthy ageing, as walking aids the ability to actively engage in both daily and social activities, which improve health and wellbeing, quality of life and independent living (Age UK, 2015a).

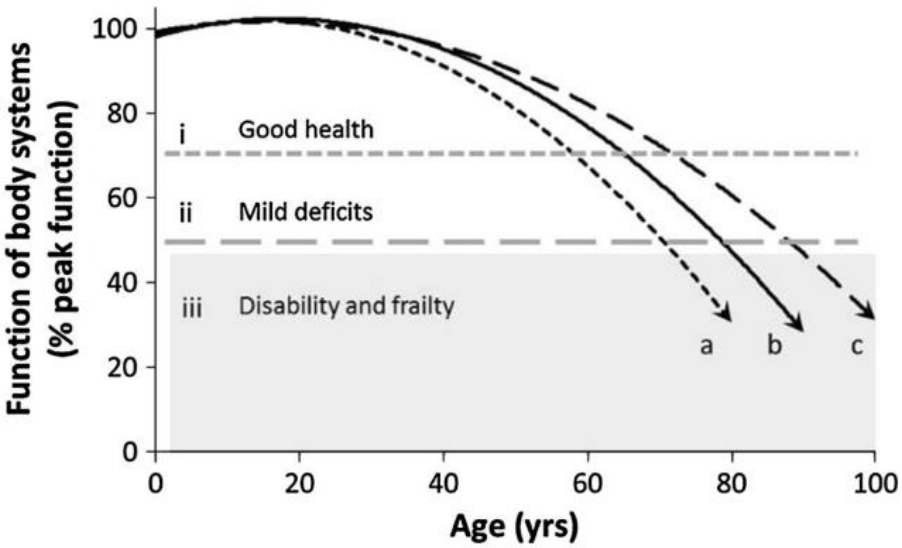


Figure 1.2. Ageing trajectories and physical activity on the musculoskeletal function (McPhee *et al.*, 2016). *Note:* **a**) accelerated ageing, **b)** normal ageing and **c)** healthy ageing.

Locomotive process of gait involves several tasks, for example equilibrium maintenance during walking and ability to meet environmental demands (e.g. walking on uneven ground) (Patla *et al.*, 1990, Woollacott and Tang, 1997). As such, gait is not limited to straightforward walking, as the neuromuscular system is challenged to negotiate environmental demands with dual task (DT) walking, such as turning, avoiding obstacles, stepping over objects (Harris *et al.*, 2008). Walking is a complex motor task which is generally performed automatically by adults (Hausdorff *et al.*, 2001). However, compared to young adults, older adults' walking is often no longer automatic requiring more cognitive attention to motor control as the efficiency of the neurological system is reduced (Woollacott and Shumway-Cook, 2002, Yogev-Seligmann *et al.*, 2008). Decline in cognitive function ability is a normal process of ageing, for example reduced processing speed, attention and executive function abilities (Salthouse, 2010, Harada *et al.*, 2013). There is also identified physiological and neurological changes in the brain, with declines in grey and white matter volume and neurotransmitter levels that contribute to observed changes in cognition with ageing (Harada *et al.*, 2013). Motor tasks, like gait have found to coincide with declined cognitive function (van Iersel *et al.*, 2008). As previously mentioned, gait is a complex motor task, which requires information processing for attention, memory and planning (Theill *et al.*, 2011) and motivation and judgement (Amboni *et al.*, 2013). Memory ageing effects have found to be associated with slowed information processing speed (Luszcz and Bryan, 1999) and an inability to disregard irrelevant information (Darowski *et al.*, 2008), for example.

The role of cognitive function (executive function and attention) evidently influences gait when a secondary task is implemented (Woollacott and Shumway-Cook, 2002, Alexander and Hausdorff, 2008, Yogev-Seligmann *et al.*, 2008, Yogev-Seligmann *et al.*, 2010). For example, young adults have a reduced walking speed when performing a secondary task (e.g. counting backwards in 3s) (Ebersbach *et al.*, 1995, Abernethy *et al.*, 2002, Woollacott and Shumway-Cook, 2002, Beauchet *et al.*, 2005, Yogev-Seligmann *et al.*, 2010). Even in healthy older adults an attention demanding task such as stepping over an obstacle or walking whilst talking causes a gait alteration such as, reducing walking speed, increasing double-support time and gait variability (Woollacott and Shumway-Cook, 2002, Brunt *et al.*, 2005, Hausdorff *et al.*, 2008, Laessoe *et al.*, 2008, Yogev-Seligmann *et al.*, 2008, Plummer-

D'Amato *et al.*, 2011, Peper *et al.*, 2012). Underlying mechanisms of DT reactions are not fully understood (Hausdorff *et al.*, 2008), however such tasks highlight information on the automaticity of gait and fall risk (Zijlstra *et al.*, 2008), which may not be identified during straightforward walking. Walking whilst performing a secondary task, may create a conflict regarding task prioritisation and most so when information processing is limited (Pashler, 1994 and Tombu and Jolicoeur, 2003). Both healthy young and older adults prioritise gait stability above a cognitive DT when no instructions regarding task prioritisation were provided (Bloem *et al.*, 2001a, Bloem *et al.*, 2001b, Bloem *et al.*, 2006, Yogev-Seligmann *et al.*, 2010).

DT interference using capacity theory, suggests the resources of information processing requirements are flexible but limited (Abnerthy, 1988, Tombu and Jolicoeur, 2003, Fraizer and Mitra, 2008), which may result in performance deterioration in either the walking or secondary task and or both tasks (Woollacott and Shumway-Cook, 2002). Also, the flexibility of information process as many factors which may influence resource allocation, for example motivation and task difficulty (Abnerthy, 1988, Woollacott and Shumway-Cook, 2002). In addition, bottleneck theory has also been used to explain dual task interference, which requires serial or sequential processing of two concurrent tasks (Ruthruff *et al.*, 2001). As such, to complete one task, the secondary task is temporarily postponed, which results in declined performance of the secondary task (Kelly *et al.*, 2012). For example, a number of DT gait studies using verbal tasks (cognitive DT) have shown a reduction in walking speed compared to NW (Yogev *et al.*, 2005, Springer *et al.*, 2006, Hollman *et al.*, 2007). As such the attention for a cognitive task and walking is split and allocated arbitrarily to each task, thus the additional cognitive task draws attention away from walking resulting in a change to the gait (Yogev-Seligmann *et al.*, 2010).

In addition, the cognitive motor interference theory refers to performing simultaneously a motor and cognitive task, which interferes with performance of one or both tasks (Schott *et al.*, 2016). This theory was proposed as a new approach to evaluate brain function for adults with mild cognitive impairment (Montero-Odasso *et al.*, 2014). For example, older adults who stop walking whilst talking are at greater risk of falling (Lundin-Olsson *et al.*, 1997), which demonstrates the cognitive load effect on gait. The interdependence relationship between gait and cognition in older adults is clearly evident with individuals with cognitive impairment and dementia, as walking speed becomes slower for single and DT performances (Camicioli *et al.,* 1998, van Iersel *et al.,* 2004, Allan *et al.,* 2005, Holtzer *et al.,* 2006, Pettersson *et al.,* 2007, Montero-Osasso *et al.,* 2009). Differences between young, older adults and older adults with mild cognitive impairment have found to occur when DT walking because of age, education, task prioritisation and cognitive reserve (Stern, 2009, Schaefer, 2014, Belghali *et al.,* 2017). Klotzbier and Schott (2017) found increasing cognitive task difficulty using the Trial-Walking Test (modified by Schott, 2015) allocation of information processing resource becomes more directed towards the cognitive task, neglecting the walking task. Both young (42 participants with a mean age of 23.9 ± 1.98 yrs) and older adults (43 participants with a mean age of 68.2 ± 6.42 yrs) adapted a safe strategy and prioritised the walking task over the cognitive in the Trial-Walking Test (Klotzbier and Schott, 2017). This resource allocation strategy could lead to a fall risk, especially in older adults and older adults with cognitive impairment (Montero-Odasso *et al.,* 2012, Muir *et al.,* 2012). As such, increased motor interferences in older adults and age associated cognitive function decline (e.g. reduced cognitive attention), has been suggested the reason for fall risk during DT walking (Liu-Ambrose *et al.,* 2008, Mak *et al.,* 2014). Consequently, DT walking is thought to be affected by a reduced cognitive reserve for older adults compared to young adults (Verghese *et al.,* 2010, Perrochon *et al.,* 2013, Klotzbier and Schott, 2017). Cognitive reserve is described as the increase efficiency and capacity of existing neural pathways and/or the recruitment of new pathways, for example counter-acting age-related cognitive changes without cognitive deficit development (Belghali *et al.,* 2017, Franzmeier *et al.,* 2017, Gelfo *et al.,* 2017).

Therefore, identifying the onset of a reduction in function and/or the presence of disability associated with ageing is a key factor especially for screening community-dwelling older adults (Gill, 2010). Evaluating physical functionality ensures guidance for effective treatment interventions and in older adults provides classification of the ageing process in terms of vigorous to frailty status (Bierman, 2001). For this reason, there is developing evidence that measuring physical functionality such as normal overground walking, dual task, stepping and obstacle negotiation, not only indicates general health (Cesari *et al.*, 2005) and quality of life (Ferrucci *et al.*, 2000) but also predicts adverse events such as falls, dementia and mortality (Hausdorff *et al.*, 2001, Studenski *et al.*, 2003, Verghese *et al.*, 2007, Swanenburg *et al.*, 2010). Therefore, functionality acts as a marker for current and future health.

The effects of the ageing process and screening on older adults' functionality and mobility has predominantly been analysed through developmental research, using either a cross-sectional or longitudinal study design. Longitudinal ageing studies were designed to address the current and emerging concerns associated with the ageing process in a particular geographical location, for example community-dwelling older adults in Herefordshire, England (Martin *et al.*, 2008). Although population cohorts differ between the two designs, developmental research aims to identify causes and consequences of functionality and mobility in the ageing process.

Quantitative measures of functionality and mobility are useful for clinical practice and longitudinal ageing studies. They allow objective evaluation of functional mobility status including stratification of severity and illustrate gait quality in the ageing process. There are numerous ways to gauge overall functionality. For example, rating systems (e.g. functional mobility scale (FMS)) (Graham *et al.*, 2004), timed functionality (e.g. timed up and go (TUG)) (Podsiadlo and Richardson, 1991), video analysis (Sowers *et al.*, 2006), spatial-temporal analysis (GAITRite, CIR Systems, Pennsylvania, USA) (Verlinden *et al.*, 2013) and three-dimensional analysis (Winter *et al.*, 1990). The majority of functionality and mobility assessments in older adult research focuses on gait and balance as these have a stronger association with fall risk (Stevens *et al.*, 2014). As previously stated gait is not limited to straight-line walking, it also signifies other functional tasks such as dual task walking, stepping, obstacle negotiation and turning. Furthermore, a review identified (Duffy *et al.*, 2014) that six key functionality and mobility assessments were used for the majority of longitudinal ageing studies worldwide, which were: timed up and go, NW, DT walking, stepping and/or obstacle negotiation and turning. Longitudinal ageing research is heavily reliant on spatial-temporal parameters and currently no three-dimensional analysis has evolved in longitudinal ageing research, cross-sectional research also predominantly focuses on these six functional/mobility walking tasks.

1.5. The Gait Cycle

Human gait is a complex locomotive bipedal pattern, which consists of maintaining centre of gravity in a continuous changing base of support that alternates between single- and double-support (Harris *et al.*, 2008). The gait cycle is the time interval between two consecutive occurrences of initial contact with the ipsilateral (same limb) foot (Figure 1.3.) (Perry and Burnfield, 2010). The gait cycle is divided into two phases: stance and swing. Stance phase (approximately 60 percent of the gait cycle (%GC)) designates the period of time when the foot is in contact with the ground, which begins at initial contact. Swing phase (approximately 40 %GC) is the period of time when the foot is in the air, to cause limb advancement. The swing phase initiates at toe-off. The stance is sub-divided into three-periods: initial double-support, single-support and second double-support. Double-support is the bilateral foot contact on the ground and this occurs at the start and end of the stance phase, with the middle portion (single-support) being the single foot contact. Initial double-support and the event initial contact commences the gait cycle. Single-support begins when the contralateral (opposite) foot is in the swing phase.

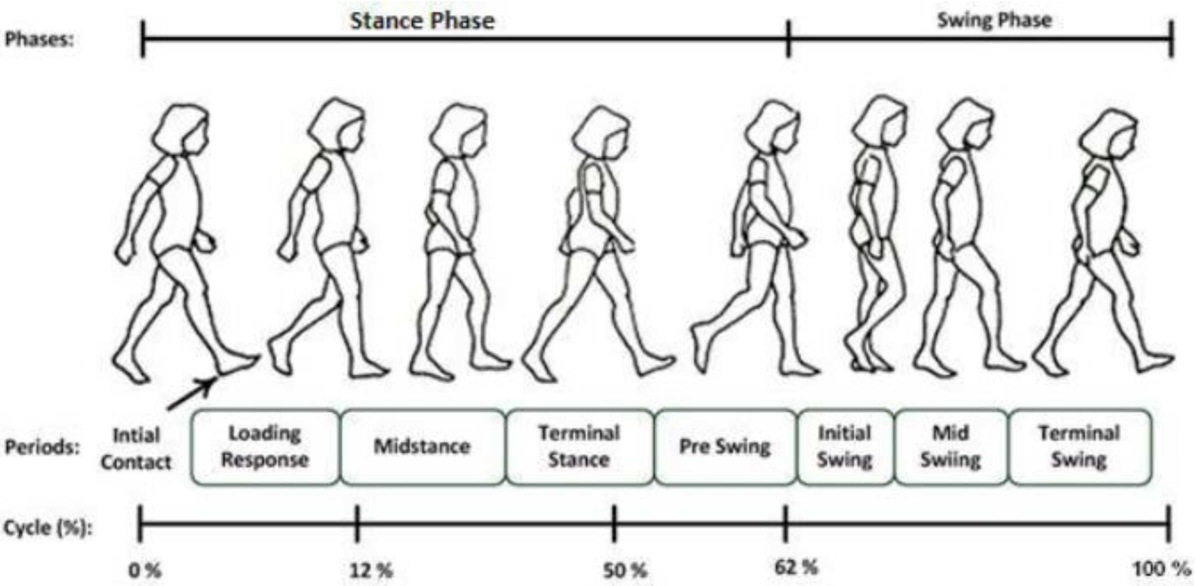


Figure 1.3. The Gait Cycle (Perry and Burnfield, 2010).

Figure 1.4. illustrates the typical joint kinematics and kinetics during normal walking (NW). Kinematics describes joint motion, without reference to forces (Levine *et al.*, 2012). Kinetics describes forces

(causes of joint ranges of motion), joint moments (a force applied away from a joint) and powers (muscle tension by the shortening velocity) (Levine *et al.*, 2012). Joint moments produce rotational accelerations, which occur when a force is exerted at a certain distance from a joint. Note the greater the distance from the joint, the greater the joint moment (Baker, 2013). Figure 1.4B. illustrates the joint moments, which represent the total moment exerted by a force, as a product of force magnitude and the perpendicular force from the joint centre (Baker, 2013). An example of a joint kinematic and kinetic during NW is the sagittal hip joint; peak hip extension and an external hip extension moment at terminal stance, with peak hip flexion and an external hip flexion moment in mid-terminal swing (Figure 1.4.). For hip power (Figure 1.4B.), loading response to midstance, energy is generated by the hip extensors to reduce hip flexion which allows hip extension single-support (limb weight acceptance); energy is then absorbed by the hip flexors to decelerate thigh rotation and at terminal stance to pre swing, energy is generated by the hip to accelerate the lower limb upward and forward to allow for toe-clearance and the next gait cycle (ipsilateral stride) (Winter *et al.*, 1990).

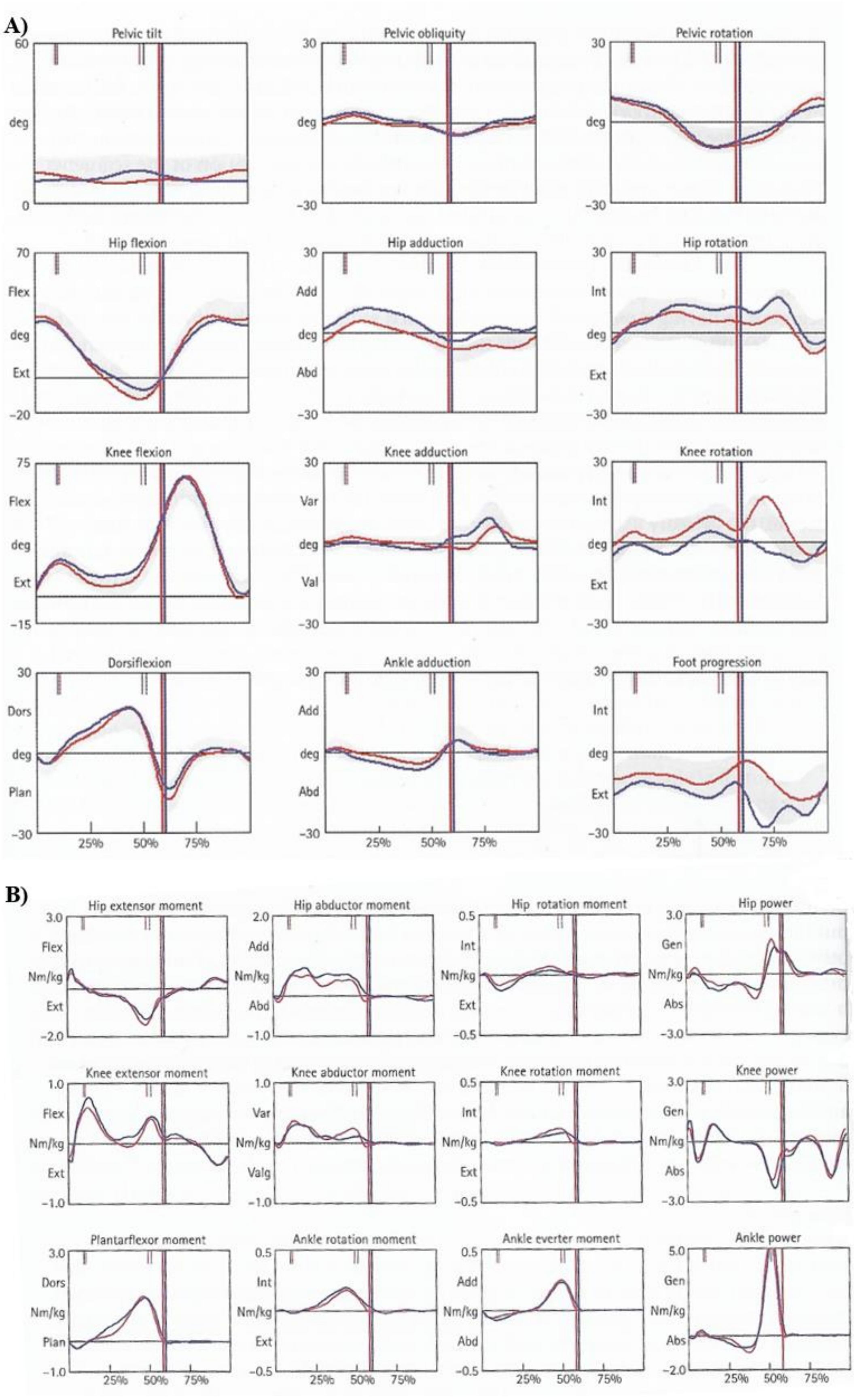


Figure 1.4. A) Joint kinematics and **B)** Joint kinetics for a single healthy person during overground normal walking (Baker, 2013). *Note:* Red line is the right limb and blue line is the left limb.

1.6. Normal Walking

Age-related gait adaptations have been identified in older adults and when compared to young adults found reduced walking speed for overground self-selected walking speed (Winter *et al.*, 1990, Kerrigan *et al.*, 1998, DeVita and Hortobagyi, 2000, Kerrigan *et al.*, 2001, Riley *et al.*, 2001, Byrne *et al.*, 2002, Monaco *et al.*, 2009, Anderson and Madigan, 2014). Walking speed has found to be associated with altered joint kinematics and kinetics in older adults (Kerrigan *et al.*, 1998, Kerrigan *et al.*, 2001, Riley *et al.*, 2001, Chung and Wang, 2010, Anderson and Madigan, 2014). Consequently, gait parameters are typically speed-dependent. Although, Alcock *et al.* (2013) revealed alterations in gait speed (reduction of 1.2 percent per year), they did not fully explain the altered gait mechanics associated with ageing. Speed-dependent variables were foot clearance, ankle plantarflexion (kinematic and moment) and hip power generation. As such, numerous joint kinematic and kinetic alterations in the ageing process are independent of gait speed, such as reduced hip extension and increased anterior pelvic tilt.

In addition, spatial-temporal parameters are altered for older adults compared to young adults, for example older adults have an increased double-support time, step time and stride width (Winter *et al.*, 1990, Elble *et al.*, 1991, Winter, 1992, Lajoie *et al.*, 1996, Begg *et al.*, 2007, Mills *et al.*, 2008, Mariani *et al.*, 2010, Schulz *et al.*, 2010) and reduced stride/step length (Winter *et al.*, 1990, Judge *et al.*, 1996, DeVita and Hortobagyi, 2000, Nutt, 2001, Paróczai *et al.*, 2006, Monaco *et al.*, 2009). This pattern is thought to be adopted as a safe 'cautious gait' strategy to reduce fall risk in older adults. In addition, a timid and reserved cautious gait pattern is also associated with excessive age-related walking changes and older adults who have a fear of falling (Arfken *et al.*, 1994, Nutt, 2001, Herman *et al.*, 2005, Pirker and Katzenschlager, 2017). For example, older adults with sensory or motor deficits who display a cautious gait pattern, typically have a reduced walking speed, wider base of support, reduced arm swing and stooped posture, which is associated with fall history (Pirker and Katzenschlager, 2017). The most excessive variant of cautious gait is known as phobic gait disorder, this affects people with extreme fear of falling and may result in inability to walk (Pirker and Katzenschlager, 2017). Older adults with no neurodegenerative disease, typically have a cautious gait pattern called dysrhythmicity, which is known as unstable gait pattern and the inability to maintain walking rhythm, that may cause or aggravate a fear of falling and lack of confidence (Herman *et al.*, 2005).

Age-related differences for older adults predominantly occur at the hip and ankle joint when compared to young adults (McGibbon, 2003, Silder *et al.*, 2008, Anderson and Madigan, 2014), for example increased anterior pelvic tilt, reduced hip extension and ankle plantarflexion power generation for older adults (Winter *et al.*, 1990, Judge *et al.*, 1996, Kerrigan *et al.*, 1998, DeVita and Hortobagyi, 2000, Kerrigan *et al.*, 2001, Riley *et al.*, 2001, Lee *et al.*, 2005, Silder *et al.*, 2008, Monaco *et al.*, 2009, Anderson and Madigan, 2014). These age-related differences may be due to ankle plantarflexion strength and hip range of motion (RoM) for older adults.

1.6.1. Research Aim

There is currently no research which has provided a normative database for older adults during overground walking using three-dimensional motion analysis. Consequently, the biomechanics of age and gait is not fully understood. Previous research has predominantly stated older adults exhibit reduced hip extension during the gait cycle compared to younger adults. Few studies have analysed gait within an older adult population (Ko *et al.*, 2010a, Ko *et al.*, 2011), however none of these have explored the effect on age. Furthermore, gait analysis for older adults is limited to the sagittal and coronal plane, none to date have investigated the effect of age in the transverse plane (e.g. hip rotation). As such, a creation of a normative database within an older adult population would allow the effects of age on gait to be examined and would not assume age can be categorised into a single group regardless of the ageing process. The creation of a normative database will direct future research and provide clinicians with evidence to determine effective interventions and rehabilitation.

1.7. Toe-Clearance

One third of adults aged 65 years and above will experience a fall each year, with 53% occurring from a trip whilst walking (Blake *et al.*, 1988, Winter *et al.*, 1990, Berg *et al.*, 1997, Scott *et al.*, 2007). An important gait parameter during walking is toe-clearance (also known as minimum toe-clearance (MTC)) (Tinetti *et al.*, 1988, Winter, 1991, Winter, 1992, Berg *et al.*, 1997, van Dieën *et al.*, 2005) and is linked to trips. Toe-clearance is the vertical height of the toe above the ground during the swing phase (Winter, 1992). MTC occurs at a critical time point during the swing phase, where not only the toe closely approaches the ground (1-2 cm above the ground) but the speed of the foot and toe is near the maximum and the body's centre of mass is located to the anterior stance foot and outside the base of support in the direction of progression (Winter, 1992) (Figure 1.5.). Therefore, at MTC the risk of a fall is at its highest (van Dieën *et al.*, 2005) and been found to be related to trip risk in older adults (Begg *et al.*, 2007, Best and Begg, 2008, Begg *et al.*, 2014).

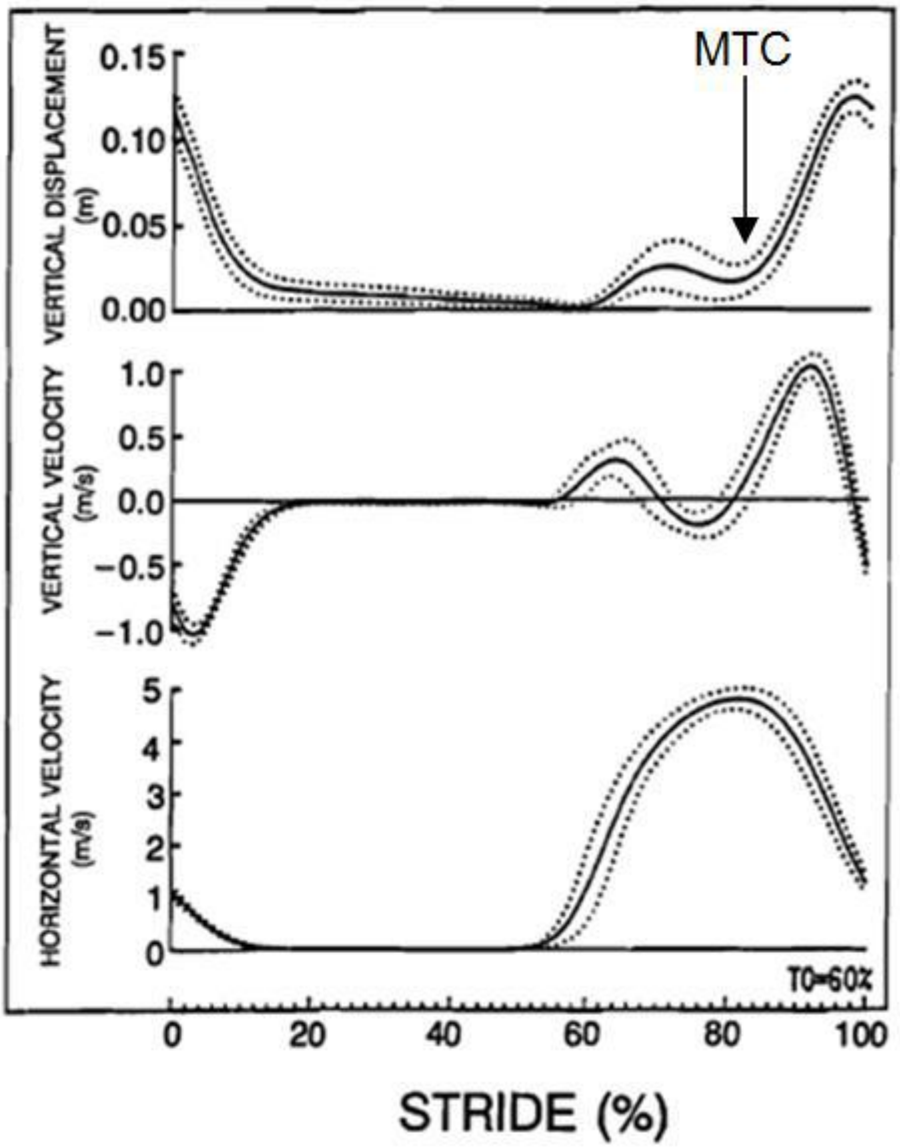


Figure 1.5. Winter *et al.* (1992) averaged displacement (vertical) and velocities (vertical and horizontal) of the toe over one gait cycle for eleven subjects (5 females; 6 males; age range 21-28 yrs (mean 24.9 yrs)). *Note:* dashed line indicates standard deviation, with the arrow highlight MTC. *Abbreviations:* minimum toe-clearance (MTC).

A systematic review (Barrett *et al.*, 2010) revealed comparing young to older adults does not reveal alterations to MTC central tendencies (e.g. mean and median) or disruptions during overground and treadmill normal walking (NW). Although the literature implies there is no age effect on central tendencies for MTC during NW the above studies compared young to older adults. Although, no age effect has been found for MTC, toe-clearance success has been associated with two additional events first maximum toe-clearance (MxT1) and second maximum toe-clearance (MxT2) during the swing phase (Nagano *et al.*, 2011). Nagano *et al.* (2011) reported only MxT2 had an age effect when comparing young to older adults during overground and treadmill NW. MxT2 occurrence coincides with peak dorsiflexion (Winter, 1991). As such, muscle weakness of the dorsiflexors may have contributed to reduced MxT2 for the older adult participants.

To date, currently only two studies have investigated manual DT walking on MTC (Schulz *et al.*, 2010, Santhiranayagam *et al.*, 2015). Schulz *et al.* (2010) investigated MTC during three DT (carrying a 9kg laundry basket, carrying a tray with cups of water and walking whilst talking). Results indicated increase in MTC vertical displacement for carrying a 9kg laundry basket, no change to carrying a tray with cups of water and reduced for walking whilst talking when compared to overground NW (Figure 1.6.). The researchers concluded DT walking maybe independent of fall risk. However, this study was undertaken in adults aged 22-58 years old, as such MTC may have altered if an older adult population had been explored. Whereas, Santhiranayagam *et al.* (2015) used a manual DT and reported no significant difference between younger (15 young adults; 4 females; 11 males; 26.1 ± 3.8 yrs) and older adults (15 older adults; 7 females, 8 males; 73.1 ± 5.6 yrs) MTC. Data capture for the walking tasks were performed on a treadmill. The walking speed for both age groups for the preferred NW task were considerably slower compared to normative overground comfortable walking speed data (Bohannon, 1997). For example, the young adults in Santhiranayagam *et al.* (2015) study had a preferred walking speed of 1.06 ± 0.14 $m{\cdot}s^{-1}$, whereas the normative walking speed data (Bohannon, 1997) found young adults (20-29 yrs) walked at comfortable speed of 1.41 ± 1.8 $m{\cdot}s^{-1}$ (females) and 1.39 ± 1.5 $m{\cdot}s^{-1}$ (males). Limitations of treadmill include not being equivalent to overground walking (Row Lazzarini and Kataras, 2016), which is clearly evident when comparing walking speed and treadmills also may artificially reduce gait variability (Chien *et al.*, 2015). Therefore, the use of a treadmill in Santhiranayagam *et al.* (2015) study may have contributed to no differences in gait patterns.

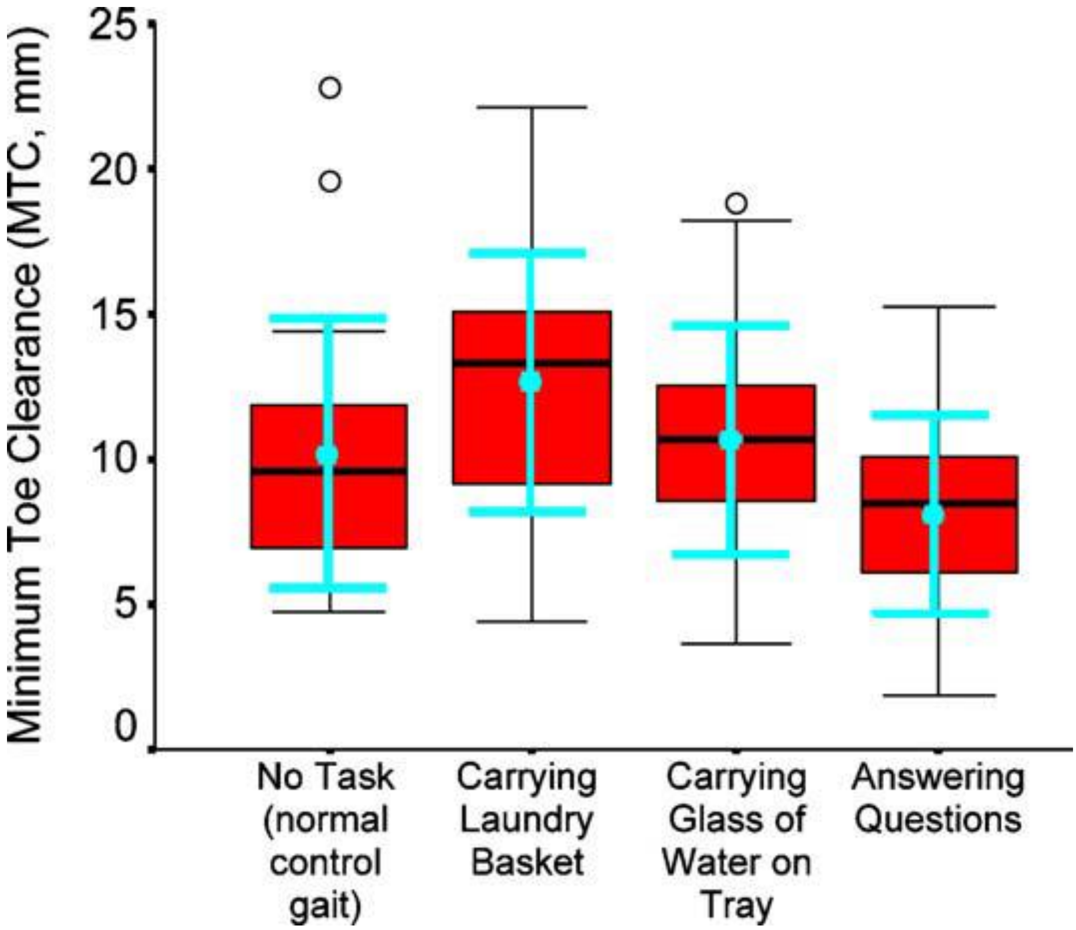


Figure 1.6. Schulz *et al.* (2010) mean minimum toe-clearance for 10 adults (5 females; 5 males; age range 22-58 yrs (44 ± 13 yrs)).

1.7.1. Research Aim

Previous research which has compared toe-clearance parameters (MTC, MxT1 and MxT2) has focused on comparing young to older adults (e.g. ≤ 25 vs. ≥ 65 yrs). Gait differences are observed year on year in older adults (Ashton-Miller, 2005). As such a different research approach would be to investigate the age effect on toe-clearance parameters within a group of older adults. In addition, manual DT walking research and the effects on toe-clearance parameters is very limited. Firstly, the current studies focus on MTC and disregard the additional toe-clearance parameters (MxT1 and MxT2). Secondly, one study is limited to adults below the age of 58 years and the other study compares young to older adults, therefore both studies disregard the ageing process. Consequently, research is required to explore whether toe-clearance parameters are affected by task and related to age when performing NW and manual DT walking.

1.8. Arm Swing

There is a paucity of upper body gait analysis, as research primarily focuses on lower body gait analysis although, arm swing is essential for efficient locomotion (Ortega *et al.*, 2008, Bruijn *et al.*, 2010). For normal walking, the pendulum-like motion of the arm swinging in opposition to the legs aids balance, by counteracting the angular momentum generated by the lower body (Elftman, 1939) and reducing the lateral displacement of the centre of mass (Ortega *et al.*, 2008). The mechanism of arm swing counteracts free vertical moments caused by the lower body (i.e. torque about the vertical axis of the body) (Pontzer *et al.*, 2009), as angular arm acceleration has found to be equal to the torso (Elftman, 1939). Arm swing has been considered to be a passive swing as a result of thoracic movement (Jackson *et al.*, 1978, Kubo *et al.*, 2004, Gutnik *et al.*, 2005). However, surface electromyography revealed arm swing is partly active (Pontzer *et al.*, 2009, Barthelemy and Nielsen, 2010, Kuhtz-Buschbeck and Jing, 2012). Shoulder muscle activity has been suggested to induct changes in arm swing direction (Barthelemy and Nielsen, 2010, Kuhtz-Buschbeck and Jing, 2012). A systematic review (Meyns *et al.*, 2013) concluded determining arm swing with muscle control extent or passive movement remains unclear.

Irrespective of the determinants of arm swing, swinging the arms during locomotion has been suggested to aid gait stability (steady-state gait with small perturbations) and energy consumption (Meyns *et al.*, 2013). Gait stability is thought to be the distinction between steady state gait and recovery ability caused by large perturbations (Meyns *et al.*, 2013). For gait stability, walking can be divided into two phases (initial and recovery phase) (Bruijn *et al.*, 2010). The initial phase depends upon a steady system and intrinsic mechanical properties, such as inertia stiffness, with the recovery dependent on active control. For example, Ortega *et al.* (2008) concluded arm swing contributes to lateral stabilisation when young and older adults were compared with and without arm swing. However, Pijnappels *et al.* (2010) and

Bruijn *et al.* (2010) revealed negative effects on arm swing during steady state gait stability (i.e. decreases in energy expenditure when walking without arm swing). These studies also found arm swing movement helps recover walking after perturbations (Bruijn *et al.*, 2010, Pijnappels *et al.*, 2010). As such, a lateral velocity mechanism of arm swing can regulate dynamic balance (Curtze *et al.*, 2011), with an increased walking speed associated with increased arm swing amplitude (range between peak flexion and extension) for young adults (Bruijn *et al.*, 2008, Liang *et al.*, 2014).

Koo and Lee (2016) investigated the use of arm swing on gait ability for forty-five healthy young adults (30 females; 15 males; aged 20.8 ± 1.6 yrs), performing three walking conditions; **1)** walking with normal arm swing, **2)** walking with a constraint on dominant arm swing and **3)** walking without arm swing (constraint on both arms). Gait ability was assessed by measuring the following parameters: walking speed, stride length, cadence, step time, single- and double-support. The results revealed walking without arm swing caused decreased walking speed, stride length and an increased cadence and double-support compared to walking with arm swing. Similar findings were also reported for Ford *et al.* (2007) when exploring arm constrains for walking in healthy young adults (10 participants; 7 females; 3 males; aged range 21-24 yrs). Walking speed reduction with arm constraints may be a result of a decreased propulsion during gait (Koo and Lee, 2016). Kubo *et al.* (2006) reported walking speed increases when young adults (10 females; 4 males; 26.8 ± 4.2 yrs) walk with an arm swing, because of the generated increased rotation between the thorax and pelvis as a consequence of walking with an arm swing. As such, the role of the upper body during gait is to aid balance and allow pelvic rotation in the transverse plane to transmit to the upper body as a compensation of rotation in the contralateral arm (Umberger, 2008).

Walking without an arm swing, increases the metabolic expense mechanisms, which causes an increased trunk muscle force generation to reduce the occurrence of excessive trunk twisting and to allow a straight gait trajectory path (Ortega *et al.*, 2008). In addition, walking without an arm swing causes increased lateral oscillations for the centre of mass (Shibukawa *et al.*, 2001). Therefore, gait requires more metabolic energy and mechanical work (Donelan *et al.*, 2001). As such, external stabilisation mechanisms compensate for walking without arm swing, to prevent increased metabolic for both young and older adults (Ortega *et al.,* 2008). External stabilisers during walking have found metabolic cost is equal regardless of walking with or without arm swing (Ortega *et al.,* 2008). External stabilisation withstands trunk twisting, which is a substitution for arm swing, for example straight-line walking without arm swing the trunk is forward facing and as such lateral force stabilisers are applied with the centre of mass to ensure no twisting moment is created (Ortega *et al.,* 2008). However, twisting moment about the vertical axis, external stabilisers are not aligned with the centre of mass, this creates a moment which twists the trunk to a forward-facing orientation (Ortega *et al.,* 2008). Therefore, external stabilisers provide a more cost effective mechanism to control for trunk movement when walking without arm swing.

Whole body kinematics and ground reaction forces were captured for twenty-one healthy young adults (age range 21-32 yrs) and twenty healthy older adults (age range 66-81 yrs) walking at 80 %, 100 % and 120 % of their preferred walking speed to calculate centre of mass accelerations and work done on the centre of mass by the limbs (Hernández *et al.,* 2009). The older adults had a reduced mediolateral centre of mass accelerations during double-support compared to the young adults, which the researchers suggested changes may also be present in the coronal joint kinetics. For example, the leading limb assists forward progression of the trailing limb through vertical support and mediolateral shift of the centre of mass (Hernández *et al.,* 2009). Consequently, the control of mediolateral accelerations during mid-terminal stance (i.e. transition from single to double-support) may be an important age-related factor. Research (Winter, 1995) suggested these age-related reductions in mediolateral centre of mass acceleration during push-off were attributed to the muscle potential of the coronal plane, for example hip adductors/abductors. Consequently, centre of mass control differs between healthy young and older adults. Future work should be considered to explore centre of mass control once the role of arm swing and the affects of age are established within an older adult population as opposed to comparing young to older adults.

Similar, to lower body gait analysis, research has primarily assessed the age effect of arm swing between young and older adults. With older adults reported to have a reduced arm swing in comparison to young adults (Elble *et al.*, 1991, Krasovsky *et al.*, 2014, Mirelman *et al.*, 2015), reduction in arm swing may increase the risk of falls amongst older adults (Mirelman *et al.*, 2015). Arm swing analysis within older adult populations is also limited. The majority of research has focused on arm swing on age-related neurodegenerative diseases such as Parkinson's disease with arm swing shown to be reduced and associated with increase fall risk (Lewek *et al.*, 2010, Plate *et al.*, 2015, Mirelman *et al.*, 2016). In addition, the effect of arm swing on dual task walking is currently limited to one study (Mirelman *et al.*, 2005), which explores the effect of a cognitive dual task (subtracting in three's) between sixty healthy adults aged thirty-three to seventy-seven. The results revealed arm swing asymmetry increased during dual task walking for the oldest group (61-77 yrs) compared to the other groups (30-40 yrs; 41-50 yrs; 51-60 yrs).

1.8.1. Research Aim

There is a paucity of research exploring older adult arm swing with research typically reporting older adults having a reduced arm swing compared to young adults during normal walking (NW). In addition, research has also reported similar findings for DT walking in older adults when compared to young adults, although current research is currently limited to cognitive DTs. Research has yet to investigate the effect on arm swing for a manual DT. A reduced arm swing has been suggested to increase fall risk, with reduced arm swing being a marker for Parkinson's disease. In addition, arm swing for DT walking has only explored the effects on cognitive tasks such as, for example, counting backwards in three's (Mirelman *et al.*, 2015). However, manual DT walking such as carrying an object whilst walking reflects a more concurrent everyday activity. In addition, walking tasks such as obstacle clearance have yet to be explored for the effects on arm swing. As such, arm swing assessment within an older adult population would be more appropriate along with evaluating arm swing during different walking challenges.

1.9. Landing Forces

Negotiating a changing environment is necessary for independent living. For an older adult performing a task such as stair descent is important for functional mobility. Tasks which involve stepping result in serious injuries amongst older adults, for example hip fractures (Garcia *et al.*, 2006, Jacobs, 2016). Consequently, ground reaction forces (GRF), have been evaluated to determine the state of gait locomotion (Jacobs *et al.*, 1972). The higher the force magnitude, for example stepping down from a curb, results in higher shock absorption and dissipated force on the musculoskeletal system, consequently, increasing the risk of joint pathology or injury (Dufek and Bates, 1990, McNitt-Gray, 1991, Irmischer *et al.*, 2004, Elvin *et al.*, 2007).

There is contradictory evidence for the effects of age on GRF during NW for older adults. Yamada and Maie (1988) investigated GRF on sixty-six male participants (23-78 years old). Older men had a reduced first and second vertical peak GRF, higher minimum mid-stance peak and reduced anterior-posterior GRF. In addition, gait alterations (walking speed, step length and GRF) occurred from the age of fifty years which suggests such gait parameters to determine the age effect should be explored from above the age of fifty. However, Toda *et al.* (2015) reported no significant difference for first peak and minimum mid-stance peak, with a significant reduction for second GRF vertical peak for older adults compared to young adults.

When compared to NW, the first vertical peak force was increased for obstacle clearance (Christina and Cavanagh, 2002). In addition, changes also occur for anterior-posterior GRF, and although braking impulse is similar to NW, propulsive impulse is lower (Christina and Cavanagh, 2002, Riener *et al.*, 2002). Older adults demonstrated a safer step gait strategy during step negotiations compared to young adults due to alterations on GRF and lower propulsion. Older adults exhibiting a reduced propulsion are considered to be displaying a more cautious gait pattern (Simoneau *et al.*, 1991), which utilises friction creation at foot contact and foot-off (Christina and Cavanagh, 2002). This may reflect an increase in joint stiffness (Christina and Cavanagh, 2002), which may increase slip likelihood.

1.9.1. *Research Aim*

The effect of age on GRF for both NW and obstacle clearance tasks has shown some contradictions in the literature. With research comparing young to older adults, again this disregards the ageing process. As such, the effects of landing forces within an older adult population are unclear.

1.10. Conclusion

Older adults' functionality is a key marker for current and future health as this provides determinants for health during the ageing process. Functional movement such as NW or obstacle clearance are important daily tasks which require systematic actions of the musculoskeletal system. Older adults have reduced musculoskeletal function resulting from physical and neuromuscular changes. Numerous gait and functional movement adaptations of older adults are attributed to spatial-temporal alterations for example, reduced walking speed. Also, alterations in joint kinematics and kinetics have been demonstrated in the ageing process, for example increased anterior pelvic tilt. Consequently, older adults have been found to strategically modify their gait pattern to potentially minimise fall risk.

The majority of research has compared young adults (20-40 years) to older adults (≥ 50 years). However, this approach assumes older adults can be categorised into a single group regardless of the ageing process. As shown in this literature review, gait pattern alterations such as reduced step length and GRF occur from the age of fifty years. Furthermore, cross-sectional studies solely analysing older adults' gait and functional movements possess limitations, such as small sample sizes (≤30 older adult participants) and comparing healthy older adults to pathological older adults (e.g. Parkinson's disease). Therefore, the extent of gait functionality within older adults is unknown.

1.11. Aims of the Current Thesis

The overall aim of this thesis was to explore the effects of age on gait and functional movement characteristics in community-dwelling older adults. As identified in this literature review, gait is a complex motor task which is not limited to straight-line walking, as it requires the ability to walk around an ever changing environment, for example obstacle clearance. As such, gait ability for older adults highlight many functions necessary for independent living. Consequently, participants performed a variety of functional walking tasks varying in task complexity on an overground walkway. This was achieved using three-dimensional motion capture. Further details can be found in Chapter Two: Methodology.

The objectives that were addressed within the chapters of this thesis were to:

1. Create a normative gait database for an older adult population.
2. Describe normal gait in older adults.
3. Explore the effects of age and/or walking speed on gait and functional walking tasks.
4. Identify whether changes to gait in older adults are a consequence of age and/or task complexity.

The specific aims of each thesis chapters and current gaps in the literature are outlined in Table 1.1.

Table 1.1. Mapping identified literature gaps onto chapter aims.

Chapter	Gaps in the Literature	Research Aims	Chapter Aims
4	• Investigated older adult gait by comparing young to older adults. • Within an older adult population, none to date have explored the effect on age, as such the biomechanics of age and gait is not fully understood. • No normative database for older adults during overground walking using three-dimensional motion analysis.	• A creation of a normative database within an older adult population to examine the age effect on gait. • A normative database in an older adult population accounts for the ageing process, as opposed to categorising into a single age group, for example older adults (65-80 years).	• The aim of this study was to examine the effects of age on gait parameters within an older adult population.
5	• Minimum toe-clearance focused on comparing young to older adults. • Additional toe-clearance parameters were yet to be investigated to determine the effect on dual task walking. • Mechanisms underlying control of minimum toe-clearance during normal and dual task walking tasks for older adults were currently unknown.	• Investigate toe-clearance parameters within an older adult population, to account for the ageing process when older adults perform normal and manual dual task walking.	• The aim of this study was to establish if toe-clearance events decreased with age and task and if the joint kinematics of the ipsilateral and contralateral limb adapt to performing a dual task. • A secondary aim was to determine if fall history affected toe-clearance parameters.
6	• Current research for older adults on the effects of arm swing have compared to young adults. • The effects of arm swing on manual dual task walking and obstacle clearance tasks have yet to be investigated. • Research currently focuses on arm swing (elbow position relative to the shoulder) and little is known on the role of forearm swing, especially in an older adult population.	• Exploring arm swing within an older adult population during different walking challenges, such as normal walking and obstacle clearance.	• The aim of this study was to explore the effect of walking task on arm swing for an older adult population. • The secondary aim of the study was to establish if walking task affected forearm swing for the older adult population.

Chapter	Gaps in the Literature	Research Aims	Chapter Aims
7	• Contradictory evidence regarding ground reaction forces, as current research compared young to older adults. • It is unclear what the biomechanical strategy older adults adopt for joint kinetics when task complexity increases, for example obstacle clearance.	• Explore the age effect in an older adult population on landing forces when performing normal walking and obstacle clearance tasks.	• The aim of this study was to determine the alterations on landing forces and joint kinetics for obstacle clearance when compared to normal walking in an older adult population.

Chapter Two: Methodology and Pilot Work

The overall methodology of this thesis is provided below. Each chapter (4-7) has a specific methodology associated with the chapter.

2.1. Research Design and Setting

The overall aim of this study was to explore the effects of age on gait and functional movement characteristics in community-dwelling older adults. A gait database was established to determine the normative effects of age on walking for a community-dwelling older adult population, which was achieved using a cross-sectional design. Each participant attended the Biomechanics Laboratory at the University of Essex once, within a four month data collection period.

2.2. Pilot Work

The aim of the study was twofold to establish: **1)** protocol feasibility and **2)** reliability of lower body marker placement for normal walking. Four healthy older adults participated, age range 55-64 years (1 female; 3 males; 59.3 ± 4.4 yrs). The pilot study demonstrated that the protocol was feasible for four walking tasks (normal, manual dual task, stepping onto and stepping over an obstacle). The turning task was not feasible due to technical limitations and therefore was excluded from the main study. Highest reliability occurred in coronal and transverse planes, with most parameters of the sagittal plane within acceptable limits. One participant was affected by misplacement of the pelvis (Appendix Three: Pilot Work).

2.3. Sample Size

Minitab 17 (Statistical Software, Coventry, UK) was used to estimate the minimum sample size at various confidence intervals required to create a normative gait database for the study. The sample size power calculation was performed using sample sizes of gait normative databases for children (Chester *et al.*, 2007, Pinzone *et al.*, 2014, Kennedy *et al.*, 2016). Forty participants were required to have a ninety-five percent confidence of an actual standard deviation within two degrees of measured standard deviation (e.g. pelvic tilt during gait) and ninety-seven participants would be required if this was reduced to within one degree (Stratford and Goldsmith, 1997, Pinzone *et al.*, 2014). Although, ninety-seven participants were required to create a normative gait database within a one degree confidence interval; the power calculation was performed using narrow child age bands (e.g. 3-5 yrs), as no current older adult normative gait databases exist. Consequently, as many participants feasible to attend the University within the allocated data collection period was the aim, in order to establish a representation of an older adult population with a range of ages (pre-retirement 55-64 yrs, 65-74 yrs and ≥ 75 yrs) and to improve the statistical power of the normative gait database. Also, barriers are encountered when recruiting older adults; for example, transportation obstacles (Ory *et al.*, 2002, Gonzalez *et al.*, 2007, Crawford Shearer *et al.*, 2010), with drop-out rates of 14.34 %. Therefore, a higher recruitment number than required allowed for any potential data collection barriers.

2.4. Participants

All participants were recruited from local communities in Essex and Suffolk and from the University of Essex. Recruitment strategies included telephone and email contact, social media, face-to-face contact with potential participants (e.g. over 50 clubs, ageing societies and community centres) and the media (Newspapers, Television and Radio interview) (East Anglian Daily Times, 2014, ITV News, 2014) (Figure 2.1.). The inclusion criteria for this study was as follows; all participants must live independently and be independent walkers (able to walk at least 10 m unaided), with no surgical procedures occurring in the last six months and be aged fifty-five years old or older. As previously stated the United Kingdom has an ageing population with an estimated 23.6 million people aged fifty years and older, which is a third of the total UK population (Office for National Statistics, 2016). It is becoming more evident that different age ranges (55-64 yrs, 65-74 yrs and ≥ 75 yrs) demonstrate different results in an older adult population (Schoenborn *et al.*, 2006, Taekema *et al.*, 2011, Poortvliet *et al.*, 2013, Ogliari *et al.*, 2015, Stijntjes *et al.*, 2016). As such, recruiting participants from the age of fifty-five years and older allows the database to assess the distribution and effect of age on gait for adults who are ‘near’ to being an older adult (55-64 yrs) and pre-retirement age, as well as older adults within the UK retirement age (65-74 yrs and ≥ 75 yrs). Although, the study research design was a cross-section normative gait database, future work was designed with a ten year longitudinal ageing study to establish the ageing effect within this data collected older adult population. Therefore, the inclusion criteria allowed for a representative sample of a community-dwelling older adult population to be included within the gait database. In addition, criteria with excessive restrictions have been found to limit sample sizes which effect statistical power and representation of the target population (Cassidy *et al.*, 2001, Yancey *et al.*, 2006). As such, a convenience sample of one-hundred and fifty-eight community-dwelling older adults, age range of 55-86 years (101 females; 57 males; 65.7 ± 6.8 yrs; 168.6 ± 9.2 cm; 74.0 ± 14.8 kg) volunteered for the study. Ethical approval was granted by the University of Essex Ethics Committee and all participants gave written informed consent.

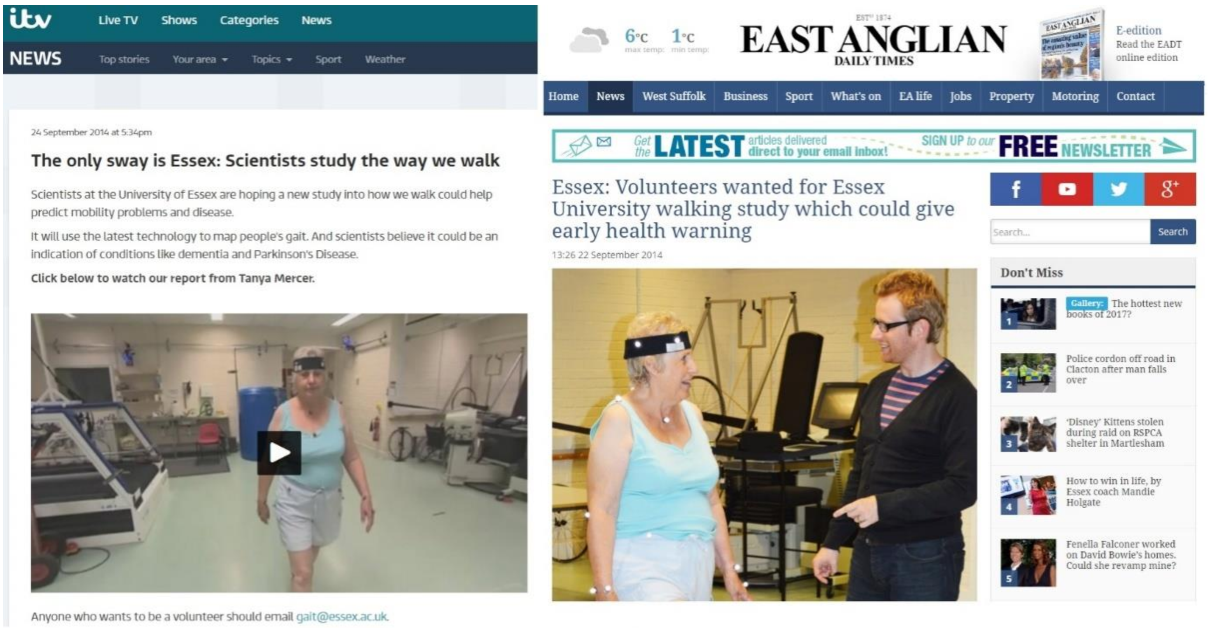
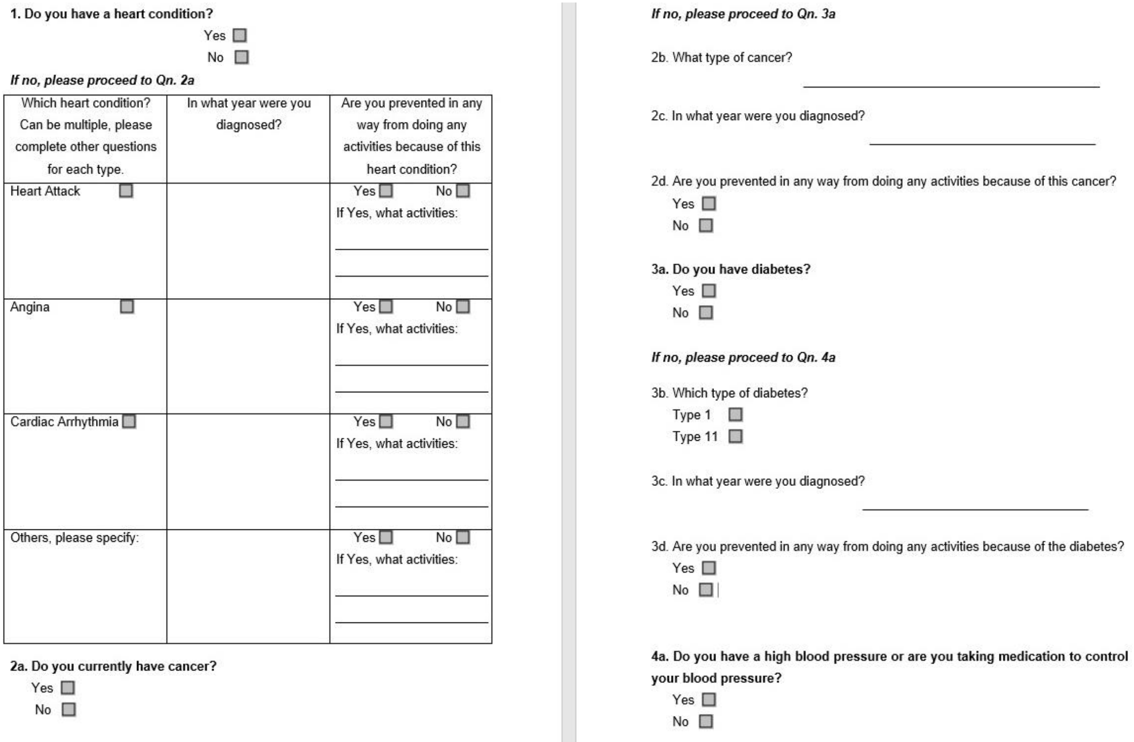


Figure 2.1. Sample of using the media to recruit participants.

2.5. <u>Questionnaire</u>

Prior to data collection, all participants were given the EAGLES (Essex Ageing and Gait Longitudinal Study) questionnaire (Figure 2.2.) which focused on health (including fall history), functionality, physical activity, leg dominance and socio-economic status (Appendix Five: EAGLES Questionnaire).

The health and socio-economic status questions were derived from pre-existing questionnaires used by longitudinal ageing studies (English Longitudinal Study of Ageing (ELSA, 2014b), Australian Longitudinal Study of Ageing (ALSA, 2014) and Baltimore Longitudinal Study of Aging (BLSA, 2014)). Physical activity and functionality questions were also derived from longitudinal ageing studies (ELSA (ELSA, 2014b) and ALSA (ALSA, 2014)) and the Community Healthy Activities Model Program for Seniors (CHAMPS) (Stewart *et al.*, 2001a) and physical activity questionnaire (Hagstromer *et al.*, 2006). The footedness questionnaire (Elias *et al.*, 1998) was used to assess participants' lateral dominance (consistent preference for the use of one side of the body) (Hanke and Tiberio, 2006). The EAGLES questionnaire provides information regarding 'who the participants are' including their general health, functionality and physical activity level.



1. Do you have a heart condition?
Yes ☐
No ☐
If no, please proceed to Qn. 2a

Which heart condition? Can be multiple, please complete other questions for each type.	In what year were you diagnosed?	Are you prevented in any way from doing any activities because of this heart condition?
Heart Attack ☐		Yes ☐ No ☐ If Yes, what activities:
Angina ☐		Yes ☐ No ☐ If Yes, what activities:
Cardiac Arrhythmia ☐		Yes ☐ No ☐ If Yes, what activities:
Others, please specify:		Yes ☐ No ☐ If Yes, what activities:

2a. Do you currently have cancer?
Yes ☐
No ☐

If no, please proceed to Qn. 3a

2b. What type of cancer?

2c. In what year were you diagnosed?

2d. Are you prevented in any way from doing any activities because of this cancer?
Yes ☐
No ☐

3a. Do you have diabetes?
Yes ☐
No ☐

If no, please proceed to Qn. 4a

3b. Which type of diabetes?
Type 1 ☐
Type 11 ☐

3c. In what year were you diagnosed?

3d. Are you prevented in any way from doing any activities because of the diabetes?
Yes ☐
No ☐

4a. Do you have a high blood pressure or are you taking medication to control your blood pressure?
Yes ☐
No ☐

Figure 2.2. Examples of the health questions used in the EAGLES (Essex Ageing and Gait Longitudinal Study) Questionnaire.

2.6. Protocol

Figure 2.3. illustrates the protocol design of the study, expansion of task details is below. A familiarisation period was provided for all tasks except the mini-mental state examination (MMSE). All participants received a minimum two-minute rest period between each task.

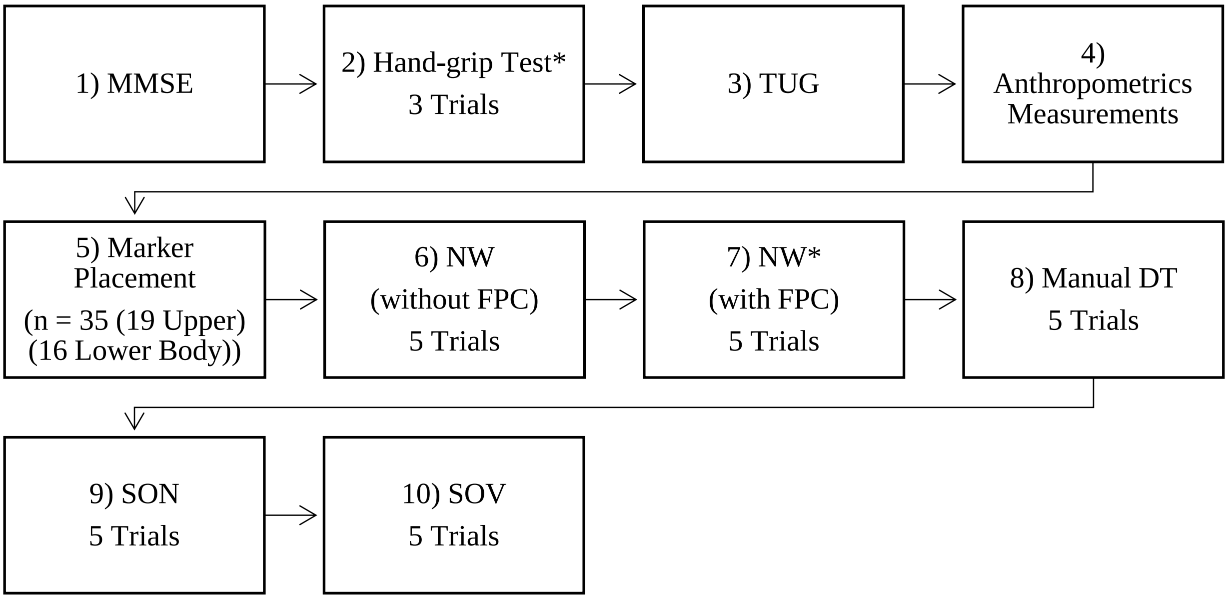


Figure 2.3. Illustration of protocol design for the study (tasks 6–10 used three-dimensional motion capture). * Trials collected for the right and left limb. *Abbreviations:* Mini-Mental State Examination (MMSE), Timed Up and Go (TUG), Normal walking (without Force Plate Contact) (NW (without FPC), Normal Walking (with Force Plate Contact) (NW (with FPC)), Manual Dual Task Walking (manual DT), Stepping Onto an Obstacle (SON) and Stepping Over an Obstacle (SOV).

2.6.1. Laboratory Set-Up

Figure 2.4. illustrates the layout of the Biomechanics laboratory for data collection for this study. A seven camera Vicon T20 infrared motion capture system (Oxford, UK) sampling at 100 Hz, with a floor-mounted Kistler 9281CA force plate (Winterthur, Switzerland) sampling at 1000 Hz were used to derive the three-dimensional motion analysis for the walking tasks. Two pairs of Brower timing gates (Utah, USA) were positioned (2.28 m apart) in the middle of a 10 m walkway (to allow participants

sufficient distance to achieve optimal walking speed) and were used to calculate the walking speed for the walking tasks. Timing gates were used as a real-time feedback aid for the researcher, to determine if participants were walking slower for the NW task with FPC compared to without FPC. If participants were slower, then the walking trial was repeated.

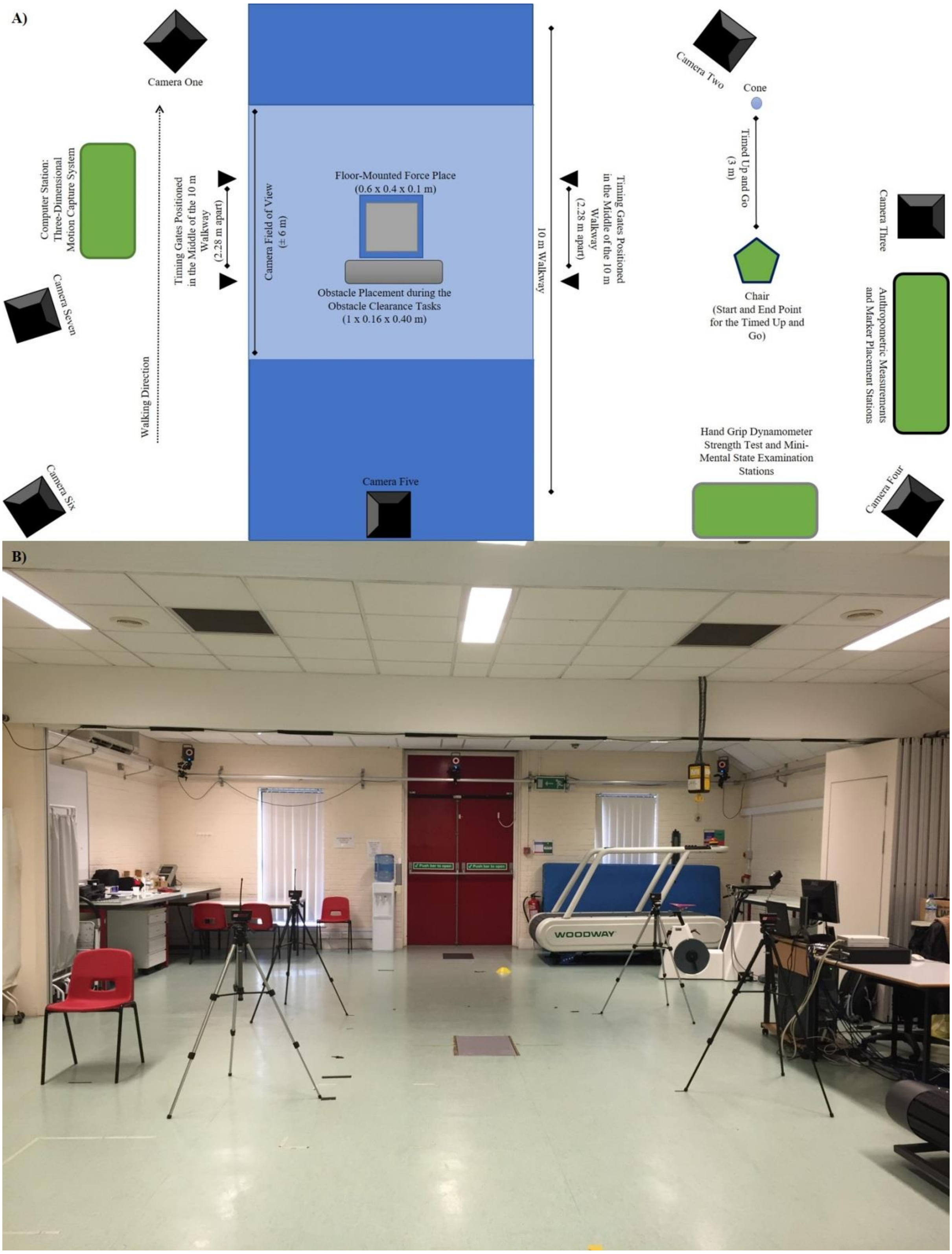


Figure 2.4. Laboratory set-up of the Biomechanics Laboratory University of Essex for data collection: **A)** layout overview for the entire study protocol including layout dimensions and **B)** photograph of the laboratory (taken from the end of the 10 m walkway).

2.6.2. *Clothing*

Participants were instructed to wear tight compressive non-reflective clothing (e.g. Lycra clothing), such as a vest top or t-shirt and a pair of short shorts, which would minimise extraneous movement (e.g. marker movement artefacts). Participants were also instructed to wear comfortable footwear that reflected their everyday use, for example footwear participants use to go to the supermarket. Overall, participants typical wore flat shoes with none or minimal sole wedge (e.g. trainers). Clinical studies (Oeffinger *et al.*, 1999, Menant *et al.*, 2008, Wolf *et al.*, 2008, Zhang *et al.*, 2013) typically capture gait barefoot, however footwear provides protection against surface abrasions and infections from mechanical debris (Squadrone *et al.*, 2009, Menant *et al.*, 2008), with a systematic review suggesting older adults should wear footwear with low heels and firm slip-resistant soles for inside and outside the home (Menant *et al.*, 2008). Walking in footwear has a significant impact on gait parameters, which is associated with increased walking speed, stride length and dorsiflexion at heel contact for example (Oeffinger *et al.*, 1999, Wolf *et al.*, 2008, Lythgo *et al.*, 2009, Moreno-Hernandez *et al.*, 2010, Wirth *et al.*, 2011, Tsai and Lin, 2013, Zhang *et al.*, 2013). This older adult population were habitually accommodated to wearing footwear, as such barefoot walking would not be considered normal. Therefore, wearing footwear in this study reflects a more real-world setting.

2.6.3. *Mini-Mental State Examination*

Prior to the functional and walking tasks, all participants completed a cognitive mental status examination called the MMSE (Folstein *et al.*, 1975). MMSE is widely used with older adults to determine cognitive change and dementia (Tombaugh and McIntyre, 1992, Harvan and Cotter, 2006, Hotte *et al.*, 2010, Moraes *et al.*, 2010). The MMSE includes eleven questions on mental function (Appendix Three: MMSE (Mini-Mental State Examination)), which takes ≤ 10 minutes to conduct and is scored immediately by the researcher. The MMSE is divided into two sections (totalling a score of 30): **1)** vocal responses only revolving orientation, memory and attention (out of 21) and **2)** assesses the ability to name, follow verbal and written commands, write a sentence spontaneously and copy a polygon shape (out of 9). Table 2.1. reveals the interpretation of MMSE score, with research suggesting a cut-off score of 24 as normal cognitive function and a score ≤ 24 cognitive decline (Folstein *et al.*, 1975, Hensel *et al.*, 2007, Stein *et al.*, 2012). Any participants scoring ≤ 24 were advised to make an appointment with their GP for a formal cognition screening.

Table 2.1. Interpretation of the Mini-Mental State Examination Score (Folstein *et al.*, 1975).

Score	Degree of Impairment	Formal Psychometric Assessment	Day-to-Day Functioning
25-30	Within normal range	Not required.	If score was towards the lower range (score of 25) a mild deficit may occur. However, this is only likely to affect highly demanding activities of daily living.
20-24	Mild	Formal assessment may be helpful to determine pattern and extent of deficits.	May require the need for support, supervision and assistance.
10-19	Moderate	Formal assessment may be helpful if specific clinical indications are present.	Clear impairment. May require 24-hour support and supervision.
0-10	Severe	Not testable.	Clear impairment. Likely to require 24-hour support and supervision.

2.6.4. Functional Measures

To establish baseline functionality, two simple functionality measures were collected. These were hand-grip test and timed up and go (TUG). All participants reported their dominant hand (the hand the participants write with). The hand-grip test was conducted on the participants' dominant hand first then repeated on their non-dominant hand. All participants performed the test using the Takei Hand-Grip Dynamometer Analogue 5001 (Niigata, Japan) in a standardised protocol (American Society of Hand Therapists clinical recommendations) (Fess, 1992). Standard instructions (via the same researcher) in the same verbal command were provided to all participants to minimise performance influence (Fess, 1992), but ensuring maximum force was reached. Three trials were performed for each hand; with a 15 second rest between each trial and all participants alternated hands between each trial (starting with their dominant hand), in accordance with the literature (Mathiowetz, 1990, Harth and Vetter, 1994, Hanten *et al.*, 1999, Werle *et al.*, 2009). Hand-grip strength was recorded in kg to the nearest 0.1 kg.

The TUG was recorded in seconds (to the nearest 0.1 s), using an iPhone stopwatch application (iPhone 5, California, USA). All participants were instructed to stand-up from the chair (same chair for all participants and the chair did not have arms to assist standing), walk 3 metres (at a self-selected normal walking speed), turn around a cone and walk to the chair and sit down. Participants performed the TUG once. The TUG is recommended for not only screening falls (≥ 13.5 s) (Panel on Prevention of Falls in Older Persons and British Geriatrics, 2011, Barry *et al.*, 2014), but also an indicator for current health (Barry *et al.*, 2014).

2.6.5. Anthropometric Measurements

Anthropometric measurements were obtained for all participants to aid the Plug-in Gait Marker Model (PiG) scaling and biomechanical modelling of segments (Vicon, 2010). The following anthropometric measurements are required for the PiG: height (mm), body mass (kg), upper body measurement – shoulder offset (mm), elbow width (mm), wrist width (mm) and hand thickness (mm) and lower body measurements – leg length (mm), knee width (mm) and ankle width (mm) (Table 2.2.). PiG automatically calculates the following anthropometrics: anterior superior iliac spine (ASIS) trochanter distance (mm), thigh rotation offset (°) and shank rotation offset (°) (Vicon, 2010).

Table 2.2. Description of the required anthropometric measurements for the Plug-in Gait Marker Model (Vicon, 2010).

Anthropometric Measurements	Description
Height (mm)	Measured the height of all participants using SECA stadiometer (Hamburg, Germany) to the nearest 0.01 m. Although height is a required input for the Vicon motion capture system, it is not required for the Plug-in Gait Marker Model.
Body Mass (kg)	Body mass of the all participants was recorded using SECA scales (Hamburg, Germany) to the nearest 0.1 kg. The Plug-in Gait Marker Model required the body mass measurement to determine kinetics.
Upper Body Measurements:	All participants were measured standing and all measurements were recorded for the right and left limb.
Shoulder Offset (mm)	With a calliper to the nearest 0.01 m, the vertical distance from the base of the acromion to the shoulder joint was measured.
Elbow Width (mm)	With a calliper to the nearest 0.01 m, the distance between the distal epicondyles of the humerus along the flexion axis was measured.
Wrist Width (mm)	With a calliper to the nearest 0.01 m, the anterior to posterior thickness of the wrist between the distal head of the ulna and radius was measured.
Hand Thickness (mm)	With a calliper to the nearest 0.01 m, the anterior to posterior thickness between the dorsum and palmar surfaces of the hand was measured.

Anthropometric Measurements	**Description**
Lower Body Measurements:	All participants were measured standing and all measurements were recorded for the right and left limb.
Leg Length (mm)	With a tape measure to the nearest 0.01 m, the distance between the anterior superior iliac spine and medial malleolus, via the knee joint was measured.
Anterior Superior Iliac Spine Trochanter Distance (mm)	This is automatically calculated by the Plug-in Gait Marker Model using the following equation (Vicon, 2010): $distance\ (mm) = \left(0.1288 * leg\ length\ (mm)\right) - 45.56$
Thigh Rotation Offset (°)	This is automatically inputted by the Plug-in Gait Marker Model as zero, as the model assumes the thigh marker has been placed exactly in the sagittal plane between the hip and knee joint centre.
Knee Width (mm)	With a calliper to the nearest 0.01 m, the distance between the lateral and medial femoral epicondyles was measured.
Shank Rotation Offset (°)	This is automatically inputted by the Plug-in Gait Marker model as zero, as the model assumes the tibia marker is placed exactly in the sagittal plane between the knee and ankle joint centre.
Ankle Width (mm)	With a calliper to the nearest 0.01, the distance across the malleoli was measured.

2.6.6. *Three-dimensional Motion Capture*

2.6.6.1. *Calibration Procedure*

Prior to each data collection, all cameras were checked to ensure all markers were detectable in camera field of view (Figure 2.4.). A dynamic calibration was performed by waving the Vicon wand (five reflective spheres (14 mm with a thread of 3 mm) attached, (Oxford, UK)) to determine capture volume (dynamic calibration took place in the camera field of view starting at the floor-mounted force-plate (Kistler 9281CA, Winterthur, Switzerland) (Figure 2.3.)). Then a static calibration was performed by placing the Vicon wand (Oxford, UK) on the top right corner of the floor-mounted force plate ((capture volume origin) Kistler 9281CA, Winterthur, Switzerland) to determine the laboratory global coordinates of the walkway. Residual error of less than 2 mm was accepted for each camera.

2.6.6.2. *Plug-in Gait Marker Model*

Thirty-five passive reflective markers (14 mm with a 3 mm thread) were placed on the upper (n = 19) and lower (n = 16) body in accordance to the PiG (Vicon, 2010) (Figure 2.5., Table 2.3.). Marker movement artefacts are highly probable as the skin is shown to move as much as 25 mm over the skeleton during gait, due to the inherent elastic properties and muscle bulk shape changes under the skin (Macleod and Morris, 1987). Accurate marker placement was required to minimise marker and skin/clothing movement artefacts and, reduce misplacement errors. For example, 5 mm misplacement of the lateral epicondyle knee marker equates to a 2-degree angle error (Szczerbik and Kalinowska, 2011). To ensure accurate marker placement by the researcher, pilot work was executed to determine the inter-rater reliability of marker placement prior to data collection (Chapter Two: Methodology 2.2. Pilot Work). An explanation of the PiG model is given in Appendix One, with implications of using the PiG model for gait analysis in Appendix Two.

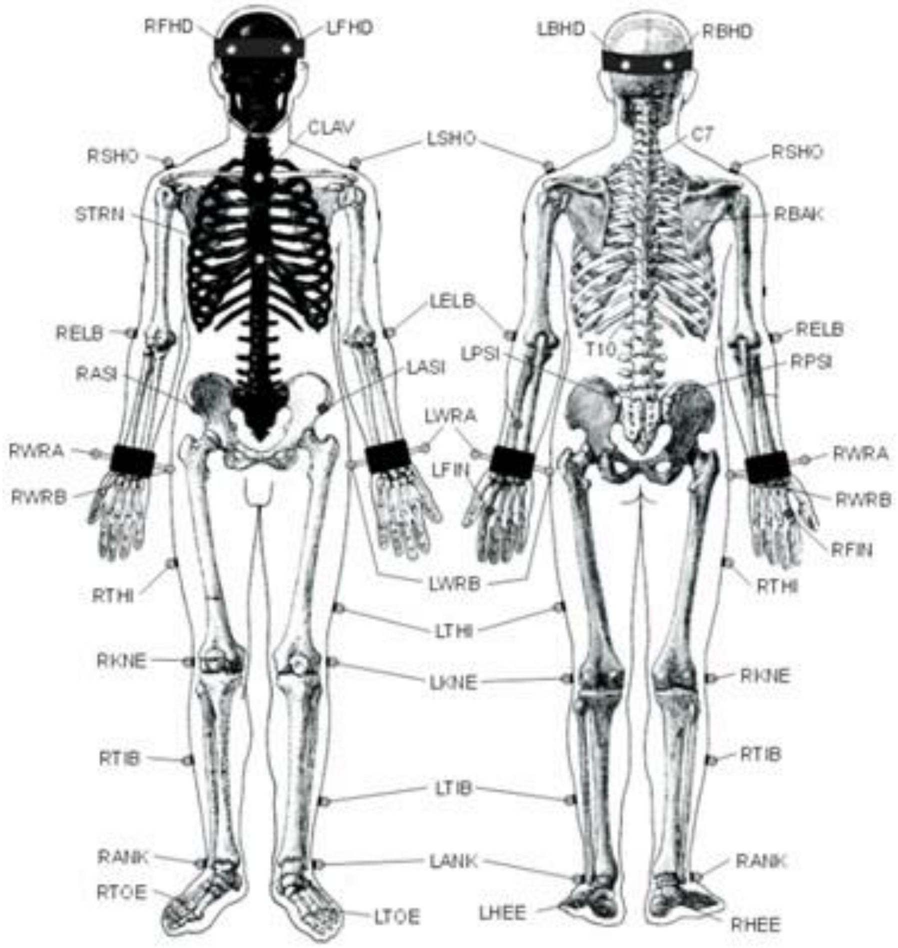


Figure 2.5. Marker placement for the Plug-in Gait Marker Model (Vicon, 2010). *Table 2.3. for full names of marker abbreviations.*

Table 2.3. Plug-in Gait marker placement (Vicon, 2010).

Marker Placement	Description
Upper Body Markers	
Head Markers RFHD/LFHD – right and left front head markers RBHD/LBHD – right and left back head markers	The four head markers were fixed to a headband to ensure the rear markers are level with the front markers. The headband is then placed around the middle of the frontal bone and the occipital bone for each participant. The front head markers are positioned over the temple on each side of the frontal bone, with back head markers applied horizontal to the front head markers.
Torso Markers	All torso markers were applied onto the participants' vest top or t-shirt using double-sided body tape.
C7 – 7^{th} cervical vertebrae	Marker was placed on the spinous process of the 7^{th} cervical vertebrae.
T10 – 10^{th} thoracic vertebrae	Marker was placed on the spinous process of the 10^{th} thoracic vertebrae.
CLAV – clavicle	Marker was placed on the jugular notch where the clavicle meets the sternum.
STRN – sternum	Marker was placed on the xiphoid process of the sternum.
RBAK – right back	Marker was placed on the right mid-scapula.
Arm Markers	All arm markers were applied directly onto the skin for each participant using double-sided body tape. If a participant wore a t-shirt the sleeves were rolled up in a similar position of a vest top and taped to prevent the sleeves from moving.
RSHO/LSHO – right and left shoulder	Markers were placed on the acromioclavicular joints.
RELB/LELB – right and left elbow	Markers were placed on the lateral epicondyles of the right and left humerus.
RWRA/LWRA – right and left wrist at the distal radius	Markers were placed on the styloid process of the right and left radius.
RWRB/LWRB – right and left wrist at the distal ulna	Markers were placed on the styloid process of the right and left ulna.
RFIN/LFIN – right and left finger	Markers were placed on the shaft of the right and left second metacarpals.

Marker Placement	Description
Lower Body Markers	
Pelvis Markers	To minimise marker movement around the pelvis, all participants were instructed to tuck their vest top or t-shirt into their shorts. The pelvis markers were applied onto the shorts for each participant.
RASI/LASI – right and left anterior superior iliac spine	Markers were placed directly over the right and left anterior superior iliac spine.
RPSI/LPSI – right and left posterior superior iliac spine	Markers were placed directly over the right and left posterior superior iliac spine.
Leg Markers	
RKNE/LKNE – right and left lateral epicondyle of the knee	Markers were placed onto the skin directly over the lateral epicondyle of the right and left knee.
RTHI/LTHI – right and left thigh	Depending on the participants' shorts length and if the shorts could be made shorter, the markers were either placed directly onto the skin or onto the participants' shorts. The right and left thigh markers were placed off the belly of the vastus lateralis muscle. To ensure alignment between the greater trochanter and the lateral epicondyle of the knee, the greater trochanter was located and from the hand moved two widths down towards the lateral epicondyle of the knee. Once two hand widths down, place the marker off the belly of the vastus lateralis muscle.
RANK/LANK – right and left lateral malleolus of the ankle	Markers were placed onto the skin directly over the lateral malleolus of the right and left ankle.
RTIB/LTIB – right and left tibia	Markers were placed onto the skin off the belly of the right and left tibialis anterior muscle. To ensure alignment between the lateral epicondyle of the knee and the lateral malleolus of the ankle, the marker was placed half-way off the belly of the tibialis anterior muscle.
Foot Markers	All foot markers were applied directly onto the participants' footwear.
RTOE/LTOE – right and left toe placed on the second metatarsal	Markers were placed on the head of the second metatarsal for the right and left toe.
RHEE/LHEE – right and left heel	Markers were placed on the right and left calcaneous, ensuring horizontal alignment between the heel marker and the toe marker.

2.6.6.3. Walking Tasks

Following the static trial, participants were familiarised with the 10 m walkway and each walking task (Figure 2.3.): 1) NW (without force plate contact), 2) NW (with force plate contact – right contact then left contact), 3) manual dual task (DT) walking and obstacle clearance – 4) stepping onto an obstacle (SON) and 5) stepping over an obstacle (SOV). Due to the methodological limitations associated with speed-controlled studies, for example difficulty in generalising findings (Astephen Wilson, 2012), it was decided not to control walking speed. Instead, participants were instructed to walk 'at their preferred walking speed'. For DT walking participants held a full cup of water (200 ml, in their dominant hand) and were instructed to walk without spilling the water. To date, no standardised manual dual task has been proposed (Asai *et al.*, 2014). As such, this task was chosen as it replicates a real-world setting. For the obstacle clearance tasks (SON and SOV) the obstacle (Reebok Stepper (100 x 16 x 40 cm), Adidas Group, Herzogenaurach, Germany)) was placed horizontally before the force plate on the 10 m walkway (Figure 2.4.), with reflective markers placed on all corners of the obstacle. Participants were instructed to step onto the obstacle then off onto the force plate for SON and step over the obstacle and step onto the force plate for the SOV walking tasks. No instruction was given regarding leading leg for the obstacle clearance tasks; participants self-selected. All walking tasks were recorded using three-dimensional motion capture, with five trials for each walking task. Kinematic analysis was recorded for all tasks with kinetic analysis occurring for normal walking (with force plate contact) and the obstacle clearance tasks.

2.7. Data Processing

Processing of all trials for all walking tasks was performed using Vicon Nexus (v 1.8.5, Oxford, UK). Reconstruction of the markers and auto-labelling of marker trajectories were performed. Each trial was then visually inspected and unlabelled marker trajectories were manually labelled. Gaps in marker trajectories of up to 10 sample frames were joined with linear interpolation filtered with a quintic spline filter (Woltring; mean square error of 10). This cut-off frequency was selected to attenuate noise without distorting high-frequency marker movement at heel contact (Sinclair *et al.*, 2013b). A low-pass 4^{th} order

Butterworth filter at 10 Hz was applied to the force plate data (raw analogue signal). For NW (with force plate contact) and the obstacle clearance tasks gait cycle events of initial contact (on the force plate) and toe-off (on the force plate) were identified using a Nexus sub-routine which checks for the crossing threshold value (10 N) of the amplitude of the vertical component of the ground reaction force when the ankle and toe markers lie within the bounds of the force plate. Visual inspection was used to verify these events and manual gait cycle events (heel contact and toe-off) occurred for all NW (without force plate contact) and DT walking, obstacle clearance tasks prior to the obstacle and next initial contact for NW (with force plate contact) and obstacle clearance tasks. Gait cycle events which were manually identified used frame by frame visual inspection of the lowest trajectory frame (closet to the ground) of the heel marker for heel contact and the next frame after the lowest trajectory frame (closet to the ground) of the toe marker for toe-off. The dynamic PiG model was then applied and PiG bones, gait cycle events, marker trajectories, joint kinematics and kinetics were exported using ASCII files in a .csv format. Joint kinetic modelling was calculated using the local coordinate frame of the distal segment in the hierarchical kinetic chain (Vicon, 2010), as such Vicon Nexus (v 1.8.5, Oxford, UK) exports PiG as external moments (e.g. sagittal hip moment: flexion = positive and extension = negative).

2.8. <u>Data and Statistical Analysis</u>

Processed walking trials for all walking tasks were exported into Vicon Polygon (v 4.3.1, Oxford, UK) to generate the spatial-temporal parameters (cadence (steps/min), step time (s), stride time (s), double-support time (s), single-support time (s), limp index (s), foot-off (percentage of the gait cycle (%GC)), opposite foot contact (%GC), opposite foot-off (%GC), walking speed ($m \cdot s^{-1}$), step length (m), stride length (m) and step width (m)) for each task, which were subsequently exported in a .csv format (Table 2.4.).

Table 2.4. Definition of calculated spatial-temporal parameters (Vicon, 2017).

Spatial-Temporal Parameter	Definition
Cadence	Strides per minute. Right and left cadence was calculated for a single stride.
Step Time	Time between contralateral and the ipsilateral foot contact.
Stride Time	Time between ipsilateral foot strikes.
Double-support Time	Time between ipsilateral foot contact to contralateral foot-off and contralateral foot contact to ipsilateral foot-off.
Single-support Time	Time between contralateral foot-off and contralateral foot contact.
Limp Index	The time the ipsilateral foot is on the ground and divides it by the time the contralateral foot is on the ground during ipsilateral gait cycle.
Foot-off	Percentage of the gait cycle of the ipsilateral foot-off.
Opposite Foot Contact	Percentage of the gait cycle of the contralateral initial contact.
Opposite Foot-off	Percentage of the gait cycle of the contralateral foot-off.
Walking Speed	Stride length divided by stride time.
Step Length	Distance from the ipsilateral toe marker to the contralateral toe marker.
Stride Length	Distance from the ipsilateral toe marker position at first ipsilateral foot contact and second contact.
Step Width	Distance from the contralateral toe marker position onto the ipsilateral first foot contact and second foot contact.

Questionnaire responses were coded (0 = no and 1 = yes) (e.g. do you have a cardiovascular condition? Answer = 1), to allow for subsequent statistical analysis in Excel (Microsoft Office 2010, Tokyo, Japan). The Compendium of Physical Activities were developed from physical activity and survey results of observational studies, which codes physical activity metabolic equivalent intensity (MET) levels (Ainsworth *et al.*, 1993). The MET ratio of metabolic work rate to metabolic rest rate is defined as $1.0 \cdot kg^{-1} \cdot h^{-1}$ with 1.0 MET considered resting metabolic rate at quiet sitting. MET physical activity levels range from 0.9 sleeping to 18 running at 10.9 mph METs (Ainsworth *et al.*, 2000). Physical activity for this study was measured using METs, similar to CHAMPS (Stewart *et al.*, 2001a) and physical activity questionnaires (Hagstromer *et al.*, 2006). MET intensity level for physical activity as light (< 3.0 METs), moderate (3.0-6.0 METs) and vigorous (> 6.0 METs) (Pate *et al.*, 1995).

Walking speed ($m \cdot s^{-1}$) for all walking tasks were derived from the Brower timing gates (Utah, USA), these were calculated in Excel (Microsoft Office 2010, Tokyo, Japan) using the following equation:

$$Walking\ speed\ (m \cdot s^{-1}) = \frac{distance\ between\ timing\ gates\ (2.28\ m)}{time\ to\ walk\ through\ the\ timing\ gates\ (s)}$$

All remaining data analysis was completed in custom-made Python code (Python v. 2.7.10, Delaware, USA) or Excel (Microsoft Office 2010, Tokyo, Japan). Statistical analysis was performed using IBM SPSS (v. 23, Chicago, USA) and Python code (Python v. 2.7.10, Delaware, USA) for whole and group analysis. Specific chapter data and statistical analysis refer to the associated chapter (4-7).

Chapter Three: The Older Adult Population

Summary

Demographics: One-hundred and fifty-eight community-dwelling older adults, age range 55-86 years (101 females; 57 males; 65.7 ± 6.8 yrs) were recruited into this population. Within the population, sixty-nine participants (44 females; 25 males) were grouped into the 55-64 years age group, seventy-three participants (46 females; 27 males) were grouped into the 65-74 years age group and sixteen participants (11 females; 5 males) were grouped into the over 75 years age group. The main current employment status for this population was retired (108 older adults), which is similar to other longitudinal ageing studies, for example the English Longitudinal Study of Ageing.

Cognitive Function: All participants scored within the normal cognitive function range (25-30) in accordance to the mini-mental state examination, mean score of 29 ± 1. Older adults aged 75 years and above had a reduced cognitive function score compared to the 55-64 years and 65-74 years age groups (28 ± 2 vs. 29 ± 1 and 29 ± 2). Similar findings were also reported by longitudinal ageing studies, for example The Irish Longitudinal Study on Ageing (median score of 29 for the mini-mental state examination).

Functional Measures: An increase in age was associated with an increased timed up and go. For example, the over 75 years age group (9.0 ± 1.8 s) had a similar time to that reported for community-dwelling older adults above the age of 70 years. Similarly, normal walking speed (1.34 ± 0.18 $m \cdot s^{-1}$) for this population was comparable to that reported in the literature. However, compared to longitudinal ageing studies this study had a notable faster walking speed. For example, the English Longitudinal Ageing Study reported a walking speed of 1.01 ± 0.3 $m \cdot s^{-1}$ for older adults aged 60-64 years.

Self-Reported Health: On average older adults perceived themselves as having excellent health. Thirty-six percent of the population reported having arthritis. Similar results were reported in the English Longitudinal Ageing Study. Nineteen percent of the population encountered at least one fall in the last 12 months. Fall occurrence was most commonly reported for the 55-64 years age group. This

finding contradicts the English Longitudinal Study of Ageing, which reported for their population (≥ 50 yrs) an increase in age was associated with an increase in falls.

Self-Reported Mobility: All age groups reported some form of difficulty in performing everyday tasks. For example, difficulty in crouching was reported for 36 % of this population. Compared to the English Longitudinal Study of Ageing reported higher rates of difficulty in performing everyday tasks (e.g. around 50 % of their population).

Physical Activity: Health government guidelines, suggest older adults should participant in 2 and a half hours each week moderate intensity physical activity (3.0-6.0 METs (metabolic equivalent intensity)), for example gardening. The majority of this population reported spending, either 1-3 or 3-6 hours performing everyday tasks such as gardening (1.5-6.0 METs). In addition, walking for errands (91 % of the population) and exercise (79 % of the population) was performed at least one hour each week. Therefore, for physical activity, typically this population met the health government guidelines.

Conclusion: As such, this older adult population were relatively healthy and high-functioning and for the most part was comparable to other ageing population studies.

3.1. Participant Demographics

Table 3.1. describes the participant characteristics for this population. One-hundred and fifty-eight community-dwelling older adults, age range 55-86 years (101 females; 57 males; 65.7 ± 6.8 yrs) were recruited into this study. The majority of participants in this older adult population were female, which coincides with ageing research studies for older adult participation (Anderson *et al.,* 2016, ELSA, 2016). Overall, males in older adult research are typically underrepresented (Batra *et al.*, 2012, Melchior *et al.,* 2014, Ory *et al.,* 2015, Smith *et al.,* 2015), which are associated with barriers such as research activity and male gender roles (Anderson *et al.,* 2016).

Within the population, sixty-nine participants (44 females; 25 males) were grouped into the 55-64 years age group, seventy-three participants (46 females; 27 males) were grouped into the 65-74 years age group and sixteen participants (11 females; 5 males) were grouped into the over 75 years age group.

The Longitudinal Aging Study Amsterdam (Stijntjes *et al.*, 2016) also recruited participants aged 55 years and above and use similar age group breakdowns to this population (55-64 yrs, 65-74 yrs, 75-85 yrs). Whereas, the English Longitudinal Study of Ageing (ELSA) (ELSA, 2016) and The Irish Longitudinal Study on Ageing (TILDA) (Cronin *et al.*, 2013) recruited participants aged 50 years and above and typically use decade or sub-decade age group breakdowns, for example 50-60 years and 60-64 years.

One-hundred and fifty older adults, age range 55-86 years (96 females; 54 males; 65.7 ± 6.5 yrs) responded to the Essex Ageing and Gait Longitudinal Study (EAGLES) Questionnaire, with a no-response from eight older adults, age range 55-77 years (5 females; 3 males; 66.1 ± 10.8 yrs). The main current employment status for this population was retired (108 older adults) (Table 3.1.), which coincides with longitudinal ageing studies such as the ELSA (ELSA, 2016).

Table 3.1. The older adult population; participant characteristics.

	Whole Group N = 158	55-64 yrs N = 69	65-74 yrs N = 73	≥ 75 yrs N = 16
Females	101	44	46	11
Males	57	25	27	5
Employment Status				
Working	41	30	11	0
Unable to Work	1	1	0	0
Retired	108	34	60	14
Body Mass (kg)	74.0 ± 14.8	74.1 ± 16.0	73.5 ± 14.5	75.6 ± 11.2
Height (cm)	168.6 ± 9.2	169.6 ± 9.2	168.1 ± 9.6	166.8 ± 6.3
BMI ($kg.m^{-2}$)	25.9 ± 4.3	25.7 ± 4.8	25.9 ± 3.7	27.3 ± 4.5
MMSE	29 ± 1	29 ± 1	29 ± 2	28 ± 2
Hand-grip (kg)				
Dominant	31.0 ± 11.0	32.8 ± 10.3	29.7 ± 9.3	29.3 ± 18.6
Non-dominant	28.8 ± 9.5	31.0 ± 10.3	28.0 ± 8.7	23.1 ± 6.1
TUG (s)	7.8 ± 1.4	7.5 ± 1.2	7.8 ± 1.3	9.0 ± 1.8
Footedness				
Equal	33	15	13	5
Right Always	64	24	32	8
Right Usually	46	24	21	1
Left Always	3	1	2	-
Left Usually	4	1	3	-
Walking Speed ($m \cdot s^{-1}$)				
NW (without FPC)	1.34 ± 0.18	1.36 ± 0.17	1.35 ± 0.17	1.19 ± 0.21

Abbreviation: Body Mass Index (BMI), Mini-Mental State Examination (MMSE), Timed Up and Go (TUG), Normal Walking without Force Plate Contact (NW (without FPC).

3.2. Anthropometric Measures

Table 3.1. shows the mean body mass index (BMI) for the entire older adult population and age groups. The overall mean BMI for this population was 25.9 ± 4.3 $kg.m^{-2}$, which was lower than the (ELSA) older adult population (recruited older adults 50 years and above) (females: 28.5 $kg.m^{-2}$ and males: $kg.m^{-2}$) (ELSA, 2016). For this population, an increase in age was associated with an increase in BMI, which contradicts ELSA's findings. Although, reported BMI for all age groups (55-64 yrs: 25.7 ± 4.8 $kg.m^{-2}$; 65-74 yrs: 25.9 ± 3.7 $kg.m^{-2}$; ≥ 75 yrs: 27.3 ± 4.5 $kg.m^{-2}$) was lower than ELSA's age groups (55-59 yrs: 28.7 (females) and 28.5 (males) $kg.m^{-2}$; 60-64 yrs: 28.8 (females and males) $kg.m^{-2}$; 65-69 yrs: 28.7 (females) and 28.5 (males) $kg.m^{-2}$; 70-74 yrs: 28.5 (females) and 28.3 (males) $kg.m^{-2}$; 75-79 yrs: 28.5 (females) and 27.8 (males) kg/m^2; ≥ 80 yrs: 27.3 (females) and 27.4 (males) $kg.m^{-2}$) (ELSA,

2016), both older adult populations on average are categorised as overweight; in accordance to the ELSA BMI guidelines (ELSA, 2016).

3.3. Cognitive Function

All participants scored within the normal range (25-30) for the mini-mental state examination (MMSE) (Folstein *et al.*, 1975), mean score was 29 ± 1 for this older adult population. An increase in age was associated with a decrease in cognitive function. The over 75 age group on average scored 28 ± 2, which was lower than the 55-64 years and 65-74 years age groups (29 ± 1 and 29 ± 2) (Table 3.1.). Similar findings were also reported by ELSA (ELSA, 2016). The Irish Longitudinal Study on Ageing (TILDA) (recruited older adults 50 years and above) reported a median MMSE of 29, with the entire cohort scoring between 28-30 (Cronin *et al.*, 2013).

3.4. Functional Measures

3.4.1. *Hand-Grip Strength*

Average grip strength for this population was 31.0 ± 11.0 kg (dominant) and 28.8 ± 9.5 kg (non-dominant), which was higher than TILDA's population (24.5 ± 14.0 kg) (Cronin *et al.*, 2013). However, TILDA only report the overall average hand-grip strength as opposed to specifying the grip strength for both the dominant and non-dominant hand. Strength declined with age for both the dominant (55-64 yrs: 32.8 ± 10.3 kg; 65-74 yrs: 29.7 ± 9.3 kg; ≥ 75 yrs: 29.3 ± 18.6 kg) and non-dominant hand (55-64 yrs: 31.0 ± 10.3 kg; 65-74 yrs: 28.0 ± 8.7 kg; ≥ 75 yrs: 23.1 ± 6.1 kg) (Table 3.1.). Note the rate of decline was greater for the non-dominant hand, especially for the over 75 years age group (Figure 3.1. and Table 3.1.). Dodds *et al.* (2014) combined grip strength measurements (using a hand dynamometer) of 49,964 participants (age range from 5 to ≥ 95 yrs) from twelve population studies in the Great Britain, for example ELSA (age range 52-89 yrs), Hertfordshire Cohort Study (age range 59-73 yrs) and Hertfordshire Ageing Study (age range 63-73 yrs). The normative values for grip strength were divided into sub decades (e.g. age 5, 10 and 15 yrs) by gender. As such, direct comparisons are not possible. However, similar to this population, grip strength declines (rate of decline starts from the age of 50 yrs).

For example, females aged 75 had a mean grip strength of 21.4 ± 5.4 kg, which is similar to the over 75 years group of this population (23.1 ± 6.1 kg).

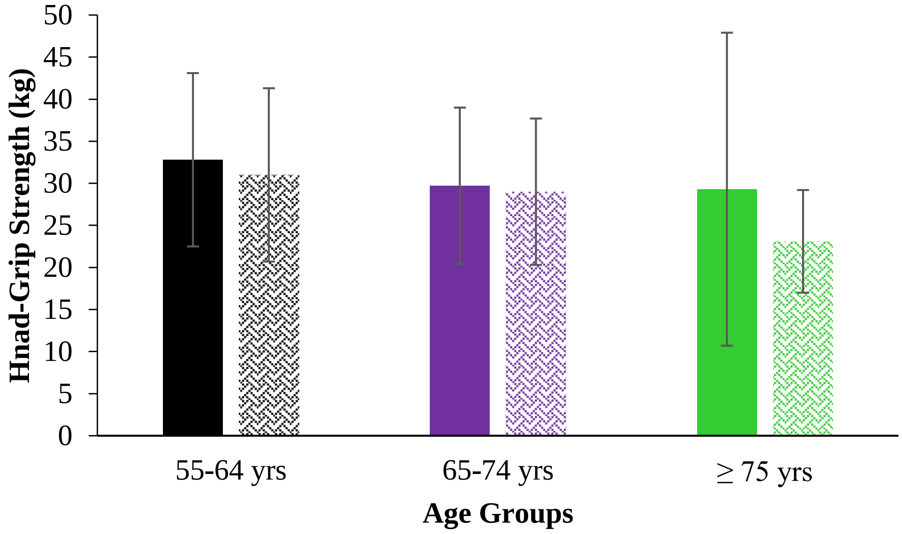


Figure 3.1. Hand-grip strength (mean ± SD) for this older adult population (n = 158) for the dominant and non-dominant hand. Age groups (55-64 yrs: n = 69; 65-74 yrs: n = 73 and ≥ 75 yrs: n = 16). *Note:* solid colour indicates dominant hand and patterned filled colour indicates non-dominant hand.

3.4.2. Timed Up and Go

An increase in age was also associated with an increased timed up and go (TUG) (55-64 yrs: 7.5 ± 1.2 s; 65-74 yrs: 7.8 ± 1.3 s; ≥ 75 yrs: 9.0 ± 1.8 s) (Figure 3.2. and Table 3.1.). For the over 75 years age group, TUG was comparable to community-dwelling older adult over the age of 70 (9.0 ± 1.8 s vs. 9.5 ± 1.7 s) (Srygley *et al.*, 2009). Overall TUG average for this population was 7.8 ± 1.4 seconds, which was comparable to TILDA's population (8.3 ± 2.1 s) (Cronin *et al.*, 2013). However, this population is notably faster in the TUG, compared to other community-dwelling older adult populations (≥ 50 years), which reported medians from 9.0-13.1 seconds (IQR 2.1-7.8 seconds) (Austin *etal.*, 2007, Yamada and Ichihashi, 2010, de Moraes *et al.*, 2011).

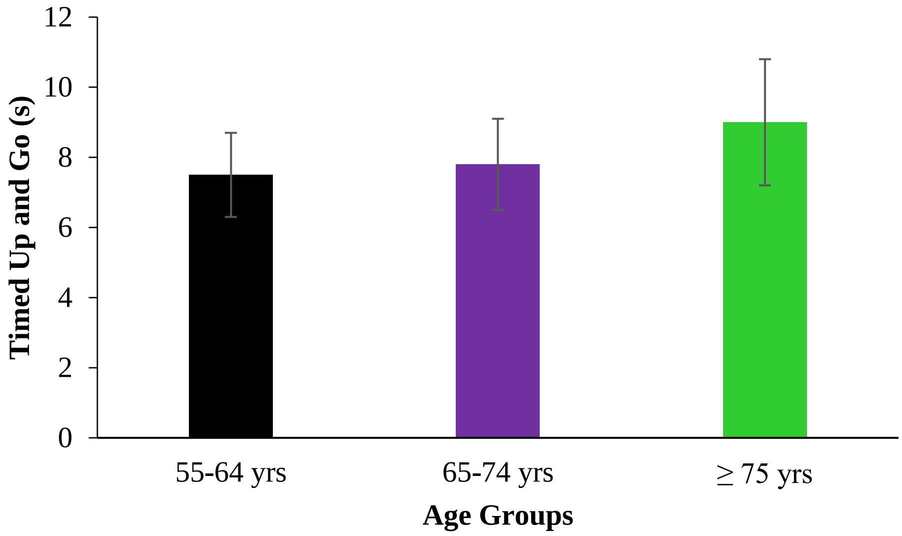


Figure 3.2. Timed Up and Go (mean ± SD) for this older adult population (n = 158). Age groups (55-64 yrs: n = 69; 65-74 yrs: n = 73 and ≥ 75 yrs: n = 16).

3.4.3. <u>Normal Walking Speed</u>

Normal walking speed (1.34 ± 0.18 $m \cdot s^{-1}$) for this population (Table 3.1.) was comparable to that reported in the literature (Shumway-Cook *et al.*, 2007, Bohannon and Williams Andrews, 2011). It was however notably faster for all age groups compared to eight community-dwelling older adult populations (ageing studies) (Cooper *et al.*, 2011a). For example, walking speed reported from ELSA was 1.01 ± 0.3 $m \cdot s^{-1}$ for the 60-64 years age group. The older adult populations reported in (Cooper *et al.*, 2011a) used walkways ranging from 2.4-6 m and all used a standing start. Whereas, a rolling start was used to measure walking speed at the mid-point of the walkway, which is likely to explain the faster speed identified. Although, the older adult population reported in Cooper *et al.* (2011a) have a notably slower walking speed compared to this older adult population, they also showed an increase in age demonstrates a slower walking speed. For example, this population (Figure 3.3.) the over 75 years had a walking speed of 1.19 ± 0.21 $m \cdot s^{-1}$ compared to 1.36 ± 0.17 $m \cdot s^{-1}$ and ELSA the 75-79 years had a walking speed of 0.81 ± 0.2 $m \cdot s^{-1}$ compared to 1.01 ± 0.3 $m \cdot s^{-1}$ for the 60-64 years age group.

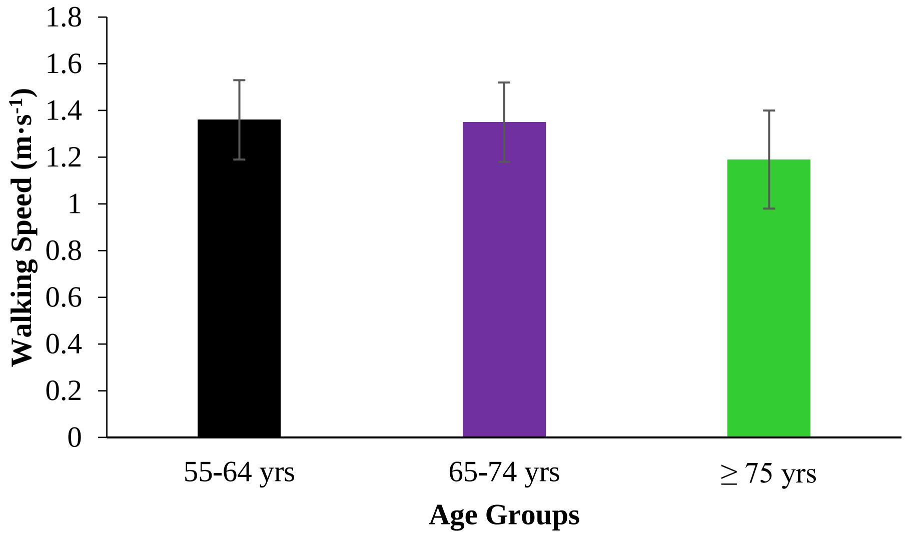


Figure 3.3. Normal walking speed (mean ± SD) for this older adult population (n = 158). Age groups (55-64 yrs: n = 69; 65-74 yrs: n = 73 and ≥ 75 yrs: n = 16).

3.5. Self-Reported Health

Table 3.2. shows the participants responses to the health section of the EAGLES Questionnaire. Health perception was scored out of 100 (ELSA, 2014a); 0-20 = perceived poor health, 21-40 = perceived fair health, 41-60 = perceived good health, 61-80 = perceived very good health and 81-100 = perceived excellent health. On average, both the whole and age groups, the older adults perceived themselves as having excellent health, for example entire population self-rated score of 82 ± 15. The majority of the ELSA population self-rated their health as good or very good (ELSA, 2016), with the TILDA population self-rating as good, very good or excellent for the entire cohort (Cronin *et al.*, 2013).

In accordance with the health government guidelines for older adults (Office of Disease Prevention and Health Promotion, 2017a), older adults are a varied population in terms of health and physical activity (e.g. loss of fitness with age). Typically, an older adult may have one or more chronic conditions which will vary in severity (Office of Disease Prevention and Health Promotion, 2017a).

3.5.1. Health Conditions

For this population, 36 % reported having arthritis (Table 3.2.) and the highest reported age group was the 65-74 years (32 participants with arthritis out of 71 participants) (Figure 3.4. and Table 3.2.). The ELSA population also reported arthritis as the most commonly reported condition (45.4 % for women and 30.3 % for men) (ELSA, 2016). In the UK, osteoarthritis treatment occurs in 33 % of middle-aged older adults (≥ 45 yrs), with 49 % of women and 42 % of men over the age of 75 years seeking treatment (Arthritis Research UK, 2014). The next most commonly reported condition for this population was blood pressure (28 %) and closely followed by cholesterol (25 %), with the highest reported age group being the 65-74 years (Table 3.2.). Blood pressure has found to increase throughout an adult's lifespan (Lewington *et al.*, 2002). Approximately, 38 % of females and 45 % of males between the age of 65-69 years of age in England have been diagnosed with hypertension (Age UK, 2015b). Compared to this statistic, this population only had 23 older adults (15 % of the population) aged between 65-74 years reporting high blood pressure.

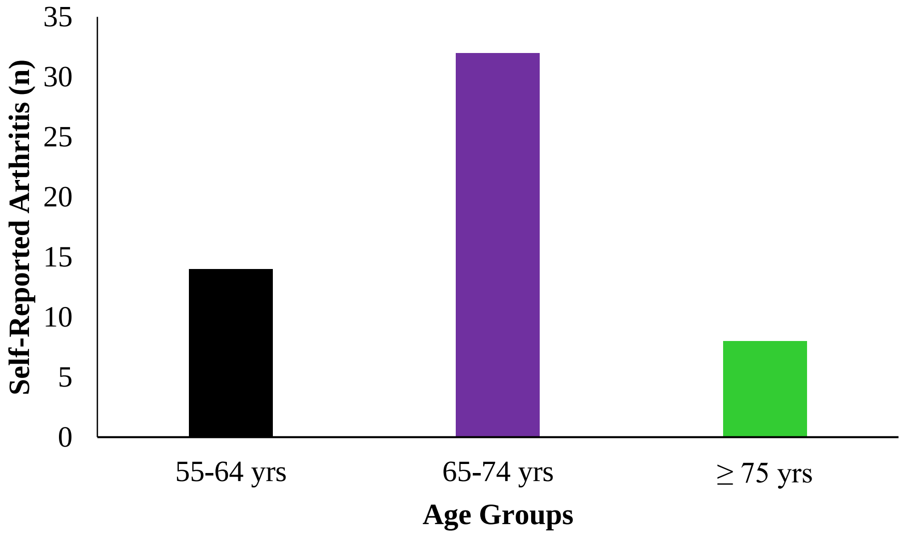


Figure 3.4. Self-reported arthritis (results from EAGLES (Essex Ageing and Gait Longitudinal Study) Questionnaire) for this older adult population (n = 150). Age groups (55-64 yrs: n = 65; 65-74 yrs: n = 71 and ≥ 75 yrs: n = 14).

3.5.2. Medication

The literature reports, older adults have 1-3 forms of medication per day (Srygley *et al.*, 2009, Donoghue *et al.*, 2013), which was comparable to this population (average 2 per day) (Table 3.2.).

3.5.3. Falls

For this population, 19 % reported encountering at least one fall in the last 12 months. The majority (18 participants) reported only one fall. One participant in the 55-64 years age group reported 12 falls in the last 12 months (Table 3.2.). The TILDA population also reported low fall rates (Donoghue *et al.*, 2013). However, ELSA reported 32.0 % of the population had fallen within the last two years (ELSA, 2016). The ELSA (2016) population also illustrated an increase in age associated with an increase in fall rating, for example 25.6 % 60-64 years to 47.3 % 80 years and above. This was not found within our population, as most falls were reported within the 55-64 years age group (Figure 3.5.).

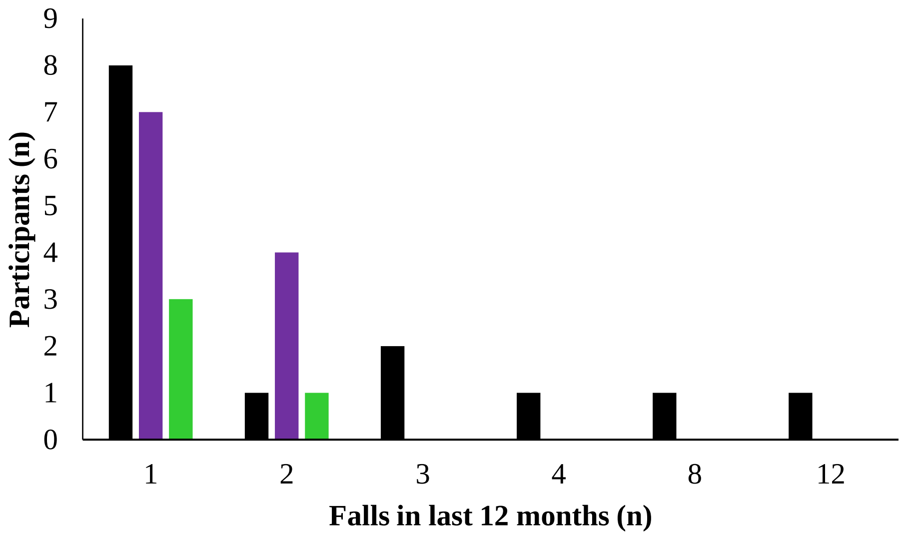


Figure 3.5. Number of reported falls in the last 12 months (results from EAGLES (Essex Ageing and Gait Longitudinal Study) Questionnaire for this older adult population (n =150). Age groups (55-64 yrs: n = 65; 65-74 yrs: n = 71 and ≥ 75 yrs: n = 14). *Note:* 55-64 yrs (black line), 65-74 yrs (purple line) and ≥ 75 yrs (green line).

3.5.4. Musculoskeletal

Twelve participants reported undergoing surgery in the last 12 months. Eleven participants reported having a joint replacement, with eleven participants reported having screws and plates within joints (Table 3.2.). This figure was relatively low in comparison to the National Joint Registry (2016), for example 70,000 total knee joint replacements occur for older adults aged 65 years and above in England and Wales.

3.5.5. Hearing and Vision

Twelve percent of the population reported wearing a hearing aid, with the 65-74 years age group identified as the most common group. The ELSA population (ELSA, 2016) has similar results, in addition researchers reported an increase in hearing aids worn with age. One hundred and forty-five participants reported wearing glasses for either reading, distance or both. Twelve percentage of the population have been diagnosed with a cataract, with the majority reported within the 65-74 years age group (Table 3.2.). In comparison to ELSA (Whillans and Nazroo, 2016) reported 20.6 % of their population had cataracts with moderate visual impairment.

3.5.6. Lifestyle Behaviours

Only three participants reported to be current smokers, with both (ELSA, 2016) and TILDA (Cronin *et al.*, 2013) studies reporting similar findings for their older adult populations. An increase in age was associated with a decrease in smoking. Forty-four percent of the population reported to drink a few times a week. Rate of alcohol consumption did however decrease with age. Highest rate of alcohol consumption was reported in the 55-64 years age group. Fourteen percent reported consuming alcohol daily (Table 3.2.), similar findings have been reported in the literature (ELSA, 2016).

Table 3.2. Participant responses to the health section of the EAGLES Questionnaire.

	Whole Group N = 150	55-64 yrs N = 65	65-74 yrs N = 71	≥ 75 yrs N = 14
Health Perception	82 ± 15	82 ± 16	83 ± 13	82 ± 18
Cardiovascular 9%	14	5	7	2
Cancer 1%	2	-	1	1
Diabetes 3%	5	2	3	-
Blood Pressure 28%	42	13	23	6
Cholesterol 25%	37	11	20	6
Osteoporosis 6%	9	2	5	2
Arthritis 36%	54	14	32	8
Respiratory 9%	13	4	8	1
Stroke 5%	7	1	5	1
Parkinson's Disease <1 %	1	-	1	-
Medication (per day) 68%				
1	22	12	6	4
2	23	10	10	3
3	21	4	15	2
4	10	3	7	-
5	8	2	5	1
6	7	3	4	-
7	5	1	3	1
8	4	1	2	1
9	1	-	1	-
12	1	-	1	-
Falls (last 12 months) 19%				
1	18	8	7	3
2	6	1	4	1
3	2	2	-	-
4	1	1	-	-
8	1	1	-	-
12	1	1	-	-
Musculoskeletal				
Surgery (last 12 months) 8%	12	4	7	1
Joint Replacements 7%	11	3	4	4
Screws and Plates 6%	9	4	5	-
Current Injuries 13%	19	9	9	1
Previous Injuries 35%	52	28	20	4
Previous Fractures 3%	4	1	2	1
Hearing Loss (hearing aid) 13%	19	1	13	5
Vision (glasses) 96%	145	64	67	14
Cataracts (current/previous) 37%	5/13	-/1	4/10	1/2
Loss of Sight 1%	2	1	-	1
Macular Degeneration 1%	2	-	2	-
Glaucoma (previous) 3%	4	1	2	1

	Whole Group N = 150	55-64 yrs N = 65	65-74 yrs N = 71	≥ 75 yrs N = 14
Smoking (per day) 2%				
9	1	1	-	-
10	1	1	-	-
12	1	-	1	-
Alcohol Consumption 91%				
Daily	22	10	12	-
Few Times a Week	67	33	27	7
Weekly	15	11	4	-
Monthly	12	3	7	2
Less than Monthly	20	3	13	4

3.6. Self-Reported Physical Activity

3.6.1. Mobility

Six participants (3 participants (55-64 years) and 3 participants (65-74 years)) reported they were unable to walk half a mile unaided, with two participants (1 participant (55-64 years) and 1 participant (≥ 75 years)) reported climbing stairs required assistance. On average, older adults reported climbing 6-10 stairs before 6pm and 1-5 stairs after 6pm on weekdays and weekends. All age groups reported difficulty in performing everyday tasks such as pulling/pushing objects and crouching, with the majority rating these tasks as a little or some difficulty. The 65-74 years age groups reported the most difficulty in performing such tasks, for example 42 participants reported having a little difficulty in reaching/lifting above the head, compared to 6 participants in the 55-64 years age group and 0 in the over 75 years age group. Two participants (1 participant (65-74 years) and 1 participant (≥75 years)) reported they were unable to perform a task (crouching and reaching/lifting above the head) (Table 3.3.). Difficulty in crouching accounted for 36 % of this population, whereas ELSA (2016) reported high rates (58.2 % women and 47.3 % men). In addition, the ELSA population reported the prevalence of mobility difficulty increased with age (ELSA, 2016). However, this was not observed for this older adult population.

3.6.2. Transportation

Walking (2.0-5.0 metabolic equivalent intensity (METs)) was reported as the most common (86 %) mode of transport for journeys less than 1 mile. With the car (1.0 MET) reported as the most common (91 %) for journeys more than 1 mile. However, 19 % of the 55-64 years age group reported walking 1-5 miles. The least common was cycling (4.0-8.0 METs) (< 1 % of the population) (Table 6.3.). ELSA (2016) reported the car was the most popular form of transport (81.7 % women and 87.8 % men), with a low response for public transport. The researchers did however report an increased use of public transport for older adults aged 70 years and above, this population also found similar findings (Table 3.3.).

3.6.3. Television Viewing

The number of hours of television viewed per week was similar across age groups. The majority watched 1-2 hours before 6 pm and 2-3 hours after 6 pm weekdays and weekends (Table 3.3). Whereas, (ELSA, 2016) reported average television views of 15-16 hours per week. As such, this population watches more television.

3.6.4. Activities at Home

According to the health government guidelines (Office of Disease Prevention and Health Promotion, 2017a), older adults should participate in 2 and a half hours each week moderate intensity physical activity (3.0-6.0 METs) (Pate *et al.*, 1995), for example gardening. Reported activities at home ranged from 1.0-6.0 METs. The majority of this population reported spending, either 1-3 or 3-6 hours per week performing activities such as preparing food (2.0-2.5 METs), food/clothes shopping (2.3 METs), cleaning the house (2.3-4.0 METs) and gardening (1.5-6.0 METs). As such, this older adult population typically met the moderate intensity physical activity guidelines and MET intensity level (Pate *et al.*, 1995). Twenty-five percent of the population did however report a minimum of 6-10 hours a week using a computer (1.5 METs) (Table 3.3.).

3.6.5. Physical Activity

As previously stated, the health government guidelines (Office of Disease Prevention and Health Promotion, 2017a), highlighted older adults should participate in 2 and a half hours moderate intensity (3.0-6.0 METs) physical activity each week. Alternatively, the guidelines (Office of Disease Prevention and Health Promotion, 2017a), state older adults should participate in an hour and fifteen minutes of vigorous intensity (> 6.0 METs) (Pate *et al.*, 1995) physical activity per week, for example swimming. Reported physical activities ranged from 1.0-18.0 METs. Thirty-three percent of the population performed aerobic classes (3.0-10.0 METs), with the majority participating for 1-3 hours per week. Low participation was found for moderate-vigorous activities such as running (4.5-18.0 METs) and swimming (4.0-11.0 METs), which accounted for 21 % and 23 % of the population. However, walking for errands (2.0-12.0 METs), ninety-one percent of the population undertook a minimum of one hour per week, with forty-six percent of those older adults walking 1-3 hours a week. Similarly, seventy-nine percent of the population walked for exercise (2.5-9.0 METs) for at least one hour per week. Unlike, other forms of exercise this activity was rated the most common for older adults over 75 years of age (12 out of 14 participants) (Table 3.3.).

Correspondingly, the ELSA (2016) study highlighted a small portion of the population was physically inactive (22.3 women and 15.0 % men), with an increase in age demonstrating a decrease in physical activity. Whereas, the TILDA study reported low rates of physical activity amongst population (Cronin *et al.*, 2013), with Stewart *et al.* (2001b) identified out of 173 community-dwelling older adults only 20 % of women and 25 % of men met the physical activity guidelines. Therefore, for this population the majority of older adults met the physical activity guidelines (Office of Disease Prevention and Health Promotion, 2017a).

Table 3.3. Participant responses to the physical activity section of the EAGLES Questionnaire.

	Whole Group N = 150	55-64 yrs N = 65	65-74 yrs N = 71	≥ 75 yrs N = 14
Walk ½ mile				
Able 96%	144	62	68	14
Unable 4%	6	3	3	-
Climb stairs				
Without Assistance 99%	148	64	71	13
With Assistance 1%	2	1	-	1
Stairs Weekday Before 6pm (Number)				
1-5 3%	45	27	16	2
6-10 33%	50	16	29	5
11-15 10%	15	5	9	1
16-20 4%	6	2	3	1
> 20 1%	2	2	-	-
Stairs Weekday After 6pm				
None 1%	2	-	2	-
1-5 62%	93	42	43	8
6-10 15%	22	9	12	1
11-15 <1%	1	-	1	-
Stairs Weekend Before 6pm				
1-5 29%	43	22	17	4
6-10 36%	54	21	29	4
11-15 9%	13	5	8	-
16-20 5%	7	3	3	1
> 20 <1%	1	-	1	-
Stairs Weekend After 6pm				
None 1%	2	-	2	-
1-5 61%	91	40	43	8
6-10 14%	21	11	9	1
11-15 3%	4	-	4	-

	Whole Group N = 150	**55-64 yrs** N = 65	**65-74 yrs** N = 71	**≥ 75 yrs** N = 14
Difficulty with everyday activities				
Pulling/Pushing Objects				
A Little 16%	24	11	31	3
Some 8%	12	3	10	2
A Lot 3%	4	3	7	1
Crouching				
A Little 37%	55	22	28	5
Some 8%	12	5	3	4
A Lot 5%	8	5	2	1
Unable <1%	1	-	1	-
Lifting 4kg				
A Little 23%	34	15	16	3
Some 3%	4	2	2	-
A Lot 5%	7	2	4	1
Reaching/Lifting Above the Head				
A Little 6%	10	6	42	-
Some 3%	4	2	4	-
A Lot 1%	2	1	2	1
Unable <1%	1	-	-	1
Writing/Handling Small Things				
A Little 12%	18	3	14	1
Some 2%	3	-	2	1
A Lot <1%	1	-	-	1

	Whole Group N = 150	55-64 yrs N = 65	65-74 yrs N = 71	≥ 75 yrs N = 14
Transportation				
Car (1.0 MET)				
Less than 1 mile 11%	16	8	6	2
1-5 miles 41%	62	25	30	7
5+ miles 91%	136	61	62	13
Walk (2.0-5.0 METs)				
Less than 1 mile 86%	129	57	62	10
1-5 miles 35%	53	28	22	3
5+ miles 3%	4	4	-	-
Public Transportation (1.0 MET)				
Less than 1 mile 1%	2	-	1	1
1-5 miles 20%	30	8	18	4
5+ miles 10%	15	4	10	1
Cycle (4.0-8.0 METs)				
Less than 1 mile 3%	5	1	3	1
1-5 miles 6%	9	5	4	-
5+ miles 1%	2	-	2	-

	Whole Group N = 150	55-64 yrs N = 65	65-74 yrs N = 71	≥ 75 yrs N = 14
Television Viewing (1.0 MET)				
Television 99%	148	63	71	14
No Television <1%	1	1	-	-
Television Viewing Weekday Before 6pm				
None 36%	54	32	20	2
< 1 hour 37%	56	22	28	6
1-2 hours 18%	27	8	17	2
2-3 hours 5%	7	-	4	3
3-4 hours 3%	4	2	1	1
Television Viewing Weekday After 6pm				
< 1 hour 7%	10	4	5	1
1-2 hours 27%	40	21	17	2
2-3 hours 33%	49	23	20	6
3-4 hours 27%	41	12	26	3
> 4 hours 5%	8	3	3	2
Television Viewing Weekend Before 6pm				
None 34%	51	24	23	4
< 1 hour 33%	49	20	25	4
1-2 hours 21%	32	13	17	2
2-3 hours 8%	12	4	5	3
3-4 hours 2%	3	1	1	1
> 4 hours <1%	1	1	-	-
Television Viewing Weekend After 6pm				
< 1 hour 3%	5	-	4	1
1-2 hours 20%	30	12	16	2
2-3 hours 35%	52	27	20	5
3-4 hours 34%	51	21	26	4
> 4 hours 6%	10	3	5	2

	Whole Group N = 150	55-64 yrs N = 65	65-74 yrs N = 71	≥ 75 yrs N = 14
Activities at Home				
Preparing Food (2.0-2.5 METs)				
None 2%	3	2	1	-
1 hour 5%	8	2	5	1
1-3 hours 13%	20	9	10	1
3-6 hours 25%	38	19	17	2
6-10 hours 25%	37	18	16	3
10-15 hours 12%	18	7	10	1
> 15 hours 17%	26	8	12	6
Shopping Food (2.3 METs)				
None 1%	2	1	1	-
1 hour 11%	16	9	6	1
1-3 hours 56%	84	42	38	4
3-6 hours 25%	38	12	19	7
6-10 hours 6%	10	1	7	2
Shopping Clothes (2.3 METs)				
None 10%	15	8	5	2
1 hour 54%	81	38	38	5
1-3 hours 24%	36	13	17	6
3-6 hours 8%	12	6	6	-
6-10 hours 3%	5	-	4	1
> 15 hours <1%	1	-	1	-
Cleaning House (2.3-4.0 METs)				
None 5%	7	4	3	-
1 hour 18%	27	12	14	1
1-3 hours 40%	60	28	25	7
3-6 hours 21%	32	13	16	3
6-10 hours 13%	19	6	11	2
10-15 hours 1%	2	2	-	-
> 15 hours 2%	3	-	2	1

	Whole Group N = 150	55-64 yrs N = 65	65-74 yrs N = 71	≥ 75 yrs N = 14
Laundry (2.0-2.3 METs)				
None 18%	27	11	11	5
1 hour 19%	29	12	15	2
1-3 hours 43%	65	32	29	4
3-6 hours 15%	23	9	12	2
6-10 hours 3%	4	1	3	-
10-15 hours <1%	1	-	-	1
> 15 hours <1%	1	-	1	-
Gardening (1.5-6.0 METs)				
None 23%	34	17	11	6
1 hour 18%	27	13	11	3
1-3 hours 29%	44	20	20	4
3-6 hours 17%	26	11	15	-
6-10 hours 7%	11	4	7	-
10-15 hours 3%	4	-	4	-
> 15 hours 3%	4	-	3	1
Caring for Family (2.5-4.0 METs)				
None 34%	51	22	22	7
1 hour 17%	25	12	12	1
1-3 hours 22%	33	13	16	4
3-6 hours 10%	15	6	8	1
6-10 hours10%	15	8	7	-
10-15 hours 2%	3	1	2	-
> 15 hours 5%	8	3	4	1
Computer (1.5 METs)				
None 6%	10	-	8	2
1 hour 4%	6	2	4	-
1-3 hours 17%	25	10	14	1
3-6 hours 20%	30	11	16	3
6-10 hours 25%	37	19	15	3
10-15 hours 13%	19	7	9	3
> 15 hours 15%	23	16	5	2

	Whole Group N = 150	**55-64 yrs** N = 65	**65-74 yrs** N = 71	**≥ 75 yrs** N = 14
Play Instrument (1.8-4.0 METs)				
None 94%	141	59	68	14
1 hour 1%	2	2	-	-
10-15 hours 4%	6	3	3	-
> 15 hours <1%	1	1	-	-
Read (1.0-1.3 METs)				
None <1%	1	1	-	-
1 hour 6%	10	3	7	-
1-3 hours 27%	40	18	19	3
3-6 hours 27%	40	21	16	3
6-10 hours 24%	36	12	20	4
10-15 hours 10%	15	7	6	2
> 15 hours 5%	8	3	3	2
Physical Activity				
Visit Friends/Family (1.5 METs)				
None 6%	10	4	6	-
1 hour 13%	20	11	5	4
1-3 hours 29%	44	18	23	3
3-6 hours 28%	42	19	19	4
6-10 hours 15%	22	9	12	1
10-15 hours 5%	7	2	4	1
> 15 hours 2%	5	2	2	1
Senior Centre (1.5-3.5 METs)				
None 97%	146	65	69	12
1-3 hours 3%	4	-	2	2

	Whole Group N = 150	**55-64 yrs** N = 65	**65-74 yrs** N = 71	**≥ 75 yrs** N = 14
Volunteer Work (1.5-4.0 METs)				
None 54%	81	39	37	5
1 hour 13%	20	8	9	3
1-3 hours 15%	22	8	9	5
3-6 hours 10%	15	6	8	1
6-10 hours 5%	8	2	6	-
10-15 hours 1%	2	1	1	-
> 15 hours 1%	2	1	1	-
Religious Activities (1.0-4.0 METs)				
None 86%	129	59	62	8
1 hour <1%	1	-	-	1
1-3 hours 8%	13	5	6	2
3-6 hours 2%	3	-	2	1
6-10 hours 2%	3	1	1	1
> 15 hours <1%	1	-	-	1
Attend a Club (1.5-3.5 METs)				
None 39%	58	28	25	5
1 hour 9%	14	8	4	2
1-3 hours 29%	43	17	22	4
3-6 hours 17%	26	10	14	2
6-10 hours 5%	7	2	5	-
> 15 hours 1%	2	-	1	1
Attend Concert and/or Movie (1.5 METs)				
None 33%	50	21	23	6
1 hour 30%	45	20	23	2
1-3 hours 23%	34	19	11	4
3-6 hours 12%	18	5	13	-
6-10 hours 2%	3	-	1	2
Bingo (1.5 METs)				
None 99%	148	65	70	13
1 hour 1%	2	-	1	1

	Whole Group **N = 150**	**55-64 yrs** **N = 65**	**65-74 yrs** **N = 71**	**≥ 75 yrs** **N = 14**
Dance/Aerobic Class (3.0-10.0 METs)				
None 67%	100	43	46	11
1 hour 1%	2	2	-	-
1-3 hours 23%	35	16	16	3
3-6 hours 5%	8	1	7	-
6-10 hours 3%	5	3	2	-
Play Golf (3.0-4.5 METs)				
None 97%	145	65	66	14
1-3 hours <1%	1	-	1	-
3-6 hours 2%	3	-	3	-
6-10 hours <1%	1	-	1	-
Play Racket Sport (4.0-12.0 METs)				
None 94%	141	65	63	13
1-3 hours 4%	6	-	5	1
3-6 hours 2%	3	-	3	-
Play Sport (2.5-12.5 METs)				
None 89%	133	57	64	12
1 hour 3%	4	1	2	1
1-3 hours 3%	5	4	1	-
3-6 hours 3%	4	2	2	-
6-10 hours 3%	4	1	2	1
Running (4.5-18.0 METs)				
None 87%	131	54	65	12
1 hour 7%	10	5	4	1
1-3 hours 4%	6	4	1	1
3-6 hours 2%	3	2	1	-

	Whole Group N = 150	**55-64 yrs** N = 65	**65-74 yrs** N = 71	**≥ 75 yrs** N = 14
Walk to Errands (2.0-12.0 METs)	14	8	4	2
None 9%	32	16	14	2
1 hour 21%	70	29	36	5
1-3 hours 47%	24	10	12	2
3-6 hours 16%	8	1	4	3
6-10 hours 5%	1	-	1	-
10-15 hours <1%	1	1	-	-
> 15 hours <1%				
Walk for Exercise (2.5-9.0 METs)	31	13	16	2
None 21%	27	16	8	3
1 hour 18%	45	16	24	5
1-3 hours 30%	17	5	10	2
3-6 hours 11%	21	9	10	2
6-10 hours 14%	7	6	1	-
10-15 hours 5%	2	-	2	-
> 15 hours 1%				
Riding a Bike (3.0-12.5 METs)	97	34	51	12
None 65%	27	15	11	1
1 hour 18%	16	11	4	1
1-3 hours 11%	4	3	1	-
3-6 hours 3%	5	2	3	-
6-10 hours 3%	1	-	1	-
10-15 hours <1%				
Swimming (4.0-11.0 METs)	114	45	56	13
None 76%	14	7	6	1
1 hour 9%	21	12	9	-
1-3 hours 14%	1	1	-	-
6-10 hours <1%				

	Whole Group N = 150	**55-64 yrs** N = 65	**65-74 yrs** N = 71	**≥ 75 yrs** N = 14
Yoga (2.5 METs)				
None 71%	107	44	52	11
1 hour 6%	10	4	6	-
1-3 hours 18%	27	13	11	3
3-6 hours 4%	6	4	2	-
Weights Training (3.0-6.0 METs)				
None 83%	125	52	62	11
1 hour 7%	11	6	5	-
1-3 hours 7%	11	4	4	3
3-6 hours 2%	3	3	-	-
Hairdressing (2.5 METs)				
None 99%	149	65	70	14
1 hour <1%	1	-	1	-
Speech Exercises (1.5-2.0 METs)				
None 99%	149	65	70	14
3-6 hours <1%	1	-	1	-
Singing (1.5-2.0 METs)				
None 99%	148	64	70	14
3-6 hours <1%	1	1	-	-
6-10 hours <1%	1	-	1	-
Horse Riding (2.6-8.0 METs)				
None 99%	149	64	71	14
6-10 hours <1%	1	1	-	-
Sailing (3.0-6.0 METs)				
None 99%	148	63	71	14
3-6 hours <1%	1	1	-	-
6-10 hours <1%	1	1	-	-

3.7. Conclusion

The ELSA (2014a) study identified older adults who engage in healthy and active lifestyle behaviours are associated with positive outcomes. Involvement in social and cultural activities are associated with reduced depression, higher perceived health, improved physical and cognitive function and lower mortality risk (de Leon *et al.*, 2003, Glass *et al.*, 2006, Niti *et al.*, 2008, Chiao *et al.*, 2011, Thomas, 2011). In addition, not smoking, consuming alcohol within the recommended limits and increased physical activity were associated with improved physical and mental health for older adults (LaCroix *et al.*, 1991, Blow *et al.*, 2000, Gow *et al.*, 2012). This may explain why for this population participants were relatively healthy and high-functioning older adults.

Chapter Four: The Effects of Age on Gait in an Older Adult Population

ABSTRACT

Introduction: Age-related gait adaptations have been identified in older adults. Older adults tend to walk slower, have a reduced step length and altered hip and ankle joint kinematics and kinetics. However, the age effect has only been explored by comparing young to older adults. Therefore, the objective was to explore the effect of age on gait within an older adult population.

Methods: 158 community-dwelling older adults, age range 55-86 years (65.7 ± 6.8 yrs) participated, walking at a self-selected comfortable walking speed. Research has identified different age ranges within an older adult population demonstrate different results, as such participants were grouped into three age groups (55-64 yrs; 65-74 yrs; ≥ 75 yrs). Mini-mental state examination and timed up and go were measured as a function baseline. Three-dimensional motion analysis was used to capture spatial-temporal parameters, joint kinematics and kinetics of the right limb. One-way between-subjects' ANOVAs were performed to determine the age effect on gait parameters (spatial-temporal parameters, joint kinematics and kinetic peaks). Statistical parametric mapping was used to compare the joint kinematic and kinetic waveforms for each age group to determine if differences within the gait cycle were phase-specific (e.g. knee flexion throughout midstance increased) and/or highlighted different gait cycle locations which were not analysed as the typical gait peaks. Correlation between age and joint kinematics and kinetics were performed, whilst controlling for walking speed.

Results: Reduced walking speed, stride/step length and a slower timed up and go was present for older adults aged 75 years and over. Hip extension range of motion was reduced during late stance, with a reduced hip extension torque and power generation in late stance, with reduced knee power generation and absorption and ankle power generation for the 75 years and older age group. No significant differences were found between the 55-64 years and 65-74 years age groups. When controlling for walking speed, age was not significantly correlated to joint kinematics and kinetics, except knee valgus moment (second peak).

Conclusion: No age-related gait adaptations occurred between the 55-64 years and 65-74 years age groups. This suggests for this older adult population gait parameters are relatively stable up to the age of 74 yrs. Age-related gait adaptations occur from the age of 75 in this population, which suggests the age effect shifted. Age effect was predominantly present in the joint kinetics. A reduced hip extension, leads to a reduced stride length and walking speed for the over 75 years age group. The hip joint for the over 75 years age group, also displayed a reduced hip extension torque and power generation in late stance, with reduced knee power generation and absorption and ankle power generation. These age-related changes for the over 75 years age group are associated with a reduction in walking speed. Reduction in walking speed is commonly reported in older adult research. It is found to be associated with a reduction in joint power and altered joint moments, which is thought to be caused by age-related declines of the musculoskeletal system, such as muscle weakness. Therefore, this altered gait pattern (e.g. altered joint kinetics and walking speed) for the over 75 years age group may influence the success of toe-clearance. Consequently, future work is required to investigate toe-clearance parameters within this older adult population.

Keywords: Older Adults; Overground Gait; Falls; Walking Speed; Three-dimensional Analysis

4.1. Introduction

Walking is an important daily task which requires synchronised actions of the musculoskeletal system to function independently. Changes in gait are used to assess health status and indicate adverse events such as falling and mortality in older adults (Maki, 1997, Ferrucci *et al.*, 2000, Hausdorff *et al.*, 2001, Studenski *et al.*, 2003, Cesari *et al.*, 2005, Morris *et al.*, 2005, Verghese *et al.*, 2006, Verghese *et al.*, 2007, Verghese *et al.*, 2009, Studenski *et al.*, 2011). Therefore, alterations to gait may act as markers for current and future health.

Older adults have a reduced self-selected comfortable walking speed when compared to young adults (Winter *et al.*, 1990, Kerrigan *et al.*, 1998, DeVita and Hortobagyi, 2000, Kerrigan *et al.*, 2001, Riley *et al.*, 2001, Byrne *et al.*, 2002, Monaco *et al.*, 2009, Anderson and Madigan, 2014). A reduction in walking speed has also been associated with altered joint kinematics and kinetics in older adults (Kerrigan *et al.*, 1998, Kerrigan *et al.*, 2001, Riley *et al.*, 2001, Chung and Wang, 2010, Anderson and Madigan, 2014). However, alterations in joint kinematics and kinetics occur with similar walking speeds (DeVita and Hortobagyi, 2000, Silder *et al.*, 2008), with Alcock *et al.* (2013) revealing that a reduced walking speed (1.2 % per year) does not explain altered gait as a consequence of ageing. In addition, older adults have been found to have an increased double-support time, step time and stride width (Winter *et al.*, 1990, Elble *et al.*, 1991, Winter, 1992, Lajoie *et al.*, 1996, Begg *et al.*, 2007, Mills *et al.*, 2008, Mariani *et al.*, 2010, Schulz *et al.*, 2010) and reduced stride/step length compared to young adults (Winter *et al.*, 1990, Judge *et al.*, 1996, DeVita and Hortobagyi, 2000, Paróczai *et al.*, 2006, Monaco *et al.*, 2009). This pattern is thought to be adopted as a safe ‘cautious gait’ strategy to reduce fall risk in older adults.

For joint kinematics and kinetics, age-related differences for older adults are commonly reported at the hip and ankle joint (McGibbon, 2003, Silder *et al.*, 2008, Anderson and Madigan, 2014). Reduced hip extension range of motion (RoM) has been reported for older adults compared to young adults (Kerrigan *et al.*, 1998, Kerrigan *et al.*, 2001, Lee *et al.*, 2005, Monaco *et al.*, 2009, Anderson and Madigan, 2014). In addition, studies have identified plantar-flexor kinetics such as peak torque and power generation are reduced in older adults during gait (Winter *et al.*, 1990, Judge *et al.*, 1996, Kerrigan *et al.*, 1998, DeVita and Hortobagyi, 2000, Riley *et al.*, 2001, Silder *et al.*, 2008, Monaco *et al.*, 2009). Increased hip extensor power (DeVita and Hortobagyi, 2000, Silder *et al.*, 2008, Monaco *et al.*, 2009) or hip flexor power has been associated as a compensation for reduced plantar-flexor kinetics (Judge *et al.*, 1996, Goldberg and Neptune, 2007, Monaco *et al.*, 2009, Cofre *et al.*, 2011). These studies suggested older adults walk with an increased hip flexion and reduced plantar-flexor peak torque compared to young adults as an age-related compensation for limited plantarflexion strength and hip extension and ankle plantarflexion RoM. In addition, older adults have been reported to have an increased anterior pelvic tilt (Judge *et al.*, 1996, Kerrigan *et al.*, 1998, Lim *et al.*, 2013) and reduced knee flexion in the swing phase compared to young adults (Finley *et al.*, 1969, Murray *et al.*, 1969, Hageman and Blanke, 1986, Winter *et al.*, 1990, Elble *et al.*, 1991, Nigg *et al.*, 1994, Ostrosky *et al.*, 1994, Judge *et al.*, 1996, Kerrigan *et al.*, 1998, DeVita and Hortobagyi, 2000, Kerrigan *et al.*, 2001, McGibbon and Krebs, 2004, Boyer and Andriacchi, 2016). Knee kinetics has also found to be reduced for older adults (Kerrigan *et al.*, 1998, Schloemer *et al.*, 2017).

Researchers have predominantly investigated older adult gait by comparing to young adults, which does not consider the ageing process. As such, the older adults are categorised into a single age group. Few studies have analysed gait within an older adult population (Ko *et al.*, 2010a, Ko *et al.*, 2011), however none of these have explored the age effect. Furthermore, gait analysis for older adults is limited to sagittal and coronal plane, none to date have investigated the effect of age in the transverse plane (e.g. hip rotation). The aim of this study was to examine the effects of age on gait parameters within an older adult population. It was hypothesised that an age effect would occur for walking speed, stride/step length, stride width and double-support time. It was also hypothesised an increase in age would show an effect on reduced range of motion for joint kinematics and altered joint kinetics for the hip and ankle joint. In addition, another approach to investigating the age effect for joint kinematics and kinetics would be to explore phase effects rather than solely gait peaks. The traditional approach to gait analysis is to analyse the peaks and troughs of the time series data. Statistical Parametric Mapping (SPM) allows a robust statistical method for understanding the phase-specific effect (Pataky *et al.*, 2013) and as such this method was used to establish if an age effect was not only present for joint kinematic peaks but also for a phase in the gait cycle (e.g. increased pelvic obliquity throughout terminal stance), for example.

4.2. <u>Methods</u>

Detailed methodology is provided in Chapter Two. Participants performed normal walking with a right foot force plate contact. Five successful trials were collected, which had no force plate targeting. A custom-made Python code (Python v. 2.7.10, Delaware, USA) was used to analyse the data. Joint kinematics and kinetics (lower body) were normalised to one gait cycle (100 %), using linear interpolation to 101 data samples. The calculations of the peak joint kinematics and kinetics were guided by Winter *et al.* (1990) and Winter (1992). The ipsilateral limb (right) peaks were calculated during the stance and swing phase for joint kinematics in the sagittal plane at the pelvis (tilt), hip (flexion/extension), knee (flexion/extension) and ankle (plantar/dorsiflexion) and in the coronal plane at the pelvis (obliquity), hip (abduction/adduction) and knee (varus/valgus) and in the transverse plane at the pelvis (rotation) and hip (rotation) and for joint kinetics moments in the sagittal plane at the hip (flexion/extension), knee (flexion/extension) and ankle (plantar/dorsiflexion) and in the coronal plane at the hip (abduction/adduction) and knee (varus/valgus) and powers (hip, knee and ankle). On completion of data analysis, participants were grouped into age groups (55-64 yrs, 65-74 yrs and ≥ 75 yrs).

Statistical analysis was performed using IBM SPSS v.23 software (Chicago, USA). One-way between-subjects' ANOVAs, with spatial-temporal parameters, peak joint kinematics and kinetics as the dependent variable and age groups as the between factors was executed. The ANOVAs were followed by pre-planned comparisons, based on Bonferroni adjusted post-hoc Tukey tests. Pearson's R correlations between age, walking speed and joint kinematics and kinetics were performed for the whole population. In addition, a partial correlation was performed to find the association between age and joint kinematics and kinetics when controlling for walking speed. Statistical significance was considered at $p < 0.05$.

Statistical Parametric Mapping (SPM) analyses were performed using open-source SPM1d code (v. 0.3) (spm1D, 2015), which was installed in Python 2.7.10 and implemented in Enthought Canopy 1.7.4.3348 (Enthought Inc., Austin, USA). SPM one-way between-subjects' ANOVAs with a Bonferroni correct threshold of 0.0167 (0.05/3) were used to examine whether the mean kinematic angle and kinetics (moments and powers) of the waveform patterns per joint differed significantly between the age groups (alpha rate of 0.05). Age group post-hoc analysis was only performed when significance was achieved. For each SPM ANOVA, a statistical parametric map (SPM {F}) was created by calculating the F-statistic at each point of the gait curve (Pataky, 2010). Then, Random Field Theory determined the critical threshold (α = 0.05) of equally smooth random data was expected to cross (Pataky, 2011). Field smoothness was derived from time-varying gradients of the residuals to determine significance (Pataky, 2016). If the SPM {F} crossed the critical threshold, a supra-threshold cluster was created (grey shading), indicating a significant difference ($p < 0.05$) between the joint pattern (e.g. hip flexion/extension angle) in a specific location of the gait cycle (e.g. midstance) (Figure 3.1.). If the SPM {F} crossed the critical threshold, post-hoc SPM {t} maps were calculated for between-group (age groups) comparisons.

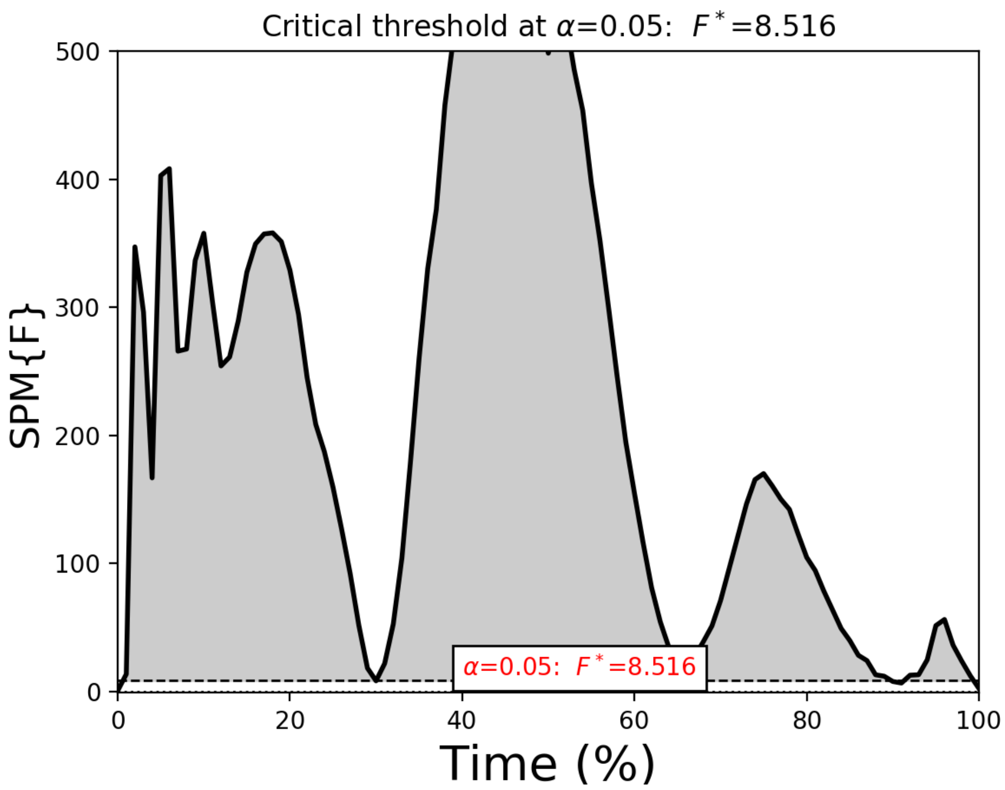


Figure 3.1. Example of a Statistical Parametric Mapping (SPM) one-way between-subjects ANOVA output (spm1D, 2017). *Note*: the supra-threshold cluster (grey shading) indicates where the significance occurred in the gait cycle.

4.3. <u>Results</u>

Following data collection, eighteen participants, age range 60-77 years (11 females, 7 males; 68.3 ± 8.5 yrs; 169.9 ± 8.2 cm; 73.6 ± 14.5 kg) were excluded from the study (Figure 4.2.). Therefore, one-hundred and forty participants, age range 55-86 years (65.4 ± 6.5 yrs) were included in the study (Table 4.1.). There was no significant difference for MMSE score between age groups ($F_{2,137}$ = 1.917, P = 0.151). There was a significant difference for TUG time (in seconds) between age groups ($F_{2,137}$ = 4.534, P = 0.012), the over 75 years age group (8.8 ± 1.9 s) were significantly slower compared to the 55-64 years (7.6 ± 1.2 s) and 65-74 years (7.8 ± 1.3 s) age groups.

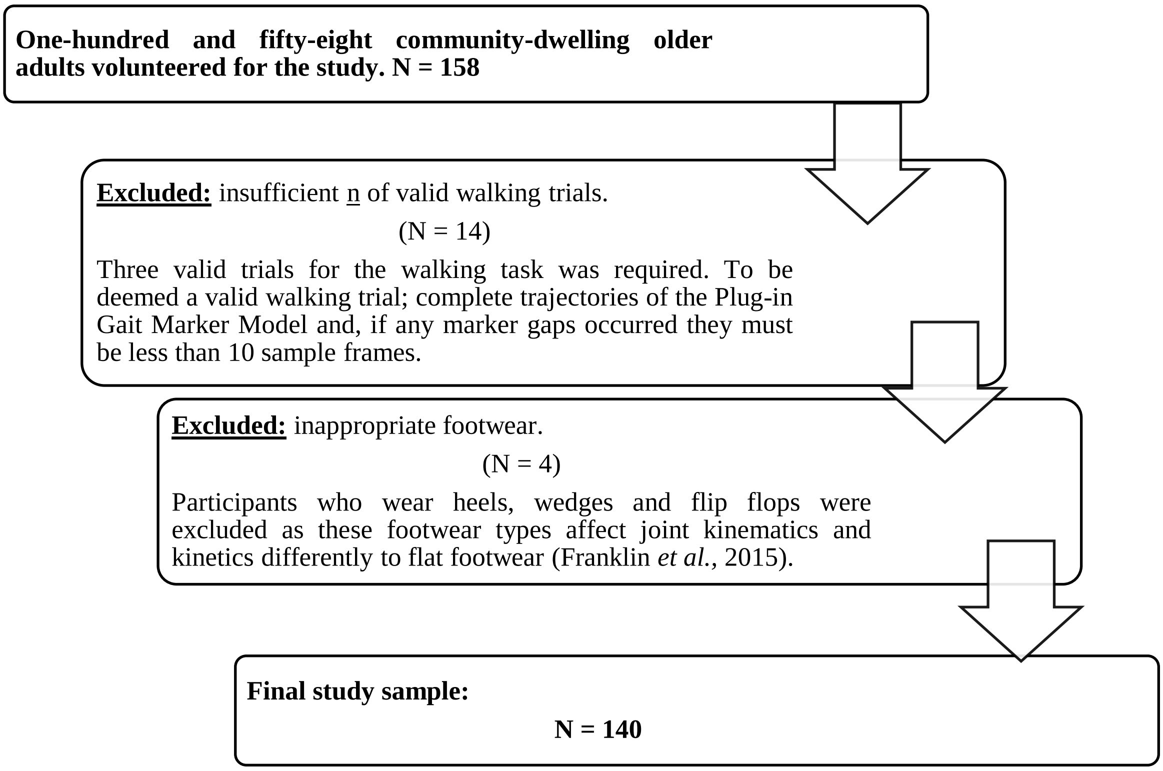


Figure 4.2. Description of participant selection and participant exclusion.

Table 4.1. Participant characteristics including MMSE score and TUG.

	Whole Group (n = 140)	**55-64 yrs (n = 63)**	**65-74 yrs (n = 65)**	**≥ 75 yrs (n = 12)**
Sex (Females/Males)	90/50	41/22	40/25	9/3
Age (yrs)	65.4 ± 6.5	59.8 ± 3.2	68.4 ± 2.7	78.5 ± 3.4
Height (cm)	168.5 ± 9.3	169.3 ± 9.2	168.2 ± 9.8	165.2 ± 6.3
Mass (kg)	74.0 ± 14.9	74.0 ± 15.6	74.1 ± 14.9	73.9 ± 11.8
MMSE Score (out of 30)	29 ± 1	29 ± 1	29 ± 1	28 ± 2
TUG (s)	7.8 ± 1.4	7.6 ± 1.2[a*]	7.8 ± 1.3[a*]	8.8 ± 1.9

[a*] ≥ 75 yrs significantly slower than 55-64 yrs and 65-74 yrs.

4.3.1. Spatial-Temporal Parameters

There was a significant age effect for double-support time, walking speed, step length and stride length (Table 4.3.). The over 75 years age group had a significantly increased double-support time compared to the 55-64 years age group, with a significantly reduced walking speed (1.24 ± 0.21 m·s^{-1}), step length (0.65 ± 0.07 m) and stride length (1.32 ± 0.14 m) compared to the 55-64 years age group (1.45 ± 0.18 m·s^{-1}, 0.73 ± 0.09 m, 1.50 ± 0.17 m) and 65-74 years age group (1.40 ± 0.17 m·s^{-1}, 0.71 ± 0.07 m, 1.46 ± 0.13 m) (Figure 4.3. and Table 4.2.).

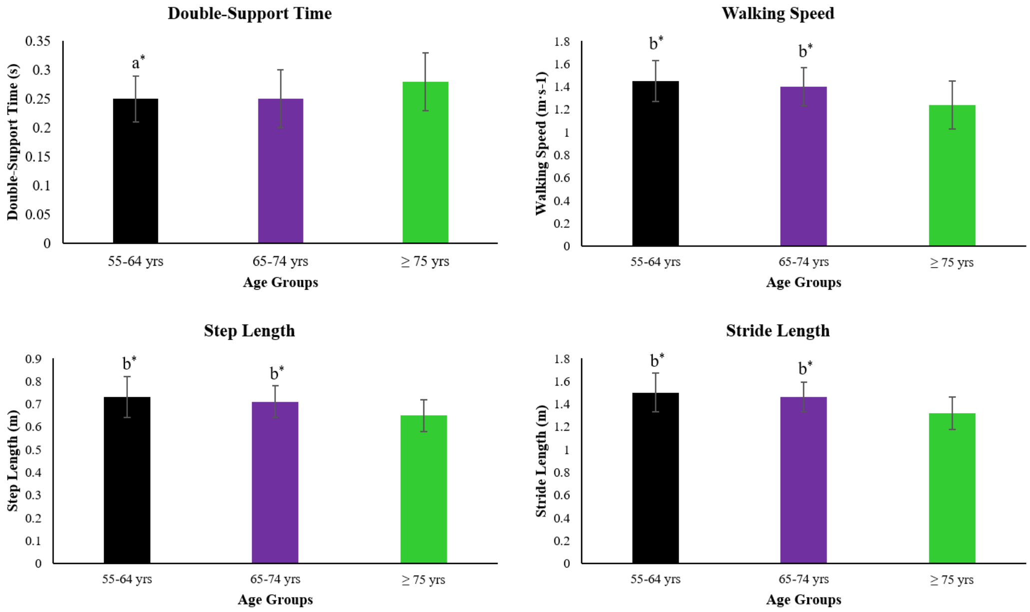


Figure 4.3. Significant spatial-temporal parameters (mean ± SD) for an older adult population during NW (n = 140). Age groups (55-64 yrs: n = 63; 65-74 yrs: n = 65 and ≥ 75 yrs: n = 12). *Note:* * significant age effect. a* 55-64 yrs significantly reduced parameter compared to ≥ 75 yrs only. b* ≥ 75 yrs significantly reduced parameter compared to 55-64 yrs and 65-74 yrs.

Table 4.2. Spatial-temporal parameters (mean ± SD) for an older adult population during NW.

Parameter	Whole Group (n = 140)	55-64 yrs (n = 63)	65-74 yrs (n = 65)	≥75 yrs (n = 12)	*F* Value
Rhythm					
Cadence (steps/min)	115.2 ± 8.8	115.7 ± 8.0	115.3 ± 9.5	111.7 ± 8.6	$F_{2,137} = 1.077$, $P = 0.343$
Step Time (s)	0.52 ± 0.04	0.51 ± 0.04	0.51 ± 0.05	0.54 ± 0.05	$F_{2,137} = 1.768$, $P = 0.175$
Stride Time (s)	1.05 ± 0.08	1.04 ± 0.07	1.05 ± 0.09	1.08 ± 0.09	$F_{2,137} = 1.227$, $P = 0.296$
Single-support Time (s)	0.41 ± 0.03	0.41 ± 0.03	0.41 ± 0.03	0.40 ± 0.03	$F_{2,137} = .121$, $P = 0.886$
Phases					
Double-support Time (s)	0.25 ± 0.05	0.25 ± 0.04[a*]	0.25 ± 0.05	0.28 ± 0.05	$F_{2,137} = 3.597$, $P = 0.030$*
Foot-off (%)	63.70 ± 10.26	64.75 ± 15.16	62.66 ± 1.85	63.77 ± 1.83	$F_{2,137} = .657$, $P = 0.520$
Limp Index (s)	1.03 ± 0.03	1.03 ± 0.03	1.03 ± 0.04	1.02 ± 0.03	$F_{2,137} = 935$, $P = 0.395$
Opposite Foot Contact (%)	50.82 ± 1.03	50.83 ± 0.78	50.86 ± 1.18	50.51 ± 1.37	$F_{2,137} = 612$, $P = 0.544$
Opposite Foot-off (%)	11.96 ± 1.87	11.71 ± 1.89	11.99 ± 1.71	13.14 ± 2.26	$F_{2,137} = 3.043$, $P = 0.051$
Pace					
Walking Speed ($m \cdot s^{-1}$)	1.41 ± 0.19	1.45 ± 0.18[b*]	1.40 ± 0.17 [b*]	1.24 ± 0.21	$F_{2,137} = 6.638$, $P = 0.002$*
Step Length (m)	0.72 ± 0.08	0.73 ± 0.09[b*]	0.71 ± 0.07 [b*]	0.65 ± 0.07	$F_{2,137} = 5.960$, $P = 0.003$*
Stride Length (m)	1.47 ± 0.16	1.50 ± 0.17[b*]	1.46 ± 0.13 [b*]	1.32 ± 0.14	$F_{2,137} = 6.856$, $P = 0.001$*
Base of Support					
Step Width (m)	0.16 ± 0.05	0.16 ± 0.05	0.16 ± 0.05	0.15 ± 0.05	$F_{2,137} = .400$, $P = 0.671$

* significant age effect. a* 55-64 yrs significantly reduced parameter compared to ≥ 75 yrs only. b* ≥75 yrs significantly reduced parameter compared to 55-64 yrs and 65-74 yrs.

4.3.2. Joint Kinematics

There was a significant age effect for peak hip extension and plantarflexion at loading response (Table 4.3.). The over 75 years age group had a significantly reduced hip extension (-5.69 ± 10.58 °) compared to the 55-64 years (-13.18 ± 8.47 °) and 65-74 years (-12.22 ± 8.41 °) age groups, with a significantly increased plantarflexion at loading response (-8.40 ± 4.47 °) compared to the 55-64 years (-4.59 ± 4.11°) and 65-74 years (-4.76 ± 4.71°) age group (Table 4.3. and Figure 4.4.). SPM phase significant age effect occurred for hip flexion/extension ($F_{1.996,135.630} = 4.874$, $P = 0.044$), hip extension was significantly reduced in terminal stance (36-48 percent of the gait cycle (%GC)) for the over 75 years age group compared to 55-64 years age group (Figure 4.5.).

In addition, significant age correlations (Table 4.4.) were found for the joint kinematics (maximum pelvic obliquity in stance, hip extension, maximum hip rotation in swing, knee extension at terminal stance and maximum ankle plantarflexion). Similarly, there were significant correlations between walking speed and joint kinematics (maximum pelvic obliquity in stance, minimum pelvic obliquity in swing, minimum pelvic rotation and maximum pelvic rotation in swing, hip extension, hip adduction and abduction in swing, knee flexion at loading response, knee extension at terminal stance, ankle plantarflexion at loading response and maximum ankle plantarflexion). When controlling for walking speed however, no significant correlations were found between age and joint kinematics.

Table 4.3. Kinematic gait peaks (mean ± SD) for an older adult population during NW.

Parameter (°)	Whole Group (n = 140)	55-64 yrs (n = 63)	65-74 yrs (n = 65)	≥ 75 yrs (n = 12)	*F* Value
Pelvic Tilt					
Range of Motion (RoM)	6.60 ± 6.24	6.13 ± 5.67	6.36 ± 6.19	10.44 ± 8.31	$F_{2,137}$ = 2.557, P = 0.081
Maximum Tilt - Stance	8.66 ± 6.24	8.24 ± 5.60	8.34 ± 6.24	12.64 ± 8.35	$F_{2,137}$ = 2.745, P = 0.068
Minimum Tilt - Stance	5.10 ± 6.40	4.67 ± 5.86	4.79 ± 6.30	9.11 ± 8.56	$F_{2,137}$ = 2.642, P = 0.075
Maximum Tilt - Swing	7.53 ± 6.18	7.09 ± 5.71	7.33 ± 6.09	10.98 ± 8.31	$F_{2,137}$ = 2.098, P = 0.127
Minimum Tilt - Swing	5.23 ± 6.32	4.66 ± 5.80	5.10 ± 6.29	8.89 ± 8.22	$F_{2,137}$ = 2.332, P = 0.101
Pelvic Obliquity					
RoM	0.55 ± 2.52	0.46 ± 2.49	0.79 ± 2.35	-0.29 ± 3.49	$F_{2,137}$ = 0.991, P = 0.374
Maximum Obliquity - Stance	4.66 ± 2.96	4.85 ± 3.04	4.82 ± 2.71	2.82 ± 3.50	$F_{2,137}$ = 2.590, P = 0.079
Minimum Obliquity - Stance	-2.37 ± 2.95	-2.66 ± 2.87	-2.13 ± 2.87	-2.16 ± 3.85	$F_{2,137}$ = 0.552, P = 0.577
Maximum Obliquity - Swing	1.43 ± 2.63	1.28 ± 2.85	1.70 ± 2.33	0.77 ± 2.99	$F_{2,137}$ = 0.807, P = 0.448
Minimum Obliquity - Swing	-3.62 ± 2.95	-4.03 ± 2.99	-3.21 ± 2.72	-3.72 ± 3.78	$F_{2,137}$ = 1.261, P = 0.287
Pelvic Rotation					
RoM	1.02 ± 3.65	0.71 ± 3.42	1.26 ± 3.47	1.40 ± 5.59	$F_{2,137}$ = 0.434, P = 0.649
Maximum Rotation - Stance	5.45 ± 4.10	5.59 ± 3.85	5.25 ± 4.22	5.79 ± 5.02	$F_{2,137}$ = 0.152, P = 0.859
Minimum Rotation	-4.35 ± 4.52	-4.97 ± 4.18	-3.83 ± 4.31	-3.96 ± 6.91	$F_{2,137}$ = 1.068, P = 0.346
Maximum Rotation - Swing	5.80 ± 4.06	5.63 ± 3.92	6.04 ± 3.98	5.34 ± 5.36	$F_{2,137}$ = 0.246, P = 0.782
Hip Flexion/Extension					
RoM	15.48 ± 7.69	15.03 ± 7.14	15.22 ± 8.06	19.25 ± 8.12	$F_{2,137}$ = 1.599, P = 0.206
Flexion - Stance	36.22 ± 7.74	36.01 ± 7.21	35.83 ± 8.19	39.39 ± 7.93	$F_{2,137}$ = 1.114, P = 0.331
Extension	-12.09 ± 8.81	-13.18 ± 8.47[a*]	-12.22 ± 8.41[a*]	-5.69 ± 10.58	$F_{2,137}$ = 3.801, P = 0.025*
Flexion - Swing	36.10 ± 7.93	35.97 ± 7.38	35.84 ± 8.13	38.19 ± 9.84	$F_{2,137}$ = 0.458, P = 0.633
Hip Abduction/Adduction					
RoM	3.73 ± 4.37	3.77 ± 4.32	3.65 ± 4.40	3.95 ± 4.79	$F_{2,137}$ = 0.027, P = 0.973
Adduction - Stance	10.18 ± 4.91	10.71 ± 4.82	9.97 ± 4.93	8.55 ± 5.25	$F_{2,137}$ = 1.090, P = 0.339
Adduction - Swing	4.04 ± 4.60	4.00 ± 5.17	3.93 ± 4.07	4.92 ± 4.44	$F_{2,137}$ = 0.237, P = 0.789
Abduction - Swing	-3.51 ± 4.83	-4.03 ± 4.92	-3.35 ± 4.52	-1.61 ± 5.83	$F_{2,137}$ = 1.331, P = 0.267

Parameter (°)	Whole Group (n = 140)	55-64 yrs (n = 63)	65-74 yrs (n = 65)	≥ 75 yrs (n = 12)	*F* Value
Hip Rotation					
RoM	-8.28 ± 7.87	-8.97 ± 6.73	-7.93 ± 8.51	-6.54 ± 9.93	$F_{2,137}$ = 0.597, P = 0.552
Maximum Rotation -Stance	-0.85 ± 9.01	-0.81 ± 7.65	-0.96 ± 9.76	-0.46 ± 11.87	$F_{2,137}$ = 0.016, P = 0.984
Minimum Rotation - Stance	-18.56 ± 8.63	-18.84 ± 8.42	-18.25 ± 8.87	-18.78 ± 9.13	$F_{2,137}$ = 0.080, P = 0.923
Maximum Rotation - Swing	1.21 ± 9.54	-0.13 ± 7.37	1.61 ± 10.72	6.02 ± 11.81	$F_{2,137}$ = 2.244, P = 0.110
Minimum Rotation - Swing	-18.80 ± 8.70	-19.17 ± 8.65	-18.04 ± 8.80	-20.95 ± 8.68	$F_{2,137}$ = 0.667, P = 0.515
Knee Flexion/Extension					
RoM	24.27 ± 3.78	24.24 ± 3.38	24.48 ± 4.04	23.26 ± 4.45	$F_{2,137}$ = 0.535, P = 0.587
Flexion at Loading Response (LR)	23.08 ± 5.80	23.21 ± 6.38	23.52 ± 4.85	19.97 ± 6.90	$F_{2,137}$ = 1.954, P = 0.146
Extension at Terminal Stance (TS)	5.67 ± 5.01	5.10 ± 4.03	5.92 ± 5.63	7.31 ± 6.02	$F_{2,137}$ = 1.131, P = 0.326
Flexion - Swing	64.60 ± 5.16	65.37 ± 4.84	64.30 ± 5.19	62.18 ± 6.06	$F_{2,137}$ = 2.170, P = 0.118
Knee Varus/Valgus					
RoM	-0.43 ± 1.83	-0.54 ± 1.70	-0.35 ± 1.98	-0.24 ± 1.81	$F_{2,137}$ = 0.241, P = 0.786
Maximum Varus - Stance	2.51 ± 2.28	2.49 ± 2.09	2.50 ± 2.54	2.69 ± 1.82	$F_{2,137}$ = 0.042, P = 0.959
Minimum Valgus - Stance	-3.30 ± 2.85	-3.54 ± 3.09	-3.23 ± 2.78	-2.45 ± 1.64	$F_{2,137}$ = 0.778, P = 0.461
Maximum Varus - Swing	2.92 ± 2.58	2.77 ± 2.48	3.12 ± 2.73	2.59 ± 2.29	$F_{2,137}$ = 0.392, P = 0.676
Minimum Valgus - Swing	-3.80 ± 3.16	-4.12 ± 3.20	-3.56 ± 3.11	-3.41 ± 3.38	$F_{2,137}$ = 0.597, P = 0.552
Ankle Plantar/Dorsiflexion					
RoM	5.21 ± 3.04	5.22 ± 2.92	5.31 ± 3.12	4.61 ± 3.39	$F_{2,137}$ = 0.267, P = 0.766
Plantarflexion (LR)	-4.99 ± 4.52	-4.59 ± 4.11[b*]	-4.76 ± 4.71 [b*]	-8.40 ± 4.47	$F_{2,137}$ = 3.902, P = 0.022*
Dorsiflexion - Stance	23.02 ± 7.30	23.50 ± 7.63	22.90 ± 6.88	21.18 ± 8.06	$F_{2,137}$ = 0.521, P = 0.595
Maximum Plantarflexion	-19.07 ± 7.97	-20.28 ± 8.17	-18.67 ± 7.77	-14.80 ± 6.70	$F_{2,137}$ = 2.588, P = 0.079

* significant age effect. a* ≥ 75 yrs significantly reduced parameter compared to 55-64 yrs and 65-74 yrs. b* ≥ 75 yrs significantly increased parameter compared to 55-64 yrs and 65-74 yrs. *Abbreviation:* Loading Response (LR) and Terminal Stance (TS).

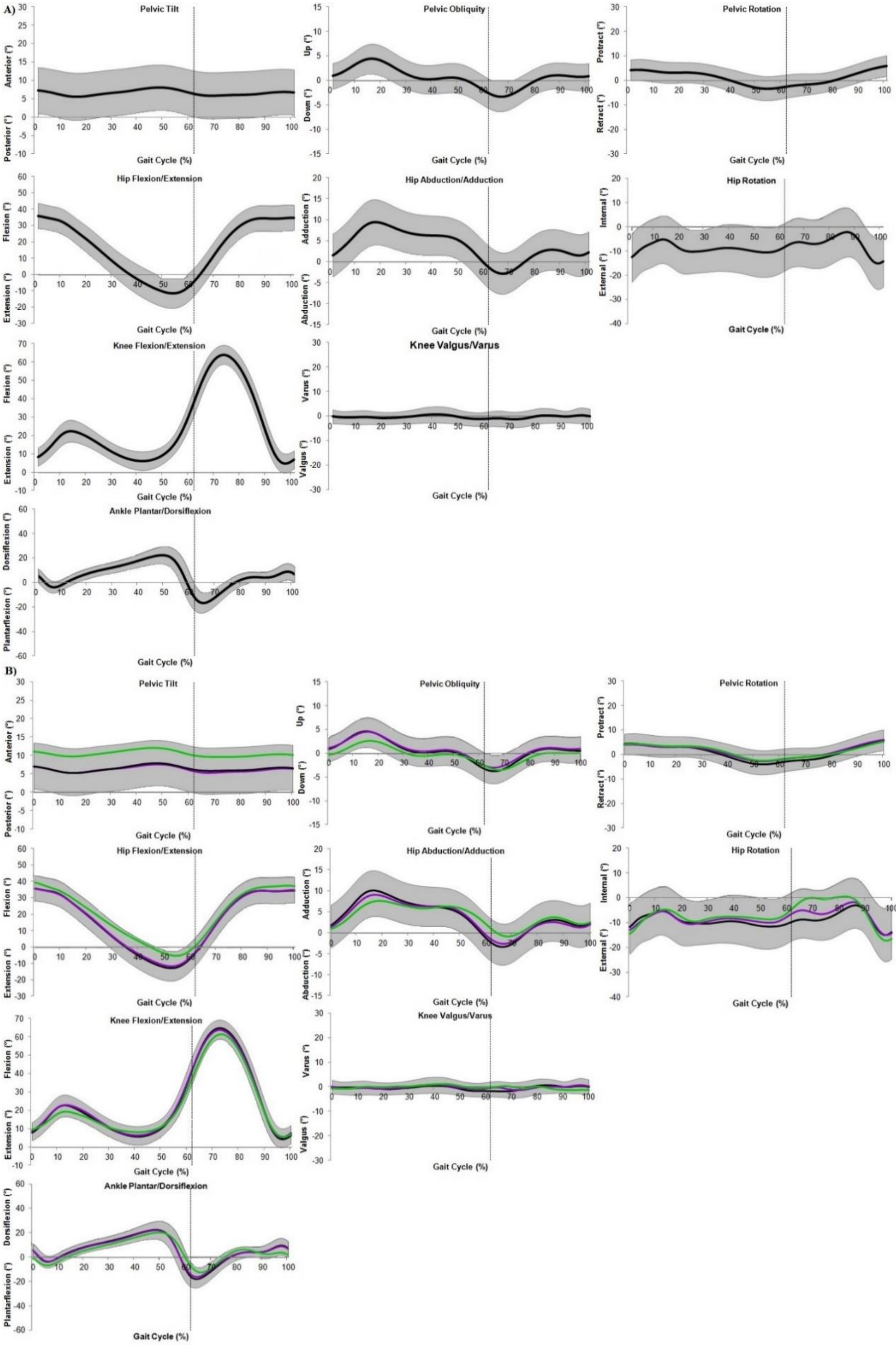


Figure 4.4. Joint kinematics (mean ± SD) for an older adult population during NW: **A)** whole group averages (n = 140) and **B)** age group averages (black line = 55-64 yrs (n = 63), purple line = 65-74 yrs (n = 65) and green ≥ 75 yrs (n = 12)).

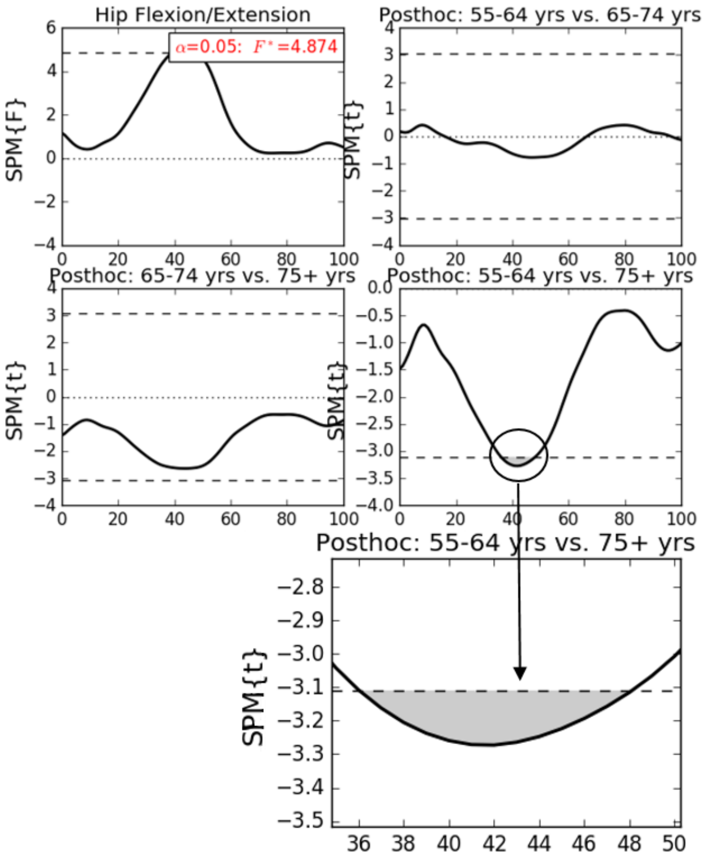


Figure 4.5. Statistical Parametric Mapping (SPM) output: age group post-hoc analysis for significant joint kinematics.

Table 4.4. Correlation between age, walking speed, peak joint kinematics and range of motion for an older adult population during walking, including a partial correlation to control for walking speed.

Parameter (º)	Age (yrs)	Walking Speed ($m \cdot s^{-1}$)	Age (yrs) Relationship When Controlling for Walking Speed ($m \cdot s^{-1}$)
Pelvic Tilt			
Range of motion (RoM)	0.126	-0.049	-0.011
Max Tilt – Stance	0.128	-0.035	-0.019
Min Tilt – Stance	0.116	-0.060	-0.004
Max Tilt – Swing	0.120	-0.053	-0.004
Min Tilt – Swing	0.132	-0.058	-0.019
Pelvic Obliquity			
RoM	-0.057	0.050	0.010
Max Obliquity – Stance	-0.299*	-0.267*	0.054
Min Obliquity – Stance	0.085	-0.114	-0.032
Max Obliquity – Swing	-0.036	-0.050	-0.022
Min Obliquity – Swing	0.135	-0.214*	-0.036
Pelvic Rotation			
RoM	0.023	-0.072	0.022
Max Rotation – Stance	-0.035	0.026	0.001
Min Rotation	0.059	-0.208*	0.054
Max Rotation – Swing	-0.076	0.243*	-0.033
Hip Flexion/Extension			
RoM	0.129	-0.077	-0.019
Flexion – Stance	0.077	0.100	-0.013
Extension	0.226*	-0.331*	-0.012
Flexion – Swing	0.042	0.075	-0.001
Hip Abduction/Adduction			
RoM	-0.002	-0.150	0.025
Adduction – Stance	-0.152	0.041	0.048
Adduction – Swing	0.045	-0.208*	0.043
Abduction – Swing	0.160	-0.244*	-0.024

Parameter (º)	Age (yrs)	Walking Speed ($m \cdot s^{-1}$)	Age (yrs) Relationship When Controlling for Walking Speed ($m \cdot s^{-1}$)
Hip Rotation			
RoM	0.132	-0.016	0.093
Max Rotation – Stance	0.074	0.005	0.030
Min Rotation – Stance	0.057	0.020	0.021
Max Rotation – Swing	0.228*	-0.021	0.146
Min Rotation – Swing	0.041	0.067	-0.016
Knee Flexion/Extension			
RoM	0.034	0.058	-0.135
Flexion (LR)	-0.009	0.345*	-0.080
Extension (TS)	0.189*	-0.267*	-0.116
Flexion – Swing	-0.136	0.098	-0.127
Knee Varus/Valgus			
RoM	0.060	-0.048	0.140
Max Varus – Stance	0.004	0.054	0.041
Min Valgus – Stance	0.116	-0.101	0.160
Max Varus – Swing	0.054	0.035	0.008
Min Valgus – Swing	0.074	-0.039	0.133
Ankle Plantar/Dorsiflexion			
RoM	0.000	0.140	-0.071
Plantarflexion (LR)	-0.142	0.258*	0.030
Dorsiflexion - Stance	-0.066	0.002	-0.022
Max Plantarflexion	0.252*	-0.232*	-0.092

* Significant correlation. *Abbreviation:* Loading Response (LR) and Terminal Stance (TS).

4.3.3. Joint Kinetics

There was a significant age effect for peak moments, hip extension moment, knee flexion moment at loading response and knee valgus moment second peak (Table 4.5.). The over 75 years age group (-0.68 ± 0.50 Nm/kg) had a significantly reduced hip extension moment and knee flexion moment at loading response (0.70 ± 0.45 Nm/kg) compared to the 55-64 years (-1.27 ± 0.54 Nm/kg, 1.02 ± 0.31 Nm/kg) and 65-74 years (-1.19 ± 0.44 Nm/kg, 1.01 ± 0.29 Nm/kg) age groups. Knee valgus moment second peak was significantly reduced for the over 75 years age group (-0.03 ± 0.15 Nm/kg) compared to the 55-64 years age group (-0.12 ± 0.10 Nm/kg). There was also a significant age effect for peak powers knee generation 0, knee absorption 3, knee generation 4 and ankle generation 2. Knee power generation 0 was significantly reduced for the over 75 years age group (0.36 ± 27 Watts/kg) compared to the 55-64 years (0.77 ± 0.51 Watts/kg) and 65-74 years (0.68 ± 0.35 Watts/kg) age group. Knee power absorption 3, knee power generation 4 and ankle power generation 2 was significantly reduced for the over 75 years age group (-1.38 ± 0.61 Watts/kg, -1.73 ± 0.72 Watts/kg, 3.23 ± 0.71 Watts/kg) compared to the 55-64 years age group (-1.88 ± 0.62 Watts/kg, -2.27 ± 0.57 Watts/kg, 4.27 ± 1.28 Watts/kg) (Table 4.5. and Figure 4.6).

The SPM group analysis found phase significant age effect occurred for hip flexion/extension moment ($F_{1.996,135.630} = 6.876$, $P = 0.001$), hip power ($F_{1.996,135.630} = 7.188$, $P = 0.005$) and knee power ($F_{1.996,135.630} = 7.160$, $P = 0.001$). The over 75 years age group hip extension moment was significantly reduced during terminal stance compared to 55-64 years (41-53 %GC) and 65-74 years (45-51 %GC) age groups. The over 75 years age group were significantly reduced in late stance (52-56 %GC (terminal stance into pre-swing)) hip power generation compared to the 55-64 years age group, with a significant reduction of knee power generation at initial contact (0-1.5 %GC) compared to the 65-74 years age group (Figure 4.7.).

In addition, significant age correlations (Table 4.6.) were found for joint kinetics (hip extension moment, hip flexor moment in swing, knee flexor moment at loading response, knee extensor moment at terminal stance, knee flexor moment at pre-swing, knee extensor moment in swing, knee valgus moment (second peak). Correlations were also found between walking speed and joint kinetics (hip flexor moment in stance, hip extensor moment, hip flexor moment in swing, hip abductor moment (first peak), hip adductor moment (second peak), all knee flexion/extension moment peaks, knee varus moment (first peak), knee valgus moment (first and second peak), all ankle plantar/dorsiflexion moment peaks, hip power generation 1, hip power absorption 2 and all knee power peaks). There was also a significant relationship between age and knee valgus moment (second peak) when controlling for walking speed.

Table 4.5. Joint kinetics gait peaks (mean ± SD) for an older adult population during walking.

Parameter	Whole Group (n = 140)	55-64 yrs (n = 63)	65-74 yrs (n = 65)	≥ 75 yrs (n = 12)	*F* Value
Moments (Nm/kg)					
Hip Flexion/Extension					
Flexor Moment – Stance	0.79 ± 0.34	0.80 ± 0.33	0.81 ± 0.35	0.61 ± 0.31	$F_{2,137}$ = 1.801, P = 0.169
Extensor Moment	-1.18 ± 0.52	-1.27 ± 0.54	-1.19 ± 0.44	-0.68 ± 0.50 [ab]	$F_{2,137}$ = 7.306, P = 0.001*
Flexor Moment – Swing	0.67 ± 0.27	0.70 ± 0.27	0.68 ± 0.26	0.50 ± 0.28	$F_{2,137}$ = 2.815, P = 0.063
Hip Abduction/Adduction					
Abductor Moment (First Peak)	1.06 ± 0.29	1.10 ± 0.27	1.07 ± 0.29	0.90 ± 0.37	$F_{2,137}$ = 2.227, P = 0.112
Adductor Moment (First Peak)	-0.01 ± 0.20	-0.01 ± 0.20	-0.01 ± 0.19	0.00 ± 0.29	$F_{2,137}$ = 0.027, P = 0.973
Abductor Moment (Second Peak)	0.55 ± 0.41	0.49 ± 0.37	0.62 ± 0.42	0.53 ± 0.52	$F_{2,137}$ = 1.687, P = 0.189
Adductor Moment (Second Peak)	-0.33 ± 0.40	-0.37 ± 0.41	-0.28 ± 0.35	-0.39 ± 0.63	$F_{2,137}$ = 0.967, P = 0.383
Knee Flexion/Extension					
Flexor Moment (LR)	0.99 ± 0.33	1.02 ± 0.31	1.01 ± 0.29	0.70 ± 0.45[ab]	$F_{2,137}$ = 5.211, P = 0.007*
Extensor Moment (TS)	0.06 ± 0.15	0.06 ± 0.13	0.07 ± 0.16	0.03 ± 0.18	$F_{2,137}$ = 0.482, P = 0.619
Flexor Moment (PSw)	0.48 ± 0.17	0.51 ± 0.18	0.48 ± 0.16	0.39 ± 0.14	$F_{2,137}$ = 2.420, P = 0.093
Extensor Moment – Swing	-0.42 ± 0.10	-0.44 ± 0.11	-0.41 ± 0.09	-0.37 ± 0.11	$F_{2,137}$ = 3.083, P = 0.050
Knee Varus/Valgus					
Varus Moment (First Peak)	0.35 ± 0.15	0.36 ± 0.15	0.34 ± 0.14	0.33 ± 0.16	$F_{2,137}$ = 0.201, P = 0.818
Valgus Moment (First Peak)	-0.14 ± 0.12	-0.14 ± 0.11	-0.14 ± 0.11	-0.11 ± 0.16	$F_{2,137}$ = 0.289, P = 0.749
Varus Moment (Second Peak)	0.29 ± 0.15	0.29 ± 0.14	0.28 ± 015	0.35 ± 0.19	$F_{2,137}$ = 1.285, P = 0.280
Valgus Moment (Second Peak)	-0.11 ± 0.12	-0.12 ± 0.10	-0.11 ± 0.12	-0.03 ± 0.15[a]	$F_{2,137}$ = 3.402, P = 0.036*
Ankle Plantar/Dorsiflexion					
Plantarflexor moment (LR)	-0.37 ± 0.15	-0.36 ± 0.12	-0.39 ± 0.18	-0.29 ± 0.12	$F_{2,137}$ = 2.205, P = 0.114
Dorsiflexor moment (TS)	1.29 ± 0.23	1.31 ± 0.22	1.29 ± 0.25	1.24 ± 0.19	$F_{2,137}$ = 0.513, P = 0.600

Parameter	Whole Group (n = 140)	55-64 yrs (n = 63)	65-74 yrs (n = 65)	≥ 75 yrs (n = 12)	*F* Value
Powers (W/kg)					
Hip Power					
H1 (Generation)	0.77 ± 0.48	0.78 ± 0.50	0.75 ± 0.46	0.79 ± 0.46	$F_{2,137}$ = 0.071, P = 0.932
H2 (Absorption)	-0.89 ± 0.42	-0.94 ± 0.46	-0.86 ± 0.36	-0.76 ± 0.53	$F_{2,137}$ = 1.251, P = 0.289
H3 (Generation)	1.99 ± 0.65	2.07 ± 0.66	1.94 ± 0.61	1.81 ± 0.83	$F_{2,137}$ = 1.119, P = 0.330
Knee Power					
K0 (Generation)	0.70 ± 0.44	0.77 ± 0.51	0.68 ± 0.35	0.36 ± 0.27[ab]	$F_{2,137}$ = 4.706, P = 0.011*
K1(Absorption)	-1.11 ± 0.73	-1.11 ± 0.66	-1.18 ± 0.80	-0.67 ± 0.60	$F_{2,137}$ = 2.589, P = 0.079
K2 (Generation)	1.15 ± 0.63	1.22 ± 0.64	1.15 ± 0.58	0.74 ± 0.78	$F_{2,137}$ = 2.856, P = 0.061
K3 (Absorption)	-1.80 ± 0.63	-1.88 ± 0.62	-1.81 ± 0.59	-1.38 ± 0.61[a]	$F_{2,137}$ = 3.302, P = 0.040*
K4 (Generation)	-2.11 ± 0.59	-2.27 ± 0.57	-2.04 ± 0.55	-1.73 ± 0.72[a]	$F_{2,137}$ = 5.430, P = 0.005*
Ankle Power					
A1 (Absorption)	-0.98 ± 0.37	-1.00 ± 0.36	-0.96 ± 0.39	-0.97 ± 0.29	$F_{2,137}$ = 0.115, P = 0.891
A2 (Generation)	4.00 ± 1.17	4.27 ± 1.28	3.89 ± 1.04	3.23 ± 0.71[a]	$F_{2,137}$ = 4.809, P = 0.010*

* significant age effect. [a] Significantly different compared to 55-64 yrs. [b] Significantly different compared to 65-74 yrs. *Abbreviation:* Loading Response (LR), Terminal Stance (TS) and Pre-Swing (PSw).

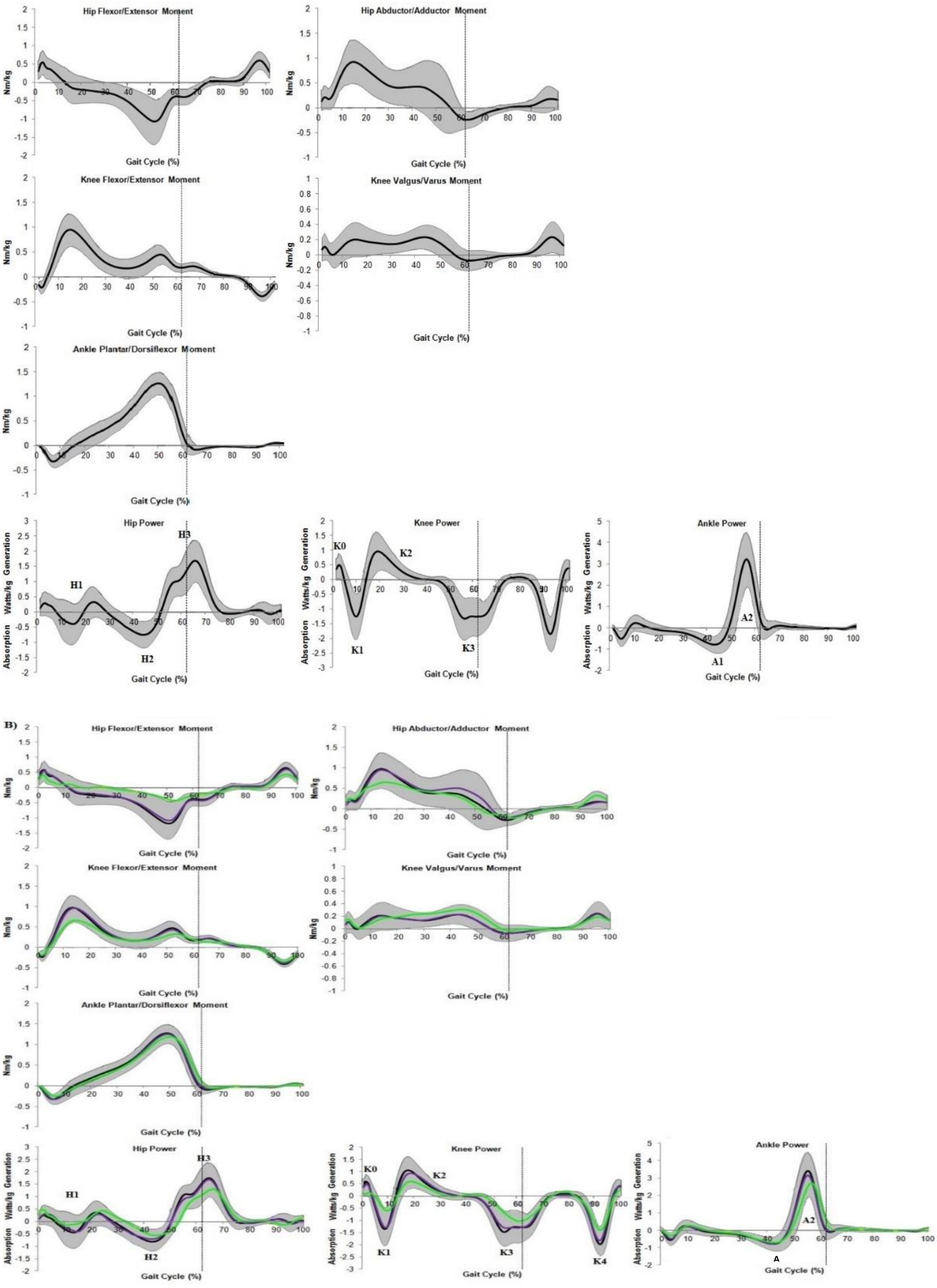


Figure 4.6. Joint kinetics (mean ± SD) for an older adult population during NW: **A)** whole group averages (n = 140) and **B)** age group averages (black line = 55-64 yrs (n = 63), purple line = 65-74 yrs (n = 65) and green ≥ 75 yrs (n = 12)). Power labels include: H1 (hip generation 1), H2 (hip absorption 2), H3 (hip generation 3), K0 (knee generation 0), K1 (knee absorption 1), K2 (knee generation 2), K3 (knee absorption 3), K4 (knee generation 4), A1 (ankle absorption 1) and A2 (ankle generation 2).

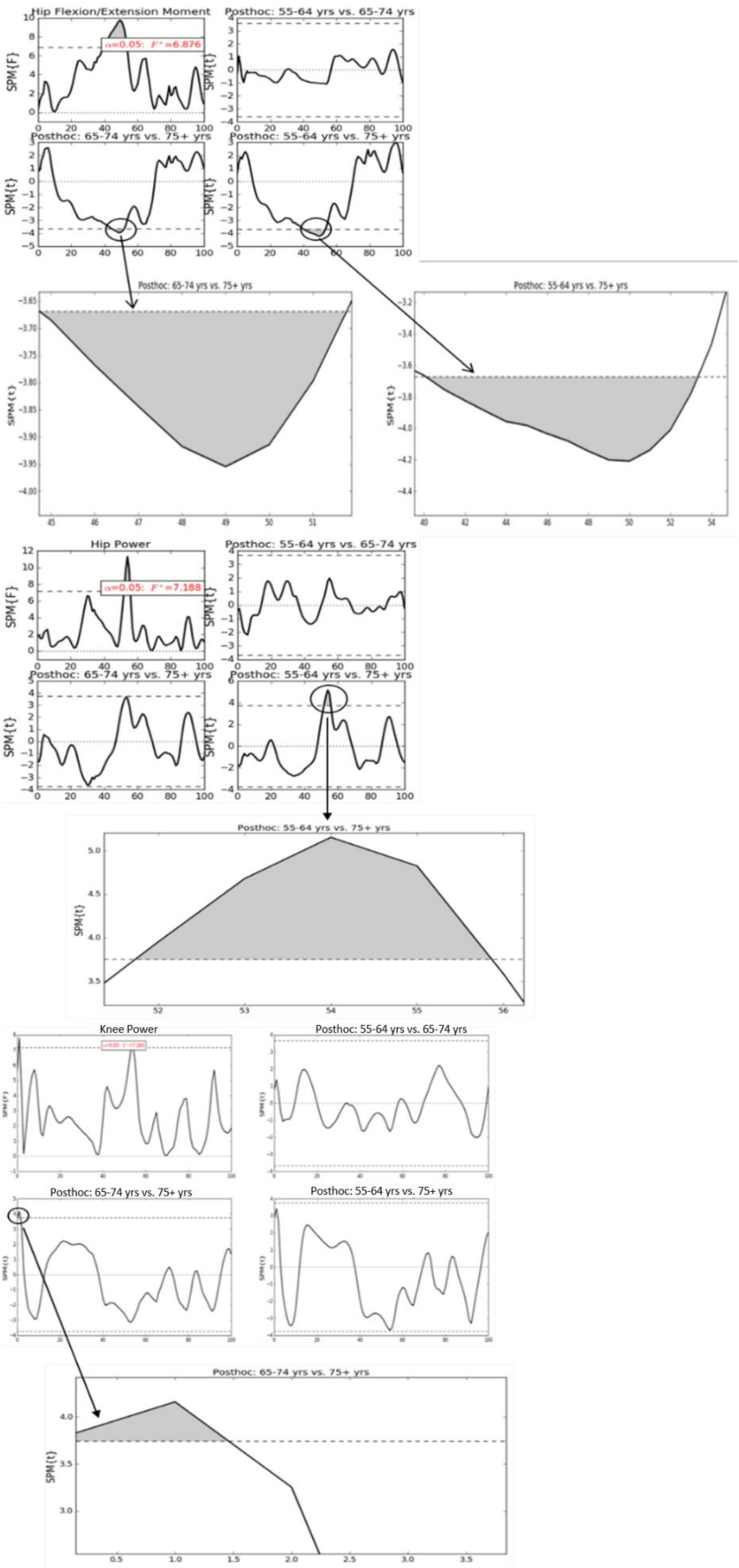


Figure 4.7. Statistical Parametric Mapping (SPM) output: age group post-hoc analysis for significant joint kinetics.

Table 4.6. Correlation between age, walking speed, peak joint kinetics and range of motion for an older adult population during walking, including a partial correlation to control for walking speed.

Parameter	Age (yrs)	Walking Speed ($m \cdot s^{-1}$)	Age (yrs) Relationship When Controlling for Walking Speed ($m \cdot s^{-1}$)
Moments (Nm/kg)			
Hip Flexion/Extension			
Flexor Moment - Stance	-0.078	0.391*	0.052
Extensor Moment	0.329*	-0.277*	0.017
Flexor Moment - Swing	-0.182*	0.303*	0.071
Hip Abduction/Adduction			
Abductor Moment (First Peak)	-0.173*	0.171*	0.128
Adductor Moment (First Peak)	-0.032	-0.018	-0.011
Abductor Moment (Second Peak)	0.094	-0.125	0.061
Adductor Moment (Second Peak)	0.061	-0.271*	-0.054
Knee Flexion/Extension			
Flexor Moment (LR)	-0.196*	0.429*	-0.038
Extensor Moment (TS)	-0.071	-0.273*	-0.109
Flexor Moment (PSw)	-0.214*	0.176*	0.064
Extensor Moment – Swing	0.193*	-0.434*	-0.072
Knee Varus/Valgus			
Varus Moment (First Peak)	-0.056	0.234*	0.080
Valgus Moment (First Peak)	0.074	-0.172*	0.351
Varus Moment (Second Peak)	0.064	0.027	0.031
Valgus Moment (Second Peak)	0.205*	-0.332*	-0.189*
Ankle Plantar/Dorsiflexion			
Plantarflexion moment (LR)	0.081	-0.325*	0.102
Dorsiflexion moment (TS)	-0.081	0.216*	0.084

Parameter	Age (yrs)	Walking Speed ($m \cdot s^{-1}$)	Age (yrs) Relationship When Controlling for Walking Speed ($m \cdot s^{-1}$)
Powers (W/kg)			
Hip Power			
H1 (Generation)	0.001	0.172*	0.062
H2 (Absorption)	0.206*	-0.405*	0.014
H3 (Generation)	-0.205*	0.541	0.117
Knee Power			
K0 (Generation)	-0.188*	0.433*	0.038
K1 (Absorption)	0.076	-0.430*	-0.011
K2 (Generation)	-0.148	0.586*	-0.049
K3 (Absorption)	0.222*	-0.497*	-0.063
K4 (Generation)	0.336*	-0.456*	0.032
Ankle Power			
A1 (Absorption)	-0.051	-0.097	-0.130
A2 (Generation)	-0.304*	0.539*	0.034

* Significant correlation. *Abbreviation:* Loading Response (LR), Terminal Stance (TS) and Pre-Swing (PSw). *Abbreviation:* Loading Response (LR), Terminal Stance (TS) and Pre-Swing (PSw).

4.4. Discussion

Alterations to gait can be used as a marker for current and future health, as walking is an important activity for independent living. Older adult gait assessment can identify an impaired pattern and predict fall risk (Kerrigan *et al.*, 1998). To date, the majority of research has investigated older adults' (typically 55-80 yrs) gait compared to younger adults (typically 20-40 yrs), with gait adaptations of older adults attributed to spatial-temporal variation (Bendall *et al.*, 1989). Predominantly this age-related adaptation is due to a decline in walking speed (1.2 % per year) with an increase in age (Alcock *et al.*, 2013). Therefore, a more suitable approach is to investigate changes to gait within an older adult population rather than viewing older adults as a single group, which disregards the ageing process. As such, age-effect gait adaptations within an older adult population remain unclear.

This study investigated gait using three-dimensional motion analysis for 140 community-dwelling older adults (age range of 55-86 yrs) during overground walking at a comfortable self-selected walking speed. The overall objective of this study was to examine the effects of age on gait parameters (spatial-temporal, joint kinematics and kinetics). There was a significant age effect for walking speed, stride/step length and double-support time. The main joint age effects were present at the hip for both kinematics and kinetics and knee and ankle for joint kinetics. A reduction in hip extension RoM, hip extension moment and hip generation was present during late stance for the older adults aged 75 years and above. Knee power generation and absorption and ankle generation was reduced for the older adults aged 75 years and above, with a phase significance occurring for knee generation at initial contact. No significant differences were found between the 55-64 years and 65-74 years age groups. Age-related gait adaptations can derive from a change in kinetic strategy which results in a reduced ability to generate joint moments thus adapting the kinematics and/or a change in kinematic strategy which results in an altered neuromuscular control thus adapting the kinetics (Sorenson and Flanagan, 2015).

Walking speed in this study was notably faster to that reported in the literature for older adults walking at a self-selected comfortable speed (Kerrigan *et al.*, 1998, Silder *et al.*, 2008, Cooper *et al.*, 2011a, Hollman *et al.*, 2011, Toda *et al.*, 2015, Toda *et al.*, 2016). For example, walking speed reported from ELSA (Cooper *et al.*, 2011a) for older adults aged 60-64 years was 1.0 ± 0.3 $m \cdot s^{-1}$ compared to 1.45 ± 0.18 $m \cdot s^{-1}$ in this current study. DeVita and Hortobagyi (2000) however reported an average walking speed of 1.43 $m \cdot s^{-1}$ for older adults, which is similar to these findings (1.41 ± 0.19 $m \cdot s^{-1}$). This fast walking speed, which was freely chosen, is indicative of relatively healthy older adults (physically fit with a good neuromuscular function). Walking speed was significantly slower for the over 75 years age group (1.24 ± 0.21 $m \cdot s^{-1}$), whereas the 55-64 years (1.45 ± 0.18 $m \cdot s^{-1}$) and 65-74 years (1.40 ± 0.17 $m \cdot s^{-1}$) age groups walking speed was similar, which suggest for this population 75 years of age was when walking speed significantly declined. This was similar to Daly *et al.* (2013) who reported greater deterioration in walking speed from the age of 70 years and above. As such, walking speed is relatively stable for this older adult population up to the age of 74 years. A reason for reducing walking speed is thought to be a compensatory strategy adopted during walking when stability is challenged (Hollman *et al.*, 2007). Furthermore, muscle force generation is essential for walking and an increase in walking speed is accompanied by an increase in joint moments (Riley *et al.*, 2001). Therefore, a slower walking speed in older adults is also likely to be a consequence of reduced strength (Bohannon *et al.*, 1996), muscle weakness (Busse *et al.*, 2006) and impairment of motor control (Kaya *et al.* 1998). It may also be indicative of a 'cautious' gait for the over 75 years age groups. Walking speed decline with ageing still remains unclear if this was a compensatory effort to improve walking ability (i.e. safety) (Winter *et al.*, 1990), related to fear of falling (Chamberlin *et al.*, 2005) or simply deterioration of muscle activity (Ko *et al.*, 2010b).

'Cautious' gait is not only indicative of a reduced walking speed but also, a change in other spatial-temporal parameters such as short step length, increased step timing variability and an increased double-support time (Menz *et al.*, 2003). Not only did the over 75 years age group have a slower walking speed but also illustrated a reduced stride/step length and an increased double-support time. Research (Elble *et al.*, 1991, Lord *et al.*, 1996a, Lord *et al.*, 1996b, Maki, 1997, Toulotte *et al.*, 2006, Kang and Dingwell, 2008b, Toebes *et al.*, 2012) has commented on a greater association of stride-to-stride variability and walking speed with fall risk for older adults. An increase in variability for older adults may be a result of slower walking speeds (Kang and Dingwell, 2008b). Although, variability was not assessed; 'cautious' gait may have been present for the over 75 years age group. However, it must be noted that the double-support time was not significantly different to the 65-74 years age group. This study hypothesised an age effect would be present for walking speed, stride/step length, stride width and double-support time. As such, this hypothesis is only partially accepted, as there was no significant difference in stride width.

Age-related differences were mainly observed in the joint kinetics. A reduced hip extension moment in late stance was found for the over 75 years age group compared to the 55-64 years and 65-74 years age groups, which is commonly reported in the literature for older adults when compared to young adults (Kerrigan *et al.*, 1998, DeVita and Hortobagyi, 2000, Monaco *et al.*, 2009). In addition, unlike peak analysis, SPM found hip power generation reduced in late stance for the over 75 years age group compared to the 55-64 years age group, this contradicts the literature which reported an increased hip power generation in older adults when compared to younger adults (DeVita and Hortobagyi, 2000, Silder *et al.*, 2008, Monaco *et al.*, 2009). Although hip power generation 3 was not significantly correlated to walking speed, both hip power generation 1 and absorption 2 were significantly correlated to walking speed. In addition, walking speed for the over 75 years age group was significantly reduced, as such a slower walking speed may indirectly have resulted in less power generation. Reduced hip power generation in late stance may impact the swing phase transition of the ipsilateral limb. As such, for the over 75 years could affect successful toe-clearance and would require compensatory strategies to occur on the contralateral limb to ensure clearance.

Reduction in knee power generation and absorption maybe a consequence of reduced knee extensor or quadricep muscle activity (Winter, 1991, Perry *et al.*, 2007). Winter (1991) suggested a reduction in knee power generation 4 leads to less energy absorption of the hamstrings during terminal swing, which reduces the rate of deceleration available prior to heel contact and as such increases the probability of a fall. However, Kerrigan *et al.* (1998) found knee power generation and absorption either exceeded or was comparable to young adults when older adults walked at a faster walking speed compared to their comfortable speed. The results revealed walking speed was associated with all knee power peaks and when controlling for walking speed there was no association with age and knee power peaks. As such, the reduction in knee power observed in this study maybe a consequence of reduced walking speed for the over 75 years age group. It would be interesting to observe if fast walking would increase joint power for the over 75 years age group.

Similar to Kerrigan *et al.* (1998), these results demonstrated ankle moment was preserved between age groups which is uncommon as a reduced dorsiflexor moment is typically observed in older adults (DeVita and Hortobagyi, 2000, Silder *et al.*, 2008, Anderson and Madigan, 2014). Ankle power generation (A2) was reduced for the over 75 years age group compared to the 55-64 years age group, which is commonly reported for older adults compared to young adults (Winter *et al.*, 1990, Judge *et al.*, 1996, Kerrigan *et al.*, 1998, DeVita and Hortobagyi, 2000, Riley *et al.*, 2001, Silder *et al.*, 2008, Monaco *et al.*, 2009, Schloemer *et al.*, 2017). Because there was no difference compared to the 65-74 years age group, it suggests these changes in this healthy population are starting to decline after the age of 65 years old. Reduced ankle power generation (A2) is associated with a reduced walking speed and suggests weakness/reduced strength in the plantarflexor muscles (Bassey *et al.*, 1988, Winter *et al.*, 1990, Winter, 1991, Bassey *et al.*, 1992, Judge *et al.*, 1996, Kerrigan *et al.*, 1998). An increase in age has found to be associated with a decline in plantarflexor muscle strength (Christ *et al.*, 1992, Gajdosik *et al.*, 1999). For example, Gajdosik *et al.*, (1999) found torque reductions of 40-45 % in the calf muscles at isokinetic velocities of 30-180 °/s for women aged 20 to 84 years.

The ankle during late stance is an important joint which aids the propulsion of the ipsilateral limb from stance to swing created from a forward and vertical acceleration of the upper body (Neptune *et al.*, 2001). As such, eliciting a slower walking speed with advanced age, which this study also found for the over 75 years age group, has found to be associated with reductions in propulsive forces during the push-off phase of walking (Franz and Kram, 2013). As previously mentioned in this discussion, biomechanical age changes for joint kinetics may be associated with age-related factors such as muscle weakness or even sarcopenia (Franz, 2016). Moreover, a reduce walking speed may also precipitate propulsive force reduction during push-off (Franz, 2016), which potentially explains the reduced ankle power generation (A2) seen for the over 75 years age group. The results also found an association with ankle power generation (A2) and walking speed, however when controlling for walking speed there was no correlation between age and ankle power generation (A2). These biomechanical changes associated with ageing and the decline in muscle-force generating capacity, account for between 48-75 % of the explained variance in ankle power (Silder *et al.*, 2008).

Walking slower for older adults and smaller propulsive forces may be indicative of stability prioritisation over mobility (Browne and Franz, 2017). Smaller propulsive forces for older adults has found to be associated with 11-35 % reduction in mechanical power by the propulsive ankle plantarflexor muscles (Devita and Hortobagyi, 2000, Franz and Kram, 2014). Therefore, ankle power generation is critical for gait, especially as ankle power generation (A2) contributes to limb advancement (i.e. toe-clearance) and centre of mass acceleration (Zelik and Adamczyk, 2016). As such, a reduced mechanical ankle power generation (A2) at push-off may affect the successfulness of toe-clearance for this over 75 years age group, which may also increase the likelihood of trip probability. As such, future work is required to investigate toe-clearance and joint mechanisms within this older adult population.

Comparable to DeVita and Hortobagyi (2000) and Schloemer *et al.* (2017), this study did not support the common hypothesis that older adults increase their hip extensor kinetics to compensate for reduced plantarflexor kinetics. Unlike previous work, this study demonstrates a more lower body proximal shift in gait adaptations (i.e. the hip), as oppose to distal changes. Therefore, the hypothesis is only partially accepted. The results demonstrate the over 75 years age group have a reduced hip extension RoM and moment and power generation in late stance, also knee power generation and absorption is reduced. This suggests for this population, the over 75 years age group, hip extensor tightness, muscle weakness and strength (Kang and Dingwell, 2008a) maybe affecting their gait pattern.

Reduced hip extension RoM was found for the over 75 years age group, which is comparable to older adult research when compared to young adults during self-selected comfortable walking speed (Winter *et al.*, 1990, Judge *et al.*, 1996, Kerrigan *et al.*, 1998, Kerrigan *et al.*, 2001, Graf *et al.*, 2005, Lee *et al.*, 2005, Monaco *et al.*, 2009, Anderson and Madigan, 2014). The results also illustrate a significant decline in hip extension was present within late stance, not just peak hip extension. However, this was only significantly different to the 55-64 years age group. Changes in hip extension leads to a decrease in stride length and walking speed with a preserved cadence which is seen for older adults when comparing to young adults (Winter *et al.*, 1990, Judge *et al.*, 1996, Kerrigan *et al.*, 1998). This was also observed for the over 75 years age group. Reduced hip extension and stride length may be a compensation for poor balance (Kerrigan *et al.*, 2001) and/or walking speed. When controlling for walking speed, there was no significant association between hip extension and age. Reduced hip extension has been found to signify not only hip tightness but hip flexion contractures, which prevents full hip extension in gait (Shimada, 1996, Lee *et al.*, 1997) and lead to a reduce stride length (Kerrigan *et al.*, 2001). Hip contractures can lead to an altered walking pattern (Kerrigan *et al.*, 2001). A reduction in hip extension and stride length may pose biomechanical threats when encountering an obstacle or uneven surface (Kerrigan *et al.*, 2001).

A compensation for reduced hip extension and stride length is an increase in anterior pelvic tilt (Judge *et al.*, 1996, Shimada, 1996, Lee *et al.*, 1997, Kerrigan *et al.*, 1998, Kerrigan *et al.*, 2001, Lee *et al.*, 2005). Although no significant difference was reported for anterior pelvic tilt for this study, averages were higher for the over 75 years age group (10.44 ± 8.31 °) compared to the 55-64 years (6.13 ± 5.67 °) and 65-74 years (6.36 ± 6.19°) age groups, so a compensatory strategy may have been employed. Kerrigan *et al.* (2001) found a reduced hip extension not only distinguished gait of healthy young (28.1 ± 4.2 yrs) but also healthy older adults (73.2 ± 5.6 yrs) when compared to older adults who fall (77.0 ± 7.8 yrs). The 75 years age group peak hip extension (-5.69 ± 10.58 °) was lower for this study compared to Kerrigan *et al.* (2001) for healthy older adults (-14.3 ± 4.4 °). Therefore, for the 75 years age group reduced hip extension may be a marker for potential fall risk.

SPM to date has not been used to investigate gait age effects within an older adult population. This analysis tool provides a novel method to evaluate the effect of age in specific-phases of the gait cycle rather than investigate gait peaks. For example, hip power statistical analysis using conventional methods (i.e. ANOVA) found no significant age effect, yet SPM identified there was a phase-specific significant age effect. SPM was designed for testing region-of-interest related hypotheses, which is valid for one-dimensional segmented regions (Pataky *et al.*, 2016). SPM provides a comprehensive statistical solution to complex biomechanical systems (Pataky *et al.*, 2013), which avoids statistical assumptions regarding the spatial-temporal field signals and is advantageous of the unified framework (Pataky, 2010).

A limitation for this study was no lower limb strength measurements. As mentioned above, muscle weakness may potentially have been the ageing factor which influenced the change in joint kinetics for the over 75 years age group. As such, an isometric and/or isokinetic dynamometer test for lower limb strength measurements would have provided knowledge of the age-related strength changes in this older adult population, which may have provided a better understanding of the impact of gait change especially for the over 75 years age group. Although, this measurement would have been beneficial, it may have impacted on other parts of the protocol, due to factors such as fatigue. In addition, this study was the relatively small number of participants aged 75 years and over, which was possibly due to the travel requirements to access the University where data was collected.

4.5. <u>Conclusion</u>

This study presents a large dataset for gait parameters (joint kinematics and kinetics) for older adults walking at a comfortable self-selected walking speed. This study identified that age-related gait adaptations occur from the age of 75 in this population. No age effect was present between 55-64 years and 65-74 years for this population suggesting that gait parameters are relatively stable up to the age of 74. As such, the age effect has shifted to 75 years of age and above. The majority of changes with age occurred at the hip. For the over 75 years age group, there was a reduction in hip extension, extension moment and power generation in late stance, along with reduced knee power generation and absorption and ankle power generation. These age-related changes for the over 75 years age group are associated with a reduction in walking speed. Reduced walking speed, altered joint moments and reduced joint powers (e.g. reduce ankle power) has found to be associated with age-related biomechanical changes such as muscle weakness. Such changes found for this adult population may influence the success of toe-clearance. For example, the over 75 years age group had a reduced hip extension and ankle power generation during late stance, which could impact the effectiveness of push-off for the limb swing advancement and if successful toe-clearance is achieved, does this age group alter their gait pattern to ensure success (i.e. hip adduction). Consequently, future work is required to investigate the age effect on toe-clearance parameters and determine the joint kinematics mechanisms associated with the swing phase.

Chapter Five: The Effect of Age on Toe-Clearance Parameters in an Older Adult Population

Abstract

Introduction: Tripping is the main cause of falling in older adults, with the greatest risk of tripping whilst walking occurring at minimum toe-clearance. Toe-clearance is a complex biomechanical interaction with research identifying that successful toe-clearance is associated with two additional events; first, maximum toe-clearance (following toe-off) and second, maximum toe-clearance (after 90 % of the gait cycle). These parameters have been found to influence the amplitude of minimum toe-clearance. Furthermore, everyday walking usually involves a secondary task. The aim of this study was to determine if toe-clearance parameters were affected by walking task and if there was an age-related association in an older adult population. A secondary aim was to determine if fall history affected toe-clearance parameters.

Methods: 158 community-dwelling older adults, age range 55-86 years (65.7 ± 6.8 years) participated and walked at their self-determined comfortable walking speed. All participants were grouped into three age categories: 55-64 years, 65-74 years and over 75 years. Three-dimensional motion analysis was used to capture the trajectories of the shoe-mounted toe markers and the joint kinematics of both the limbs. Three toe-clearance events were identified: first maximum toe-clearance, minimum toe-clearance and second maximum toe-clearance during normal and manual dual task (carrying a cup of water) walking. Mixed ANOVAs, were performed to determine the age and task effect on toe-clearance parameters.

Results: No age effect occurred for first maximum toe-clearance and minimum toe-clearance. There was however a significant age effect for second maximum toe-clearance. The over 75 years age group had a significantly lower second maximum toe-clearance (73.7 ± 14.9 mm) compared to the 55-64 years (88.8 ± 17.4 mm) and 65-74 years (89.3 ± 15.5 mm) age groups. All toe-clearance events were significantly lower for manual dual task compared to normal walking. For example, minimum toe-clearance was significantly lower for manual dual task walking (12.4 ± 4.9 mm) compared to normal walking (14.7 ± 6.0 mm). There were significant differences between manual dual task and normal walking in joint kinematics at the toe-clearance events. For example, hip adduction was significantly increased at minimum toe-clearance for dual task walking (7.5 ± 4.1 °) compared to normal walking (1.7 ± 4.4 °). There was no significant difference for fall history on toe-clearance parameters for NW or DT walking and as such no further analysis on fall history was executed.

Conclusion: Age did not affect minimum toe-clearance. Age did however effect second maximum toe-clearance. This parameter is associated with peak dorsiflexion, as such for the over 75 years may have a limited dorsiflexion range due to weak ankle plantarflexor muscles which was found in Chapter Four. As such, a reduced dorsiflexion with an increase in age may compromise toe-clearance and potentially increase the risk of tripping. The reduction in toe-clearance during manual dual task walking, suggested this task placed older adults at a greater risk of tripping. Gait adaptations were present when performing the manual dual task, with significantly altered joint kinematics for the ipsilateral and contralateral limbs. This study represents one of the largest databases of toe-clearance parameters for older adults walking overground at a self-selected speed, which identified toe-clearance parameters were affected by age and walking task. However, successful toe-clearance was achieved, as such additional gait factors may be influencing the successfulness of toe-clearance, for instance increased arm swing. Future work is therefore required to determine the role of arm swing in this older adult population and how arm swing is influenced by walking task.

Keywords: Gait; Toe-Clearance; Biomechanics; Falls

5.1. Introduction

Falls cost the UK National Health Service £1.7 billion/year (Age UK, 2010) and approximately 14,000 deaths associated with falling (Martin, 2008). Tripping accounts for the majority of all falls in older adults (Blake *et al.*, 1988, Berg *et al.*, 1997, Zhou *et al.*, 2002). Experiencing a trip, whilst walking is an event when the foot during the swing phase makes an unanticipated contact with an obstacle, an object, or the supporting surface causing instability and if unrecovered results in a fall (Chen *et al.*, 1991, Galna *et al.*, 2009, Nagano *et al.*, 2011). While consequences of ageing such as reduced walking speed and stride length may increase the probability of tripping for older adults (van Dieën *et al.*, 2005) research primarily focuses on toe-clearance, specifically minimum toe-clearance (MTC). MTC occurs at a critical time point during the swing phase where the toe closely approaches the ground, which can be as low as 10 mm above the ground (Winter *et al.*, 1990). Also, during this phase the horizontal toe velocity is near maximum and the body's centre of mass is located anterior to the stance foot and outside the base of support in the direction of progression (Winter, 1992). Consequently, foot trajectories during swing phase must not only maintain progression in the direction of travel which is reflected in step length but also incorporate a vertical displacement component which is sufficient to accommodate changes in the elevation of the supporting surface (Winter, 1992, Begg *et al.*, 2007).

The predictive value of MTC has been investigated in relation to trip risks in older adults (Begg *et al.*, 2007, Best and Begg, 2008, Begg *et al.*, 2014). A systematic review (Barrett *et al.*, 2010) revealed comparing young to older adults does not cause alterations to MTC central tendencies (e.g. mean and median) or disruptions during overground and treadmill normal walking (NW). Seven studies (Winter *et al.*, 1990, Elble *et al.*, 1991, Bunterngchit *et al.*, 2000, Mills and Barrett, 2001, Begg *et al.*, 2007, Khandoker *et al.*, 2008, Mills *et al.*, 2008) reported no significant difference in MTC central tendency measures between young and older adults (Table 5.1.). Only one study (Menant *et al.*, 2009) showed an age effect where mean MTC was significantly greater for older females (78.5 ± 4.2 yrs) compared to younger females (27.4 ± 2.5 yrs) when investigating the influence of swing gait across various walking surfaces (overground: levelled, wet and irregular walkways). Although the literature implies there is no age effect on central tendencies for MTC during NW the above studies compared young to older adults.

Table 5.1. The studies investigating minimum toe-clearance for young and older adults during overground and treadmill normal walking.

Studies	Older Adults	Younger Adults	Walking Surface	Footwear	Walking Speed	Minimum Toe-clearance (mm)
Winter *et al.* (1990)	N = 15 Mean age 68.0 yrs. Age range 62-78 yrs. 5 Females/10 Males Height (cm) = 172.0 ± 9.0 Mass (kg) = 77.2 ± 13.4	N = 12 Mean age 24.6 yrs. Age range 21-28 yrs. 5 Females/7 Males Height (cm) = 173.0 ± 10.0 Mass (kg) = 69.2 ± 10.4	Overground	Not stated	Self-selected	Elderly 11.1 ± 5.3 mm Young adults 12.7 ± 5.9 mm
Elble *et al.* (1991)	N = 20 74.7 ± 6.6 yrs. Age range 65-87 yrs. 11 Females/9 Males Height (cm) = 165.0 ± 8.0 Mass (kg) = 65.9 ± 10.3	N = 20 30.0 ± 6.1 yrs. Age range 20-39 yrs. 10 Females/10 Males Height (cm) = 171.0 ± 10.0 Mass (kg) = 68.4 ± 16.1	Overground	Barefoot	Self-selected normal and fast walking	Normal walking Elderly 16.0 ± 7.0 mm Young adults 14.0 ± 4.0 mm Fast walking Elderly 21.0 ± 8.0 mm Young adults 17.0 ± 7.0 mm
Bunterngchit *et al.* (2000)	N = 10 72.0 ± 4.35 yrs. Age range ≥65 yrs. 5 Females/5 Males Height (cm) = 170.0 ± 7.1 Mass (kg) = 77.0 ± 15.1	N = 10 26.0 ± 4.4 yrs. 5 Females/5 Males Height (cm) = 166.0 ± 4.1 Mass (kg) = 55.0 ± 5.3	Overground	Shod	Cadence of 100 steps/min	Elderly 19.2 ± 3.7 mm Young adults 20.0 ± 2.2 mm
Mills and Barrett (2001)	N = 8 68.9 ± 0.4 yrs. Age range 65-74 yrs. 8 Males Height (cm) = 172.9 ± 1.6 Mass (kg) = 82.2 ± 3.7	N = 10 24.9 ± 0.9 yrs. Age range 20-30 yrs. Height (cm) = 176.8 ± 1.4 Mass (kg) = 74.2 ± 2.3	Overground	Shod	Self-selected	Elderly men 21.0 ± 2.0 mm Young men 20.0 ± 3.0 mm

Studies	Older Adults	Younger Adults	Walking Surface	Footwear	Walking Speed	Minimum Toe-clearance (mm)
Begg *et al.* (2007)	N = 16 72.1 ± 4.9 yrs. 16 Females Height (cm) = 159.0 ± 6.0 Mass (kg) = 65.7 ± 7.1	N = 17 26.4 ± 4.9 yrs. 17 Females Height (cm) = 166.0 ± 6.0 Mass (kg) = 65.1 ± 9.9	Treadmill	Shod	Self-selected	Elderly females 14.8 ± 7.6 mm Young females 15.6 ± 6.2 mm
Khandoker *et al.* (2008)	N = 27 69.1 ± 5.12 yrs. 27 Healthy Females Height (cm) = 165.0 ± 7.8 Mass (kg) = 66.8 ± 8.4 N = 10 72.2 ± 3.1 yrs. 10 Females with falls risk Height (cm) = 166.0 ± 12.0 Mass (kg) = 66.9 ± 8.6	N = 30 28.4 ± 6.4 yrs. 30 Healthy Females Height (cm) = 171.0 ± 12.0 Mass (kg) = 71.2 ± 15.0	Treadmill	Shod	Self-selected	Healthy elderly 12.5 ± 4.7 mm Elderly females with falls risk 20.2 ± 5.1 mm Healthy young 14.6 ± 5.2 mm
Mills *et al.* (2008)	N = 9 71.1 ± 3.4 yrs. 9 Males Height (cm) = 172.0 ± 6.0 Mass (kg) = 82.7 ± 11.6	N = 10 25.8 ± 3.1 yrs. 10 Males Height (cm) = 176.0 ± 7.0 Mass (kg) = 74.4 ± 9.1	Treadmill	Shod	Self-selected	Elderly males 13.8 ± 2.1 mm Young males 14.9 ± 1.6 mm

Successful toe-clearance has been found to be associated with two additional events during the swing phase; first maximum toe-clearance (MxT1) and second maximum toe-clearance (MxT2) (Nagano *et al.*, 2011). A complex biomechanical interaction of toe-clearance parameters were observed, which influenced the amplitude at MTC (Begg *et al.*, 2007). Nagano *et al.* (2011) investigated the age effects of these toe-clearance events (MxT1: occurring after toe-off, MTC and MxT2: after 90% in the gait cycle) during overground and treadmill NW for 11 older (4 females, 7 males; 73.8 ± 7.2 yrs) and 11 young adults (4 females, 7 males; 22.5 ± 2.9 yrs). The only age effect present was at MxT2, revealing a reduced MxT2 for the older adults. MxT2 event coincides with peak dorsiflexion (Winter, 1991) and Nagano *et al.* (2011) suggested reduced MxT2 was attributed to weak dorsiflexor muscles for the older adults.

Walking while performing another task is commonly observed during daily activities and this is known as dual task (DT) walking, this may create a conflict and a need to determine which task receives priority (Pashler, 1994, Tombu and Jolicoeur, 2003). It has been observed walking speed is reduced when performing a DT compared to NW (Smith *et al.*, 2016). Manual DT's (e.g. holding an object when walking) is argued to be more ecological valid in laboratory settings than cognitive DT's (e.g. counting backwards in 7's), as such tasks are performed more frequently in daily activities. Santhiranayagam *et al.* (2015) used a manual DT and reported no significant difference between younger (15 young adults; 4 females; 11 males; 26.1 ± 3.8 yrs) and older adults (15 older adults; 7 females, 8 males; 73.1 ± 5.6 yrs) MTC.

Previous research which has compared MTC displacements has focused on comparing young to older adults (e.g. ≤ 25 vs. ≥ 65 yrs). Gait differences are observed year on year in older adults (Ashton-Miller, 2005). As such a different research approach would be to investigate toe-clearance parameters within a group of older adults. Therefore, the objective of this study was to determine if toe-clearance parameters (MxT1, MTC and MxT2) were affected by task and if they were related to age for an older adult population performing two walking tasks (NW and manual DT walking).

MTC on average is 10-20 mm above the ground, with a horizontal toe velocity of 4.3 $m \cdot s^{-1}$ (Winter *et al.*, 1990, Barrett *et al.*, 2010). MTC is a fine complex endpoint motor control, which consists of a seven link-segment chain and 12 degrees-of-freedom over 2 m long, starting from the contralateral stance limb which moves up the leg across the pelvis down towards the foot of the ipsilateral swing limb (Winter *et al.*, 1990). Winter (1992) found kinematics of the sagittal plane ipsilateral swing phase of the knee and ankle and contralateral stance phase hip abduction influence the trajectory of the swing foot, thereby influence MTC. Mills *et al.* (2008) suggested exploring the kinematics at the time of MTC may provide insight into MTC control. The preceding ipsilateral limb of the stance phase prior to the swing phase is likely to influence the swing foot kinematics and therefore MTC, which is not currently established in the literature. Furthermore, if this kinematic adaptation occurs when performing a dual task, it may provide a mechanism for underlying control of MTC during functional tasks.

As observed in chapter four, there was an age effect on gait for this older adult population during normal walking. The over 75 years age group altered their hip biomechanics and spatial-temporal parameters. Typically, using a joint kinetic strategy for gait alterations during normal walking. The results revealed this age group had a significantly reduced hip extension range of motion during terminal stance, with a reduced hip moment and power in late stance. In addition, knee power generation and absorption and ankle power generation prior to toe-off was reduced. This was suggested to be a consequence of ageing caused by muscle tightness and weak muscles (Kang and Dingwell, 2008a). Muscle force generation and walking speed affect joint moments (Riley *et al.*, 2001). Therefore, a slower walking speed accompanied with a reduced muscle force generation for older adults especially around late stance, will reduce the propulsive forces during the push-off phase (i.e. toe-off) of walking which initiates swing (Franz and Kram, 2013). As such, these gait alterations observed in chapter four for the over 75 years age group could impact the effectiveness of the push-off phase (i.e. toe-off) for limb swing advancement and consequently affect the successfulness of toe-clearance.

Therefore, the aim of this study was to establish if toe-clearance events decreased with age and task and if the joint kinematics of the ipsilateral and contralateral limb adapt to performing a DT. It was hypothesised toe-clearance events would significantly reduce for DT walking compared to NW. Also, as gait changes are observed with an increase in age in older adults, age would be significantly associated with an increase toe-clearance event for both walking tasks to avoid a trip. In addition, joint kinematics would adapt to the DT when walking. A secondary aim was to determine if fall history affected toe-clearance parameters. It was hypothesised older adults who had previously fallen would have significantly reduced toe-clearance parameters for both walking tasks compared to older adults with no falls.

5.2. Methodology

The overall methodology is provided in Chapter Two. All participants performed five trials for normal walking (NW without force plate contact) and manual dual task (DT) walking (carrying a cup of water). Walking speed ($m \cdot s^{-1}$) was derived from the Brower timing gates (Utah, USA) and calculated in Excel (Microsoft Office 2010, Tokyo, Japan). All remaining analysis was completed using a custom-made Python code (Python v. 2.7.10, Delaware, USA). Toe marker trajectories (Z axis) and joint kinematics (lower body) were normalised to one gait cycle (100 %), using linear interpolation to 101 data samples. The lowest value of the normalised right vertical toe displacement in the stance phase (prior to toe-off) was found for both walking tasks, this value was then subtracted against the normalised vertical displacement to zero the trajectories prior to toe-off (Figure 5.1.), which is in accordance with Winter *et al.* (1990) toe-clearance analysis method. From this, three toe-clearance events were identified (Figure 5.2.): MxT1 (following toe-off), MTC (approximately 80 % in the gait cycle which is the lowest value preceding MxT1) and MxT2 (after 90 % in the gait cycle).

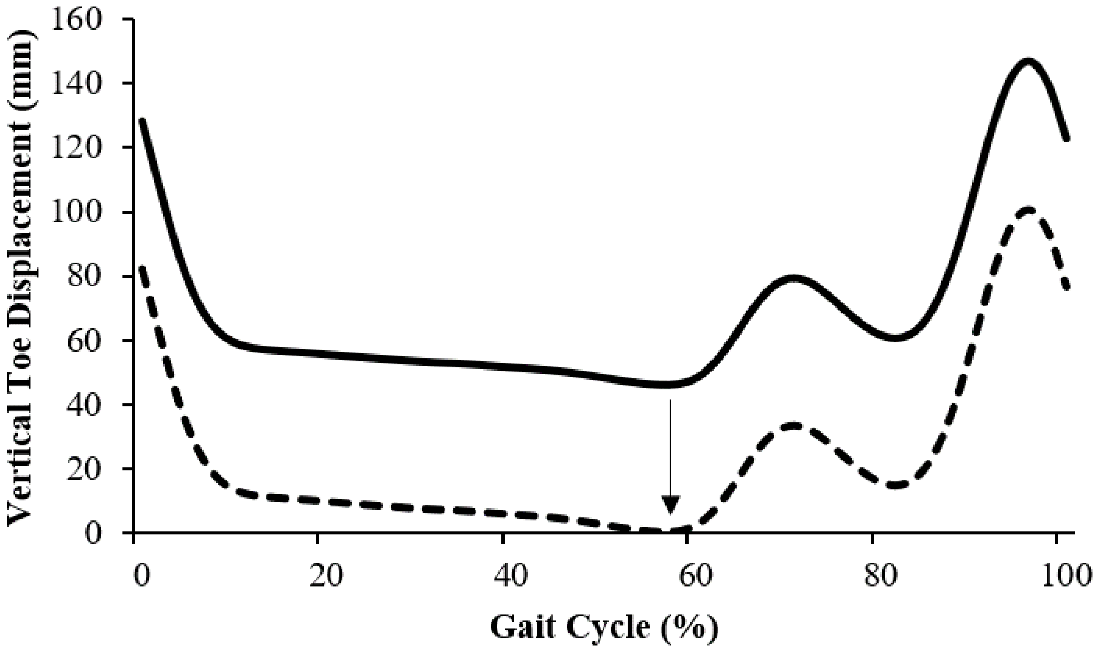


Figure 5.1. Example of the zeroing of the vertical toe displacement prior to toe-off. *Note:* Solid black line indicates the raw vertical toe displacement data prior to zeroing. The minimum value before toe-off was then found and subtracted against the entire raw vertical toe displacement, in order to zero the trajectories prior to toe-off. Dashed line indicates the zeroed vertical toe displacement trajectory.

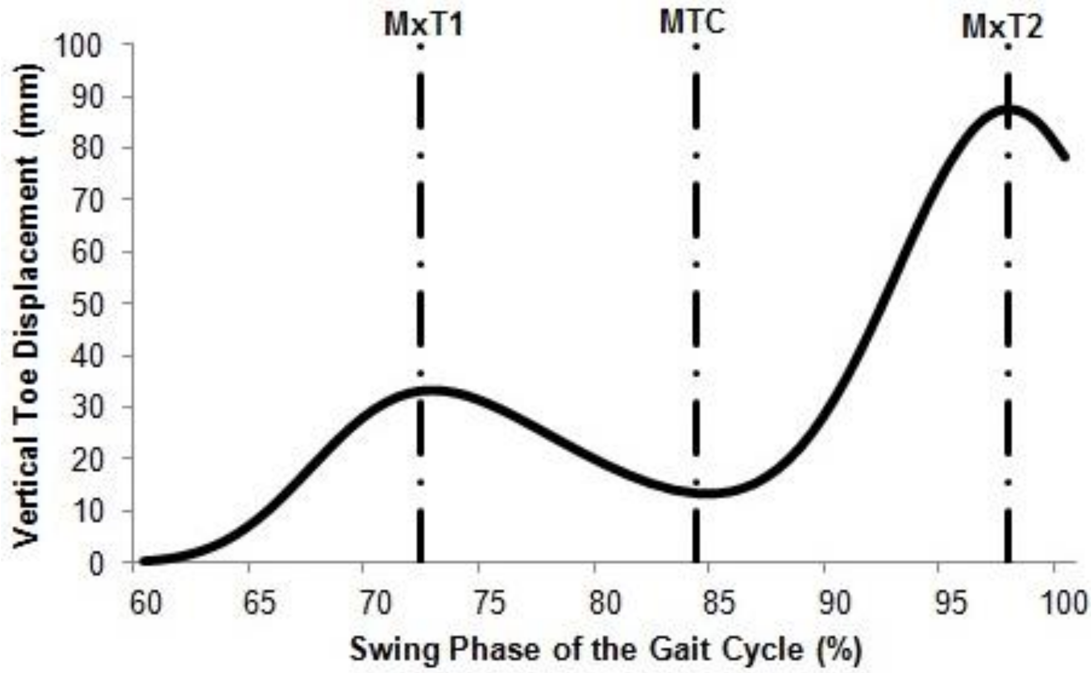


Figure 5.2. The three toe-clearance events (MxT1 (first maximum toe-clearance), MTC (minimum toe-clearance) and MxT2 (second maximum toe-clearance)).

The calculations of the joint kinematics were guided by Winter (1992). For the ipsilateral limb, joint kinematics were calculated during the stance and swing phase in the sagittal plane at the hip (flexion/extension), knee (flexion/extension) and ankle (plantar/dorsiflexion) and in the coronal plane at the hip (abduction/adduction) (Figure 5.3.). Peaks were also calculated at the toe-clearance events. For the contralateral limb, during stance phase (ipsilateral is in swing phase) in the sagittal plane at the hip (flexion/extension), knee (flexion/extension) and ankle (plantar/dorsiflexion) and in the coronal plane at the hip (abduction/adduction) (Figure 5.3.).

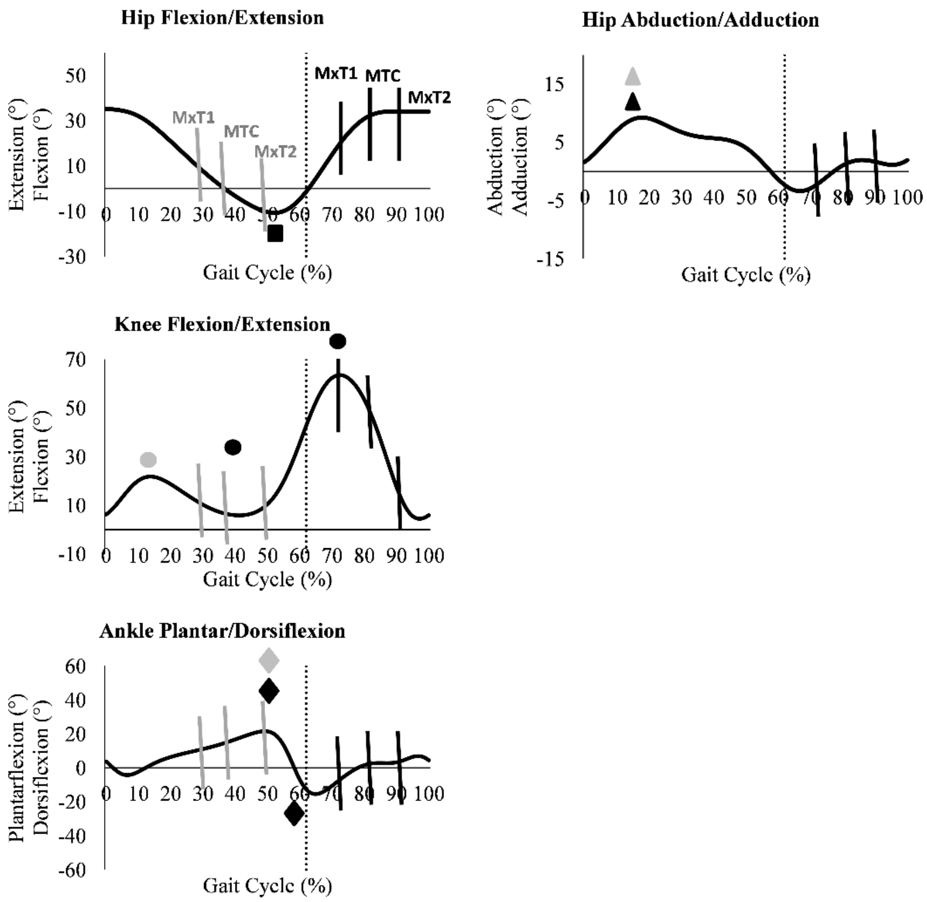


Figure 5.3. Representative sagittal and frontal plane kinematics during the gait cycle for normal walking. The dashed vertical line indicates toe-off and the start of the swing phase. The black lines and symbols indicate data for the ipsilateral (right) limb during stance and swing phases. The grey lines and symbols indicate data for the contralateral (left) limb during the stance phase when the ipsilateral limb was in swing. Solid lines indicate the kinematics taken during each toe-clearance event. MxT1 (First Maximum Toe-Clearance), MTC (Minimum Toe-Clearance) and MxT2 (Second Maximum Toe-Clearance).

■ Peak hip extension at terminal stance

▲ Peak hip adduction during stance phase

● Minimum knee flexion at terminal stance and peak knee flexion in swing phase

◆ Peak ankle dorsiflexion during stance and peak plantarflexion in stance phase

▲ Peak hip adduction during stance phase

● Peak knee flexion during stance phase

◆ Peak ankle dorsiflexion during stance phase

Statistical analysis was performed using IBM SPSS v.23 software (Chicago, USA). Shapiro-Wilk test for normality was executed (Appendix Six: Shapiro-Wilk Test of Normality for Chapter Five). Mixed ANOVAs, with the toe-clearance displacements and walking speed as the dependent variable, age groups (55-64 yrs, 65-74 yrs and ≥75 yrs) as the between factors, and the two walking tasks (NW and DT) as the within factors were also used. Sphericity was assessed using Mauchly's Test of Sphericity Epsilon Greenhouse-Geisser. All task variable values for the Green-house Geisser epsilon were greater than 0.75 therefore Huynh-Feldt corrected value was used (Atkinson and Nevill, 2001). The ANOVAs were followed by pre-planned comparisons based on Bonferroni adjusted post-hoc Tukey tests. A regression analysis was used to establish if age and walking speed predicted MTC. A coefficient of variation (CV%) was used to assess the variability of the toe-clearance events between tasks and age groups. Pearson's R correlations between age and toe-clearance events and walking speed were calculated for both walking tasks for the whole population. A paired t-test was used to ascertain if there were any significant differences between NW and DT for the joint kinematics. An independent t-test was used to determine if there was a difference for NW and DT toe-clearance parameters for fall history (older adults who had previously fallen in the last 12 months vs. older adults who had no falls in the last 12 months). Fall history was derived from the EAGLES Questionnaire results. An alpha level of 0.05 (two-sided) was employed to indicate statistical significance

5.3. Results

Following data collection, twenty-nine participants, age range 57-73 years (15 females, 14 males; 65.0 ± 7.6 yrs; 170.6 ± 9.7 cm; 74.5 ± 14.2 kg) were excluded from the study (Figure 5.4.). Therefore, one-hundred and twenty-nine participants, age range 55-83 years (65.9 ± 6.6 yrs) were included in the study and were divided into three age groups (Table 5.2.). There was no significant difference for fall history on toe-clearance parameters for NW or DT walking and as such no further analysis on fall history was executed.

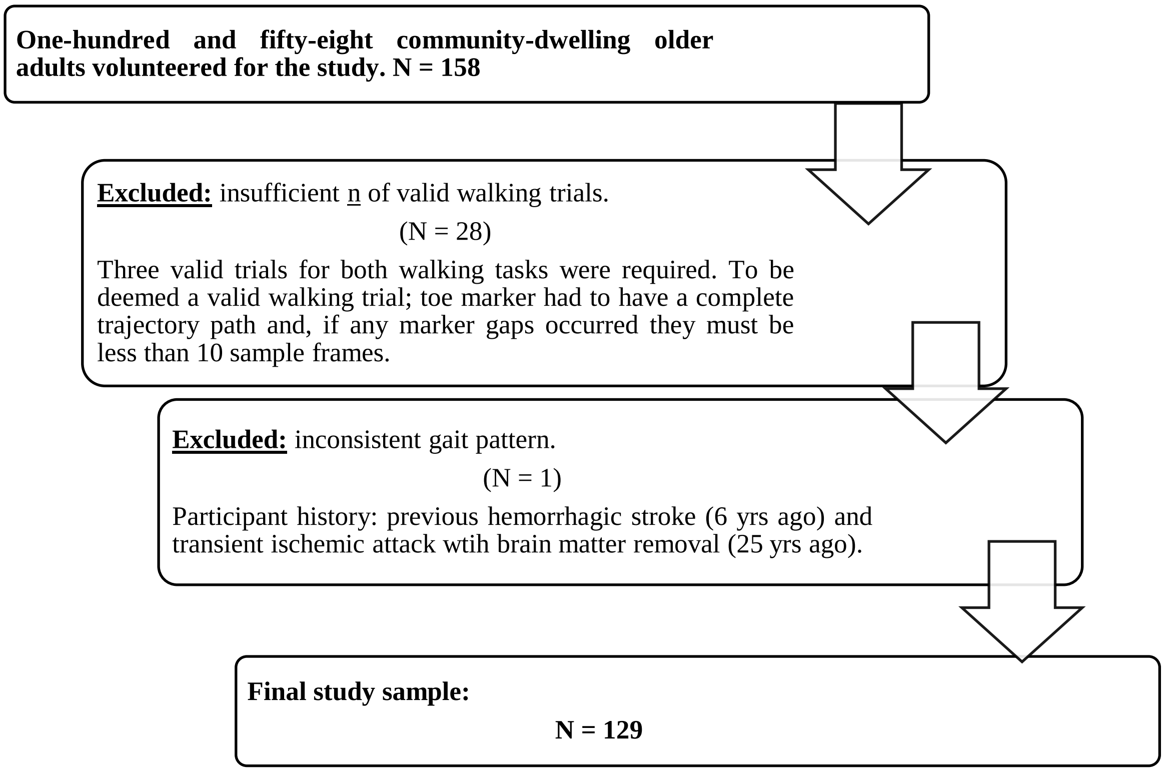


Figure 5.4. Description of participant selection and participant exclusion.

Table 5.2. Participant characteristics.

	Whole Group (n = 129)	**55-64 yrs (n = 54)**	**65-74 yrs (n = 61)**	**≥75 yrs (n = 14)**
Sex (Female/Male)	86/43	35/19	40/21	11/3
Age (yrs)	65.9 ± 6.6	59.9 ± 3.2	68.3 ± 2.8	78.3 ± 2.8
Height (cm)	168.2 ± 9.0	169.3 ± 9.4	167.6 ± 9.2	166.2 ± 6.2
Mass (kg)	73.9 ± 15.0	73.9 ± 16.0	73.4 ± 14.9	76.2 ± 11.6

There was no significant age effect for the toe-clearance events MxT1 ($F_{2,126}$ = 2.734, P = 0.069) and MTC ($F_{2,126}$ = 0.024, P = 0.977). However, there was a significant age effect for MxT2 ($F_{2,126}$ = 6.021, P = 0.003). The over 75 years age group (73.70±14.95 mm) had a significantly lower toe-clearance at MxT2 compared to 55-64 years (88.81±17.45 mm) and 65-74 years (89.36±15.55 mm) age groups, with no significant difference between the 55-64 years and 65-74 years age groups (Figure 5.5.).

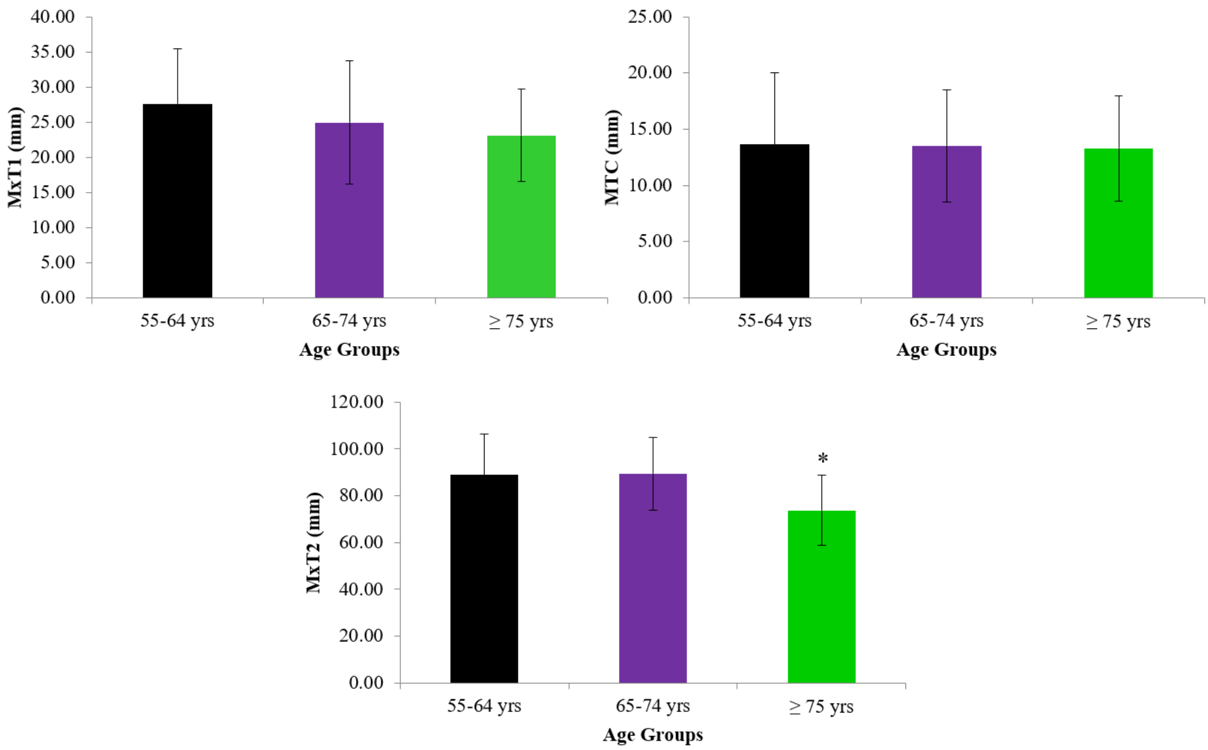


Figure 5.5. Age groups (mean ± SD) for toe-clearance parameters. Age groups (55-64 yrs: n = 54; 65-74 yrs: n = 61 and ≥ 75 yrs: n = 14). *Note:* *Significant age effect for the ≥ 75 yrs compared to the 55-64 yrs and 65-74 yrs age groups. *Abbreviations:* MxT1 (First Maximum Toe-Clearance), MTC (Minimum Toe-Clearance) and MxT2 (Second Maximum Toe-Clearance).

With an increase in age, walking speed significantly reduced for both walking tasks (Table 5.3.). Similarly, there was a significant age effect for walking speed ($F_{2,126}$ = 7.597, P = 0.001). The over 75 years age group (NW: 1.19 ± 0.20 m·s^{-1} and DT: 1.15 ± 0.17 m·s^{-1}) had a significantly reduced walking speed compared to the 55-64 years (NW: 1.36 ± 0.19 m·s^{-1} and DT: 1.36 ± 0.20 m·s^{-1}) and 65-74 years (NW: 1.36 ± 0.17 m·s^{-1} and DT: 1.36 ± 0.17 m·s^{-1}). There was however no significant interaction between age and walking task ($F_{2,126}$ = 1.082, P = 0.342) (Figure 5.6.).

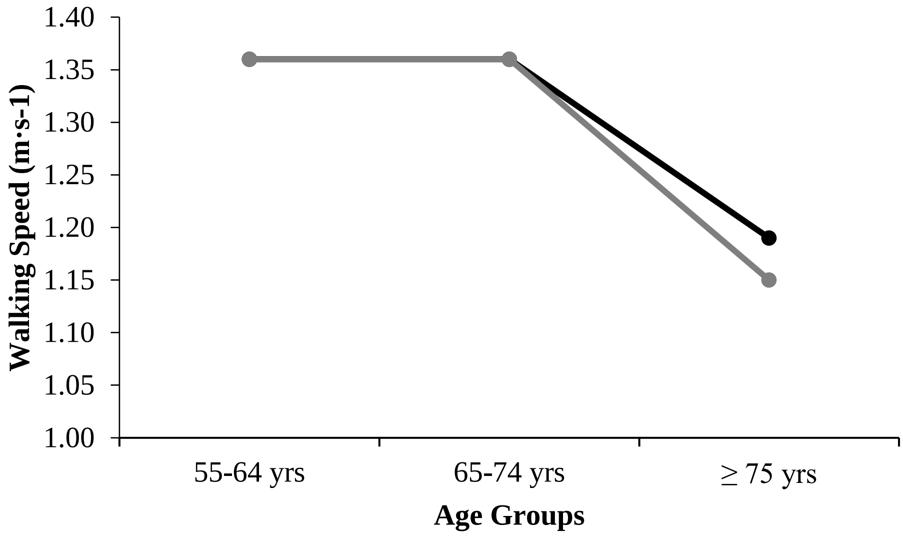


Figure 5.6. Age and walking task interaction for average (mean) walking speed. Age groups (55-64 yrs: n = 54; 65-74 yrs: n = 61 and ≥ 75 yrs: n = 14). *Note:* black line = NW and grey line = DT walking.

Age and walking speed was not significantly correlated with MTC for either task however there was a significant negative correlation with age for MxT2 for both walking tasks. MTC significantly correlated to MxT1 and MxT2 for both walking tasks (Table 5.3.). The regression analysis identified age and walking speed for both walking tasks did not predict MTC (NW r^2 = 0.000 and DT r^2 = 0.001). There was also no significant difference between walking speed for NW and DT walking ($F_{1,126}$ = 1.937, P = 0.166). The average walking speed for NW was 1.34 ± 0.19 m·s^{-1} and DT was 1.33 ± 0.18 m·s^{-1} (Table 5.4.). There was a significant within-subject effect for task for all toe-clearance events (MxT1: $F_{1,126}$ = 19.847, P = < 0.001, MTC: $F_{1,126}$ = 21.672, P = < 0.001 and MxT2: $F_{1,126}$ = 6.224, P = 0.014). Vertical toe displacement for all toe-clearance events were significantly lower for DT compared to NW (Figure 5.4., Table 5.4.).

Table 5.3. Correlations between age, toe-clearance parameters, and walking speed for both normal and manual dual task walking.

	Age	MxT1	MTC	MxT2	Walking Speed
Normal Walking					
Age	1.000	-0.114	-0.001	-0.180*	-0.272*
MxT1		1.000	0.586*	0.144	0.056
MTC			1.000	0.330*	-0.016
MxT2				1.000	0.538*
Walking Speed					1.000
Manual Dual Task Walking					
	Age	**MxT1**	**MTC**	**MxT2**	**Walking Speed**
Age	1.000	-0.166	-0.024	-0.232*	-0.279*
MxT1		1.000	0.478*	0.137	0.085
MTC			1.000	0.316*	0.028
MxT2				1.000	0.559*
Walking Speed					1.000

* Significance at < 0.05.

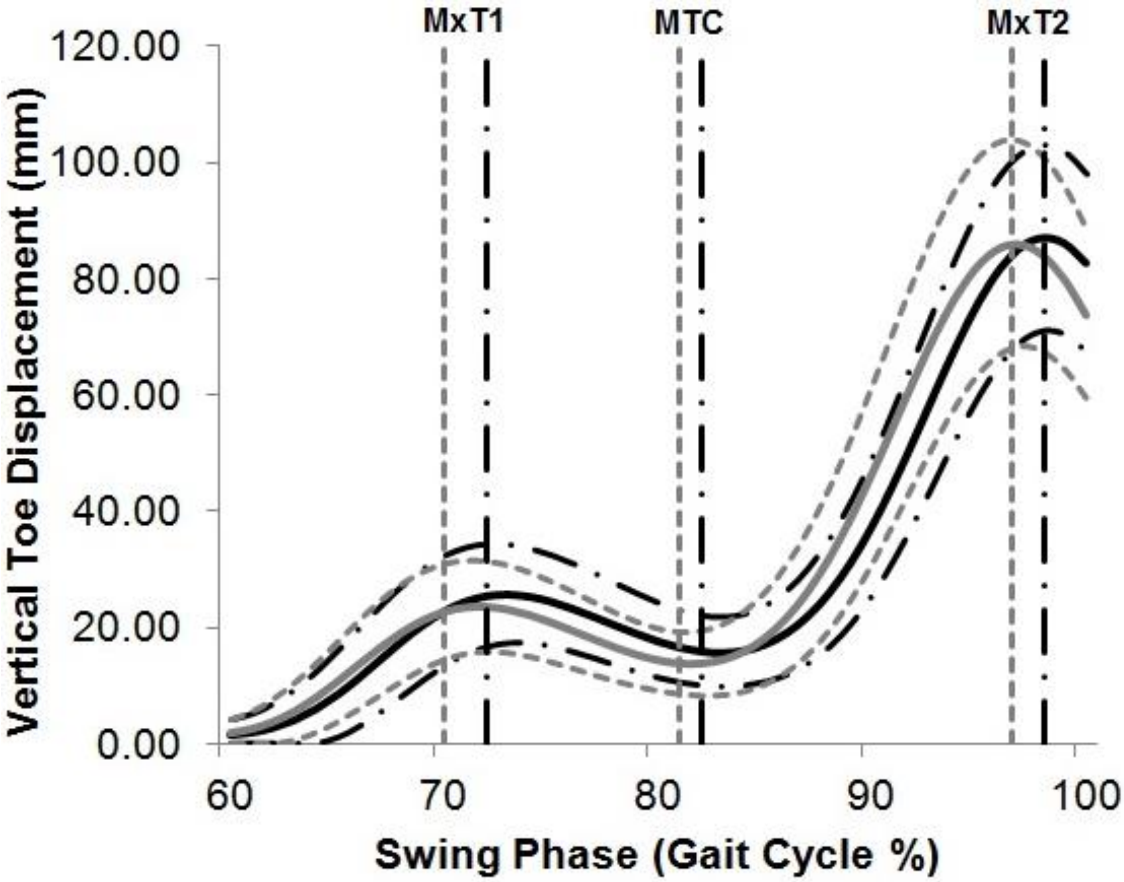


Figure 5.7. Whole group average (n = 129) for vertical toe displacement for normal and dual task walking. *Note:* grey line (mean solid and standard deviation dashed lines) signifies dual task walking; the black line (mean solid and standard deviation dashed lines) signifies normal walking. Toe-clearance events indicated with vertical dashed lines (grey: dual task and black: normal walking).

Table 5.4. Comparison between normal and dual task walking for toe-clearance events and walking speed for the whole group.

Parameter	NW	DT
MxT1 (mm)	27.1 ± 8.6	24.6 ± 7.7*
MTC (mm)	14.7 ± 6.0	12.4 ± 4.9*
MxT2 (mm)	88.1 ± 16.0	86.7 ± 17.9*
Walking Speed ($m \cdot s^{-1}$)	1.34 ± 0.19	1.33 ± 0.18

* Significant task effect. *Abbreviations:* MTC (Minimum Toe-clearance), MxT1 (First Maximum Toe-clearance) and MxT2 (Second Maximum Toe-Clearance).

There was no significant interaction between walking task and age for all toe-clearance events (Table 5.5.). Figure 5.8. illustrates the variability for toe-clearance parameters between age groups for each walking task. For NW, all age groups had the highest variability for MTC (CV%: 55-64 yrs = 46 %; 65-74 yrs = 37 % and ≥ 75 yrs = 34 %), with lowest variability occurring at MxT2 (CV%: 55-64 yrs = 19%; 65-74 yrs = 16 % and ≥ 75 yrs = 22 %). Whereas, for DT walking the 55-64 years age group had the highest MTC variability, 47 % compared to 33 % (65-74 yrs) and 19 % (≥ 75 yrs). Variability at MxT1 and MTC was similar for 55-64 years and 65-74 years age groups for both walking tasks (e.g. MxT1: 55-64 yrs NW = 28 % and DT = 28 %). Overall, variability was lower for the over 75 years age group for all toe-clearance parameters for both walking tasks, except for MxT2 during NW (Figure 5.8).

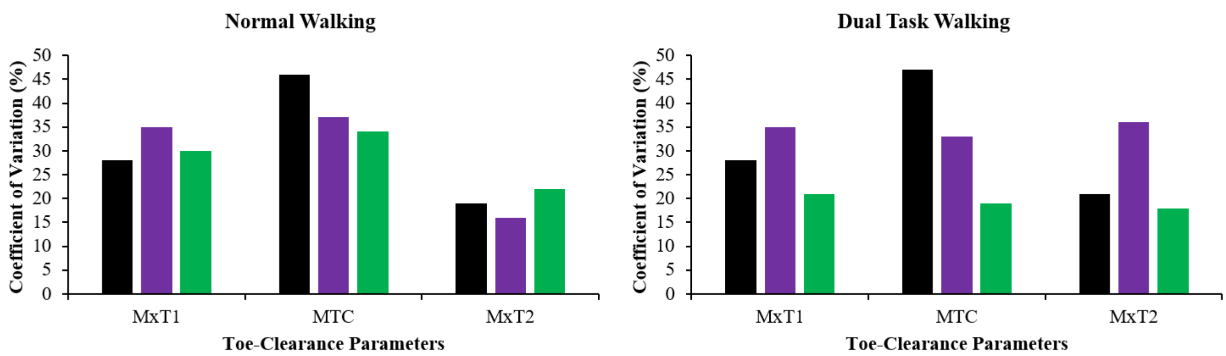


Figure 5.8. Variability of toe-clearance parameters for three age groups. Age groups (55-64 yrs: n = 54; 65-74 yrs: n = 61 and ≥ 75 yrs: n = 14). *Note:* black line = 55-64 yrs, purple line = 65-74 yrs and green line = ≥ 75 yrs.

There was no significant difference between NW and DT walking for the kinematics at the ipsilateral or contralateral limb (Table 5.6.). However, for the ipsilateral limb (Table 5.7A.); at MxT1 hip flexion was significantly reduced during DT walking compared to NW. At MTC there was significantly reduced hip flexion, and significantly increased hip adduction, knee flexion, and ankle dorsiflexion for DT walking compared to NW. At MXT2, hip adduction and knee flexion were significantly reduced, and ankle dorsiflexion was significantly increased during DT walking compared to NW (Table 5.7A.). For the contralateral limb, (Table 5.7B.) hip flexion during stance at the time of MxT1 was significantly increased for DT walking compared to NW. At the time of MTC, hip and knee flexion were significantly increased for DT walking compared to NW. At the time of MxT2 there was significantly reduced hip extension, knee flexion, and ankle dorsiflexion during DT walking compared to NW. In addition, there was significantly increased hip adduction for DT walking compared to NW.

Table 5.5. Age effect and interaction (age and task) on toe-clearance parameters.

	Normal walking			Dual task walking		
	55-64 yrs	**65-74 yrs**	**≥ 75 yrs**	**55-64 yrs**	**65-74 yrs**	**≥ 75 yrs**
MxT1	28.8 ± 8.2 mm	26.2 ± 9.1 mm	24.6 ± 7.5 mm	26.4 ± 7.3 mm	23.7 ± 8.3 mm	21.7 ± 5.5 mm
MTC	14.7 ± 6.7 mm	14.7 ± 5.5 mm	14.3 ± 4.8 mm	12.5 ± 5.9 mm	12.4 ± 4.1 mm	12.3 ± 4.4 mm
MxT2	89.1 ± 16.5 mm	89.8 ± 14.5 mm	76.7 ± 17.1 mm	88.5 ±18.5 mm	88.9 ±16.6 mm	70.7 ± 12.4 mm

Abbreviations: MTC (Minimum Toe-clearance), MxT1 (First Maximum Toe-clearance) and MxT2 (Second Maximum Toe-Clearance).

Table 5.6. Joint kinematics for the ipsilateral and contralateral limb during swing and stance phases prior to toe-clearance events.

	Normal walking **Mean ± SD**	Dual task walking **Mean ± SD**
Ipsilateral limb		
Hip adduction (peak) during stance phase	9.8 ± 4.3°	10.5 ± 3.8°
Hip extension at Terminal stance	-11.0 ± 9.3°	-11.0 ± 8.5°
Knee extension at Terminal Stance	5.5 ± 4.6°	6.0 ± 4.4°
Knee flexion (peak) during swing phase	64.1 ± 4.2°	64.1 ± 4.8°
Ankle dorsiflexion (peak) during stance phase	22.6 ±.6.1°	22.4 ± 6.0°
Ankle plantarflexion (peak) during stance phase	-17.2 ± 6.1°	-17.6 ± 6.8°
Contralateral limb		
Hip adduction (peak) during stance phase	10.2 ± 3.9°	10. 1 ± 4.6°
Knee flexion (peak) during Loading response	21.8 ± 6.0°	21.7 ± 5.7°
Ankle dorsiflexion (peak) during stance phase	22.4 ± 4.8°	22.2 ± 2.4°

Table 5.7. Joint kinematics for the ipsilateral and contralateral limb during swing and stance phases at the time of toe-clearance events.

A. Ipsilateral limb (swing limb)	**Normal walking Mean ± SD**	**Dual task walking Mean ± SD**	**B. Contralateral limb (stance limb)**	**Normal walking Mean ± SD**	**Dual task walking Mean ± SD**
Hip flexion during swing phase...	**Whole group**	**Whole group**	**Hip flexion/extension during stance phase...**	**Whole group**	**Whole group**
at MxT1	20.5 ± 8.4°	18.2 ± 8.0°*	at MxT1	17.9 ± 7.9°	18.6 ± 8.0°*
at MTC	32.7 ± 7.4°	31.4 ± 7.4°*	at MTC	5.3 ± 9.2°	6.1 ± 9.1°*
at MxT2	34.0 ± 7.8°	32.9 ± 7.8°*	at MxT2	-9.2 ±10.3°	-8.1 ± 9.4°*
Hip abd/adduction during swing phase...			**Hip abd/adduction during stance phase...**		
at MxT1	-2.0 ± 4.5°	-1.8 ± 4.0°	at MxT1	9.2 ± 4.3°	9.5 ± 4.7°
at MTC	1.7 ± 4.4°	7.5 ± 4.1°*	at MTC	7.3 ± 3.8°	7.5 ± 4.1°
at MxT2	1.8 ± 4.2°	1.8 ± 3.9°	at MxT2	5.5 ± 3.9°	6.2 ± 3.9°*
Knee flexion during swing phase...			**Knee flexion during stance phase...**		
at MxT1	63.6 ± 4.2°	63.6 ± 4.9°	at MxT1	17.6 ± 7.0°	18.1 ± 5.7°
at MTC	44.5 ± 7.0°	46.9 ± 7.3°*	at MTC	9.7 ± 6.0°	10.6 ± 5.0°*
at MxT2	5.5 ± 4.5°	4.2 ± 4.5°*	at MxT2	9.2 ± 4.4°	8.2 ± 4.9°*
Ankle planter/dorsiflexion during swing phase...			**Ankle planter/dorsiflexion during stance phase...**		
at MxT1	-5.4 ± 2.9°	-5.7 ± 3.1°	at MxT1	7.2 ± 2.8°	7.2 ± 2.5°
at MTC	2.8 ± 2.1°	3.9 ± 2.7°*	at MTC	12.3 ± 2.5°	12.5 ± 2.3°
at MxT2	3.6 ± 3.1°	8.2 ± 3.4°*	at MxT2	22.7 ± 3.1°	21.8 ± 2.9°*

* Significant difference between walking tasks. *Abbreviations:* MTC (Minimum Toe-clearance), MxT1 (First Maximum Toe-clearance) and MxT2 (Second Maximum Toe-Clearance).

5.4. Discussion

Understanding MTC during walking is important because it occurs at a critical instance in the gait cycle and if a trip occurs at or near this point stability cannot be regained unless there is rapid and safe placement of the swing foot (Winter, 1992). This study investigated toe-clearance events and the joint kinematics of 129 older adults (55-84 yrs) for NW and DT walking. Past research, compares younger to older adults and as such older adults are classified into one age group. However, walking speed has been reported to decline by 1 % per year from the age of 60 (Mills *et al.*, 2008). Therefore, it is more appropriate to look at a wide range of ages of older adults. The objectives of this study were to determine if toe-clearance parameters were affected by task and if they were related to age in an older adult population. The results suggested that age and walking speed were not predictors of MTC. The results agreed with the hypothesis given that DT walking reduced toe-clearance events. For age, there was no difference in toe-clearance events, except for MxT2 for both walking tasks, disagreeing with the hypothesis. There was also no significant interaction between task and age for the toe-clearance events. Toe-clearance was relatively stable with age for both normal and DT walking. This may be because our cohort was healthy, active and motivated, and so volunteered to come to the laboratory for gait testing.

The MTC reported here for older adults during NW were similar to the young adults of Begg *et al.* (2007) (MTC of 15.6 ± 6.2 mm) and Mills *et al.* (2008) (median 14.9 mm IQR 4.3 mm) who used different methods to calculate MTC, but higher than that reported by Schulz *et al.* (2010) (MTC of 10.3 ± 3.2 mm) whose participants were younger than those in this current study. This is likely due to the method employed by Schulz *et al.* (2010) which selected the smallest MTC from hundreds of virtual markers on the shoe as opposed to one toe marker.

Age was not significantly correlated with MxT1 and MTC for both normal and DT walking. This agrees with previous studies which compared MTC for younger and older adults (Winter, 1992, Begg *et al.*, 2007, Sparrow *et al.*, 2008, Schulz *et al.*, 2010, Nagano *et al.*, 2011, Santhiranayagam *et al.*, 2015) and found no significant difference between the two age groups. This suggests that MxT1 and MTC for a healthy older adult population remain relatively constant across a wide range of ages. Thus, regardless of an age-related reduction in the ability to recover from a trip (van Dieën *et al.*, 2005), healthy older adults do not increase their toe-clearance.

This study also showed that age was significantly negatively correlated for MxT2. At MxT2, the foot reaches peak dorsiflexion (Winter, 1991) and as such can be influenced by weaker dorsiflexor muscles and reduced ankle range of motion as a consequence of age (Prince *et al.*, 1997, Perry *et al.*, 2007). Although, MxT2 is not linked to trip risk during over ground levelled walking, sufficient dorsiflexion is a vital factor for successful MTC (Nagano *et al.*, 2011). As such, reduced dorsiflexion with an increase in age may compromise toe-clearance and potentially increase the risk of tripping. The results also revealed a significant age effect for MxT2, demonstrating the over 75 years age group were significantly lower compared to the 55-64 years and 65-74 years age groups. In normal ageing, musculoskeletal function has found to be reduced which results in physiological and neuromuscular changes for older adults (Prince *et al.*, 1997, Faulkner *et al.*, 2007). The reduction of toe-clearance at MxT2 may potentially be associated with reduced and/or changes in joint range motion with reduced muscle strength, for example weak dorsiflexor muscles; which is a consequence of ageing (Kang and Dingwell, 2008b). In Chapter Four, the results revealed the ankle moment was preserved with age during NW. However, for the over 75 years age group, ankle power generation in late stance (i.e. at toe-off) was reduced, which was likely the result of muscle weakness in the plantarflexor muscles. Inflexible plantarflexor muscles such as those caused by ageing factors such as reduce muscle elasticity and muscle weakness have found to potentially inhibit full ankle dorsiflexion (Malone and Pfeifle, 2016). Therefore, for the over 75 years age group the reduced ankle power generation which potentially was a consequence of age-related biomechanical change may be indirectly preventing the ankle dorsiflexors to reach maximum dorsiflexion range at MxT2.

The correlation between MTC and MxT1 has only been reported for treadmill walking and this was suggested to be an adaptation to the different walking surface and seen as a common response to potential destabilising walking conditions (Nagano *et al.*, 2011). This association is a ‘response’ to a change in surface because it implies that the preceding gait events may influence MxT1, which in turn will influence MTC. For example, interventions to increase strength and enhance push-off in stance would positively affect Mx1T and therefore MTC (Nagano *et al.*, 2011). This correlation was also present in this current study for older adults when walking on the ground and when performing a DT, suggesting that this was not a response to walking on a treadmill but a common feature of gait in older adults. There was also a significant correlation between MTC and MxT2 for both normal and DT walking suggesting the preceding event (MTC) influences the latter (MxT2). This event (MxT2) is not a critical determinant of tripping (Nagano *et al.*, 2011) but in obstacle negotiation this event (because it occurs later in the gait cycle) and the relationship with MTC may become more important to ensure safe obstacle clearance.

In addition, as MxT2 occurs later in the swing phase it is likely that this will be an important toe-clearance parameter to measure during obstacle crossing. Figure 5.9. illustrates the supporting limb during obstacle clearance, which may require an increased support time to allow for sufficient flexor angle to successfully clear the obstacle (Liao *et al.*, 2014). Lower extremity muscle influences crossing ability in community-dwelling older adults (Lamoureux *et al.*, 2002, Lamoureux *et al.*, 2003). Liao *et al.* (2014) reported ankle dorsiflexor strength was the primary factor for obstacle clearance stride length and velocity. As such, more so than levelled normal walking, the ankle dorsiflexors are required to contract to a sufficient flexion angle in a quick transition from ankle plantarflexion at toe-off, to prevent tripping and allow for safe obstacle clearance (Liao *et al.*, 2014). Therefore, an increase in age may cause inadequate dorsiflexion control due to muscle weakness for instance.

Future work should be considered to evaluate toe-clearance parameters during obstacle clearance for this older adult population. Moreover, obstacle clearance is a complex walking task, which requires the ability to move the centre of mass forward and away from the supporting limb. It has been reported older adults with Parkinson’s Disease reduce stride length and velocity when crossing the obstacle as a safe strategy for forward weight shifting. Consequently, for this older adult population as previously stated gait is affected by age for the over 75 years age group. Typically, age-related gait changes include reduce hip extension and altered joint kinetics (e.g. reduced knee power generation). As such, this could affect the ability to safely clear an obstacle for this older adult population due to reduce propulsive muscle force power at toe-off for instance. Therefore, an increased arm swing may be required for momentum and to aid power generation at toe-off to allow for successful obstacle clearance. As such, arm swing should be investigated to determine the role during gait and different walking challenges.

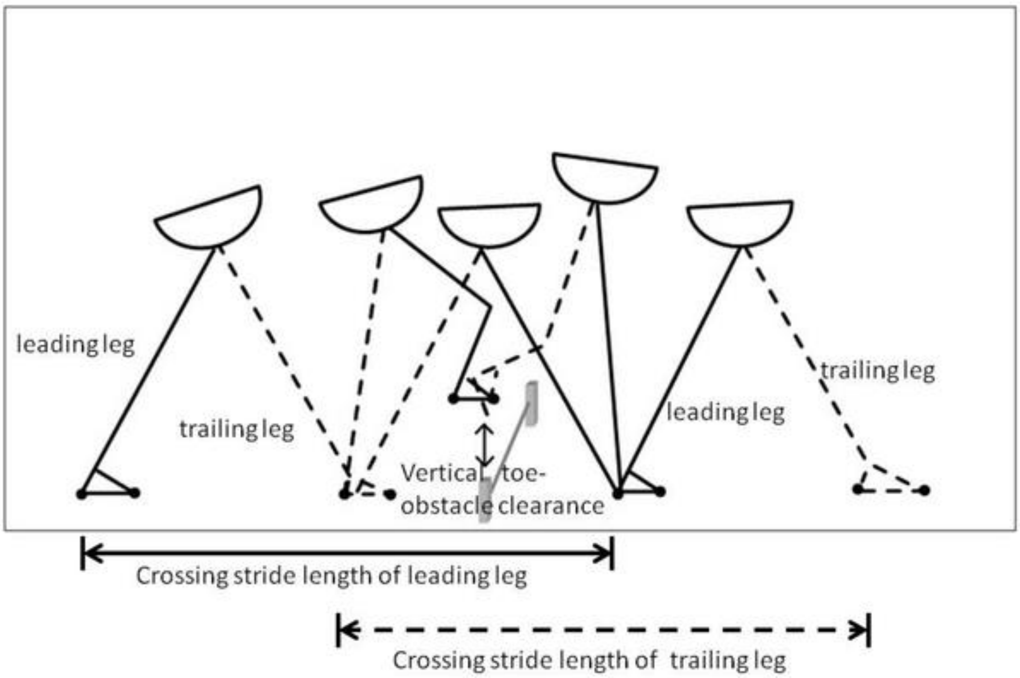


Figure 5.9. Toe-clearance during obstacle clearance (Liao *et al.*, 2014).

The coefficient of variation was notably larger for MTC across all age groups compared to MxT1 and MXT2. An increased variability in MTC would suggest that the toe is closer to the ground during some steps. Thus, an understanding of MTC variability during walking could lead to an understanding of the mechanisms responsible for tripping and falling in older adults. However, the variability of the toe-clearance events in this study was derived from 5 steps, whereas other studies have used multiple steps whilst walking on a treadmill. These studies (Begg *et al.*, 2007, Khandoker *et al.*, 2008, Mills *et al.*, 2008) has observed greater variability for MTC in older adults compared to younger adults.

When performing a DT, walking speed reduces and this reduction can be explained by several neuropsychological theories on human information processing (such as capacity-sharing theory, the Bottleneck Theory or the Multiple Resource Models Theory) (Pashler, 1994, Ruthruff *et al.*, 2001, Tombu and Jolicoeur, 2003). However, a reduction in walking speed was not seen in this study when performing a manual DT suggesting that attentional resources were not limited in capacity. This contradicts previous findings (Santhiranayagam *et al.*, 2015) using the same DT, but walking speeds reported by Santhiranayagam *et al*. (2015) for NW (0.94 ± 0.42 $m \cdot s^{-1}$) and DT walking (0.42 ± 0.08 $m \cdot s^{-1}$) were much lower compared to the current study and possibly because a motorized treadmill was used. Walking speed was significantly and positively correlated with MxT2 for DT walking only suggesting that a strategy to increase MxT2 may be to increase walking speed. This event is more critical for obstacle negotiation and so may not be crucial for walking on flat ground. Furthermore increasing walking speed when performing a DT is likely to negate the success of the secondary task, and hurrying may actually increase the likelihood of a fall following a trip (Pavol *et al.*, 1999), suggesting a different compensatory strategy may be required to increase MxT2.

We have shown that age was not correlated with MTC or MxT1 during walking yet there was a significant reduction for all toe-clearance parameters when performing a DT compared to NW, thus increasing the chance of a trip. A reason why MTC parameters were lower during DT walking is because this task (carrying a cup of water) reduces arm swing. Arm swing is a mechanism which has been suggested to stabilise the body and help achieve lateral balance during walking (Ortega *et al.*, 2008). As such, this manual DT may have reduced walking stability which impacted on toe-clearance, resulting in lower vertical displacement for all events. Despite the challenge of this task sufficient toe-clearance height was achieved to ensure successful ground clearance. However, the reduction in MTC may become more challenging to safe walking (avoiding trips) with a more demanding DT or if the walking surface is not flat as it was in the laboratory.

This study also attempted to show if the contralateral and ipsilateral kinematics when walking adapts when performing a DT, as this may have provided a mechanism for underlying control of MTC during walking. The results suggest that gait kinematics were altered during the toe-clearance events only and not in the gait cycle phases before. During swing phase for the ipsilateral limb the joint kinematics during DT walking altered to ensure a safe (not tripping), all be it lower toe-clearance, by increasing, hip adduction (suggestive of greater hip hiking), knee flexion, and ankle dorsiflexion (suggestive of shortening the effective length of the limb). There was also a significant reduction in hip flexion, which suggests a shorter step length when performing the DT. This, in isolation, is paradoxical; a reduced toe-clearance with kinematics that should increase it but combined with the contralateral stance phase during the time of the toe-clearance events suggest the ipsilateral limb was adapting to the contralateral limb. For example, the contralateral limb had significantly greater knee and hip flexion during stance at the same time as MTC, thus lowering the body during the stance phase and requiring an adaption from the ipsilateral limb to ensure safe toe-clearance. The changes in kinematics are relatively small, but Winter (1992) stated that changes of this magnitude have a large impact on the swing foot.

This study used a manual DT which may not have been demanding enough (in this population) to elicit differences between age groups in toe-clearance events. The amount of deficits that occur while walking depends on the demands of the secondary task (Beurskens and Bock, 2012). A cognitive DT may have prompted greater changes to toe-clearance events than seen in this current study. Future work may wish to compare DT paradigms (cognitive and manual tasks) to assess their impact upon toe-clearance. A further approach to take is one where participants must coordinate two sources of visual information processing similar to everyday demands, such as navigating along a crowded shopping centre or looking for signs while walking along a street (Beurskens and Bock, 2013). The impact this has upon toe-clearance and trip risk has yet to be studied.

As previously mentioned, variability for this study was derived from 5 steps, which was one gait cycle per trial. As such, variability analysis was limited to CV%. In addition, variability is typically assessed using treadmill walking to allow for consecutive gait cycles (e.g. 100 steps) (Begg *et al.*, 2007). However, as previously mentioned in Chapter One, gait patterns alter during treadmill walking and typically older adults adopt a safe gait strategy when treadmill walking, for example reduced walking speed (Row Lazzarini and Kataras, 2016). Therefore, future work should be considered to assess toe-clearance parameters for multiple gait cycles either by adding additional motion capture cameras into the gait laboratory to allow for an increased field of view for the walkway or use inertial sensors on the shoe to calculate toe-clearance (Dadashi *et al.*, 2014), as this would not be limited by room distance and would allow for numerous gait cycles during data collection. Furthermore, horizontal toe-clearance was not investigated. De Asha and Buckley (2015) investigated the relationship between MTC and swing-foot velocity for amputees and found increases in MTC on the prosthetic limb was related to toe-clearance of the modulated contralateral limb (intact limb), which occurred at swing-limb ankle. Swing-foot velocity was defined as maximum velocity in the anterior-posterior direction of the foot-segment centre of mass. As such, future work could be conducted to determine the effect on horizontal toe displacements for an older adult population.

Furthermore, unlike Chapter Four, this study did not investigate toe-clearance waveforms using Statistical Parametric Mapping (SPM). This methodology as yet to be investigated for toe-clearance displacements and for this study it was unlikely to provide any more information than what was found for the toe-clearance parameters. However, future work should be considered especially if horizontal toe displacement will be investigated.

5.5. Conclusion

This study presents one of the largest database of toe-clearance parameters for older adults walking at a self-selected speed. Age was not correlated with minimum toe-clearance, illustrating its stability across a wide age range (55-84 yrs). However, age effect was present for second maximum toe-clearance, which may have been affected by the over 75 years age group inability to reach full dorsiflexion due to weak ankle plantarflexion, which was observed in Chapter Four. Manual dual walking significantly reduced toe-clearance parameters in older adults suggesting this dual task places this older adult population at a greater risk of tripping. The mechanism for maintaining a successful toe-clearance has yet to be determined but these results suggest that kinematics of both the limbs are significantly altered to ensure a safe toe-clearance. For instance, the ipsilateral limb at minimum toe-clearance had an increase hip adduction during dual task, this was to compensate for the increased hip flexion and hip abduction of the contralateral limb. Consequently, both age and walking task affected toe-clearance parameters in this older adult population, which caused gait alterations as described in this chapter. Yet, for age and walking task, successful toe-clearance was achieved as no incidents of tripping occurred. Therefore, additional factors must be influencing the success of toe-clearance. As suggested in this discussion, an increased arm swing may aid the momentum for the swing limb to advance to next initial heel contact. As such, future work should be considered to investigate the role of arm swing in this older adult population and determine the effect of walking task on arm swing.

Chapter Six: Does Walking Task Affect Arm Swing in an Older Adult Population

ABSTRACT

Introduction: Arm swing is essential for efficient locomotion. Typically, research has explored the effect of arm swing in older adults by comparing to young adults, as such the influence of age remains unknown. Furthermore, in order to maintain independent living older adults must adapt to changing demands, such as dual task walking or obstacle clearance. Therefore, the aim of this study was to explore the effect of walking task on arm swing for an older adult population. In addition, forearm swing has yet to be investigated when performing additional tasks. Consequently, the secondary aim of the study was to establish if walking task affected forearm swing for the older adult population.

Methods: 158 community-dwelling older adults, age range 55-86 years (65.7 ± 6.8 years) participated and walked at their self-selected comfortable walking speed for four walking tasks: normal walking, manual dual task walking (carrying a cup of water), stepping onto and off an obstacle and stepping over an obstacle. Three-dimensional motion analysis was used to capture arm and forearm swing. Arm and forearm swing amplitude and asymmetry were calculated for all walking tasks.

Results: Age was not significantly correlated to either arm or forearm swing amplitude. Walking task affected arm swing amplitude. For instance, when stepping over an obstacle there was an increased arm swing (23.58 ± 7.97 °) compared to normal walking (20.89 ± 7.89 °) and stepping onto and off an obstacle (21.12 ± 7.62 °) for the dominant arm. Whereas, for the non-dominant arm, both obstacle clearance walking tasks had a significantly increased arm swing compared to normal walking (e.g. stepping onto and off an obstacle: 24.87 ± 8.11° stepping over an obstacle: 25.96 ± 7.85 ° vs. normal walking: 22.21 ± 6.23 °). Forearm swing was not affected by walking task. Arm swing during all walking tasks was found to be asymmetrical. Walking speed was significantly reduced for the obstacle clearance tasks, compared to normal and manual dual walking (stepping onto and off an obstacle: 1.16 ± 0.25 $m \cdot s^{-1}$ and stepping over an obstacle: 1.20 ± 0.21 $m \cdot s^{-1}$ compared normal walking: 1.35 ± 0.19 $m \cdot s^{-1}$ and manual dual task: 1.34 ± 0.19 $m \cdot s^{-1}$).

Conclusion: Age does not influence arm swing for this older adult population. Arm swing was however affected by the walking task. With an increase in task complexity there was an increased arm swing for the obstacle clearance tasks. An increased arm swing may aid gait stability when balance is challenged during obstacle clearance tasks. As such, arm swing may also be compensation for landing forces (e.g. propulsive force) during obstacle clearance. Therefore, future work should explore landing forces and joint kinetics during obstacle clearance for this older adult population.

Keywords: Arm Swing, Forearm Swing, Biomechanics, Gait, Asymmetry

6.1. Introduction

Gait analysis typically focuses on the lower body, yet arm swing (peak flexion and extension at the shoulder joint) has found to be essential for efficient locomotion (Bruijn *et al.*, 2010) and aids gait stability (Ortega *et al.*, 2008, Bruijn *et al.*, 2010, Nakakubo *et al.*, 2014, Punt *et al.*, 2015). During normal walking (NW), the arm swings in opposition to the lower body in pendulum-like motion, in order to assist with balance from angular momentum created by the lower body (Elftman, 1939). Conserving energy consumption during gait can be achieved through arm swing, through reduction in ground reaction forces upon foot contact (Buchthal and Fernandez-Ballesteros, 1965, Pontzer *et al.*, 2009).

For young adults, arm swing amplitude (range from peak flexion to peak extension) has found to increase with an increased walking speed (Bruijn *et al.*, 2008, Liang *et al.*, 2014). Whereas, older adults typically have a reduced arm swing when compared to young adults (Elble *et al.*, 1991, Krasovsky *et al.*, 2014, Mirelman *et al.*, 2015). As such, a reduction in arm swing could increase the risk of falls amongst older adults (Mirelman *et al.*, 2015). Research on arm swing role for older adults is limited, as research typically focuses on comparing young to older adults (Kuhtz-Buschbeck *et al.*, 2008, Ortega *et al.*, 2008, Krasovsky *et al.*, 2014, Mirelman *et al.*, 2015, Plate *et al.*, 2015) or exploring the effects on Parkinson's Disease (Lewek *et al.*, 2010, Plate *et al.*, 2015, Mirelman *et al.*, 2016). Therefore, a better approach would be to explore the effects of arm swing during walking within an older adult population to explore the association with age.

Furthermore, walking around a changing environment such as dual task walking is necessary for independent living. There is a paucity of arm swing research for dual task walking and especially obstacle negotiation tasks. Research (Kuhtz-Buschbeck *et al.*, 2008, Krasovsky *et al.*, 2014, Mirelman *et al.*, 2015, Plate *et al.*, 2015) has predominantly explored the effects of arm swing on young adults performing cognitive tasks such as counting backwards in 3s and found arm swing reduces when dual task walking. Consequently, the effects of arm swing on manual dual task walking (DT) and obstacle tasks remain unknown.

In addition, toe-clearance was found to be affected by both age and walking task, as seen in Chapter Five. An age effect, revealed second maximum toe-clearance was reduced for the over 75 years age group. This parameter coincides with peak ankle dorsiflexion (Winter, 1991). Furthermore, Chapter Four found for this age group ankle power generation was reduced at toe-off, which indicates reduced ankle plantarflexor muscles. This is suggested to be due to a consequence of ageing. Weakness of the ankle dorsiflexor muscles have found to inhibit full dorsiflexion range of motion (Malone and Pfeifle, 2016), which may be the reason for reduced second maximum toe-clearance. Moreover, toe-clearance parameters were significantly lower for DT walking. This may have been affected by an interrupted motor control for performing a secondary task, which may have compromised walking stability. As such, arm swing may have aided walking stability for both the older age group (≥ 75 yrs) and during manual DT walking, to ensure safe toe-clearance for instance.

Therefore, the aim of this study was to explore the effect of walking task on arm swing for an older adult population. It was hypothesised that arm swing amplitude would increase with an increase in task complexity, with arm swing decreasing with an increase in age. In addition, arm swing has typically been investigated on elbow position in relation to the shoulder (Knutsson, 1972, Nieuwboer *et al.*, 1998, Wood *et al.*, 2002, Lewek *et al.*, 2010). However, arm swing incorporates elbow kinematics (Kuhtz-

Buschbeck *et al.*, 2008). As such, it would also be appropriate to evaluate the position of the hand (i.e. end effector) in relation to the elbow when quantifying arm swing. Therefore, the secondary aim of this study was to establish if walking task affected forearm swing for the older adult population. It was hypothesised, that forearm swing for the contralateral arm (i.e. not holding the cup) would increase for dual task walking compared to normal and obstacle clearance and an increase in age would be associated with an increased forearm swing.

6.2. <u>Methods</u>

The overall methodology is provided in Chapter Two. Participants performed the hand-grip test. All participants performed five trials for normal walking (NW without force plate contact), manual dual task (DT) walking (carrying a cup of water), stepping onto and off an obstacle (SON) and stepping over an obstacle (SOV). Walking speed ($m \cdot s^{-1}$) was derived from the Brower timing gates (Utah, USA) and calculated in Excel (Microsoft Office 2010, Tokyo, Japan). All remaining analysis was completed using a custom-made Python code (Python v. 2.7.10, Delaware, USA). Shoulder, elbow and finger trajectories (Y and Z axis) for the dominant and non-dominant arm were normalised to one gait cycle (100 %), using linear interpolation to 101 data samples, for each walking task. Arm and forearm swing amplitude was not calculated for the dominant arm during DT walking, as all participants carried the cup of water in their dominant hand. Arm swing was defined as the range between maximum and minimum angle of the line formed between the shoulder and elbow with respect to the vertical axis about the shoulder. Forearm swing was defined as the range between maximum and minimum angle of the line formed between the elbow and finger with respect to the vertical axis about the elbow. Arm swing and forearm swing amplitudes (degrees) (Figure 6.1.) were calculated using the following equation:

$$\alpha = \cos^{-1} \frac{(Z_1 - Z_2)}{\sqrt{(Y_1 - Y_2)^2 + (Z_1 - Z_2)}}$$

Arm Swing: 1 = Shoulder trajectory and 2 = Elbow trajectory

Forearm Swing: 1 = Elbow trajectory and 2 = Finger trajectory

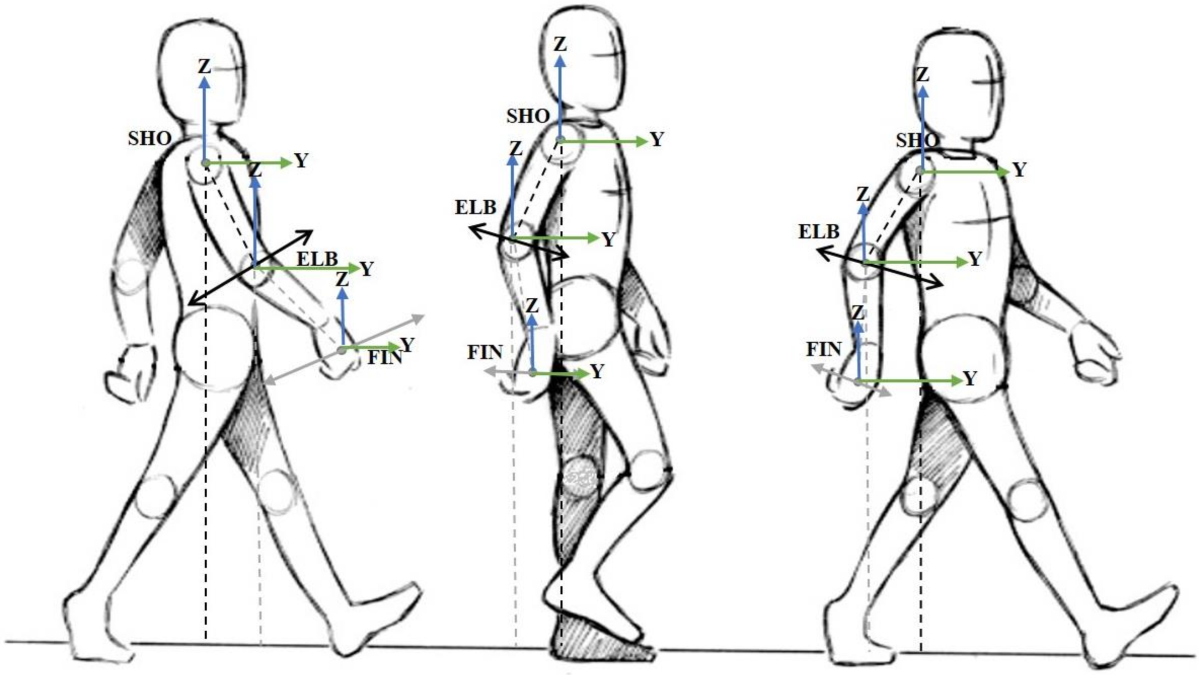


Figure 6.1. Arm swing and forearm swing schematic. Black line indicates arm swing and grey line indicates forearm swing. *Abbreviations:* Shoulder marker (SHO), Elbow marker (ELB) and Finger Marker (FIN).

Symmetry was typically measured using the symmetry index (Robinson *et al.*, 1987, Herzog *et al.*, 1989, Becker *et al.*, 1995, Karamanidis *et al.*, 2003, Nolan *et al.*, 2003). However, the symmetry index requires a reference value and as such healthy populations have no obvious reference side and averaging values can filter out differences between sides (Zifchock *et al.*, 2006). In addition, the symmetry index can be affected by artificial inflation. Whereas, the symmetry angle does not require a reference value and has been suggested as an appropriate substitute for the symmetry index (Zifchock *et al.*, 2008).

Arm swing and forearm swing asymmetry were calculated using Zifchock *et al.* (2008) method:

$$ASA = \frac{(45° - (\tan^{-1}(Swing_{more}/Swing_{less})}{90°} * 100\ \%$$

A value of 0 indicates complete symmetry (Hogan and Sternad, 2009), with Lewek *et al.* (2010) suggesting a cut-off threshold of 7.4 %. Asymmetry for arm and forearm swing was not calculated for DT walking, due to the known arm asymmetry of walking whilst holding a cup of water.

Statistical analysis was performed using IBM SPSS v.23 software (Chicago, USA). Mixed ANOVAs, with the arm and forearm swing amplitudes and walking speed as the dependent variable, age groups (55-64 yrs, 65-74 yrs and ≥ 75 yrs) as the between factors, and the four walking tasks (NW, DT, SON and SOV) as the within factors were also used. The ANOVAs were followed by pre-planned comparisons based on Bonferroni adjusted post-hoc Tukey tests. Pearson's R correlations between age and hand grip were performed for the whole population, with correlations between age, walking speed and arm swing and forearm swing amplitude for all walking tasks. Pearson's R correlations between age and asymmetry was only calculated for the NW, SON and SOV walking tasks. In addition, a partial correlation was performed, to find the association between age and arm swing and forearm swing amplitude and asymmetry, when controlling for walking speed.

6.3. <u>Results</u>

Following data collection, 60 participants, age range 60-75 years (34 females, 26 males; 67.1 ± 7.5 yrs; 168.7 ± 10.0 cm; 75.0 ± 14.7 kg) were excluded from the study (Figure 6.2.). Therefore, ninety-eight participants, age range 55-83 years (64.9 ± 6.2 yrs) were included in the study (Table 6.1.). Two participants were unable to perform the obstacle clearance tasks due to their gait mobility (1 female; 80 yrs and 1 male; 73 yrs). As such, participants were not grouped into the three age groups and mixed ANOVAs were not executed. Therefore, only one-way within-subjects' ANOVAs were performed on walking speed, arm swing and forearm swing amplitude to determine walking task effect. The ANOVAs were followed by pre-planned comparisons, based on Bonferroni adjusted post-hoc Tukey tests.

Age was significantly correlated with hand-grip strength for the non-dominant hand (Table 6.1.). There was a significant task effect on walking speed ($F_{1.800,171.037}$ = 131.566, P = 0.000). Walking speed significantly increased for NW and DT walking compared to the obstacle clearance tasks (NW: 1.35 ± 0.19 $m \cdot s^{-1}$ DT: 1.34 ± 0.19 $m \cdot s^{-1}$ vs. SON: 1.16 ± 0.25 $m \cdot s^{-1}$ and SOV: 1.20 ± 0.21 $m \cdot s^{-1}$), with significant differences found between NW and DT walking (1.35 ± 0.19 $m \cdot s^{-1}$ vs. 1.34 ± 0.19 $m \cdot s^{-1}$). There was however, significant differences between the obstacle clearance tasks, as SON was significantly slower than all walking tasks (Figure 6.3. and Table 6.1.).

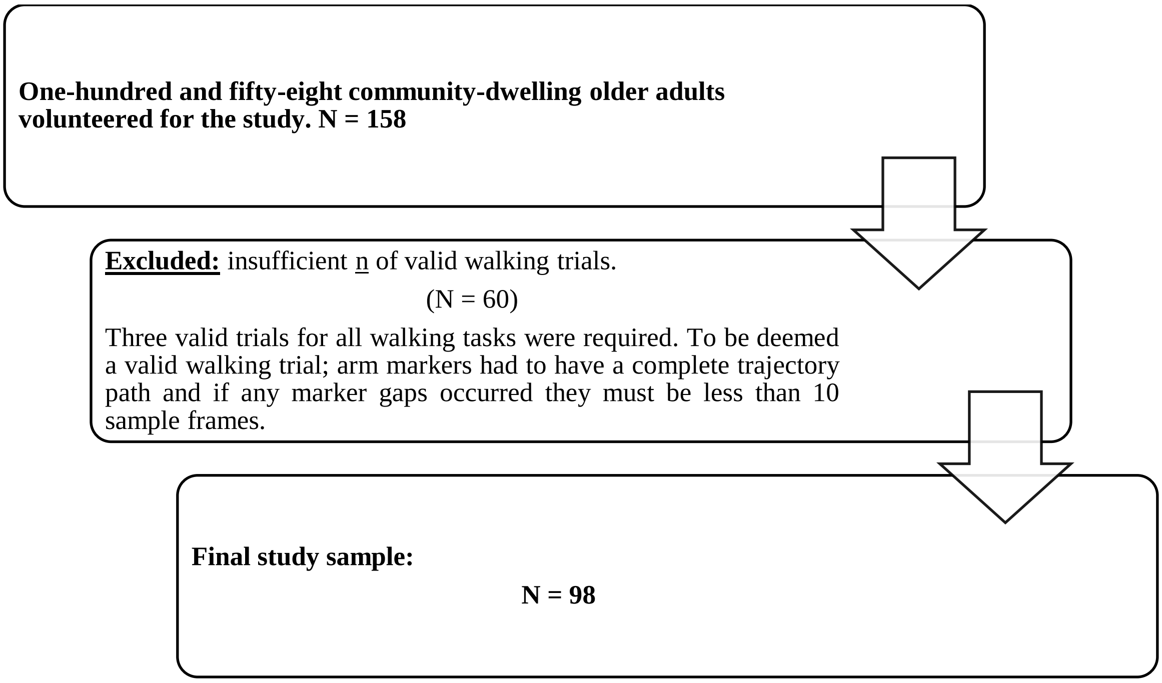


Figure 6.2. Description of participant selection and participant exclusion.

Table 6.1. Participant Characteristics.

	Whole Group (n = 98)
Sex (Females/Males)	67/31
Age (yrs)	64.9 ± 6.2
Height (cm)	168.4 ± 8.7
Mass (kg)	73.4 ± 14.9
Hand-grip (kg)	
Dominant Hand	30.9 ± 11.5
Non-dominant Hand	28.6 ± 9.5*

* Significant correlation.

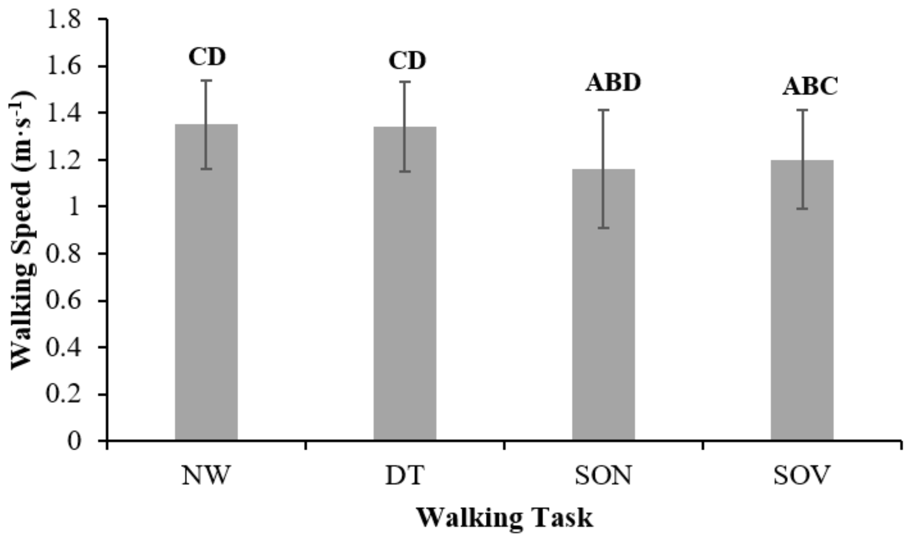


Figure 6.3 Walking speed for this older adult population (n = 98) performing four walking tasks. *Note:* letters indicate the significant differences (A = NW, B = DT, C = SON and D = SOV). *Abbreviations:* NW (Normal Walking), DT (Dual Task Walking), SON (Stepping Onto and Off an Obstacle) and SOV (Stepping Over and Obstacle).

6.3.1. <u>Arm Swing</u>

There was a significant within-subject walking task effect for arm swing amplitude (Table 6.2. and Figure 6.4.). For the dominant arm, SOV obstacle clearance walking task had a significantly increased arm swing compared to NW and SON walking tasks. For the non-dominant arm, the obstacle clearance task had a significantly increased arm swing compared to NW (SON: 24.87 ± 8.11° SOV: 25.96 ± 7.85 ° vs. NW: 22.21 ± 6.23 °). No significant difference was found between NW and DT walking for the non-dominant arm or between the obstacle clearance tasks. There was however a significant increase in non-dominant arm swing for SOV compared to DT walking (SOV: 25.96 ± 7.85 ° vs. DT: 22.59 ± 7.45 °). All walking tasks were found to be asymmetrical for arm swing. In addition, age was not significantly correlated to arm swing amplitude or asymmetry for any walking task (Table 6.3.). There were however significant correlations between walking speed and arm swing amplitude (NW, SON and SOV and DT (non-dominant arm swing)).

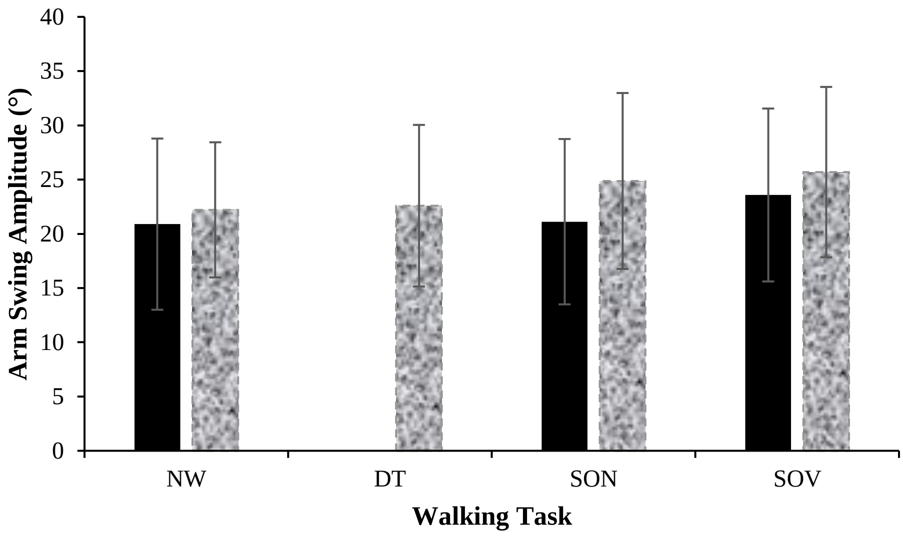


Figure 6.4. Arm swing amplitude (mean ± SD) for this older adult population (n =98) during four walking tasks. *Note:* black line = dominant hand and grey shaded line = non-dominant hand.

6.3.2. Forearm Swing

There was no significant walking task effect for forearm swing (Table 6.2.). All walking tasks were found to be asymmetrical for forearm swing. In addition, age was significantly correlated to the non-dominant arm during NW and SOV walking tasks for forearm swing (Table 6.3.). Walking speed was significantly correlated to forearm swing amplitude (NW, SON and SOV with DT (non-dominant arm)) and asymmetry (DT and SOV). There were however no significant correlations with age when controlling for walking speed.

Table 6.2. Arm swing and forearm swing amplitude and asymmetry during all walking tasks for an older adult population.

	NW	DT	SON	SOV	*F Value*
Arm Swing Amplitude (°)					
Dominant	20.89 ± 7.89[b]	-	21.12 ± 7.62[b]	23.58 ± 7.97[a]	$F_{2.514,238.784}$ = 199.790, P = 0.000*
Non-dominant	22.21 ± 6.23	22.59 ± 7.45	24.87 ± 8.11[a]	25.96 ± 7.85[ac]	$F_{2.233,212.108}$ = 12.157, P = 0.000*
Forearm Swing Amplitude (°)					
Dominant	47.31 ± 21.37	-	44.02 ± 17.83	43.17 ± 18.16	$F_{2.321,220.520}$ = 154.981, P = 0.000*
Non-Dominant	55.59 ± 17.05	55.80 ± 20.00	56.60 ± 20.12	55.98 ± 18.87	$F_{2.188,207.893}$ = 24.830, P = 0.906
Arm Swing Asymmetry (%)	9.3 ± 8.2 ^	-	11.7 ± 7.2 ^	10.5 ± 7.5 ^	
Forearm Swing Asymmetry (%)	10.1 ± 9.5 ^	-	13.5 ± 8.6 ^	12.8 ± 8.5 ^	

* Significant difference. [a] Significantly different to NW, [b] Significantly different to SOV [b] Significantly different to DT. ^ Walking tasks which were asymmetrical based on a cut-off of 7.4 % (Lewek *et al.*, 2010). *Abbreviations:* NW (Normal Walking), DT (Manual Dual Task Walking), SON (Stepping Onto and Off an Obstacle) and SOV (Stepping Over an Obstacle).

Table 6.3. Correlations between age and walking speed for arm swing, forearm swing amplitude and asymmetry for all walking tasks, with partial correlation controlling for walking speed.

		NW	
	Age	**Walking Speed**	**Age Controlling for Walking Speed**
Arm Swing Amplitude (°)			
Dominant	0.022	0.388*	0.153
Non-dominant	-0.045	0.359*	0.066
Forearm Swing Amplitude (°)			
Dominant	-0.183	0.491*	-0.49
Non-dominant	-0.231*	0.524*	-0.97
Arm Swing Asymmetry	0.022	-0.063	0.004
Forearm Swing Asymmetry	0.108	-0.163	0.065
		DT	
	Age	**Walking Speed**	**Age Controlling for Walking Speed**
Arm Swing Amplitude (°)			
Dominant	0.151	-	0.218
Non-dominant	-0.046	0.243*	0.030
Forearm Swing Amplitude (°)			
Dominant	0.130	-	0.197
Non-dominant	-0.253*	0.308*	-0.176
		SON	
Arm Swing Amplitude (°)			
Dominant	0.027	0.323*	0.131
Non-dominant	0.001	0.358*	0.115
Forearm Swing Amplitude (°)			
Dominant	-0.129	0.393*	-0.019
Non-dominant	-0.113	0.358*	-0.012
Arm Swing Asymmetry	0.085	0.054	0.105
Forearm Swing Asymmetry	0.176	-0.172	0.135

		SOV	
Arm Swing Amplitude (°)			
Dominant	0.029	0.277*	0.140
Non-dominant	0.003	0.293*	0.117
Forearm Swing Amplitude (°)			
Dominant	-0.040	0.330*	0.084
Non-dominant	-0.211*	0.327*	-0.110
Arm Swing Asymmetry	0.169	-0.142	0.129
Forearm Swing Asymmetry	0.099	-0.214*	0.027

* Significant correlation. *Abbreviations:* NW (Normal Walking), DT (Manual Dual Task Walking), SON (Stepping Onto and Off an Obstacle) and SOV (Stepping Over an Obstacle).

6.4. Discussion

Arm swing has been suggested to be a mechanism which stabilises the body and helps with lateral balance during walking (Ortega *et al.*, 2008). Typically, research exploring the association of age on arm swing has focused on comparing young to older adults which assumes older adults can be categorised into a single group. As such, a more appropriate method was to analyse arm swing within an older adult population. In addition, independent living for older adults requires the ability to adapt to a changing environment for example stair negotiation. It has been found, an increase in task complexity is associated with an increase in fall risk amongst older adults (Müller and Sternad, 2009). Consequently, arm swing has found to be an important mechanism in the recovery phase after a trip (Pijnappels *et al.*, 2010). Therefore, the aim of this study was to explore the effect of walking task on arm swing for an older adult population. In addition, arm swing movement has also been quantified with elbow to hand movement (i.e. forearm swing). As such, the secondary aim of this study was to establish if walking task affected forearm swing for the older adult population. Overall, this study found arm swing amplitude increased for the obstacle clearance walking tasks compared to NW and DT walking. Age was not found to significantly correlate with either arm swing or forearm swing. Consequently, the hypotheses were only partially accepted.

NW arm swing amplitude for this study (dominant arm: 20.89 ± 7.89 ° and non-dominant arm: 22.21 ± 6.23 °) was comparable to values reported in the literature for young adults (Krasovsky *et al.*, 2014, Plate *et al.*, 2015) (e.g. dominant: 19.8 ± 11.4 ° and non-dominant: 22.6 ± 7.5 °). However, when compared to older adults in the literature (Krasovsky *et al.*, 2014, Mirelman *et al.*, 2015, Mirelman *et al.*, 2016), this study had lower arm swing amplitude values (≥ 25 °). These discrepancies in the literature may be due to the method of data collection. For example, Krasovsky *et al.* (2014) used a Vicon Motion Capture System with a similar marker placement, however their study was conducted on a treadmill. It has been reported walking on a treadmill is not equivalent to overground walking for older adults (Row Lazzarini and Kataras, 2016). Treadmill walking causes increased step width (Dean *et al.*, 2007, Rosenblatt and Grabiner, 2010, Kubinski *et al.*, 2015), energy expenditure (Dean *et al.*, 2007, Parvataneni *et al.*, 2009, Dal *et al.*, 2010, Berryman *et al.*, 2012, Kubinski *et al.*, 2015) and impacts on walking coordination (Carpinella *et al.*, 2010). It has also been reported that vertical and anterior-posterior gait smoothness deteriorates when walking on a treadmill (Row Lazzarini and Kataras, 2016) and so arm swing may also differ when walking on a treadmill compared to walking on the ground.

Arm swing for NW was asymmetrical (9.3 ± 8.2 %), as Lewek *et al.* (2010) reported the cut-off asymmetry angle to be 7.4 %. When compared to the literature (Lewek *et al.*, 2010, Mirelman *et al.*, 2015, Mirelman *et al.*, 2016), asymmetry for this study was slightly higher, for example Mirelman *et al.* (2015) asymmetry was reported to be 8.2 ± 3.2 % for older adults aged 61-77 years old. However, arm swing asymmetry was lower than values (> 20) reported by Plate *et al.* (2015), although this study used the asymmetry index, which was 2.4 times the mean of Lewek *et al.* (2010) healthy older adults. It was suggested asymmetry of arm swing amplitude may not compromise normal gait (Plate *et al.*, 2015). In addition, Mirelman *et al.* (2015) for middle-aged adults (41-50 yrs) reported asymmetry of 9.4 ± 4.6 %, which was similar to this study. This researcher concluded both NW and DT walking were fairly symmetrically, despite being over the suggested 7.4 % cut-off (Lewek *et al.*, 2010). Nevertheless, arm swing amplitudes were asymmetrical for this study.

For DT walking, there was no significant difference for non-dominant arm swing when compared to NW for this study, which contradicts previous research (Mirelman *et al.*, 2015). Mirelman *et al.* (2015) suggested a lower arm swing amplitude for DT walking (cognitive DT: counting in 3s) was due to a reduced walking speed. However, this study found no significant difference between NW and DT walking for walking speed. A number of other dual task gait studies which have used verbal tasks have shown a reduction in walking speed compared to normal walking (Yogev *et al.*, 2005, Springer *et al.*, 2006, Hollman *et al.*, 2007). The allocation of attention may differ between manual- and cognitive-tasks (Asai *et al.*, 2014). For example, the attention for a cognitive task and walking is split and allocated arbitrarily to each task, thus the additional cognitive task draws attention away from walking resulting in a change to the gait (Yogev-Seligmann *et al.*, 2010). However, when a manual-task is used reductions in walking speed were less apparent compared to cognitive dual tasks and this may be because both walking and the manual task are both within the motor control system (Yogev-Seligmann and Hausdorff, 2008). The manual task used in this present work may have not been demanding enough to elicit a change in walking speed. For example, the manual task used Asai *et al.* (2014) involved carrying a ball on a tray one-handed. This task resulted in a significant reduction in walking speed compared to NW. When using cognitive tasks consideration needs to be made to ensure the task is challenging enough to load the attentional system, but it should not cause undue stress or anxiety to the participants (Yogev-Seligmann and Hausdorff, 2008). Similar considerations should be made for manual tasks; however, these tasks are not as commonly used in dual task studies and this may be because no standard manual dual-task currently exists (Asai *et al.*, 2014). It is suggested that a manual dual task should replicate a 'real world' action and one which participants would encounter daily. Furthermore, walking speed may not have been a sensitive enough measure to distinguish changes in gait between NW and DT. For example, Asai *et al.* (2014). reported a significant difference in lower trunk oscillations, measured via accelerometery, between manual and cognitive task (a reduction in oscillations for the manual task and an increase in the cognitive task) even though walking speed was comparable between the two tasks. As such, this manual DT may have not been sufficient enough to cause a change in arm swing. Therefore, in Chapter Five, this reduction in toe-clearance parameters for DT walking was unlikely to have been affected by arm swing. This biomechanical change was possibly caused by ageing cognitive decline associated with performing a secondary task.

The obstacle clearance task did however challenge the older adults in this study, as there was a significant effect on arm swing amplitude. Arm swing was increased for the obstacle clearance tasks compared to NW. This suggests as task complexity increases, so does arm swing. An increased arm swing may aid gait stability during obstacle clearance tasks (Ortega *et al.*, 2008, Bruijn *et al.*, 2010, Nakakubo *et al.*, 2014, Punt *et al.*, 2015), as such tasks place higher demands on balance to negotiate safe step clearance (Deshpande *et al.*, 2009). Arm swing has found to aid metabolic walking cost (Ortega *et al.*, 2008, Umberger, 2008, Collins *et al.*, 2009a), but also counteracts vertical angular momentum (Elftman, 1939, Herr and Popovic, 2008, Park, 2008, Collins *et al.*, 2009a, Bruijn *et al.*,

2010) contributing to lateral stabilisation. It has been identified emphasising arm swing during walking aids overall global gait stability, not only for young and middle-aged adults (Lulic *et al.*, 2008, Hu *et al.*, 2012), but also for older adults (Nakakubo *et al.*, 2014). This arm swing mechanisms may have been adopted by this older adult population to counteract angular momentum during the obstacle clearance tasks and to potentially ensure safe toe-clearance during such tasks. In addition, walking speed was found to significantly reduce for obstacle clearance compared to NW and DT. For instance, obstacle clearance requires greater swing time which is likely to result in a greater stance time for the supporting limb (Patla and Rietdyk, 1993, Chou and Draganich, 1997), thus a slower walking speed. Inappropriate co-ordination of the body segments when crossing an obstacle is likely to perturb balance resulting in a fall (Greenspan *et al.*, 1994, Nevitt and Cummings, 1994). Furthermore, as seen in Chapter Four walking speed is influenced by reduced joint moments (Riley, 2001) and during NW the over 75 years age group had reduced joint powers generations which may affect the propulsive forces at toe-off. As such, arm swing may also be compensation for landing forces during obstacle clearance. Therefore, future work should explore landing forces and joint kinetics during obstacle clearance for this older adult population.

Age was not associated with arm swing and forearm swing for any walking tasks, when controlling for walking speed. Despite previous research (Krasovsky *et al.*, 2014, Mirelman *et al.*, 2015, Plate *et al.*, 2015) reporting an age association for arm swing. These studies did however, compare young to older adults. Walking speed for all tasks was significantly correlated to all walking tasks, previous research also found similar findings (Lewek *et al.*, 2010, Mirelman *et al.*, 2015). This suggests arm and forearm swing are independent of walking speed. A reduced walking speed is therefore associated with an increased dominant arm swing for the obstacle clearance tasks.

A major limitation to this study was due to the technical limitations of the biomechanical laboratory design. This laboratory has low ceilings with a beam running along the horizontal axis of the walkway, which meant motion capture cameras had to be positioned underneath the mounting rig (Figure 2.4.). This was not a desirable motion capture set-up as it reduces the field of view in the vertical axis and consequently full body motion analysis becomes harder to track. As a result, 60 participants were excluded because of excessive marker trajectory gaps in the upper body and this also meant age effect was prevented. Future work could explore these walking tasks in older adults above the age of seventy to determine the effect and if this potentially impacts on fall risk. Furthermore, gait stability and centre of mass analysis (Nakakubo *et al.*, 2014) may illustrate that older adults display unstable walking patterns despite making cautious gait alterations (Yack and Berger, 1993).

6.5. Conclusion

Age was not associated with arm swing amplitudes. Forearm swing was affected by walking task. Walking task did reveal a significant task effect. Demonstrating, an increase in task complexity resulted in an increased arm swing for the obstacle clearance tasks. Reduced walking speed was found for the obstacle clearance tasks compared to normal and dual task walking. An increased arm swing may aid gait stability when balance is challenged during obstacle clearance tasks. As such, arm swing may also be compensation for landing forces (e.g. propulsive force) during obstacle clearance. Therefore, future work should explore landing forces and joint kinetics during obstacle clearance for this older adult population.

Chapter Seven: The Kinetics of Landing Following Obstacle Clearance in an Older Adult Population

ABSTRACT

Introduction: Stepping onto or over an obstacle is a commonly performed task required to negotiate the environment around us. Vertical ground reaction forces increase with an increase in task complexity, for example step negotiation (e.g. step descent). An increase in age typically illustrates reduced peak second vertical and propulsive ground reaction forces for both normal walking and obstacle clearance tasks. However, contradictory evidence occurs for first and minimum peak ground reaction forces, as research typically compared young to older adults. As such, the aim of this study was to determine the alterations on landing mechanics for obstacle clearance when compared to normal walking in older adults.

Methods: 158 community-dwelling older adults, age range 55-86 years (65.7 ± 6.8 yrs) participated and walked at their self-determined comfortable walking speed for three walking tasks: normal walking and two obstacle clearance tasks (stepping onto and off an obstacle and stepping over an obstacle). Three-dimensional motion analysis was used to capture joint kinetics, with a mounted force plate determining ground reaction forces. Five ground reaction force peaks were identified according to convention (F1-F5), with impulse calculated for vertical, braking and propulsive force for each walking task.

Results: Age was significantly correlated to braking and propulsive force for normal walking and stepping onto and off an obstacle, when controlling for walking speed. Task effect was found for all spatial-temporal parameters except double-support time. For example, stepping onto and off an obstacle illustrated a reduced stride length and increased step width compared to normal walking and stepping over an obstacle. Walking speed significantly reduced for both obstacle clearance tasks compared to normal walking (normal walking: 1.43 ± 0.18 $m \cdot s^{-1}$, stepping onto and off an obstacle: 1.11 ± 0.21 $m \cdot s^{-1}$ and stepping over an obstacle: 1.24 ± 0.20 $m \cdot s^{-1}$). There was also a significant task effect for F1, F3, F4 and F5, with stepping onto and off an obstacle having an increased first and reduced second vertical peak compared to normal walking and stepping over an obstacle (e.g. F1: stepping onto and off an obstacle clearance: 1.63 ± 0.21 BwFz vs. normal walking: 1.12 ± 0.10 BwFz and stepping over an obstacle: 1.13 ± 0.13 BwFz). Joint kinetics illustrated altered hip moments and knee and ankle power for both obstacle clearance walking tasks. For example, ankle power generation reduced for both obstacle clearance tasks compared to normal walking.

Conclusion: An increase in age was associated with a reduced braking and propulsive force for normal walking and obstacle clearance, which is potentially due to reduced power generation, which could impact on toe-clearance and consequently this population compensates by increasing their arm swing when task complexity increases. Increase in walking task demand was associated with an altered ground reaction forces. Older adults in this population typically employed a gait strategy to compensate for task by altering joint kinetics, reducing walking speed and step length and increasing base of support. First vertical peak was increased for stepping onto and off an obstacle, with a reduced second vertical peak force compared to normal walking and stepping over an obstacle. In addition, braking force was significantly lower for stepping onto and off an obstacle. Therefore, such task places higher demands on balance and increases the likelihood of slip occurrence. This may be a potential risk to the older aged adults in this population when performing such a task, as age was significantly correlated to braking force when controlling for walking speed.

Keywords: Obstacle Clearance; Older Adults; Biomechanical Strategy; Ground Reaction Forces; Joint Kinetics

7.1. Introduction

The ability to move and walk around a changing environment underline the successful achievement of many tasks necessary for independent living and as such the ability to clear obstacles is important for functional mobility. Such tasks have been identified as the most difficult tasks for older adults to perform (Williamson and Fried, 1996, Yu *et al.*, 1997, Benson *et al.*, 2002, Christina and Cavanagh, 2002, Sheehan and Gottschall, 2012), with step descent resulting in the most serious injuries (Garcia *et al.*, 2006, Jacobs, 2016).

Ground reaction forces (GRF), for example vertical force, can determine the state of locomotion (Jacobs *et al.*, 1972), which can indicate the intensity of musculoskeletal stress, by examining the external force which influences the body's centre of mass (Winter, 1991, McClay *et al.*, 1994). First vertical GRF (F1) peak results in load accommodation for foot contact to assist contralateral lower limb swing and foot contact, with the second vertical GRF (F3) peak acting as descent control, this typically lasts between mid-stance and pre-swing (McFadyen and Winter, 1988). The higher the force magnitude, for example descending a step, the more dissipated the load on musculoskeletal system to shock absorb and distribute force (Ricard and Veatch, 1990, Crossley *et al.*, 1999), which increases the risk of joint injury and pathologies (Dufek and Bates, 1990, McNitt-Gray, 1991, Irmischer *et al.*, 2004, Elvin *et al.*, 2007).

During normal walking (NW), vertical GRF is affected by age, as older adults have illustrated a reduced first and second peak force and higher minimum mid-stance peak force compared to young adults (Yamada and Maie, 1988). However, Toda *et al.* (2015) reported no significant difference for first peak and minimum mid-stance peak, with a significant reduction for second GRF vertical peak for older adults compared to young adults. It is suggested amplitude of peak vertical GRF is influenced by cadence as oppose to stride length (Martin and Marsh, 1992), which may explain why there was no difference between young and older adults at first GRF vertical peak.

Muscle force generation is important to increase joint moment during walking (Riley *et al.*, 2001). Knee extension moment during late stance has been associated with a decreased GRF at mid-stance and increased GRF during weight acceptance and push-off for young female adults (Toda *et al.*, 2015), whereas older adults adopt an increase hip extension moment to maintain an increased GRF during early stance. Toda *et al.* (2015) also reported female older adults had a positive relationship between second vertical GRF (F3) peak and ankle plantarflexion moment, which acts as the main support for push-off (Winter, 1980, Winter, 1991, Perry and Burnfield, 2010). Therefore, a reduced ankle plantarflexion moment may be the cause of reduced second GRF vertical peak for older adults. As such, there is an age-related change in landing force strategy (alteration of joint kinetics and/or vertical GRF) during NW.

For obstacle clearance tasks such as step descendent, the first vertical GRF (F1) peak is greater compared to the second vertical GRF (F3) peak when comparing to NW (Christina and Cavanagh, 2002). Differences also occur for anterior-posterior GRF, although braking impulse is similar to NW, propulsive impulse is lower (Christina and Cavanagh, 2002, Riener *et al.*, 2002).

There have been contradictory claims in the literature regarding the effects of age on GRF (vertical and anterior-posterior) for both NW and obstacle clearance tasks, as some studies (Reeves *et al.*, 2008, Silva *et al.*, 2015) have reported no differences in GRF when comparing young to older adults. With some studies, only finding differences between second vertical GRF (F3) peak (Toda *et al.*, 2015) and propulsive GRF (F5) peak (Christina and Cavanagh, 2002). The effects of body weight loading (i.e. GRF) during overground walking for healthy adults has been applied to rehabilitation protocols for individuals with gait impairment (Barela *et al.*, 2014). As such, a different research approach would be to investigate GRF parameters within a group of older adults. As previously mentioned, Chapter Four revealed altered joint kinetics (e.g. reduced joint powers) for the over 75 years age group, which is likely to be a consequence of reduced muscle strength in the ageing process. Furthermore, the majority of gait alterations for this older adult population occurred in late stance either at/or near toe-off. As a result, landing forces such as propulsion may be affected. Therefore, the aim of this study was to determine the alterations on landing mechanics and joint kinetics for obstacle clearance when compared to normal walking in an older adult population. It was hypothesised an increase in age would be associated with a decreased second vertical GRF (F3) and propulsive GRF (F5) peaks for all walking tasks. Furthermore, it was hypothesised, as task complexity increases not only would vertical GRF increase, but joint kinetics adaptations would occur for age and task. As seen in Chapter Three, this older adult population typically altered their hip joint kinetics with age, as opposed to an ankle joint kinetic strategy.

7.2. Methodology

The overall methodology is provided in Chapter Two. All participants performed five trials for NW (with force plate contact) and obstacle clearance tasks (stepping onto and off an obstacle (SON) and stepping over an obstacle (SOV)). No instruction was given regarding leading leg for the obstacle clearance tasks; participants self-selected. Data analysis was completed using a custom-made Python code (Python v.2.7.10, Delaware, USA). Joint kinetics (lower body) and the GRFs anterior-posterior (Fy) and vertical (Fz) were normalised to one gait cycle (100 %), using linear interpolation to 101 data samples. The calculations were guided by Winter *et al.* (1990) and Winter (1992) for joint kinetics and Levine *et al.* (2012) for GRF. From the normalised GRF five convention peak forces were identified (Figure 7.1.): F1 (first peak vertical force), F2 (minimum vertical peak force), F3 (second peak vertical force), F4 (braking peak force) and F5 (propulsive peak force). GRFs were normalised by body weight. Vertical impulse, braking impulse and propulsion impulse were calculated using the trapezoidal rule, using the raw ground reaction force data.

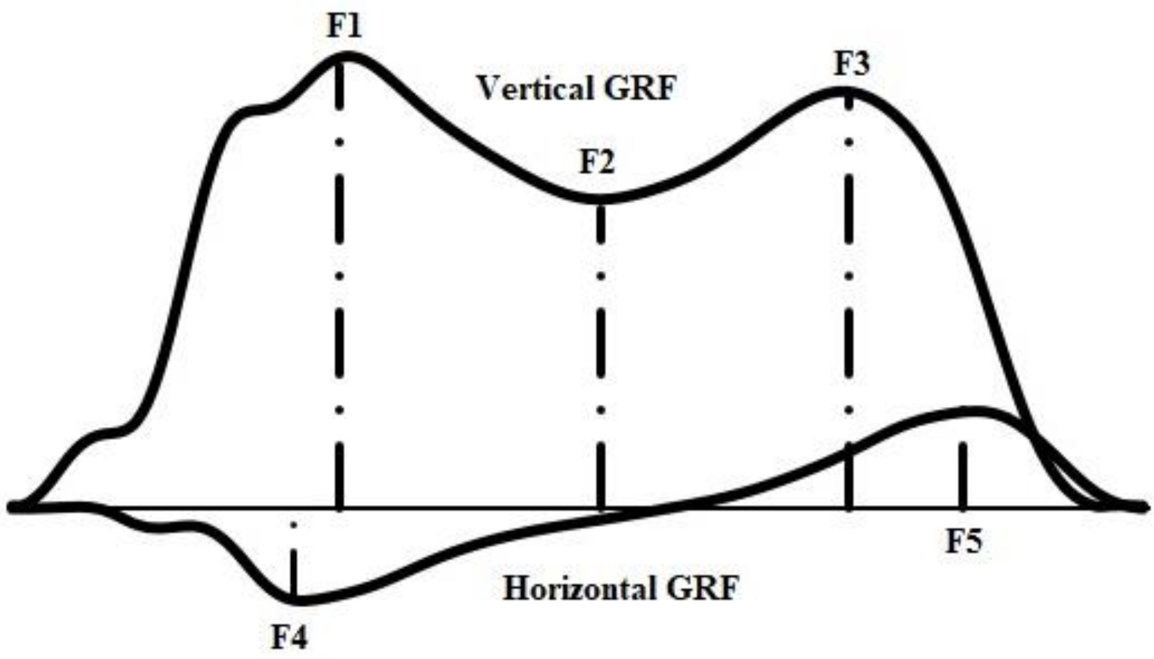


Figure 7.1. The five peak ground reaction forces (F1 (first peak vertical force), F2 (minimum vertical peak force), F3 (second peak vertical force), F4 (braking peak force) and F5 (propulsive peak force)).

Statistical analysis was performed using IBM SPSS v.23 software (Chicago, USA). Pearson's R correlation between age and walking speed for MMSE, GRFs and joint kinetics were calculated for all walking tasks for the whole population, with a partial correlation to control for walking speed between age, GRFs and joint kinetics. Mixed ANOVAs, with GRF and joint kinetic peaks as the dependent variable, age groups (55-64 yrs, 65-74 yrs and ≥ 75 yrs) as the between factors, and the walking tasks (NW, SON and SOV) as the within factors were also used. The ANOVAs were followed by pre-planned comparisons based on Bonferroni adjusted post-hoc Tukey tests.

7.3. Results

Following data collection, seventy-four participants, age range 59-75 years (40 females; 34 males; 67.0 ± 7.7 yrs; 169.4 ± 9.9 cm; 75.9 ± 14.5 kg) were excluded from the study (Figure 7.2.). Therefore, eighty-four participants, age range 55-80 years (64.6 ± 5.7 yrs) were included in the study (Table 7.1.). As such, one-way within-subjects' ANOVA's were performed to determine the walking task effect on GRF and joint kinetic peaks were performed instead of mixed ANOVAs. The ANOVAs were followed by pre-planned comparisons based on Bonferroni adjusted post-hoc Tukey tests. Two participants tripped whilst performing the SOV obstacle clearance task (1 female; 61 yrs; 1 trip and 1 female; 64 yrs; 2 trips). There was no significant correlation for MMSE with age or walking speed (Table 7.1.).

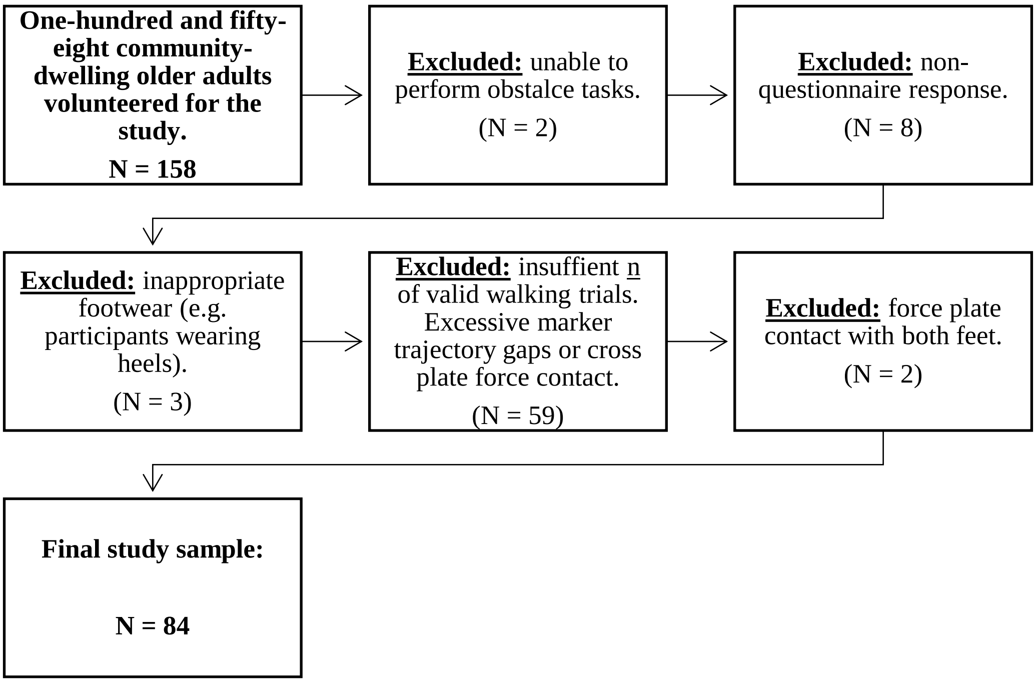


Figure 7.2. Description of participant selection and participant exclusion.

Table 7.1. Participant characteristics.

	Whole Group (n = 84)
Sex (Female/Male)	61/23
Age (yrs)	64.6 ± 5.7
Height (cm)	167.7 ± 8.5
Mass (kg)	72.3 ± 15.0
MMSE	29 ± 1

7.3.1. Spatial-Temporal Parameters

There was a significant within-subject effect for task for all spatial-temporal parameters with the exception of double-support time (Table 7.2.). Cadence was significantly higher for NW, with a reduced single-support and stride time compared to the obstacle clearance tasks (e.g. cadence for NW: 116.0 ± 9.0 steps/min vs. SON: 103.4 ± 12.8 steps/min and SOV: 101.1± 10.1 steps/min). All walking tasks were significantly different for walking speed, step time, foot-off, opposite foot contact and limp index. SON obstacle clearance task demonstrated the slowest walking speed, step time and delayed foot-off and opposite foot contact, with an increased limp index time compared to NW and SOV obstacle clearance task (e.g. walking speed - SON: 1.11 ± 0.21 $m \cdot s^{-1}$, SOV: 1.24 ± 0.20 $m \cdot s^{-1}$ and NW: 1.43 ± 0.18 $m \cdot s^{-1}$). Opposite foot-off occurred significantly earlier in the gait cycle for SOV obstacle clearance task compared to NW and SON walking tasks (SOV: 9.64 ± 2.02 % vs. NW: 11.84 ± 1.66 % and SON: 12.69 ± 2.11 %). Also, SON obstacle clearance task had a reduced step and stride length and an increased step width compared to the NW and SOV walking tasks (Table 7.2.).

Table 7.2. Task effect for spatial-temporal parameters for an older adult population.

Parameter	NW	SON	SOV	*F* Value
Rhythm				
Cadence (steps/min)	116.0 ± 9.0	103.4 ± 12.8[a]	101.1 ± 10.1[a]	$F_{1.926,126.024}$ = 93.088, P = 0.000*
Step Time (s)	0.51 ± 0.04[b]	0.47 ± 0.06[b]	0.53 ± 0.05[b]	$F_{1.757,142.319}$ = 41.543, P = 0.000*
Stride Time (s)	1.04 ± 0.08	1.18 ± 0.16[a]	1.20 ± 0.12[a]	$F_{1.736, 140.605}$ = 74.186, P = 0.000*
Single-support Time (s)	0.41 ± 0.03	0.57 ± 0.08[a]	0.56 ± 0. 06[a]	$F_{1.907,154.467}$ = 152.227, P = 0.000*
Phases				
Double-support Time (s)	0.25 ± 0.05	0.24 ± 0.06	0.24 ± 0.06	$F_{2,162}$ = 0.000, P = 1.000
Foot-off (%)	62.86 ± 2.03[b]	67.30 ± 2.91[b]	66.41 ± 2.68[b]	$F_{1.965,12.296}$ = 43.728, P = 0.000*
Limp Index (s)	1.03 ± 0.03	1.30 ± 0.09	1.25 ± 0.08	$F_{2,162}$ = 162.700, P = 0.000*
Opposite Foot Contact (%)	50.84 ± 0.85[b]	60.44 ± 2.82[b]	56.30 ± 2.13[b]	$F_{1.817,147.167}$ = 183.268, P = 0.000*
Opposite Foot-off (%)	11.84 ± 1.66[c]	12.69 ± 2.11[c]	9.64 ± 2.02[c]	$F_{2,162}$ = 29.485, P = 0.000*
Pace				
Walking Speed ($m \cdot s^{-1}$)	1.43 ± 0.18[b]	1.11 ± 0.21[b]	1.24 ± 0.20[b]	$F_{2,162}$ = 147.913, P = 0.000*
Step Length (m)	0.72 ± 0.06[d]	0.70 ± 0.07	0.73 ± 0.07[d]	$F_{1.933,156.540}$ = 16.460, P = 0.000*
Stride Length (m)	1.47 ± 0.13[d]	1.29 ± 0.12[d]	1.47 ± 0.13[d]	$F_{2.997,121.365}$ = 76.668, P = 0.000*
Base of Support				
Step Width (m)	0.15 ± 0.05[d]	0.20 ± 0.03[d]	0.14 ± 0.04[d]	$F_{1.871,151.530}$ = 45.440, P = 0.000*

* Significant task effect. [a] NW significantly different to SON and SOV; [b] all tasks significantly different; [c] SOV significantly different to NW and SON; [d] SON significantly different to NW and SOV. *Abbreviations:* Normal Walking (NW), Stepping Onto and Off an Obstacle (SON) and Stepping Over and Obstacle (SOV).

7.3.2. *Ground Reaction Forces*

There was a significant task effect at F1, F3, F4 and F5 (Table 7.3.). SON was significantly higher for F1 (SON: 1.63 ± 0.21 BwFz vs. NW: 1.12 ± 0.10 BwFz and SOV: 1.13 ± 0.13 BwFz) and lower for F3 (SON: 0.93 ± 0.15 BwFz vs. NW: 1.13 ± 0.09 BwFz and SOV: 1.08 ± 0.09 BwFz) compared to NW and SOV walking tasks. SOV was significantly higher for F5 (SOV: 0.27 ± 0.06 BwFy vs. NW and SON: 0.24 ± 0.04 BwFy) compared to NW and SON walking tasks and for F4 all walking tasks were significantly different (NW: -0.19 ± 0.05 BwFy, SON: -0.12 ± 0.03 BwFy and SOV: -0.26 ± 0.06 BwFy) (Table 7.3. and Figure 7.3.).

There was a significant task effect on vertical impulse, braking impulse and propulsive impulse. SOV obstacle clearance task was significantly higher for vertical impulse compared to NW and SON walking tasks (vertical (Fz) impulse: SOV: 0.66 ± 0.08 N·s vs. NW: 0.51 ± 0.05 N·s and SON: 0.53 ± 0.07 N·s). All walking tasks were significantly different for braking and propulsive impulse (e.g. braking impulse: NW: -0.03 ± 0.01 N·s SON: -0.01 ± 0.01 N·s and SOV: -0.04 ± 0.01 N·s) (Table 7.3.).

In addition, there were significant correlations between age and GRF peaks (F4 and F5) for NW and SON walking tasks and only F5 for SOV obstacle clearance task. There were significant correlations between age and impulse (braking impulse for NW, propulsive impulse for SON and Fz impulse for SOV walking task). There were also significant correlations between walking speed and GRF peaks (F1, F2, F3, F4 and F5) and impulse (propulsive impulse) for the NW task. There were significant correlations between walking speed and GRF peaks (F1, F2, F3, F4 and F5) and impulses (Fz impulse and propulsive impulse) for SON walking task. There were significant correlations between walking speed and GRF peaks (F2, F3, F4 and F5) and impulse (Fz impulse) for SOV walking task. When controlling for walking speed, there were significant correlations between age and F4 and F5 for NW and SON walking tasks. However, no age correlations were found for SOV when controlling for walking speed (Table 7.4.).

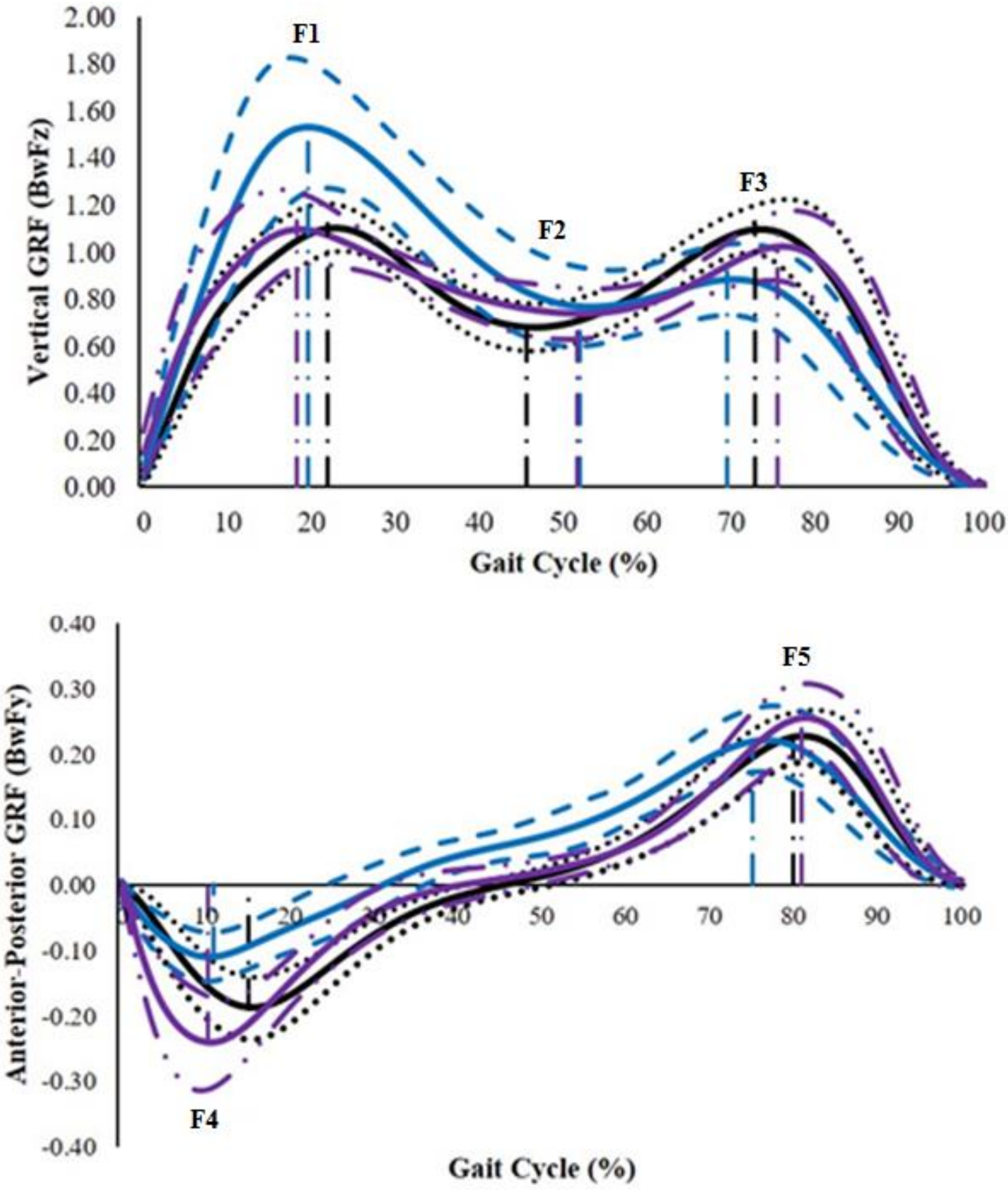


Figure 7.3. Vertical and anterior-posterior ground reaction force for an older adult population performing three walking tasks (black line: NW (normal walk), blue line: SON (stepping onto and off an obstacle), purple line: SOV (stepping over an obstacle), dashed line: standard deviation). The five peak ground reaction forces (F1 (first peak vertical force), F2 (minimum vertical peak force), F3 (second peak vertical force), F4 (braking peak force) and F5 (propulsive peak force)).

Table 7.3. Task effect on ground reaction peaks and impulse for an older adult population.

	NW	SON	SOV	*F* Value
Vertical GRF (BwFz)				
T1	1.12 ± 0.10[a]	1.63 ± 0.21	1.13 ± 0.13[a]	$F_{1.509,122.214}$ = 165.990, P = 0.000*
T2	0.67 ± 0.10	0.72 ± 0.16	0.71 ± 0.11	$F_{1.357,109.931}$ = 1.495, P = 0.229
T3	1.13 ± 0.09[a]	0.93 ± 0.15	1.08 ± 0.09[a]	$F_{1.645,133.219}$ = 35.104, P = 0.000*
Anterior-Posterior GRF (BwFy)				
T4	-0.19 ± 0.05[b]	-0.12 ± 0.03[b]	-0.26 ± 0.06[b]	$F_{2,162}$ = 93.525, P = 0.000*
T5	0.24 ± 0.04[c]	0.24 ± 0.04[c]	0.27 ± 0.06	$F_{1.949,157.890}$ = 10.226, P = 0.000*
Fz Impulse (N·s)	0.51 ± 0.05[c]	0.53 ± 0.07[c]	0.66 ± 0.08	$F_{2,162}$ = 131.805, P = 0.000*
Braking Impulse (N·s)	-0.03 ± 0.01[b]	-0.01 ± 0.01[b]	-0.04 ± 0.01[b]	$F_{1.881,152.347}$ = 124.782, P = 0.000*
Propulsion Impulse (N·s)	0.04 ± 0.01[b]	0.05 ± 0.01[b]	0.05 ± 0.01[b]	$F_{2,162}$ = 44.389, P = 0.000*

* Significant task effect. [a] SON significantly different to NW and SOV; [b] all walking tasks are significantly different; [c] SOV significantly different to NW and SON; [d] NW significantly different to SOV; [e] NW significantly different to the obstacle clearance walking tasks. *Abbreviations:* Normal Walking (NW), Stepping Onto and Off an Obstacle (SON), Stepping Over an Obstacle (SOV), First Peak Vertical Force (F1), Minimum Vertical Peak Force (F2), Second Peak Vertical Force (F3), Braking Peak Force (F4), Propulsive Peak Force (F5) and Vertical Impulse (Fz).

Table 7.4. Correlation between age, walking speed and ground reaction forces, including a partial correlation controlling for walking speed for all walking tasks.

		NW	
Parameter	**Age**	**Walking Speed**	**Age Controlling for Walking Speed**
Vertical GRF (BwFz)			
F1	-0.116	0.536*	0.003
F2	0.212	-0.846*	0.046
F3	-0.141	0.284*	-0.084
Anterior-Posterior GRF (BwFy)			
F4	0.331*	-0.596*	0.253*
F5	-0.334*	0.653*	-0.256*
Fz Impulse (N·s)	0.044	-0.696*	-0.158
Braking Impulse (N·s)	0.232*	-0.164	0.204
Propulsive Impulse (N·s)	-0.150	0.216*	-0.108
		SON	
Parameter	**Age**	**Walking Speed**	**Age Controlling for Walking Speed**
Vertical GRF (BwFz)			
F1	-0.116	0.634*	-0.015
F2	0.212	-0.686*	0.137
F3	-0.141	0.319*	-0.095
Anterior-Posterior GRF (BwFy)			
F4	0.331*	-0.583*	0.292*
F5	-0.334*	0.639*	-0.301*
Fz Impulse (N·s)	0.023	-0.691*	-0.128
Braking Impulse (N·s)	0.052	0.176	0.084
Propulsive Impulse (N·s)	-0.257*	-0.256*	-0.314

	SOV		
Parameter	**Age**	**Walking Speed**	**Age Controlling for Walking Speed**
Vertical GRF (BwFz)			
F1	-0.172	0.713*	0.021
F2	0.161	-0.796*	-0.081
F3	-0.094	0.280*	-0.023
Anterior-Posterior GRF (BwFy)			
F4	-0.001	-0.491*	-0.154
F5	-0.304*	0.540*	-0.200
Fz Impulse (N·s)	0.216*	-0.844*	-0.009
Braking Impulse (N·s)	-0.093	-0.014	-0.100
Propulsive Impulse (N·s)	-0.012	-0.143	-0.051

* Significant correlation. *Abbreviations:* Normal Walking (NW), Stepping Onto and Off an Obstacle (SON), Stepping Over an Obstacle (SOV), First Peak Vertical Force (F1), Minimum Vertical Peak Force (F2), Second Peak Vertical Force (F3), Braking Peak Force (F4) and Propulsive Peak Force (F5).

7.3.3. Joint Kinetics

There was a significant within-subject effects for walking task on joint kinetics (hip flexion moment in swing, hip extension moment, hip adduction moment first and second peak, hip abduction moment first and second peak, knee flexion moment at loading response (LR) and pre-swing (PSw), knee extension moment in swing, knee varus and valgus moment second peak, ankle plantarflexion and dorsiflexion moment, hip power generation (H1), hip power generation (H3), knee power generation (K0), knee power absorption (K1), knee power generation (K2), knee power absorption (K3), knee power generation (K4), ankle power absorption (A0), ankle power absorption (A1) and ankle power generation (A2) (Table 7.5.).

During NW, the older adult population had a significantly higher hip abduction moment during first peak, knee power generation (K0), knee power absorption (K3) and ankle power generation (A2) compared to the obstacle clearance walking tasks (e.g. ankle power generation (A2) NW: 4.07 ± 1.21 Watts/kg vs. SON: 2.75 ± 1.00 Watts/kg and SOV: 2.86 ± 1.21 Watts/kg). For SON obstacle clearance walking task, joint kinetics were significantly higher for hip adduction moment first peak and knee flexion moment at LR compared to NW and SOV walking tasks (e.g. knee flexion moment at LR SON: 1.14 ± 0.43 Nm/kg vs. NW: 0.96 ± 0.32 Nm/kg and SOV: 0.72 ± 0.32 Nm/kg). The SON obstacle clearance walking task was significantly lower for hip flexion moment in stance, knee flexion moment at PSw and hip power generation (H1) compared to NW and SOV walking tasks (e.g. hip flexion moment at stance SON: 0.25 ± 0.32 Nm/kg vs. NW: 0.83 ± 0.36 Nm/kg and 0.74 ± 0.44 Nm/kg) (Table 7.5.).

For the obstacle clearance walking tasks, SON joint kinetics was significantly reduced for hip extension moment and knee power generation (K3) compared to SOV (e.g. hip extension moment SON: 0.25 ± 0.32 Nm/kg vs. SOV: 0.74 ± 0.44 Nm/kg). The SOV obstacle clearance walking task, revealed a significantly increased hip adduction and abduction moment, knee valgus moment second peak, with a reduced knee extension moment in swing and ankle dorsiflexion moment compared to NW and SON walking tasks (e.g. ankle dorsiflexion moment SOV: 1.17 ± 0.26 Nm/kg vs. NW: 1.29 ± 0.27 Nm/kg and SON: 1.39 ± 0.30 Nm/kg). In addition, there were significant effects for all tasks for knee varus (adduction) moment second peak, ankle plantarflexion moment, hip power generation (H3), knee power absorption (K1), knee power generation (K4), ankle power absorption (A0) and ankle power absorption (A1) (e.g. ankle power absorption (A1) NW: 4.07 ± 1.21 Watts/kg, SON: 2.75 ± 1.00 Watts/kg and SOV: 2.86 ± 1.21 Watts/kg) (Table 7.5.).

In addition, there were significant correlations between age and joint kinetics (hip extension moment, knee varus (adduction) moment first peak, hip power generation (H3), knee power generation (K4) and ankle power generation (A3)) for NW, (hip flexion moment in stance, hip extension moment, hip ab/adduction second peak, knee varus moment first peak, hip power generation (H1), knee power absorption (K2), knee power generation (K3 and K4) for SON and (hip extension moment, hip ab/adduction second peak, knee valgus (abduction) second peak and knee power generation (K4)) for SOV walking task. Walking speed was significantly correlated to all joint kinetics, except ankle dorsiflexion and ankle power absorption (A1) for NW. For SON obstacle clearance walking task, walking speed was correlated to hip flexion moment in swing, hip abduction and adduction at second peak, knee flexion moment at PSw, knee extension at swing, knee varus and valgus moment second peak, ankle dorsiflexion moment and all joint powers for the hip and ankle, with correlations for knee power generation (K0 and K4) and knee power absorption (K3). For SOV obstacle clearance walking task, walking speed was correlated to all hip joint moments, knee flexion moment at LR, knee extension moment at TS and swing, knee varus moment first peak, knee valgus moment second peak and all joint powers except ankle power absorption (A1). When controlling for walking speed, there were significant correlations between age and joint kinetics (hip extension moment and knee power generation (K4)) for NW, (hip flexion moment in stance and swing, hip adduction moment at second peak, hip power generation (H1) and knee power generation (K4)) for SON and (hip flexion moment in swing and knee power generation (K4)) for SOV walking task (Table 7.6.).

Table 7.5. Joint kinetics for all walking tasks for an older adult population.

Parameter	NW	SON	SOV	*F Value*
Moments (Nm/kg)				
Hip Flexion/Extension				
Flexion Moment - Stance	0.83 ± 0.36[a]	0.25 ± 0.32	0.74 ± 0.44[a]	$F_{1.999,161.958}$ = 16.502, P = 0.000*
Extension Moment	-1.25 ± 0.50	-1.08 ± 0.31	-1.22 ± 0.33[b]	$F_{1.597,129.357}$ = 3.399, P = 0.047*
Flexion Moment - Swing	0.72 ± 0.28	0.65 ± 0.26	0.57 ± 0.27	$F_{2,162}$ = 1.654, P = 0.195
Hip Abduction/Adduction				
Maximum Moment (First Peak)	1.09 ± 0.28[a]	1.25 ± 0.38	0.98 ± 0.31[a]	$F_{1.986,160.865}$ = 13.094, P = 0.000*
Minimum Moment (First Peak)	0.03 ± 0.14	-0.04 ± 0.09[c]	-0.04 ± 0.14[c]	$F_{2,162}$ = 8.391, P = 0.000*
Maximum Moment (Second Peak)	0.61 ± 0.36[d]	0.59 ± 0.25[d]	0.77 ± 0.23	$F_{1.458,113.851}$ = 7.443, P = 0.003*
Minimum Moment (Second Peak)	-0.26 ± 0.30[d]	-0.22 ± 0.19[d]	0.24 ± 0.32	$F_{1.906,154.413}$ = 64.371, P = 0.000*
Knee Flexion/Extension				
Flexion Moment (LR)	0.96 ± 0.32[a]	1.14 ± 0.43	0.72 ± 0.32[a]	$F_{1.924,155.811}$ = 19.512, P = 0.000*
Extension Moment (TS)	0.05 ± 0.14	0.06 ± 0.14	0.05 ± 0.21	$F_{1.825,147.826}$ = 0.880, P = 0.408
Flexion Moment (PSw)	0.49 ± 0.18[a]	0.35 ± 0.13	0.44 ± 0.24[a]	$F_{2,162}$ = 5.909, P = 0.003*
Extension Moment - Swing	-0.42 ± 0.10[d]	-0.39 ± 0.10[d]	-0.35 ± 0.14	$F_{1.732,140.314}$ = 8.498, P = 0.001*
Knee Varus/Valgus				
Varus Moment (First Peak)	0.35 ± 0.14	0.28 ± 0.21	0.30 ± 0.18	$F_{1.931,156.415}$ = 0.599, P = 0.545
Valgus Moment (First Peak)	-0.14 ± 0.11	-0.12 ± 0.20	-0.13 ± 0.18	$F_{2,162}$ = 0.312, P = 0.732
Varus Moment (Second Peak)	0.29 ± 0.15[e]	0.22 ± 0.16[e]	0.32 ± 0.18[e]	$F_{1.964,159.098}$ = 5.116, P = 0.007*
Valgus Moment (Second Peak)	-0.11 ± 0.11[d]	-0.14 ± 0.11[d]	0.04 ± 0.17	$F_{3.379,136.844}$ = 34.074, P = 0.000*
Ankle Plantar/Dorsiflexion				
Plantarflexion moment	-0.37 ± 0.17[e]	0.01 ± 0.07[e]	-0.21 ± 0.20[e]	$F_{2,162}$ = 45.309, P = 0.000*
Dorsiflexion moment	1.29 ± 0.27[d]	1.39 ± 0.30[d]	1.17 ± 0.26	$F_{1.916,155.170}$ = 11.612, P = 0.000*

Parameter	NW	SON	SOV	*F Value*
Powers (Watts/kg)				
Hip Power				
H1 (Generation)	0.79 ± 0.48[a]	0.35 ± 0.44	1.00 ± 0.67[a]	$F_{1.841,149.149} = 11.729$, $P = 0.000$*
H2 (Absorption)	-0.86 ± 0.44	-0.76 ± 0.43	-0.88 ± 0.41	$F_{1.984,160.734} = 2.611$, $P = 0.077$
H3 (Generation)	2.03 ± 0.67[e]	1.80 ± 0.56[e]	1.54 ± 0.69	$F_{1.726,139.822} = 17.182$, $P = 0.000$*
Knee Power				
K0 (Generation)	0.70 ± 0.41	0.24 ± 0.24[c]	0.19 ± 0.30[c]	$F_{2,162} = 13.712$, $P = 0.000$*
K1 (Absorption)	-1.03 ± 0.66[e]	-1.81 ± 1.32[e]	-0.31 ± 0.48[e]	$F_{1.371,111.053} = 25.736$, $P = 0.000$*
K2 (Generation)	1.10 ± 0.60	1.33 ± 0.84	0.75 ± 0.56[b]	$F_{1.932,156.510} = 5.322$, $P = 0.006$*
K3 (Absorption)	-1.81 ± 0.66	-1.38 ± 0.52[c]	-1.25 ± 0.69[c]	$F_{1.833,148.491} = 18.119$, $P = 0.000$*
K4 (Generation)	-2.16 ± 0.62[e]	-1.89 ± 0.49[e]	-1.72 ± 0.63[e]	$F_{1.805,146.211} = 11.186$, $P = 0.000$*
Ankle Power				
A0 (Absorption)	-0.69 ± 0.35[e]	-6.07 ± 2.21[e]	-1.38 ± 2.14[e]	$F_{1.454,117.771} = 131.502$, $P = 0.000$*
A1 (Absorption)	-1.00 ± 0.40[e]	-0.16 ± 0.50[e]	-0.61 ± 0.35[e]	$F_{1.758,142.411} = 57.003$, $P = 0.000$*
A2 (Generation)	4.07 ± 1.21	2.75 ± 1.00[c]	2.86 ± 1.21[c]	$F_{1.934,156.685} = 21.141$, $P = 0.000$*

* Significant task effect. [a] SON significantly different to NW and SOV; [b] SON significantly different to SOV; [c] NW significantly different to SON and SOV; [d] SOV significantly different to NW and SON; [e] all walking tasks significantly different. *Abbreviations:* Normal Walking (NW), Stepping Onto and Off an Obstacle (SON) and Stepping Over and Obstacle (SOV), Loading Response (LR), Terminal Stance (TS) and Pre-Swing (PSw).

Table 7.6. Correlation between age and walking speed on joint kinetics for all walking tasks, with a partial correlation controlling for walking speed.

		NW	
Parameter	**Age**	**Walking Speed**	**Age Controlling for Walking Speed**
Moments (Nm/kg)			
Hip Flexion/Extension			
Flexion Moment - Stance	-0.024	0.437*	0.083
Extension Moment	0.275*	-0.260*	0.231*
Flexion Moment - Swing	-0.125	0.394*	-0.042
Hip Abduction/Adduction			
Maximum Moment (First Peak)	-0.083	0.108	-0.060
Minimum Moment (First Peak)	0.096	-0.034	0.091
Maximum Moment (Second Peak)	0.145	-0.248*	0.095
Minimum Moment (Second Peak)	0.114	-0.377*	0.033
Knee Flexion/Extension			
Flexion Moment (LR)	-0.182	0.335*	-0.118
Extension Moment (TS)	-0.148	-0.233*	-0.211
Flexion Moment (PSw)	-0.146	0.217*	-0.103
Extension Moment - Swing	0.152	-0.492*	0.050
Knee Varus/Valgus			
Varus Moment (First Peak)	-0.048	0.256*	0.010
Valgus Moment (First Peak)	0.222*	-0.261*	0.174
Varus Moment (Second Peak)	0.033	-0.048	0.023
Valgus Moment (Second Peak)	0.064	-0.272*	0.003
Ankle Plantar/Dorsiflexion			
Plantarflexion moment	0.004	-0.284*	-0.063
Dorsiflexion moment	-0.003	0.111	0.022

		NW	
Parameter	**Age**	**Walking Speed**	**Age Controlling for Walking Speed**
Powers (Watts/kg)			
Hip Power			
H1 (Generation)	-0.010	0.263*	0.051
H2 (Absorption)	0.197	-0.385*	0.124
H3 (Generation)	-0.219*	0.595*	-0.111
Knee Power			
K0 (Generation)	-0.223*	0.440*	-0.143
K1 (Absorption)	0.077	-0.302*	0.011
K2 (Generation)	-0.163	0.556*	-0.049
K3 (Absorption)	0.184	-0.535*	0.079
K4 (Generation)	0.349*	-0.498*	0.282
Ankle Power			
A0 (Absorption)	0.127	-0.313*	0.062
A1 (Absorption)	-0.096	-0.158	-0.136
A2 (Generation)	-0.241*	0.517*	-0.151

		SON	
Parameter	**Age**	**Walking Speed**	**Age Controlling for Walking Speed**
Moments (Nm/kg)			
Hip Flexion/Extension			
Flexion Moment - Stance	0.289*	0.202	0.334*
Extension Moment	0.235*	-0.197	0.209
Flexion Moment - Swing	0.127	0.437*	0.225
Hip Abduction/Adduction			
Maximum Moment (First Peak)	0.159	-0.051	0.153
Minimum Moment (First Peak)	-0.115	0.192	-0.086
Maximum Moment (Second Peak)	0.379*	-0.385*	0.347
Minimum Moment (Second Peak)	0.249*	-0.494*	0.195
Knee Flexion/Extension			
Flexion Moment (LR)	-0.045	-0.007	-0.047
Extension Moment (TS)	0.017	-0.207	-0.018
Flexion Moment (PSw)	0.024	-0.277*	-0.023
Extension Moment - Swing	0.047	-0.416*	-0.024
Knee Varus/Valgus			
Varus Moment (First Peak)	0.347*	0.021	0.356
Valgus Moment (First Peak)	0.154	-0.084	0.142
Varus Moment (Second Peak)	0.155	-0.261*	0.118
Valgus Moment (Second Peak)	0.174	-0.380*	0.122
Ankle Plantar/Dorsiflexion			
Plantarflexion moment	-0.036	0.001	-0.036
Dorsiflexion moment	-0.023	0.233*	0.016

		SON	
Parameter	**Age**	**Walking Speed**	**Age Controlling for Walking Speed**
Powers (Watts/kg)			
Hip Power			
H1 (Generation)	0.225*	0.228*	0.273*
H2 (Absorption)	0.027	-0.288*	-0.022
H3 (Generation)	-0.241*	0.343*	-0.199
Knee Power			
K0 (Generation)	0.003	0.326*	0.061
K1 (Absorption)	-0.138	0.061	-0.130
K2 (Generation)	-0.221*	0.151	-0.201
K3 (Absorption)	0.237*	-0.283*	0.201
K4 (Generation)	0.394*	-0.401*	0.363*
Ankle Power			
A0 (Absorption)	0.039	-0.403*	-0.031
A1 (Absorption)	-0.050	0.273*	-0.005
A2 (Generation)	-0.150	-0.247*	-0.200

	SOV		
Parameter	**Age**	**Walking Speed**	**Age Controlling for Walking Speed**
Moments (Nm/kg)			
Hip Flexion/Extension			
Flexion Moment - Stance	0.033	0.389*	0.152
Extension Moment	0.266*	-0.352*	0.192
Flexion Moment - Swing	0.083	0.470*	0.241*
Hip Abduction/Adduction			
Maximum Moment (First Peak)	0.070	-0.009	0.070
Minimum Moment (First Peak)	0.109	-0.292*	0.035
Maximum Moment (Second Peak)	0.294*	-0.443*	0.206
Minimum Moment (Second Peak)	0.249*	-0.649*	0.108
Knee Flexion/Extension			
Flexion Moment (LR)	-0.084	0.230*	-0.025
Extension Moment (TS)	-0.045	-0.228*	-0.111
Flexion Moment (PSw)	-0.028	0.072	-0.009
Extension Moment - Swing	0.018	-0.551*	-0.157
Knee Varus/Valgus			
Varus Moment (First Peak)	0.103	0.246*	0.179
Valgus Moment (First Peak)	0.130	-0.090	0.111
Varus Moment (Second Peak)	0.159	-0.013	0.161
Valgus Moment (Second Peak)	0.265*	-0.489*	0.163
Ankle Plantar/Dorsiflexion			
Plantarflexion moment	-0.002	-0.152	-0.043
Dorsiflexion moment	0.063	0.174	0.114

		SOV	
Parameter	**Age**	**Walking Speed**	**Age Controlling for Walking Speed**
Powers (Watts/kg)			
Hip Power			
H1 (Generation)	-0.118	0.444*	-0.002
H2 (Absorption)	0.146	-0.344*	0.062
H3 (Generation)	-0.119	0.664*	0.076
Knee Power			
K0 (Generation)	0.010	0.229*	0.074
K1 (Absorption)	0.024	-0.291*	-0.056
K2 (Generation)	-0.153	0.361*	-0.065
K3 (Absorption)	0.179	-0.577*	0.036
K4 (Generation)	0.341*	-0.617*	0.237*
Ankle Power			
A0 (Absorption)	0.107	-0.374*	0.010
A1 (Absorption)	-0.143	-0.140	-0.188
A2 (Generation)	-0.208	0.526*	-0.085

* Significant correlation. *Abbreviations:* Normal Walking (NW), Stepping Onto and Off an Obstacle (SON) and Stepping Over and Obstacle (SOV), Loading Response (LR), Terminal Stance (TS) and Pre-Swing (PSw).

7.4. Discussion

Clearing obstacles is important for functional mobility, in order to maintain independent living. As such, evaluating GRFs can be used to determine the state of locomotion during walking. The aim of this study was to determine the alterations on landing mechanics for obstacle clearance when compared to normal walking in an older adult population. Typically, an increase in task demand was associated with altered spatial-temporal parameters. For example, walking speed was reduced for both obstacle clearance tasks compared to normal walking (NW: 1.43 ± 0.18 $m{\cdot}s^{-1}$, SON: 1.11 ± 0.21 $m{\cdot}s^{-1}$ and SOV: 1.24 ± 0.20 $m{\cdot}s^{-1}$). Walking task effects were found for F1, F3, F4 and F5, with SON revealing an increased first (F1) and reduced second (F3) vertical peak compared to NW (e.g. F1: SON: 1.63 ± 0.21 BwFz vs. normal walking: 1.12 ± 0.10 BwFz and SOV: 1.13 ± 0.13 BwFz). In addition, braking force (F4) was reduced for SON, with an increased propulsive force (F5) for SOV. Age was correlated with braking (F4) and propulsive (F5) force for all walking tasks, however when controlling for walking speed there were only significant correlations for NW and SON walking tasks. As such, there was only a partial acceptance of the hypotheses as age was only associated with propulsive (F5) peak force. In addition, as task complexity increased, vertical GRF only increased for SON not SOV obstacle clearance task, with joint kinetics adaptations mainly occurring for task rather than age. For example, reduced dorsiflexion moment for SOV compared to NW and SON (SOV: 1.17 ± 0.26 Nm/kg vs. NW: 1.29 ± 0.27 Nm/kg and SON: 1.39 ± 0.30 Nm/kg).

GRF profiles were predominantly affected by walking task. An increase in task complexity was associated with an altered GRF profile for both vertical and anterior-posterior forces. First vertical GRF (F1) peak was significantly increased for SON compared to NW and SOV walking tasks (SON: 1.63 ± 0.21 BwFz vs. NW: 1.12 ± 0.10 BwFz and SOV: 1.13 ± 0.13 BwFz), with a reduced second vertical GRF (F3) peak (SON: 0.93 ± 0.16 BwFz, NW: 1.13 ± 0.09 BwFz and SOV: 1.08 ± 0.09 BwFz). This is in agreement with previous findings (Christina and Cavanagh, 2002, Stacoff *et al.*, 2005). In addition, Christina and Cavanagh (2002) reported stair descent was approximately 0.35 body weight higher during first vertical GRF (F1) peak and reduced body weight of approximately 0.15 during second vertical GRF (F3) peak) compared to NW. This study found on average, first vertical GRF (F1) peak had an increased 0.51 body weight and reduced body weight of 0.20 during second vertical GRF (F3) peak compared to NW. Although, this study in comparison to Christina and Cavanagh (2002) have higher body weight values compared to NW, this study had a 40 cm step height (i.e. pavement curb height) whereas the literature used a 18 cm step. In addition, this study did not state body mass of the subjects. Potentially, this study had an increased body mass, which would associate extra loading and greater GRF. Riener *et al.* (2002) investigated step ascent and descent at three step inclinations (24 °, 30 ° and 42 °) and found greater inclination caused an increased vertical ground reaction force (first vertical GRF (F1) peak).

For anterior-posterior GRF, braking force (F4) was significantly different for all walking tasks, with SON having the lowest and SOV having the highest braking force (F4) (SOV: -0.26 ± 0.06 BwFy, NW: -0.19 ± 0.05 BwFy and SON: -0.12 ± 0.03 BwFy). This finding contradicts previous findings (Christina and Cavanagh, 2002, Riener *et al.*, 2002), however these studies compared young to older adults for NW and consecutive step descent walking tasks. For SON, reduced braking force (F4) is associated with reduced friction and as such the likelihood of a slip will increase.

Similar GRF profiles occurred between NW and SOV for vertical force. However, anterior-posterior force was significantly different for braking (F4) and propulsive (F5) force. SOV had a significantly higher braking (F4) and propulsive (F5) force compared to NW (F4 SOV: -0.26 ± 0.06 BwFy vs. NW: -0.19 ± 0.05 BwFy and F5 SOV: 0.27 ± 0.06 BwFy vs. NW: 0.24 ± 0.04 BwFy). This is due to the nature of SOV obstacle task compared to NW, as the increased braking force (F4) reflects the control of landing over the obstacle and reduced momentum, whilst an increased propulsive force (F5) assists the muscle force generation to influence the control of the contralateral limb as it trails over the obstacle (Houser *et al.*, 2008) and may ensure successful foot clearance. In addition, this strategy occurs when obstacle clearing in order to adapt to the low friction footwear-floor interface, to minimise the risk of slips (Patla *et al.*, 1991). This may also be the reason for an increase in arm swing observed for this older adult population during obstacle clearance in Chapter Six.

In addition, when controlling for walking speed, age was significantly correlated to braking (F4) and propulsive (F5) force for NW and SON walking tasks. This suggests braking (F4) and propulsive (F5) force for such tasks are independent of walking speed. As such, there was a partial agreement to previous findings (Christina and Cavanagh, 2002, Toda *et al.*, 2015), as there was no significant correlation on age and vertical GRF. However, no significant correlation on age and SOV were found when controlling for walking speed, as such this task is dependent on walking speed.

Age-related gait adaptations were observed for this older adult population in Chapter Four. The over 75 years age group typically adopted a joint kinetic strategy. For instance, ankle power generation was reduced at toe-off. In late stance the ankle is an important joint to aid propulsion of ipsilateral limb from stance to swing (Neptune *et al.*, 2001). As such, an increase in age was associated with a reduced propulsive which is likely to the consequence of biomechanical changes associated with ageing, for instance a decline in muscle-force generating capacity (Silder *et al.*, 2008). Additionally, in Chapter Five an age effect was present for second maximum toe-clearance for the over 75 years age group. Although, weak dorsiflexor muscles are likely to contribute to this decline in second maximum toe-clearance for this age group. Ankle power generation is associated with weak ankle plantarflexor muscles, as such this muscle weakness may be inhibiting the ankle to achieve full dorsiflexion range at second maximum toe-clearance. Therefore, these ageing factors may have also impacted on braking (F4) and propulsive force (F5), which potentially indirectly caused these altered GRF with age. Menz *et al.* (2008) reported older adults who had fallen, had decreased ankle flexibility and toe plantarflexor strength and also reduced plantar tactile sensitivity and hallux valgus deformity. Consequently, these gait adaptations which coincide with toe-off may be an indicator of fall risk. Future work should be considered to investigate this older adult population in a longitudinal design to determine the ageing effects on gait and also incorporate centre of mass and centre of pressure to explore fall risk for these walking tasks.

For the obstacle clearance tasks GRF profiles were distinguishable. For example, SON had an increased vertical first peak force (F1) compared to SOV (1.63 ± 0.21 BwFz and 1.13 ± 0.13 BwFz) which is due to step descent height of the obstacle. As such, this higher force magnitude increases the dissipated load on the musculoskeletal system to shock absorb and distribute the force. This may explain why there is increased knee (K1) and ankle power (A0) absorption for this task (e.g. ankle power absorption (A0) -6.07 ± 2.21 Watts/kg vs. -1.38 ± 2.14 Watts/kg). An increased force magnitude increases the risk of joint injury and pathologies (Dufek and Bates, 1990, McNitt-Gray, 1991, Irmischer *et al.*, 2004, Elvin *et al.*, 2007). Furthermore, SON had a reduced braking force (F4) compared to SOV which suggests older adults performing such a task have an increased slip occurrence risk.

For SOV obstacle clearance task, older adults typically adopted a hip (flexion/extension and ab/adduction) and knee (flexion) joint moment strategy with hip power generation (H1) and knee and ankle powers, whereas SON adopted hip abd/adduction and ankle moments, knee powers (all except K2 generation) and all ankle powers strategy. These strategies employed during SON and SOV obstacle clearance tasks have also been reported in the literature for young adults and suggested to aid safe obstacle clearance (Patla and Prentice, 1995, Niang and McFadyen, 2004, MacLellan and Patla, 2006). Increased knee power aids toe elevation of the contralateral limb (Patla and Prentice, 1995, Niang and McFadyen, 2004). For SON, knee power (K3 generation) increased, whereas SOV knee power reduced yet older adults employed an increased hip adduction moment which may have aided toe elevation and allowed for safe clearance of the obstacle for the trailing limb. The joint kinetic adaptation for SOV has also been observed in lower limb amputees (Hill *et al.*, 1999), to employ a hip strategy when knee power is reduced. For this older adult population, similar findings were found in Chapter four; older adults employed a hip kinematic strategy to achieve successful toe-clearance during NW and manual dual task walking.

In addition, a cautious gait strategy was employed as task complexity increased for this older adult population, for example walking speed declined with increase in task complexity (NW: 1.43 ± 0.18 $m \cdot s^{-1}$, SOV: 1.24 ± 0.20 $m \cdot s^{-1}$ and SON: 1.11 ± 0.21 $m \cdot s^{-1}$). Compensatory strategies such as reduced walking speed, step length and increased step width occur when walking stability is challenged (Hollman *et al.*, 2007). In addition, reduced joint moments and power generations at the ankle for obstacle clearance tasks and accompanied with reduced braking force (F4) for SON are also indicative of cautious gait. This conservative gait pattern is typical strategy adopted by older adults for step clearance (Simoneau *et al.*, 1991, Christina and Cavanagh, 2002). Obstacle tasks are more likely to place higher demands on balance which necessitates much higher conscious control in older adults compared to NW (Deshpande *et al.*, 2009). This compensation mechanism may be employed by this older adult population to reduce slip risk during step descent (i.e. SON walking task).

The main limitation to this study was the same as Chapter Six. This study had a problem with technical limitations of the laboratory, which was caused by room ceiling for example. In addition, only seven motion capture cameras at 2 megapixels were available to track full body movement, as such there was not enough cameras to allow complete full body marker tracking, especially during the obstacle clearance tasks when the obstacle causes camera occlusions. Therefore, the extent of age effect during landing forces remains unknown for this older adult population. Furthermore, GRF profiles were assessed using single limb contact, as participants self-selected limb force plate contact which is typical of peak GRF research (McCrory *et al.*, 2001, Stacoff *et al.*, 2005, Toda *et al.*, 2015). In addition, obstacle clearance requires interdependent control of both the leading and trailing limb (Bovonsunthonchai *et al.*, 2015). As such, future work should investigate the motor and biomechanical control of landing forces for both the leading and trailing limb to determine the effect for older adults.

7.5. <u>Conclusion</u>

Older adults in this population typically employed a gait strategy to compensate for task by altering joint kinetics, reducing walking speed, step length and increasing base of support. An increase in age was associated with a reduced braking and propulsive force for normal walking and obstacle clearance, which is potentially due to reduced power generation, which could impact on toe-clearance and consequently this population compensates by increasing their arm swing when task complexity increases. An increase in task demand was associated with altered ground reaction forces for vertical and anterior-posterior forces. Typically, the obstacle clearance stepping onto and off and obstacle task illustrated more differences in ground reaction forces and joint kinetics in comparison to normal walking, than normal walking and stepping over an obstacle. For example, stepping onto and off an obstacle illustrated a greater first vertical peak force, with a reduced second vertical peak force compared to normal walking and stepping over an obstacle. In addition, braking force was significantly lower for stepping onto and off an obstacle. Therefore, such a task places higher demands on balance and increases the likelihood of slip occurrence. This may be a potential risk to the older aged adults in this population when performing such a task, as age was significantly correlated to braking force when controlling for walking speed.

Chapter Eight: Discussion

The overall aim of this thesis was to explore the effects of age on gait and functional movement characteristics in community-dwelling older adults, as the majority of previous research had compared young adults to older adults, thus disregarding the ageing process and assuming older adults can be categorised into a single age group. As such, the extent of the age effect on gait functionality within older adults was unknown. Four objectives of this thesis were addressed. Aims and key findings for Chapter 4-7 are presented in Table 8.1.

1. Create a normative gait database for an older adult population.

In order to explore the overall aim, a gait database was established to determine the normative effects of age on walking for a community-dwelling older adult population (aged 55 years and above). It was identified in Chapter One, that physical functionality illustrates the ability to perform everyday tasks (Cooper *et al.*, 2011b) and as such walking is not limited to straight-line gait for example, it also can indicate walking with an additional task. Consequently, a gait database was created for five walking tasks (normal walking with and without force plate contact, manual dual task walking, stepping onto and off an obstacle and stepping over an obstacle) for one-hundred and fifty-eight community-dwelling older adults, age range 55 to 86 years (65.7 ± 6.8 years).

This gait database poses similar traits to longitudinal ageing studies, as they also designed to address the current and emerging associations to the age process in a particular geographical location, for example community-dwelling older adults in Herefordshire, England (Martin *et al.*, 2008). However, unlike longitudinal ageing studies, this database is extremely novel due to the research design. Firstly, longitudinal ageing studies such as the English Longitudinal Ageing Study (ELSA, 2016) have been heavily reliant on spatial-temporal parameters (e.g. walking speed) when assessing gait. Although, this poses benefits in terms of high volume of participant recruitment and data capture, spatial-temporal parameters are not sufficient to identify biomechanical mechanisms associated with gait. Whereas, utilising three-dimensional motion analysis allows for joint kinematic and kinetic analysis, which is advantageous in potentially illustrating biomechanical mechanisms and identifying what is ‘normal’ gait for older adults. The Baltimore Longitudinal Aging Study is the current known database which includes three-dimensional motion analysis (Ko *et al.*, 2011, Jerome *et al.*, 2015). Again, this database is limited to either sub-sampling their population, measuring walking speed or reporting a particular parameter for example mechanical work expenditure (Ko *et al.*, 2010, Ko *et al.*, 2011, Jerome *et al.*, 2015).

As such, the joint kinematic and kinetics profiles that are known for children and young adults are not established for older adults. However, this has now been established for normal gait. It is clearly evident from this thesis, global measures such as walking speed are not sufficient to explore the ageing effect. As these baseline measurements do not take into account biomechanical mechanisms which are influenced by ageing. For example, using three-dimensional motion analysis for this older adult population, allowed for the identification of reduced propulsive force with age, which was associated with reduced muscle power generation during walking, caused by reduce muscle strength in the over 75 years age group. As such, current data collection protocols for longitudinal ageing studies do not capture the biomechanics of gait and therefore within their databases are unable to determine the gait changes which may be influencing reduced walking speed in ageing, for instance.

Consequently, this is one of the largest databases for older adult gait and this database represents a normative gait database which could be used as a clinical tool to compare to older adults who are prone to falling or older adults with osteoarthritis for example. This database also highlights that older adults within this population are relatively healthy and high-functioning. Therefore, future work could adopt a longitudinal design to establish the ageing process for this population. This potentially may highlight where the ageing process causes age-related gait adaptations for this population, as currently this cross-sectional design illustrates the age effect occurs at 75 years. However, using a longitudinal design this could be pinpointed to determine which age this typically occurs for this healthy population and may identify potential gait markers which may illustrate adverse ageing effects. For example, performing a manual dual task with reduce hip range of motion on the ipsilateral limb may predict fall risk.

2. Describe normal gait in older adults.

The literature suggested older adults exhibit age-related changes at the hip and ankle joint during normal walking. Chapter Four identified the effects on age during normal walking occurred from the age of 75 years and above for this older adult population. Unlike previous research, this chapter illustrated older adults in this age group altered their gait pattern with a hip joint strategy which predominantly exhibited joint kinetic alterations. There was a reduced hip extension torque and power generation in late stance for the older adults aged 75 years and above. There were no differences for normal walking joint kinematic or kinetics between older adults aged 55-64 years and 65-74 years.

3. Explore the effects of age and/or walking speed on gait and functional walking tasks.

Age does effect gait. As described above, no significant age effect was found for normal walking between the 55-65 years and 65-74 years. Consequently, age effect shifted, which suggests for this population the ageing effect occurs from the age of 75 years for normal walking. In addition, for this age group alterations typically occurred for joint kinetics. Joint kinetics have found to be associated with walking speed. The over 75 years age group, did display a 'cautious gait' pattern (e.g. reduced walking speed and step length). Consequently, this reduction in walking speed may have been a consequence of reduced muscle strength as a result of ageing as oppose to walking speed causing the alteration in joint kinetics for this age group. Although, age was not significantly correlated with joint kinematics or kinetics when controlling for walking speed. In addition, Chapter Six found no association between age and arm swing when performing various walking tasks. These walking tasks were found to be dependent on walking speed for arm swing. For example, a decrease in walking speed was found to increase arm swing for older adults during obstacle clearance walking tasks.

Whereas, for Chapter Five walking speed for normal and manual dual task walking were similar. This suggests the reduction in toe-clearance parameters displayed during dual task walking were independent of walking speed. Although, age was not correlated with minimum toe-clearance, there was an age association with second maximum toe-clearance. The over 75 years age group had a significantly reduced second maximum toe-clearance. This toe-clearance peak occurs when the foot reaches maximum dorsiflexion (Winter, 1991). It was found that an increase in age was associated with reduced second maximum toe-clearance which suggests weak dorsiflexor muscles for this population. For dual task walking compared to normal walking, the ipsilateral limb had an increased hip adduction, knee flexion and ankle dorsiflexion. This strategy may have been employed in this population to ensure successful toe-clearance of the ipsilateral limb to compensate for potentially weak dorsiflexor muscles. Also, Chapter Seven found braking and propulsive force were significantly correlated to age for normal walking and stepping onto and off an obstacle, when controlling for walking speed.

Consequently, age is associated with gait changes for older adults when performing various walking tasks, with gait parameters such as toe-clearance illustrated to be independent of walking speed. However, this thesis also highlights the importance of measuring walking speed, as throughout this thesis gait parameters have found to be independent of walking speed. In the literature, it has been reported there are age-related gait adaptations when walking. However, were these changes due to age, when comparing young to older adults. For instance, is the change in walking speed a result of the nature of task or a result of ageing musculoskeletal decline (Faulkner *et al.*, 2007, Snijders *et al.*, 2007) for instance reduced muscle-force generation capacity.

4. Identify whether changes to gait in older adults are a consequence of age and/or task complexity.

Chapter Five revealed all toe-clearance events were significantly lower for manual dual task when compared to normal walking. Suggesting such a task could increase the likelihood of a trip for this population. Although age was not associated to minimum toe-clearance, there was a negative correlation with second maximum toe-clearance and significant age effect for the over 75 years age group. Nevertheless, changes in toe-clearance parameters were primarily due to task. Similarly, Chapter Six found no age association with arm swing or forearm swing for any walking tasks. An increase in task complexity found an increase in arm swing, although this may have been a compensatory mechanism of this population as a consequence of reduced walking speed for the obstacle clearance tasks. Whereas, Chapter Seven found both task effects and age association on ground reaction force. Increase in walking task demand was associated with an altered ground reaction forces. Older adults in this population typically employed a gait strategy to compensate for task by altering joint kinetics, reducing walking speed and step length and increasing base of support. Braking force was significantly lower for stepping onto and off an obstacle. Therefore, such a task places higher demands on balance and increases the likelihood of slip occurrence. This may be a potential risk to the older aged adults in this population when performing such a task, as age was significantly correlated to braking force when controlling for walking speed.

Table 8.1. Thesis map outlining Chapter aims and key findings.

Chapter	Chapter Aims	Key Findings
4	• The aim of this study was to examine the effects of age on gait parameters within an older adult population.	• No significant differences were found between the 55-64 years and 65-74 years age groups. • Reduced walking speed, stride/step length and a slower timed up and go was present for older adults aged 75 years and over. • Hip extension range of motion was reduced during late stance, with a reduced hip extension torque and power generation in late stance, with reduced knee power generation and absorption for the 75 years and older age group. • When controlling for walking speed, age was not significantly correlated to joint kinematics and kinetics, except knee valgus moment (second peak).
5	• The aim of this study was to establish if toe-clearance events decreased with age and task and if the joint kinematics of the ipsilateral and contralateral limb adapt to performing a dual task. • A secondary aim was to determine if fall history affected toe-clearance parameters.	• Age was not significantly correlated with minimum toe-clearance. • Age was negatively correlated with second maximum toe-clearance. • The over 75 years age group had a significantly reduced second maximum toe-clearance compared to 55-64 years and 65-74 years. • All toe-clearance events were significantly lower for manual dual task compared to normal walking. • There were significant differences between manual dual task and normal walking in joint kinematics at the toe-clearance events.

Chapter	Chapter Aims	Key Findings
6	• The aim of this study was to explore the effect of walking task on arm swing for an older adult population. • The secondary aim of the study was to establish if walking task effected forearm swing for the older adult population.	• Age did not influence arm swing or forearm swing amplitude. • Walking task affected arm swing amplitude. • For the dominant arm, stepping over an obstacle had an increased arm swing compared to normal and dual task walking and stepping onto and off an obstacle. For example, stepping over an obstacle: 23.58 ± 7.97 ° compared to normal walking: 20.89 ± 7.89 ° and stepping onto and off an obstacle: 21.12 ± 7.62 °. • For the non-dominant arm, obstacle clearance task had a significantly increased arm swing compared to normal walking (e.g. stepping onto and off an obstacle: 24.87 ± 8.11° stepping over an obstacle: 25.96 ± 7.85 ° vs. normal walking: 22.21 ± 6.23 °). • Forearm swing was not affected by walking task. • All walking tasks were found to be asymmetrical.

Chapter	Chapter Aims	Key Findings
7	• The aim of this study was to determine the alterations on landing mechanics and joint kinetics for obstacle clearance when compared to normal walking in an older adult population.	• Age was significantly correlated to braking and propulsive force for normal walking and stepping onto and off an obstacle, when controlling for walking speed. • Task effect was found for all spatial-temporal parameters except double-support time. For example, stepping onto and off an obstacle illustrated a reduced stride length and increased step width compared to normal walking and stepping over and obstacle. • With walking speed significantly reduced for both obstacle clearance tasks compared to normal walking (normal walking: 1.43 ± 0.18 $m \cdot s^{-1}$, stepping onto and off an obstacle: 1.11 ± 0.21 $m \cdot s^{-1}$ and stepping over an obstacle: 1.24 ± 0.20 $m \cdot s^{-1}$). • There was also a significant task effect for F1, F3, F4 and F5, with stepping onto and off an obstacle having an increased first and reduced second vertical peak compared to normal walking and stepping over an obstacle (e.g. F1: stepping onto and off an obstacle clearance: 1.63 ± 0.21 BwFz vs. normal walking: 1.12 ± 0.10 BwFz and stepping over an obstacle: 1.13 ± 0.13 BwFz). • Joint kinetics illustrated altered hip moments and knee and ankle power for both obstacle clearance walking tasks. For example, ankle power generation reduced for both obstacle clearance tasks compared to normal walking.

8.1. Thesis Limitations and Future Research

Sample Size

The thesis was limited by the low recruitment size for the over 75 years age group. Attending the University may have posed a barrier for this age group. This may be one of the reasons why no ankle joint range of motion was found for this age group, because of the low statistical power. As such, future work should explore the effects of gait and functional movement characteristics on this age, as this was the age range changes in walking typically occurred.

Data Protocol

This thesis was limited by protocol design as muscle strength measurements for the lower extremities were not collected. As such, these observed walking speed reductions and joint power generation were only assumed to be influenced by ageing musculoskeletal decline (e.g. reduced muscle strength). Therefore, future data collection within this database should consider measuring lower limb strength using an isokinetic dynamometry machine. However, data collection protocol may have to be altered to minimise the likelihood of fatigue.

Group Analysis

Observed gait changes within this thesis could either affect or effect fall risk for this older adult population. Yet, only Chapter Five explored fall history for this older adult population. It is worth noting fall history did not affect toe-clearance for this older adult population. In addition, the Timed Up and Go is typically used for fall-screening in a clinical setting (Panel on Prevention of Falls in Older Persons and British Geriatrics, 2011, Barry *et al.*, 2014). Older adults are only classified as fall prone if the Timed Up and Go time is equal to or more than 13.5 seconds. The highest Timed Up and Go time was for the over 75 years age group (9.0 ± 1.8 s), as such this older adult population were not classified as fallers.

Technical Limitations

The main limitation for this thesis was the technical limitations of the biomechanics laboratory. Similar, to observations in the pilot study with the turning task, excessive marker trajectory gaps occurred for the obstacle clearance tasks this caused excessive participant exclusions for Chapter Six and Seven. This was due to the low ceiling height of the laboratory and a beam which is run horizontally across the room. Consequently, the mounting of the three-dimensional motion capture cameras was positioned under the rig as oppose to the top to avoid ceiling beam and occlusions. As a result, vertical field of view for full body marker tracking becomes difficult. Furthermore, the biomechanics laboratory is limited to seven motion capture cameras which means the field of view off all cameras was limited and this was the main reason only one gait cycle was captured for all walking tasks.

Whole Body Analysis

Although, joint kinematic and kinetic analysis identified an age effect on gait for this older adult population, the thesis was limited for not exploring the effects of age on centre of mass and pressure. For instance, the leading limb assists forward progression of the trailing limb through vertical support and mediolateral shift of the centre of mass (Hernández *et al.*, 2009). Consequently, the control of mediolateral accelerations during mid-terminal stance (i.e. transition from single to double support) may be an important age-related factor. Research (Winter, 1995) suggested these age-related reductions in mediolateral centre of mass acceleration during push-off were attributed to the muscle potential of the frontal plane, for example hip adductors/abductors. Therefore, an increase in age resulted in reduced braking and propulsion, which associated with reduced joint powers as a result arm swing increased with an increase in task complexity to aid walking stability and to ensure successful toe-clearance. Although, ageing factors such as reduced muscle strength may have impacted on this biomechanical change, control during these walking tasks were not investigated. Future work should be considered to explore centre of mass control and centre of pressure, to determine if an increased age reduces control.

8.2. Conclusion

The work of this thesis has highlighted age-related gait adaptations can be identified when exploring within an older adult population. This thesis presents a large dataset for gait parameters for community-dwelling older adult population, not only for normal walking but also increased task complexity (e.g. manual dual task walking), indicating mechanisms of gait. Unlike other large databases, for instance longitudinal ageing studies (e.g. English Longitudinal Study of Ageing) this study was not limited by global measures (i.e. spatial-temporal parameters – walking speed). As such, this gait database for this thesis is novel because gait for various walking tasks were captured using three-dimensional motion analysis. Age does effect gait and functional movement characteristics within this older adult population. Normal walking and toe-clearance were not affected by age for older adults aged 55-64 years and 65-74 years. Therefore, gait seems relatively stable up to the age of 74 years for this older adult population. Consequently, age effect has shifted to 75 years and above. For example, the over 75 years age group adopted a joint kinetic strategy (e.g. reduced hip extension moment in terminal stance) and altered spatial-temporal parameters (e.g. reduced walking speed) during normal walking. Furthermore, a reduction in braking and propulsive forces with an increased in age, is linked to a reduction in joint power generation which may impact on the effectiveness of toe-off and limb advancement during walking. Therefore, arm swing increases with task complexity to aid forward momentum and potentially increase walking stability, in order to achieve successful toe-clearance. These gait changes are associated with task complexity and also the consequences of ageing, for instance reduced muscle strength. For example, a reduced ankle plantarflexor generation at toe-off with age may increase the likelihood of a trip. Consequently, future work is required to determine the ageing effect within this older adult population using a longitudinal design. This thesis highlights the potential for using such a task when evaluating functionality for older adults. Walking tasks which compromise and place higher demands on balance may increase the likelihood of a fall. As such, clinicians may consider using similar walking task protocols for assessing gait mechanisms for older adults who are prone to falling for example.

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Appendix Contents

Appendix One: Plug-in Gait Marker Model

A1.1. Marker Model Assumptions

Modelling Approach

To execute Plug-in Gait Marker Model (PiG) four assumptions must be met (Vicon, 2010): **1)** the minimum required markers are the pelvis (for the lower body) and thorax (for the upper body). **2)** Static values of each walking trial are required to calculate the defined segments. **3)** Rigid segment positions are defined frame by frame for each walking trial. To define a segment, the origin of the global laboratory coordinates and three orthogonal axis directions are used. These are identified from two directions of the marker data using the Cartesian right-hand coordinate system (Pennock and Clark, 1990, Rivest, 2005): 1 – dominant direction; establishes the axes in each segment, 2 – subordinate direction which defines the plane and 3 – axis of the segment directly perpendicular to each plane (Figure A1.1.). **4)** Once all segments are defined, model outputs (kinematics and kinetics) are calculated using frame by frame positions of the segments for each walking trial.

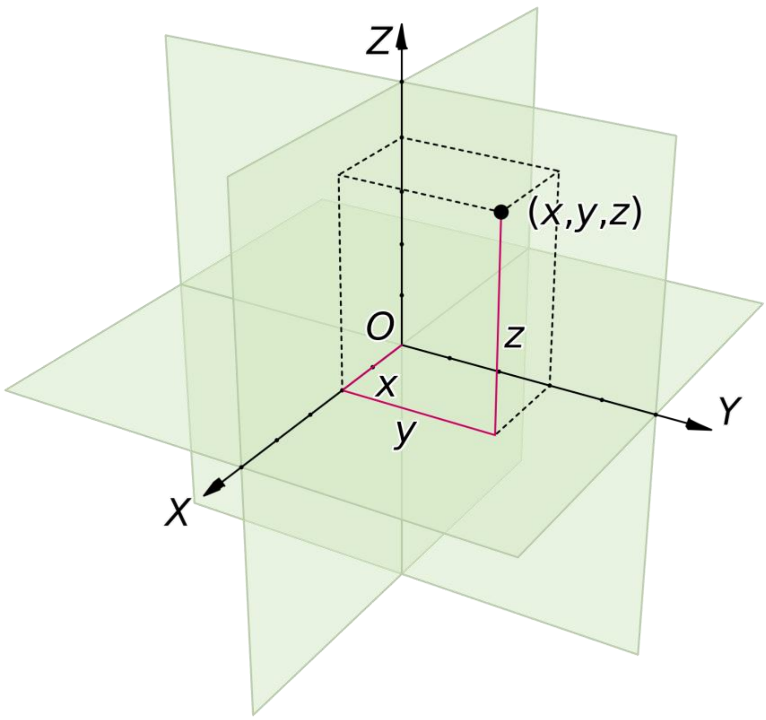


Figure A1.1. Cartesian right-handed coordinate system.

Chord Function

Chord function defines the joint centres using three assumptions to define a plane: **1)** joint centre has previously been calculated, **2)** acquires data from a known marker position and **3)** acquires data from a known marker which is perpendicular to the joint centre to calculate joint centre offset (Figure A1.2.).

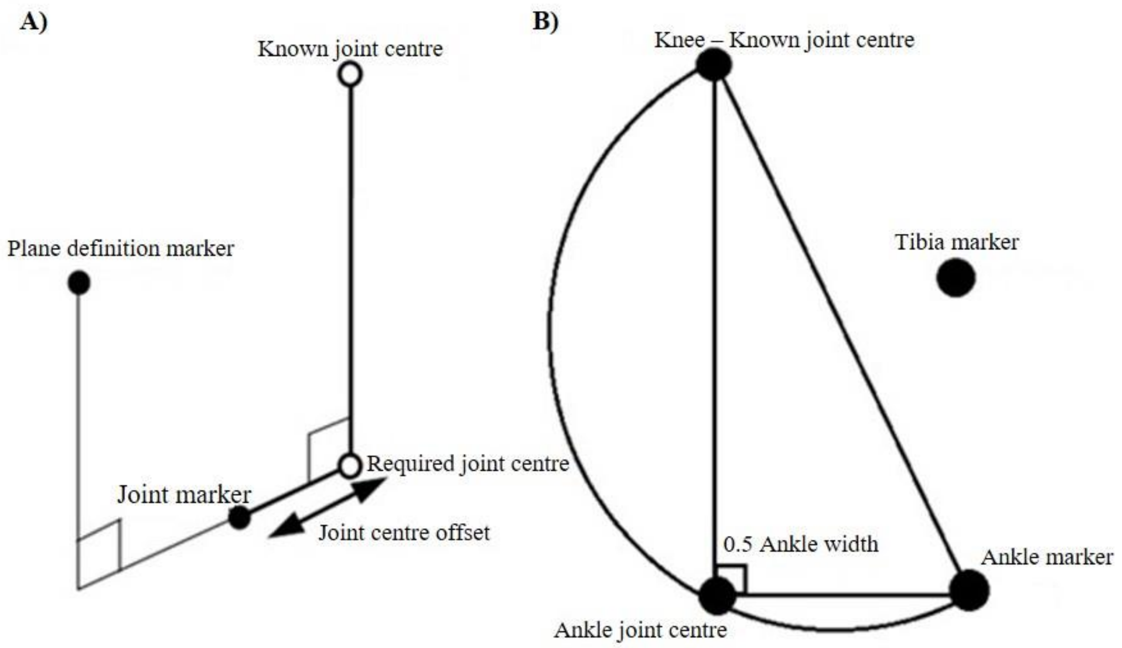


Figure A1.2. Chord function: **a)** the three points used to define a plane and **b)** example of a chord function for a lower body segment.

Fixed Values

Upper body anthropometric offset values are calculated from the measured values (anthropometric measurements) and the marker diameter using this equation:

$$Shoulder\ Offset\ (mm) = measured\ shoulder\ offset + (\frac{1}{2} marker\ diameter)$$

$$Elbow\ Offset\ (mm) = elbow\ width + \left(\frac{marker\ diameter}{2}\right)$$

$$Wrist\ Offset\ (mm) = wrist\ width + (\frac{marker\ diameter}{2})$$

$$Hand\ Offset\ (mm) = hand\ thickness + (\frac{marker\ diameter}{2})$$

Using Dempster's data (Dempster, 1955) the position of the fifth lumbar vertebrae (L5) is found, to allow segment inertia properties to be calculated and estimate whole body centre of mass. L5 is estimated using the following equation:

$$L5 = \frac{(LHJC + RHJC)}{2} + (0.0,0.0,0.828) * length(LHJC - RHJC)$$

0.828 is the ratio of the distance from the hip joint centre to the position of L5 compared to the distance between the hip joint centres of the pelvis segment. LHJC is the left hip joint centre and RHJC is the right hip joint centre.

Lower Body Model

Pelvis Segment

Pelvis origin is the midpoint between the two ASIS markers. The pelvis segment (Figure A1.3.) is defined using: **1)** dominant axis – Y axis derived from the right ASIS marker to the left ASIS marker, **2)** secondary direction – using the mean of the two PSIS markers. The scale and position of the pelvis is established by the two ASIS markers, with the PSIS markers determining anterior tilt. Accuracy is required of the ASIS markers as the positions affect the calculations of the femur segments which can impact both hip and knee joint angles.

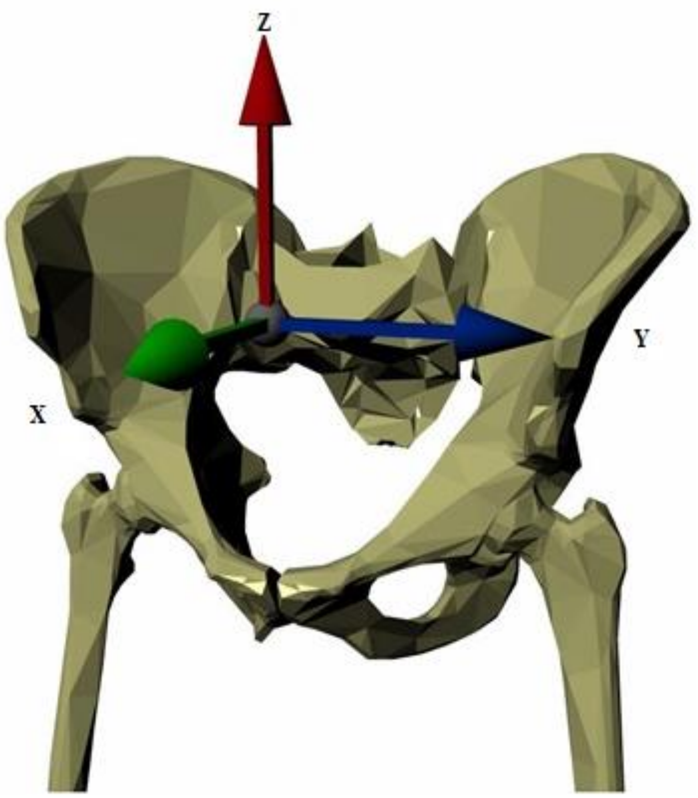


Figure A1.3. Pelvis segment displayed using the Cartesian right-hand coordinate system, with the pelvis origin indicated in the middle of the pelvis (Created using Visual 3D, v. 4.91, Philadelphia, USA).

Hip Joint Centres

The positions of the hip joint centres in the pelvis segment are defined using the Newington-Gage model (Davis *et al.*, 1991). The ASIS markers calculate the inter-ASIS distance, which determine the perpendicular positions of the hip joint centres within the pelvis segment. The calculated ASIS-trochanter distance for the right and left is subsequently used to calculate the right and left hip joint centres using the following equations:

$$C = Mean\ leg\ length * 0.115 - 15.3$$

$$X = C * \cos(theta) * \sin(beta) - (ASISTrocDist + mm) * \sin(beta)$$

$$Y = -(C * \sin(theta) - aa)$$

$$Z = -C * \cos(theta) * \cos(beta) - (ASISTrocDist + mm) * \sin(beta)$$

The value of C calculates the offset vectors for the two hip joint centres, with the right hip joint centre having a negated Y offset. 0.5 radians is theta, 0.314 radians is beta, ASIS-trochanter distance (in mm) is ASISTrocDist, marker radius is mm and half the inter-ASIS distance is aa.

Knee Joint Centres

The knee joint centre is calculated from a modified chord function for the walking trials (Figure A1.2.). For static trials, the anterior-posterior position of the knee joint centre is calculated from the position of the thigh marker with the value of thigh marker offset is zero. An accurate calculation of the knee joint centre is vital for correct kinetic modelling.

Femur Segment

The origin of the femur is at the knee joint centre. The defined femur segment (Figure A1.4.) uses: **1)** dominant axis – Z axis derived from the knee joint centre to the hip joint centre which defines the lateral orientation of the femur, **2)** secondary axis – Y axis; knee joint centre to the knee marker and **3)** X axis – femur direct anteriorly from the knee.

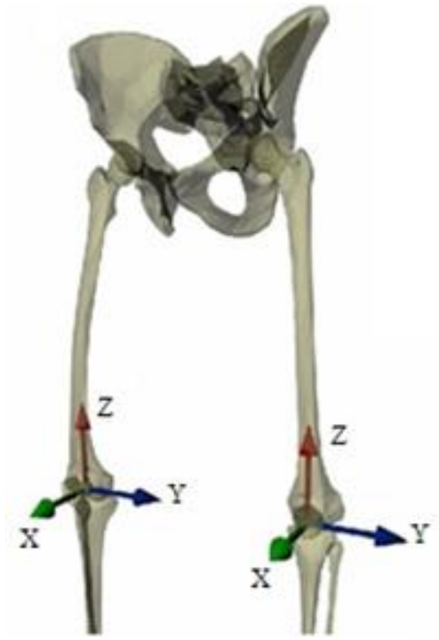


Figure A1.4. Femur segment displayed with the Cartesian right-hand coordinate system at the origin of the femur (Created using Visual 3D, v. 4.91, Philadelphia, USA).

Ankle Joint Centre

The ankle joint centre is calculated using the modified chord function for both static and walking trials (Figure A1.2.).

Tibia Segment

The model creates two tibias; torsioned and untorsioned. The torsioned tibia is defined as: **1)** origin at the ankle joint centre, **2)** X axis – forward direction at the distal end of the tibia **3)** Y axis – between the ankle joint centre and ankle marker and **4)** Z axis – in direction from the ankle joint centre to the knee joint centre, with tibial rotation offset determined by the static trial. The untorsioned tibia is determined by rotating the x and y axes of the torsioned tibia about the z axis using the negative tibial torsion, representing the proximal end and is used to calculate knee joint angles.

Shank Segment

The shank segment (Figure A1.5.) was determined by: **1)** the joining of the ankle and knee joint centres, **2)** the ankle marker passing through the ankle joint centre equally to half ankle width and marker diameter at the lateral knee epicondyle. These two axes are in the plane formed by the knee joint centre and the tibia and ankle markers, with the third axis being perpendicular.

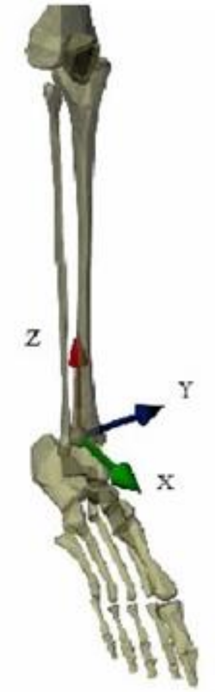


Figure A1.5. Shank segment displayed with the Cartesian right-hand coordinate system (Created using Visual 3D, v. 4.91, Philadelphia, USA).

Foot Segments

Two-foot segments are constructed using the ankle joint as the origin. The first foot segment (Figure A1.6.) uses the Z axis as the primary axis, which is the line between the toe and heel marker and Y axis (untorsioned tibia) defines the secondary Y axis.

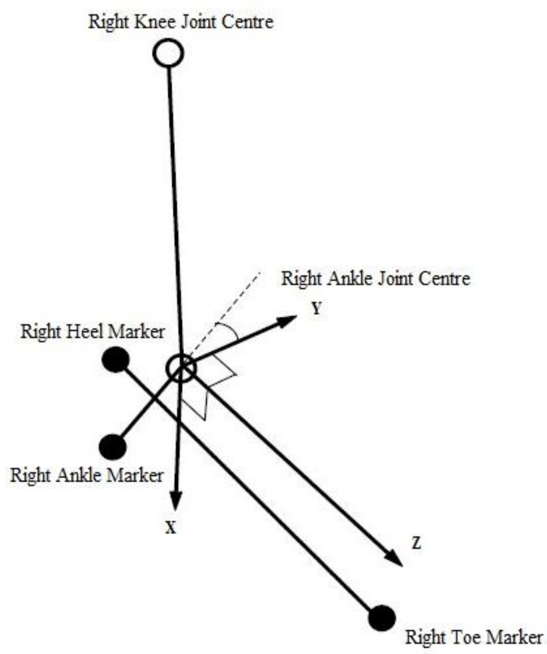


Figure A1.6. The first foot segment (Vicon, 2010).

The second foot segment (Figure A1.7.) uses the Z axis, which is the toe marker to ankle joint centre as the primary, with the Y axis of the untorsioned tibia to define the foot X and the Y axis.

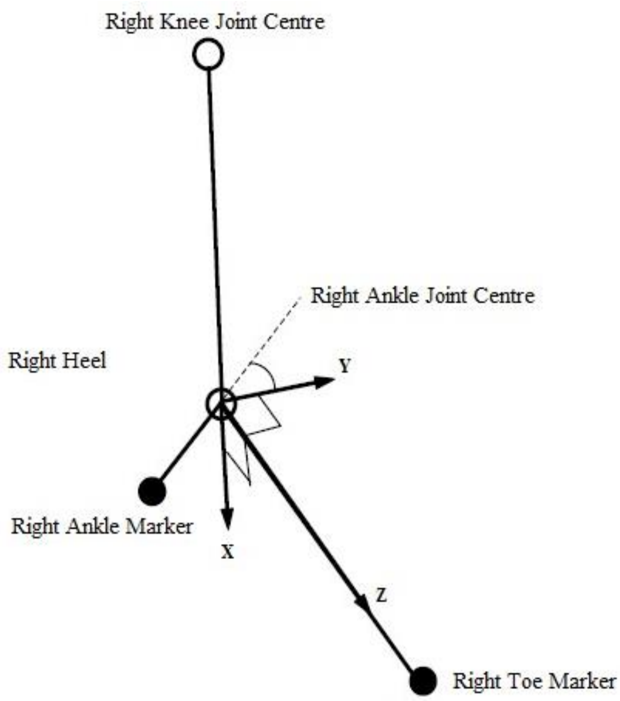


Figure A1.7. Second foot segment (Vicon, 2010).

Plantarflexion offset and rotation offset are then calculated in the static trial from the Cartesian coordinates between the two-foot segments (Figure A1.6. and A1.7.). Static plantarflexion offset occurs from the rotation in the Y axis, with the rotation offset occurring in the X axis (Figure A1.8.). This angle is calculated between the heel and toe marker for plantarflexion offset and ankle joint centre and toe marker for rotation offset. Static foot rotation with a positive value corresponds to an internal rotated foot vector and if the heel and toe markers are the same height the foot rotation axis is vertical. For the walking trials, the foot segment is determined as the equivalent process of defining the second segment in the static trial, then the plantarflexion offset and rotation offset are calculated (Figure A1.8.).

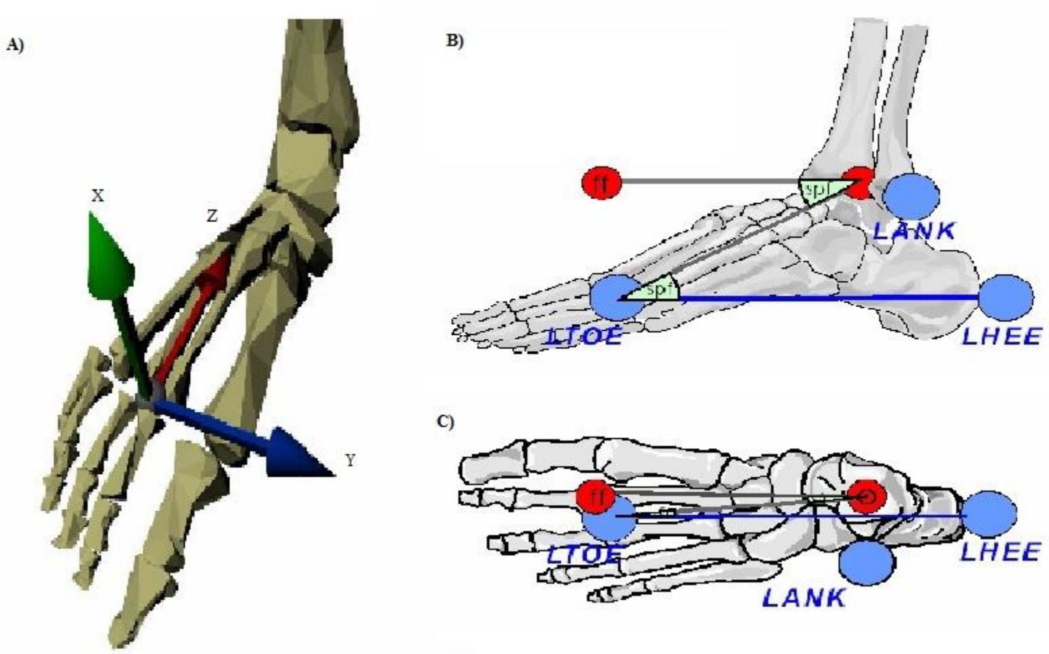


Figure A1.8. A) Foot segment displayed with the Cartesian coordinate system (Created using Visual 3D, v. 4.91, Philadelphia, USA), **B)** Static plantarflexion offset (SPF) angle in a flatfoot position (FF) and **C)** Static rotation offset angle in a flatfoot position (FF). *Note:* **B)** and **C)** process of calculating the rotation offset angle is the same for the walking trials (Vicon, 2010). *Abbreviations:* Left Toe (LTOE), Left Ankle (LANK) and Left Heel (LHEE).

Upper Body Model

Head Segment

The origin of the head (Figure A1.9.) is defined between the midpoint of the left and right front head markers. The midpoint between the left and right back marker is calculated, with the left and right side of the head calculated from both the head origin and midpoint of the back of the head. The X axis is the predominant axis, which is defined anterior-posterior in anterior direction. The secondary Y axis is medial-lateral axis from right to left. For the static trial, the Cartesian coordinate system of the head segment is calculated to the global laboratory coordinates, with Y axis rotation represented as the head offset angle. However, for the walking trials, head offset angle is rotated in the Y axis of the defined head segment.

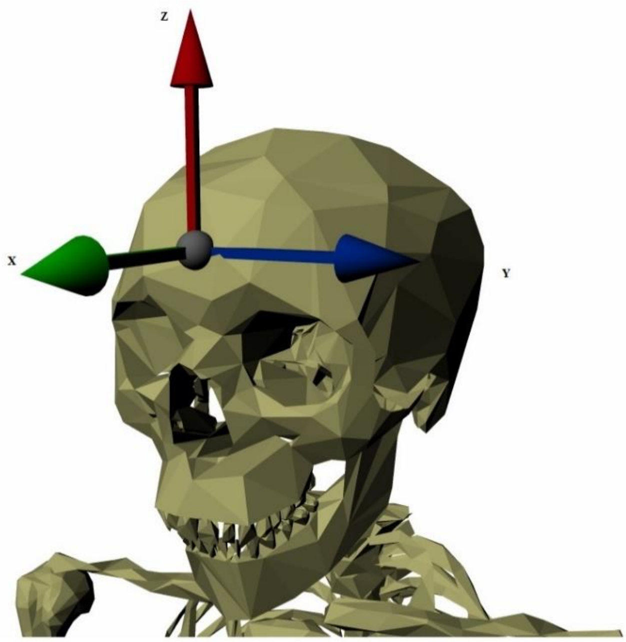


Figure A1.9. The defined head segment displayed with the Cartesian coordinate system (Created using Visual 3D, v. 4.91, Philadelphia, USA).

Thorax Segment

Orientation of the thorax is calculated first. The Z axis is the predominant axis – direction from the midpoint of the clavicle marker and C7 marker (7^{th} Cervical Vertebrae) to the midpoint of the sternum marker and T10 marker (10th Thoracic Vertebrae). The secondary direction is the X axis – midpoint of the C7 marker and T10 marker to the midpoint of the clavicle marker and sternum marker. The thorax is calculated from the clavicle marker, with backwards offset of half a marker diameter in the X axis (Figure A1.10.).

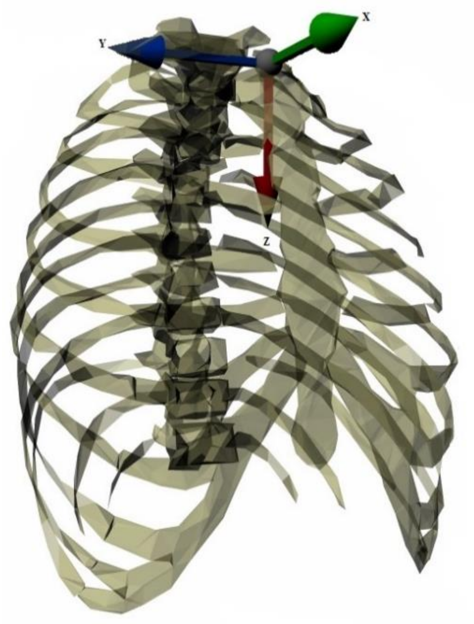


Figure A1.10. Thorax segment displayed with the Cartesian coordinate system (Created using Visual 3D, v. 4.91, Philadelphia, USA).

<u>Shoulder Joint Centre</u>

The clavicles are between the thorax origin and the shoulder joint centres, with the shoulder joint centres defined as the origins for each clavicle. A direction is defined perpendicular to the line of the thorax origin to the shoulder marker and thorax X axis. This direction is used to define the virtual shoulder marker. A chord function (Figure A1.2.) is then used to define the shoulder joint centre (Figure A1.11.).

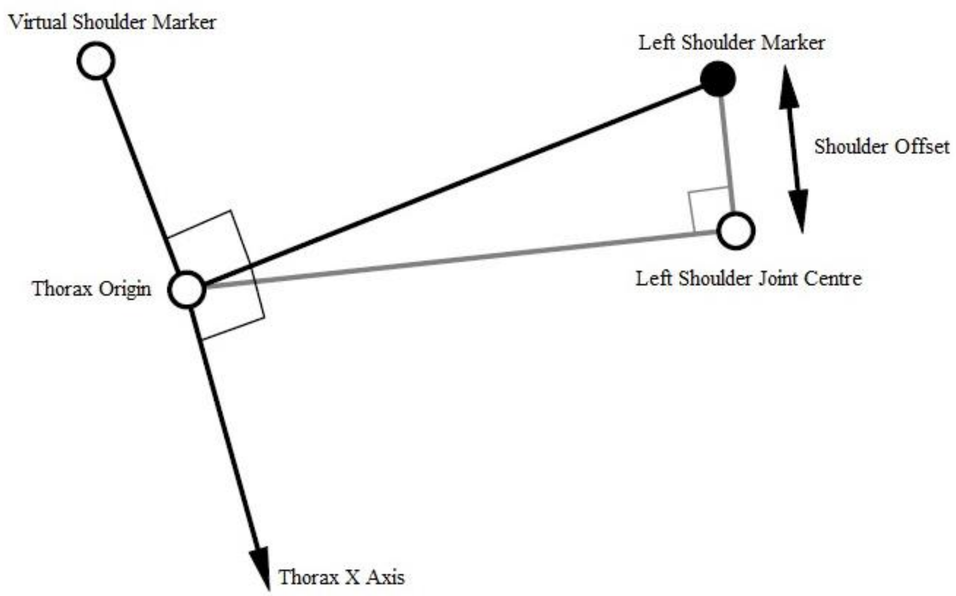


Figure A1.11. The shoulder joint centre (Vicon, 2010).

Clavicle Segment

The clavicle segment is used as an intermediate axis: X axis – forwards, Y axis – up for the left clavicle and down for the right clavicle. This is defined from the shoulder joint centre to the thorax origin as the Z axis and the virtual shoulder marker direction as the secondary axis.

Elbow Joint Centre

The elbow joint centre is defined using a chord function (Figure A1.2.) and a defined vector construction (Figure A1.12.).

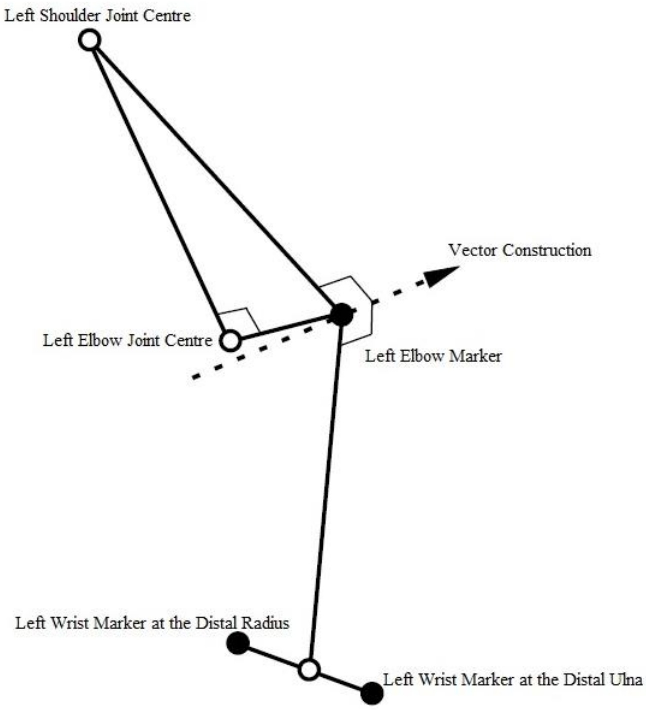


Figure A1.12. Elbow joint centre defined using the chord function. In addition, a vector is constructed which is defined by the shoulder joint centre, elbow marker and the midpoint of the distal radius and distal ulna wrist markers (Vicon, 2010).

<u>Wrist Joint Centre</u>

The wrist joint centre is defined as the offset from the midpoint of the distal radius and distal ulna wrist markers perpendicular to the line along the wrist and the wrist midpoint to the elbow joint centre (Figure 2.18.).

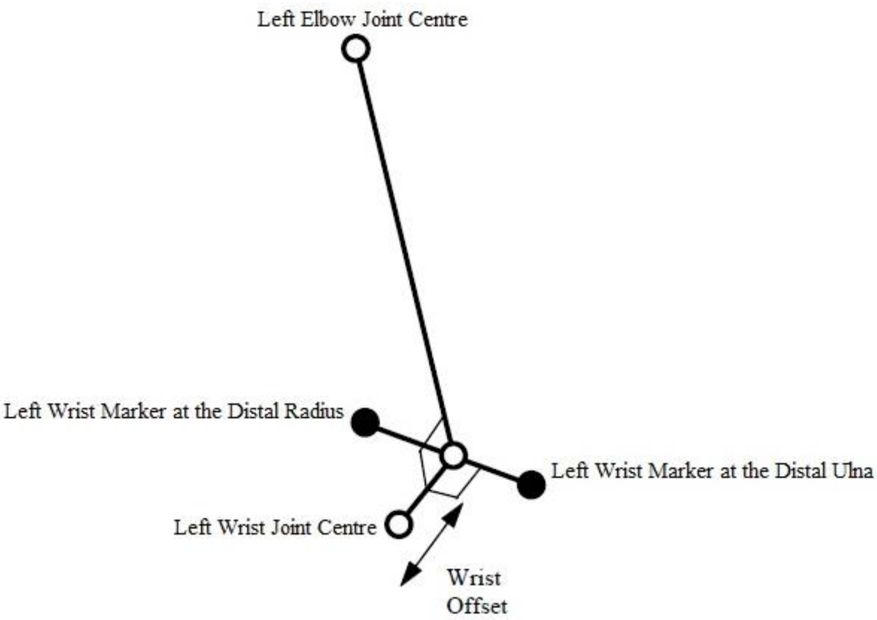


Figure A1.13. Wrist joint centre (Vicon, 2010).

Humerus Segment

The origin of the humerus is the elbow joint centre, with the primary Z axis defined from the origin to the shoulder joint centre. A secondary Y axis is defined between the elbow joint centre and the wrist joint centre (Figure A1.14.).

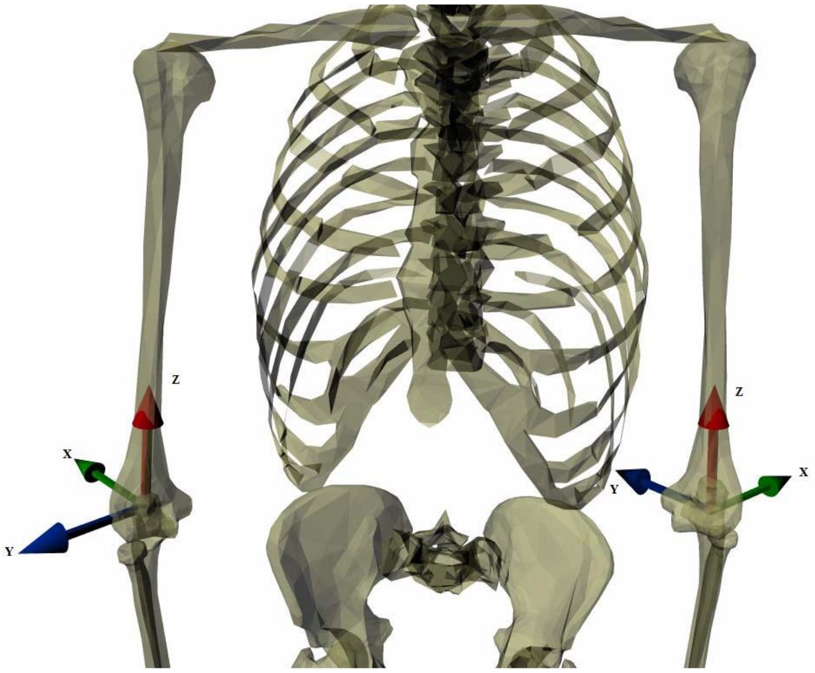


Figure A1.14. The humerus segment displayed with the Cartesian coordinate system (Created using Visual 3D, v. 4.91, Philadelphia, USA).

Radius Segment

The radius origin is the wrist joint centre. The primary Z axis is from the wrist joint centre to the elbow joint centre. The secondary Y axis is the Y axis of humerus segment. Therefore, Y axis is shared for both segments resulting in a hinge joint which is the elbow joint (Figure A1.15.).

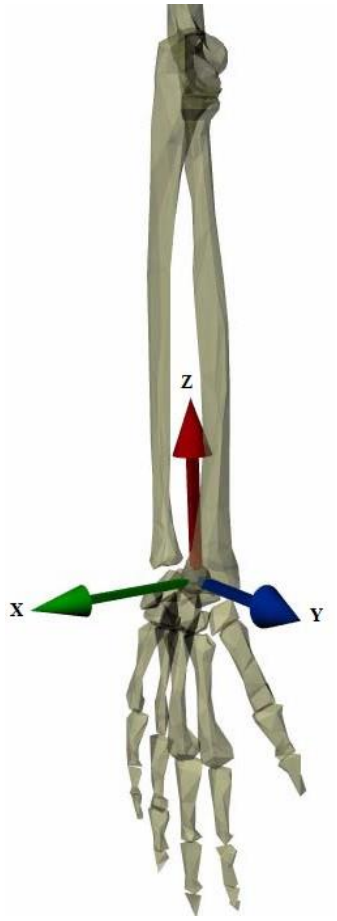


Figure A1.15. Radius segment displayed with the Cartesian coordinate system (Created using Visual 3D, v. 4.91, Philadelphia, USA).

Hand Segment

The hand origin is the third metacarpal (Figure A1.16.) and the segment is defined using a chord function (Figure A1.2.). The primary Z axis occurs in the hand origin to the wrist joint centre and the secondary Y axis the line of the distal radius and distal ulna wrist markers.

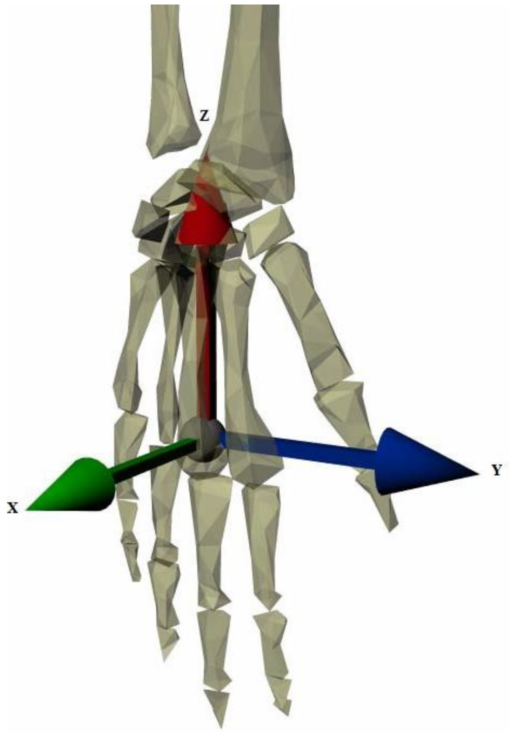


Figure A1.16. Hand segment displayed with the Cartesian coordinate system (Created using Visual 3D, v. 4.91, Philadelphia, USA).

A1.2. <u>Kinematic Modelling</u>

Joint kinematics are calculated from the cardan angles (YXZ) using the relative orientation of two segments (Kadaba *et al.*, 1990). Cardan angles are rotations which are either ordered rotations or goniometric rotations (Table A1.1.) (Kadaba *et al.*, 1990, Davis *et al.*, 1991, Vicon, 2010). Cardans angles are represented as both absolute rotations (measured relative to the laboratory axes) and relative rotations (Figure A1.17. and Table A1.2.) (Vicon, 2010). The coronal and transverse plane joint kinematics are calculated using embedded axes (Kadaba *et al.*, 1990).

Table A1.1. Description of ordered and goniometer rotations used to calculate joint kinematics (Kadaba *et al.*, 1990, Davis *et al.*, 1991, Vicon, 2010).

Rotation	**Description**
Ordered Rotation	Set of rotations carried out one after the other. *Assumptions:* 1) One segment is fixed (for absolute rotations the laboratory axes are fixed and for relative rotations the proximal segment axes are fixed). 2) Second segment moves (for absolute rotations the segment axes move and for relative rotations the distal segment moves). *Defined Joint Angle:* 1) First rotation is flexion (around the flexion axis). 2) Second rotation is abduction (around the abduction axis of the moving segment). 3) Third rotation is rotation (around the rotation axis of the moving segment).
Goniometric Rotation	One rotation fixed in a segment. *Assumptions and Defined Joint Angle:* 1) Flexion around the flexion axis of the proximal or absolute segment. 2) Rotation around the rotation axis of the distal segment. 3) Abduction axis floats and must be a right angle to the flexion and rotation axes.

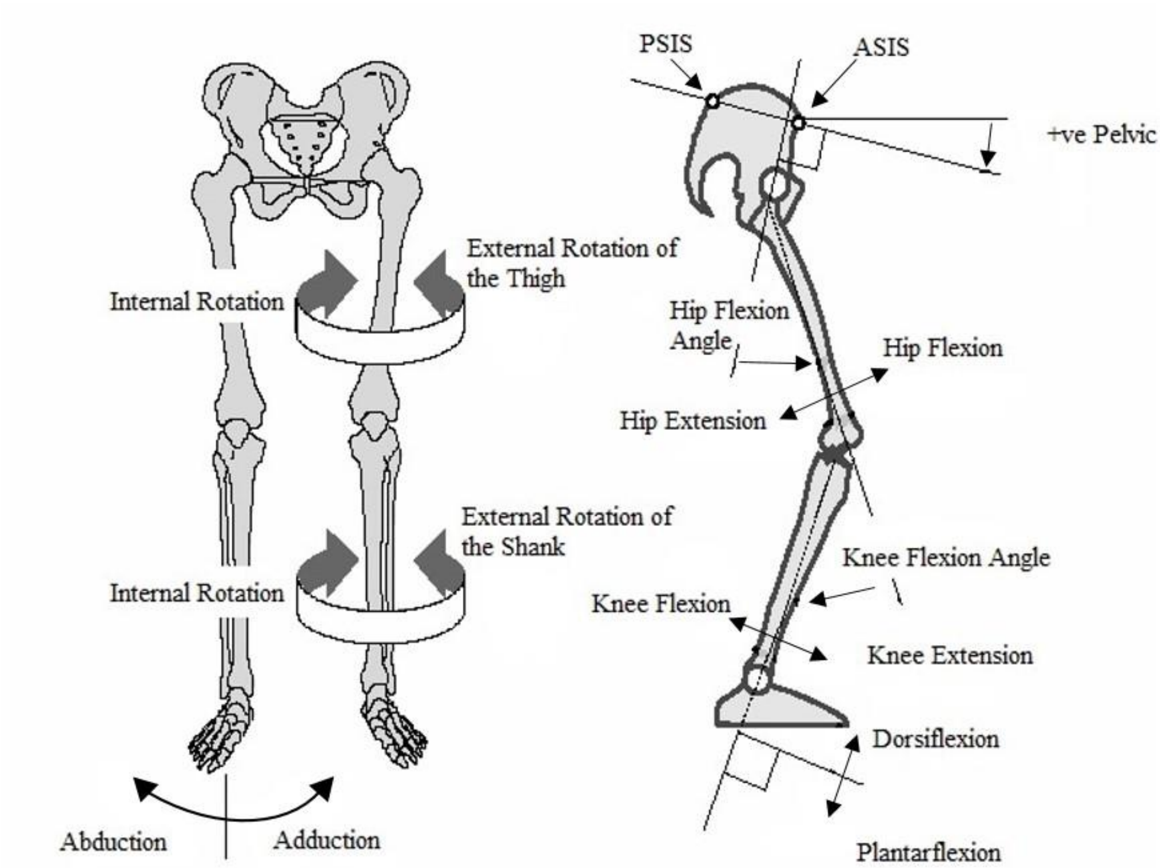


Figure A1.17. The Plug-in Gait Kinematic Modelling (Vicon, 2010).

Table A1.2. Description of the Plug-in Gait Joint Kinematics (Vicon, 2010).

Joint Kinematic	Kinematic Plane	Cardan Angle	Description
Pelvic Tilt	Sagittal	Absolute	Calculated around the laboratory transverse axis, which is measured as the angle between the projected sagittal pelvic axis and sagittal laboratory axis. Positive value = anterior pelvic tilt.
Pelvic Obliquity	Coronal	Absolute	Measured in the laboratory transverse axis and the pelvic frontal axis, between the projection into the transverse pelvic axis and projection into the laboratory transverse axis. Negative value = down pelvic obliquity (opposite side of the pelvis is lower).
Pelvic Rotation	Transverse	Absolute	Calculated around the coronal axis of the pelvic coordinate system, which is measured as the angle between the sagittal pelvic axis and the sagittal laboratory axis into the pelvis transverse plane. Negative value = external pelvic rotation.
Hip Flexion/Extension	Sagittal	Relative	Calculated around the axis parallel to the pelvic transverse axis (through the hip joint centre). The sagittal thigh axis is projected onto the hip flexion axis. Hip flexion is between the projected sagittal thigh axis and sagittal pelvic axis. Positive value = hip flexion.
Hip Abduction/Adduction	Coronal	Relative	Measured in the hip flexion axis and knee joint centre and is calculated between the long axis of the thigh and the coronal axis of the pelvis projected into this plane. Positive value = hip adduction.
Hip Rotation	Transverse	Relative	Measured around the long axis of the thigh segment and is calculated between the sagittal axis of the thigh and the sagittal axis of the pelvis projected into the plane perpendicular to the long axis of the thigh. Positive value = internal hip rotation (internal rotation of the thigh).

Joint Kinematic	**Kinematic Plane**	**Cardan Angle**	**Description**
Knee Flexion/Extension	Sagittal	Relative	The sagittal shank axis is projected into the plane perpendicular to knee flexion axis. Knee flexion is between this projection and sagittal thigh axis. Positive value = knee flexion.
Knee Varus/Valgus	Coronal	Relative	Measured in the plane of the knee flexion axis and the ankle centre, between the long axis of the shank and the long axis of the thigh. Positive value = knee varus (outward bend of the knee).
Ankle Plantar/Dorsiflexion	Sagittal	Relative	Foot vector is projected into the foot sagittal plane, which is calculated between the foot vector and the sagittal axis of the shank. Positive value = ankle dorsiflexion.

A1.3. Kinetic Modelling

The joint moments are calculated using the equation of motion for six segments of the lower body (excluding the pelvis segment) (Ramakrishnan and Kadaba, 1991), which uses the values of the external forces applied to the lower body, mass distribution within the segments, kinematics of segments and joint centre location. The assumptions for net joint moment calculation are: no external force except gravity and force plate measurements were applied and segment masses, centre of gravity and radii of gyration (Dempster, 1955) (Table A1.3.). Joint powers are calculated from the joint moment (scalar product) and angular velocity. Kinetic hierarchy starts from the foot as this segment is in contact with the force plate (Figure A1.18.) (Vicon, 2010).

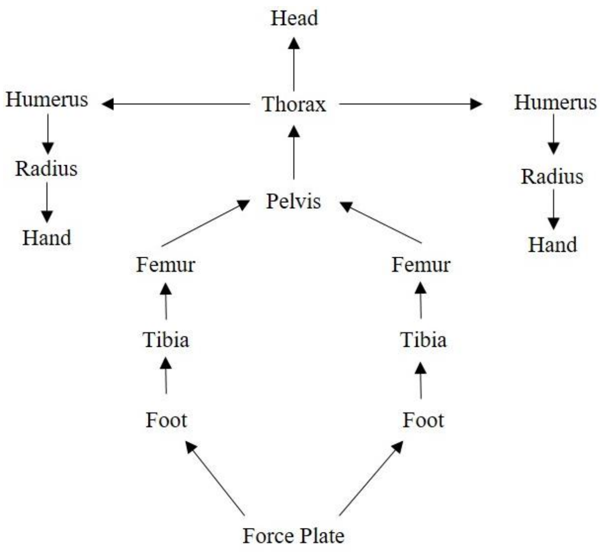


Figure A1.18. Hierarchy for calculating joint kinetics (Vicon, 2010).

Table A1.3. Dempster Data (Dempster, 1955) for kinetic hierarchy.

Force Plate Segment	Centre of Mass	Segment Mass	Radius of Gyration
Foot	0.5000	0.0145	0.475
Tibia	0.5670	0.0465	0.302
Femur	0.5670	0.1000	0.323
Pelvis	0.8950	0.1420	0.310
Thorax	0.6300	0.3550	0.310
Head	0.5200	0.0810	0.495
Humerus	0.5640	0.0280	0.322
Radius	0.5700	0.0160	0.303
Hand	0.6205	0.0060	0.223

Whole Body Centre of Mass

The centre of mass was calculated when the head or thorax segment was present even if the hand segment was not present (due to missing markers).

Appendix Two: Implications of using the Plug-in Gait Marker Model for Lower Body Gait Analysis

Adopting a hierarchical biomechanical model allows for simple marker configurations, which aids data collection demands, however they are susceptible to errors. The conventional gait model has many variations such as the Helen Hayes (Kadaba *et al.*, 1990) and the Davis Model (Davis *et al.*, 1991). For the Vicon motion capture system (Oxford, UK) the marker model is known as the Plug-in Gait (PiG) (Vicon, 2010), which uses minimal marker configuration to track three-dimensional lower body motion. As a result, joint motion is constrained with only three rotational degrees-of-freedom (DoF), which results in undesirable mathematical consequences: a) incomplete control in identifying joint centres and axes of rotation, b) body segments are not tracked independently, therefore allowing errors to cascade from the pelvis, through the thigh, shank and foot segments and consequently affecting the reliability of non-sagittal motions (Wilken *et al.*, 2012). c) Foot is modelled using only two rotational DoF and d) lack of redundant markers prevents the use of least squares techniques to control for measurement error (Collins *et al.*, 2009b, Buczek *et al.*, 2010).

Although conventional gait models such as the PiG remain the prevalent lower body model for gait laboratories especially for clinical gait analysis, the use of markers on one segment to define virtual markers that track adjacent segments in mathematically solutions implements errors. Consequently, researchers have compared the PiG model to models which have six DoF such as the six DoF model (6DoF) (Cappozzo *et al.*, 1995, Benedetti *et al.*, 1998). Greater variations have been identified in the coronal and transverse plane (Collins *et al.*, 2009b, Groen *et al.*, 2012), with only slight variations highlighted in the sagittal plane (Buczek *et al.*, 2010). These slight variations in the sagittal planes have been advocated as an effect of marker misplacement and soft tissue movement artefacts (Collins *et al.*, 2009b, Buczek *et al.*, 2010). Differences are reflected in kinetic calculations, however these are more dramatic for kinematic calculations (Charlton *et al.*, 2004). Marker misplacement results in 75 % of kinematic error (Gorton *et al.*, 2009). The 6DoF model has found to be less prone to error (Collins *et* *al.*, 2009b). Whereas, the PiG model is prone to results such as knee hyperextension caused by posterior misplacement of the lateral knee marker (Szczerbik and Kalinowska, 2011). The proximal to distal sequence utilised in identifying the segments in the PiG model affect the shank and thigh segments which postulates such model differences (Collins *et al.*, 2009b).

The coronal plane errors in the PiG model have also been attributed to the misalignment of the thigh and shank markers (Buczek *et al.*, 2010). For example, the PiG is prone to illustrate a large knee varus range of motion which resembles a knee flexion angle. This is known as crosstalk and occurs when the axes of rotations are not aligned with the joint coordinate system (Della Croce *et al.*, 2003, Schache *et al.*, 2006). Crosstalk is a reflection of marker misplacement, as there is no coupling link between the knee axes of flexion/extension and varus/valgus (Schache *et al.*, 2006). Even the 6DoF model has reported knee errors in the coronal plane (large knee valgus), however this was attributed as a soft tissue artefact; resulting from cluster marker movement of the subcutaneous fat of the thigh segment (Buczek *et al.*, 2010). Model variations also occur for the ankle joint because of foot segment definitions. The 6DoF model utilises defining the foot segment with three markers (Cappozzo *et al.*, 1995, Benedetti *et al.*, 1998), whereas the PiG model is defined by two and constructs a virtual ankle joint centre (Vicon, 2010). However, even the 6DoF model does not strictly allow the foot and ankle to calculate inversion/eversion (Collins *et al.*, 2009b). In addition, the transverse plane also reports greater external rotation for the PiG model due to model definitions of joint axes (Charlton *et al.*, 2004, Buczek *et al.*, 2010). An increased external rotation of the joints for the PiG may be a result of misalignment of the shank or thigh in the mediolateral axis (Buczek *et al.*, 2010).

Overall, the 6DoF model represents gait in all planes of motion and is therefore beneficial for utilisation. It also has less theoretical assumptions including less joint constraints and independent segment reconstruction which results in greater validity (Collins *et al.*, 2009b). However, both models are affected by soft tissue artefacts, marker misplacement and anatomical landmark identification limitations (Charlton *et al.*, 2004, Collins *et al.*, 2009b, Buczek *et al.*, 2010). To conclude PiG uses a minimal marker configuration to track three-dimensional lower body motion. Previous research has compared this model to the six DoF model for both normal and pathological gait analysis concluding great variation in the coronal and transverse plane, with sagittal plane differences attributed to marker misplacement. Marker misplacement is identified as the main limitation for gait model error. Therefore, error in a minimal marker model configuration is unlikely to change due to independent segment tracking alone. As such, the PiG is advantageous for its quick application and even with model limitations it still has successfully identified pathological gait. This may explain why it is the prevalent model used for gait analysis. Although, care must be taken when applying markers to ensure accurate placement to postulate a valid and reliable gait model.

Appendix Three: A Twofold Pilot Study: Establishing a Gait and Functionality Protocol for an Older Adult Study and Determining the Intra-Rater Reliability of Marker Placement during Normal Walking

A3.1. Introduction

Physical functionality describes a person's ability to perform everyday tasks (Cooper *et al.*, 2011b), for example walking. Older adults' physical functionality is important for health and well-being, as this is influenced by the ageing process (Guralnik and Simonsick, 1993). For this reason, there is developing evidence that measuring physical functionality such as walking and walking with an additional task (e.g. turning and obstacle negotiation); not only indicates health, wellbeing (Cesari *et al.*, 2005) and functional status (Cooper *et al.*, 2011b), but also predicts adverse events such as falls and mortality (Verghese *et al.*, 2009, Swanenburg *et al.*, 2010). As such, functionality acts as a marker for current and future health. There are many ways to gauge overall functionality, for example rating systems (e.g. functional mobility scale) (Graham *et al.*, 2004), timed functionality (e.g. timed up and go (TUG) (Podsiadlo and Richardson, 1991), video analysis (Sowers *et al.*, 2006), spatial-temporal walkways (GAITRite, CIR systems, Pennsylvania, USA) (Verlinden *et al.*, 2013) and three-dimensional gait analysis (Winter *et al.*, 1990).

The 'gold standard' currently for gait and functionality assessment is three-dimensional gait analysis, with kinematic data having a key role in movement analysis. Therefore, it is important that each biomechanics laboratory conducting three-dimensional gait analysis establish protocol feasibility, to create a standardised protocol. Also, it is necessary to ensure the analysis is reliable (determine the assessor's marker placement reliability), as quantified kinematics parameters have shown variations between data collection (McGinley *et al.*, 2009). There are numerous sources of variability within the testing procedure and these are regarded as intrinsic and extrinsic variations (Schwartz *et al.*, 2004). Intrinsic variation illustrates the inherent walking variation within the participant when performing multiple trials, which cannot be reduced. Extrinsic variations reflect assessor errors such as marker misplacement and processing errors (e.g. incorrect identification of gait cycle events) (Schwartz *et al.*, 2004, Eve *et al.*, 2006), these errors can be reduced with training and experience. The main source of error during data collection is marker misplacement which affects joint kinetics and severely affects kinematic parameters (Gorton *et al.*, 2009, McGinley *et al.*, 2009).

Intra-rater reliability studies for gait kinematics parameters have identified errors between 2-5 ° between data collection is the norm (McGinley *et al.*, 2009). Although, children and young adults reliability is well documented (McGinley *et al.*, 2009), investigations of older adult kinematic reliability is scarce. Healthy older adults display greater gait variability than young adults (Oberg *et al.*, 1993, Nigg *et al.*, 1994, Owings and Grabiner, 2004). However, variability within a healthy older adult population remains unknown. Therefore, in an older adult population does variability of gait magnitude reflect natural human variation and the ageing process or are extrinsic errors (e.g. marker misplacement) influencing this variability. A twofold pilot study was conducted: **1)** to determine protocol feasibility and establish a standardised study design and **2)** determine the intra-rater reliability of marker placement for lower body kinematics during normal walking (NW).

A3.2. <u>Methodology</u>

A3.2.1. <u>Research Design</u>

This was a twofold research design using a prospective study to establish protocol feasibility and a test re-test design to assess reliability. A single assessor was used to test all participants for both testing sessions. Research (Tsushima *et al.*, 2003, Charlton *et al.*, 2004, Schwartz *et al.*, 2004) has shown single assessors to be more reliable than multiple assessors. A one week interval between testing sessions (same protocol for both sessions) was implemented (Kadaba *et al.*, 1989, Ferber *et al.*, 2002, Maynard *et al.*, 2003, Mackey *et al.*, 2005) to reduce the likelihood of measurement change and minimise fatigue and memory bias effects (McGinley *et al.*, 2009, Thomas *et al.*, 2011). Processing of all data was not conducted until all participants completed both testing sessions. Protocol feasibility was determined if: **1**) data collection was completed within two hours and **2)** all walking tasks using three-dimensional analysis had ≤ 10 sample frame gaps for the marker trajectories. Analysis for protocol feasibility was conducted on the first testing session only, as the study design would only have one data collection session. Intra-rater reliability was determined if the kinematic parameter had ≤ 15 % variance between-sessions (Robinson *et al.*, 1993, Shechtman, 2001) and/or ≤ 5 ° measurement error (McGinley *et al.*, 2009). Measurement error was also used to assess variability as this directly relates to the measured kinematic parameter as both are expressed in degrees (Keating and Matyas, 1998).

A3.2.2. Participants

Four healthy older adults, age range 55-64 years (1 female; 3 males; 59.3 ± 4.4 yrs; 177.3 ± 7.1 cm; 87.4 ± 21.5 kg) participated in the pilot study. Recruitment and inclusion criteria for the pilot study, was the same as the main study (Chapter Two: Methodology 2.4. Participants). Participants were all instructed to wear tight compressive non-reflective clothing and flat shoes; as the study (Chapter Two: Methodology 2.5.2. Clothing). Ethical approval was granted by the University of Essex Ethics Committee and all participants gave written informed consent.

A3.2.3. Data Collection

The study was administered in the Biomechanics Laboratory at the University of Essex. A seven camera Vicon T20 infrared motion capture system (Oxford, UK) sampling at 100 Hz, with a floor-mounted Kistler 9281CA force plate (Winterthur, Switzerland) sampling at 1000 Hz were used to derive the three-dimensional motion analysis for all the walking tasks. Prior to each data capture session, the Vicon system was calibrated and a residual of < 2 mm for each camera was accepted.

A3.2.4. Protocol and Marker Placement

All participants completed the mini-mental state examination (Folstein *et al.*, 1975). Anthropometric measurements were obtained for all participants (Chapter Two: Methodology 2.6.5. Anthropometric Measurements). Two simple functionality measures (hand-grip (Fess, 1992) and TUG (Podsiadlo and Richardson, 1991)) were performed to establish baseline functionality. The hand-grip dynamometer

(Takei Analogue 5001, Niigata, Japan) was performed three times for each hand with a 15 second rest between each trial and all participants alternated hands between each trial (starting with their dominant hand), in accordance with the literature (Mathiowetz, 1990, Harth and Vetter, 1994, Hanten *et al.*, 1999, Werle *et al.*, 2009). Hand-grip strength was recorded in kg to the nearest 0.1 kg. The TUG was recorded in seconds (to the nearest 0.1 s), using an iPhone stopwatch application (iPhone 5, California, USA). All participants were instructed to stand-up from the chair (same chair for all participants and the chair did not have arms to assist standing), walk 3 metres (at a self-selected normal walking speed), turn around a cone and walk to the chair and sit down.

Thirty-five passive reflective markers were placed on the upper (n = 19) and lower (n = 16) body in accordance to the Plug-in Gait Marker Model (PiG) (Vicon, 2010). Following the static trial, participants were familiarised with their surroundings and each walking task. Five walking tasks were performed on a 10 m walkway: **1)** NW, **2)** manual dual task walking (DT), **3)** stepping onto and off an obstacle (SON), **4)** stepping over an obstacle (SOV) and **5)** turning. Due to the methodological limitations associated with speed-controlled studies, for example difficulty in generalising findings (Astephen Wilson, 2012), it was decided not to control walking speed. Instead, participants were instructed to walk 'at their preferred walking speed'. Five trials were recorded for each task using the Vicon system. The inclusion of five trials has found higher reliability indices (Diss, 2001).

For NW participants contacted the force plate with their right foot for five trials then with their left foot for five trials. Participants were instructed not to look down at the force plate during the NW task. For DT walking participants held a full cup of water (200 ml, in their dominant hand) and were instructed to walk without spilling the water. To date, no standardised manual dual task has been proposed (Asai *et al.*, 2014). As such, this task was chosen as it replicates a real-world setting. For the obstacle clearance tasks (SON and SOV) the obstacle (Reebok Stepper (100 x 16 x 40 cm), Adidas Group, Herzogenaurach, Germany)) was placed horizontally after the force plate on the walkway, with reflective markers placed on all corners of the obstacle. Participants were instructed to step onto the obstacle then step off (the other side) and continue walking for SON and step over the obstacle and continue walking for SOV. No instruction was given regarding leading leg for the obstacle clearance tasks; participants self-selected. A 90 ° step turn (turn to the opposite side of the stance limb, e.g. right foot on force plate, turn and step out with the left foot) was performed for five trials for the right and left foot. Step turns are biomechanically safer turns and the reason for turn selection (Hase and Stein, 1999).

A3.2.5. Data Processing

Processing of all trials for all walking tasks was performed using Vicon Nexus (v 1.8.5, Oxford, UK). Reconstruction of the markers and auto-labelling of marker trajectories were performed. Each trial was then visually inspected and unlabelled marker trajectories were manually labelled. Gaps in marker trajectories of up to 10 sample frames joined with linear interpolation filtered with a quintic spline filter (Woltring; mean square error of 10). Then low-pass filtered at 10 Hz using a 4th order Butterworth filter. This cut-off frequency was selected to attenuate noise without distorting high-frequency marker movement at heel contact (Sinclair *et al.*, 2013b). Gait cycle events of initial contact (on the force plate) and toe-off (on the force plate) were identified for NW and turning using a Nexus sub-routine which checks for the crossing threshold value (10 N) of the amplitude of the vertical component of the ground reaction force when the ankle and toe markers lie within the bounds of the force plate. Visual inspection was used to verify these events and manual gait cycle events were applied to the next initial contact for NW and turning and all events for DT, SON and SOV. Gait cycle events which were manually identified used frame by frame visual inspection of the lowest trajectory frame (closet to the ground) of the heel marker for heel contact and the next frame after the lowest trajectory frame (closet to the ground) of the toe marker for toe-off. The dynamic PiG model was then applied and gait cycle events, marker trajectories, kinematics and spatial-temporal parameters were exported using ASCII files in a .csv format.

A3.2.6. Data and Statistical Analysis

Required data analysis was completed using custom-made python code (Python v. 2.7.6, Delaware, USA). Kinematics for NW for both testing session were normalised to one gait cycle (100 %) using linear interpolation to 101 data samples. Average range of motion for the kinematics (sagittal plane (pelvic tilt, hip flexion/extension, knee flexion/extension and ankle plantar/dorsiflexion), coronal plane (pelvic obliquity, hip abduction/adduction and knee varus/valgus) and transverse plane (pelvic rotation and hip rotation) were calculated for all NW trials and subsequently averaged to determine the mean and standard deviation for each participant.

Statistical analysis was performed using Excel (Microsoft Office, 2010, Tokyo, Japan). Both within-testing sessions and between-testing sessions were conducted to determine the within variability of NW for both testing sessions and the variability between the two sessions. Due to the small sample size reliability was assessed using coefficient of variation (CV%), which represents typical error in a measurement and useful for repeatability of a parameter (Hopkins, 2000) and a favoured measure for gait reliability (Steinwender *et al.*, 2000, Thorpe *et al.*, 2005, Yavuzer *et al.*, 2008). CV% was calculated using this equation:

$$CV\% = \left(\frac{standard\ deviation\ (SD)}{mean}\right) * 100$$

CV% was calculated to determine the variability between each testing session and between sessions for kinematic parameters (joint range of motion) and walking speed for all participants for right and left NW. Measurement error (°) was calculated to determine the mean difference between testing sessions for the kinematic parameters and walking speed for right and left NW using the following equation:

$$Measurement\ error\ = Session\ one\ parameter - Session\ two\ parameter$$

Intra-rater reliability was accepted if the kinematic parameter had ≤ 15 % variance between-sessions (Robinson *et al.*, 1993, Shechtman, 2001) and/or ≤ 5 ° measurement error (McGinley *et al.*, 2009).

A3.3. Results

A3.3.1. Protocol Feasibility

All participants completed the entire protocol within one and a half hours. Marker trajectories sample frame gaps were within the acceptable limit for NW, DT, SON and SOV. However, turning marker trajectories were more than 15 sample frames for most of the turning trials. During data collection, each captured trial was replayed to visually inspect marker visibility. On inspection, apart from participant three for the right turning task, all participants had noticeable large marker gaps and consequently additional turning trials were recorded. No participant had five valid turning trials for both the right and left (Table A3.1.).

Table A3.1. Number of valid turning trials vs. number of recorded trials during session one.

Participant	Right Turn	Left Turn
Participant One	3/9	4/8
Participant Two	0/10	0/12
Participant Three	5/5	0/18
Participant Four	0/11	0/12

A3.3.2. Reliability

The within- (CV%) and between-session (measurement error and CV%) reliability indexes are presented in Tables A3.2-A3.3. The within reliability CV% values of all assessed parameters were < 15 % for the right and left NW, except for participant 4 right NW knee flexion/extension (CV% of 16.40). The between-session reliability revealed measurement error was < 5 ° for most parameters. However, three parameters had measurement errors of > 5 ° and these were pelvic tilt (participant 1 NW right = -6.93 ° and left = -6.81 °; participant 2 NW right = -9.28 ° and left = -9.09 °), hip flexion/extension (participant 1 NW right = -6.79 °; participant 2 NW right = -6.83 ° and left = -9.57 °) and ankle plantar/dorsiflexion (participant 4 NW left = -5.48 °) (Table A3.4. and A3.5.). The highest variability (> 15 %) for CV% was for pelvic tilt (participant 1 NW right = 49.93 and left = 48.78), hip flexion/extension (participant 1 NW right = 42.75 and left = 15.87; participant 2 NW right = 43.50), hip abduction/adduction (participant 3 right NW = 28.98), hip rotation (participant 4 NW right = -15.24 and left = -19.20) and knee flexion/extension (participant 1 NW left = 15.48) (Table A3.4. and A3.5.).

Table A3.2. Right NW within-variability for each testing session.

Parameter	Mean ± SD		CV%	
	Session One	Session Two	Session One	Session Two
Pelvic Tilt °	P1: 13.28 ± 0.05	6.35 ± 0.81	0.39	12.83
	P2: 5.52 ± 0.51	-3.76 ± 0.95	9.31	-25.23**
	P3: 4.02 ± 0.70	4.35 ± 1.05	17.35**	24.05**
	P4: 10.45 ± 0.34	9.80 ± 0.58	3.24	6.12
Pelvic Obliquity °	P1: -1.34 ± 1.94	-0.36 ± 0.51	-144.58**	-140.67**
	P2: 0.89 ± 0.24	-0.41 ± 0.26	26.69**	-63.58**
	P3: 0.18 ± 0.31	3.08 ± 0.98	168.44**	31.86**
	P4: -0.23 ± 0.24	-1.49 ± 0.56	-105.49**	-37.79**
Pelvic Rotation °	P1: 4.24 ± 0.50	5.16 ± 0.58	11.68	11.27
	P2: 4.50 ± 1.11	1.46 ± 1.23	24.70**	83.94**
	P3: 6.91 ± 1.30	5.64 ± 0.21	18.82**	3.70
	P4: 4.67 ± 0.60	3.12 ± 0.60	12.91	19.40**
Hip Flexion/Extension °	P1: 14.63 ± 0.96	7.84 ± 0.72	6.59	9.22
	P2: 14.51 ± 0.88	7.68 ± 0.89	6.07	11.64
	P3: 12.49 ± 0.68	13.40 ± 0.21	5.41	3.55
	P4: 8.41 ± 1.15	8.79 ± 1.11	13.63	12.63
Hip Abduction/Adduction °	P1: 7.02 ± 0.88	6.96 ± 0.53	12.55	7.62
	P2: 3.76 ± 0.26	3.08 ± 0.43	6.99	13.80
	P3: 7.04 ± 0.67	4.65 ± 0.89	9.45	19.20**
	P4: 2.99 ± 0.47	1.28 ± 0.75	15.62**	58.69**
Hip Rotation °	P1: -0.70 ± 0.44	1.65 ± 1.46	-62.58**	88.25**
	P2: 8.10 ± 1.04	11.65 ± 0.90	12.78	7.76
	P3: 1.89 ± 0.93	-2.65 ± 0.34	49.28**	-12.75
	P4: -15.54 ± 0.34	-12.52 ± 0.31	-2.20	-2.49
Knee Flexion/Extension °	P1: 13.39 ± 1.67	15.17 ± 1.02	12.48	6.76
	P2: 23.43 ± 0.86	23.83 ± 0.54	3.65	2.25
	P3: 25.91 ± 0.56	25.53 ± 0.93	2.15	3.66
	P4: 16.15 ± 1.41	15.51 ± 2.54	8.74	16.40*
Knee Varus/Valgus °	P1: -4.89 ± 2.77	-5.19 ± 0.93	-56.67**	-17.87**
	P2: 3.03 ± 0.98	2.21 ± 0.40	32.38**	17.94**
	P3: -2.79 ± 0.75	-4.15 ± 0.61	-27.03**	-14.62
	P4: 0.11 ± 0.58	2.74 ± 0.76	513.94**	27.68**
Ankle Plantar/Dorsiflexion °	P1: 3.80 ± 0.35	4.19 ± 0.43	9.24	10.17
	P2: 3.61 ± 0.20	3.39 ± 0.59	5.63	17.34**
	P3: 3.51 ± 0.49	3.91 ± 0.63	14.03	16.09**
	P4: 5.43 ± 0.52	1.74 ± 0.21	9.54	12.19
Walking Speed ($m \cdot s^{-1}$)	P1: 0.90 ± 0.02	1.06 ± 0.04	2.76	3.78
	P2: 1.50 ± 0.02	1.48 ± 0.03	1.54	2.19
	P3: 1.43 ± 0.02	1.49 ± 0.03	1.21	1.74
	P4: 1.28 ± 0.02	1.28 ± 0.02	1.71	1.49

** signified parameters which have > 15 % variability and ** signified parameters which were discarded from the results.*

Table A3.3. Left NW within-variability for each testing session.

Parameter	Mean ± SD		CV%	
	Session One	**Session Two**	**Session One**	**Session Two**
Pelvic Tilt °	P1: 13.28 ± 0.38	6.47 ± 0.57	2.86	8.78
	P2: 5.61 ± 0.41	-3.48 ± 0.70	7.39	-20.21**
	P3: 3.85 ± 0.39	3.46 ± 0.84	10.07	24.35**
	P4: 9.56 ± 0.51	8.56 ± 0.70	5.35	8.19
Pelvic Obliquity °	P1: 0.32 ± 0.19	0.65 ± 0.66	57.70**	102.61**
	P2: -0.69 ± 0.15	0.09 ± 0.27	-21.99**	295.82**
	P3: -0.33 ± 0.35	-3.52 ± 0.26	-108.64**	-7.51
	P4: 1.15 ± 0.34	1.59 ± 0.19	29.43**	11.86
Pelvic Rotation °	P1: -0.99 ± 1.47	-2.17 ± 2.27	-148.35**	-104.74**
	P2: -4.24 ± 1.04	-1.32 ± 0.92	-24.54**	-69.68**
	P3: -6.95 ± 1.48	-5.88 ± 1.19	-21.26**	-20.18**
	P4: -6.18 ± 0.69	-4.47 ± 0.51	-11.22	-11.34
Hip Flexion/Extension °	P1: 11.64 ± 0.27	9.29 ± 0.34	2.36	3.61
	P2: 10.84 ± 0.59	1.27 ± 0.89	5.47	69.79**
	P3: 12.81 ± 0.14	11.72 ± 1.09	1.06	9.33
	P4: 9.15 ± 0.61	9.21 ± 0.82	6.66	8.94
Hip Abduction/Adduction °	P1: 7.51 ± 0.85	8.54 ± 0.40	11.37	4.71
	P2: -0.09 ± 0.57	2.40 ± 0.19	-649.88**	7.98
	P3: 1.33 ± 0.53	0.11 ± 0.72	40.09**	678.51**
	P4: -1.16 ± 0.66	-1.22 ± 0.40	-56.73**	-32.91**
Hip Rotation °	P1: -9.88 ± 0.57	-12.20 ± 0.86	-5.77	-7.09
	P2: 11.29 ± 0.74	11.69 ± 0.95	6.60	8.11
	P3: 0.88 ± 0.45	-0.71 ± 0.55	50.34**	-77.51**
	P4: -16.63 ± 1.42	-12.66 ± 0.77	-8.52	-6.08
Knee Flexion/Extension °	P1: 15.04 ± 0.99	18.73 ± 0.40	6.56	2.14
	P2: 12.31 ± 1.12	15.19 ± 0.82	9.14	5.37
	P3: 24.43 ± 0.46	20.28 ± 0.54	1.90	2.68
	P4: 21.84 ± 0.52	22.41 ± 1.04	2.38	4.66
Knee Varus/Valgus °	P1: -5.67 ± 0.17	-6.03 ± 0.93	-2.94	-15.48
	P2: 2.50 ± 0.70	3.92 ± 0.32	28.12**	8.17
	P3: -1.48 ± 0.61	-4.42 ± 0.38	-41.12**	-8.64
	P4: 0.10 ± 0.60	1.44 ± 0.51	615.48**	35.37**
Ankle Plantar/Dorsiflexion °	P1: 4.12 ± 0.30	4.81 ± 0.36	7.29	7.47
	P2: 0.40 ± 0.44	1.07 ± 0.51	110.59**	47.84**
	P3: 1.68 ± 0.39	3.55 ± 0.53	23.33**	15.03**
	P4: 6.13 ± 0.45	0.65 ± 0.33	7.29	50.43**
Walking Speed ($m \cdot s^{-1}$)	P1: 0.96 ± 0.05	1.04 ± 0.02	5.70	2.13
	P2: 1.49 ± 0.03	1.48 ± 0.03	2.23	1.86
	P3: 1.44 ± 0.06	1.46 ± 0.03	4.33	1.93
	P4: 1.29 ± 0.05	1.26 ± 0.03	3.52	2.48

*** signified parameters which were discarded from the results.*

Table A3.4. Between-session variability for right NW.

Parameter	Mean ± SD		CV%	
	Session One	Session Two	Session One	Session Two
Pelvic Tilt °	P1: 13.28 ± 0.05	6.35 ± 0.81	-6.93*	49.93*
	P2: 5.52 ± 0.51	-3.76 ± 0.95	-9.28*	744.67**
	P3: 4.02 ± 0.70	4.35 ± 1.05	0.33	5.58
	P4: 10.45 ± 0.34	9.80 ± 0.58	-0.95	6.74
Pelvic Obliquity °	P1: -1.34 ± 1.94	-0.36 ± 0.51	0.98	-81.60**
	P2: 0.89 ± 0.24	-0.41 ± 0.26	-1.30	381.02**
	P3: 0.18 ± 0.31	3.08 ± 0.98	2.90	125.60**
	P4: -0.23 ± 0.24	-1.49 ± 0.56	-1.26	-103.65**
Pelvic Rotation °	P1: 4.24 ± 0.50	5.16 ± 0.58	0.92	13.85
	P2: 4.50 ± 1.11	1.46 ± 1.23	-3.03	71.94**
	P3: 6.91 ± 1.30	5.64 ± 0.21	-1.27	14.29
	P4: 4.67 ± 0.60	3.12 ± 0.60	-1.55	28.19**
Hip Flexion/Extension °	P1: 14.63 ± 0.96	7.84 ± 0.72	-6.79*	42.75*
	P2: 14.51 ± 0.88	7.68 ± 0.89	-6.83*	43.50*
	P3: 12.49 ± 0.68	13.40 ± 0.21	0.91	5.00
	P4: 8.41 ± 1.15	8.79 ± 1.11	0.38	3.15
Hip Abduction/Adduction °	P1: 7.02 ± 0.88	6.96 ± 0.53	-0.06	0.58
	P2: 3.76 ± 0.26	3.08 ± 0.43	-0.67	13.93
	P3: 7.04 ± 0.67	4.65 ± 0.89	-2.40	28.98*
	P4: 2.99 ± 0.47	1.28 ± 0.75	-1.71	56.70**
Hip Rotation °	P1: -0.70 ± 0.44	1.65 ± 1.46	2.35	349.80**
	P2: 8.10 ± 1.04	11.65 ± 0.90	3.55	25.42*
	P3: 1.89 ± 0.93	-2.65 ± 0.34	-4.54	-842.50**
	P4: -15.54 ± 0.34	-12.52 ± 0.31	3.02	-15.24*
Knee Flexion/Extension °	P1: 13.39 ± 1.67	15.17 ± 1.02	1.78	8.81
	P2: 23.43 ± 0.86	23.83 ± 0.54	0.40	1.18
	P3: 25.91 ± 0.56	25.53 ± 0.93	-0.38	1.04
	P4: 16.15 ± 1.41	15.51 ± 2.54	-0.64	2.86
Knee Varus/Valgus °	P1: -4.89 ± 2.77	-5.19 ± 0.93	-0.30	-4.25
	P2: 3.03 ± 0.98	2.21 ± 0.40	-0.82	22.11**
	P3: -2.79 ± 0.75	-4.15 ± 0.61	-1.36	-27.67**
	P4: 0.11 ± 0.58	2.74 ± 0.76	2.63	130.27**
Ankle Plantar/Dorsiflexion °	P1: 3.80 ± 0.35	4.19 ± 0.43	0.39	6.87
	P2: 3.61 ± 0.20	3.39 ± 0.59	-0.22	4.40
	P3: 3.51 ± 0.49	3.91 ± 0.63	0.40	7.63
	P4: 5.43 ± 0.52	1.74 ± 0.21	-3.70	72.93*
Walking Speed ($m \cdot s^{-1}$)	P1: 0.90 ± 0.02	1.06 ± 0.04	0.16	11.54
	P2: 1.50 ± 0.02	1.48 ± 0.03	-0.02	0.95
	P3: 1.43 ± 0.02	1.49 ± 0.03	0.06	2.91
	P4: 1.28 ± 0.02	1.28 ± 0.02	0.00	0.00

*Note: * signified parameters which have > 5 ° measurement error or > 15 % variability and ** signified parameters which were discarded from the results.*

Table A3.5. Between-session variability for left NW.

Parameter	Mean ± SD		Measurement Error °	CV%
	Session One	Session Two		
Pelvic Tilt °	P1: 13.28 ± 0.38	6.47 ± 0.57	-6.81*	48.78*
	P2: 5.61 ± 0.41	-3.48 ± 0.70	-9.09*	602.99**
	P3: 3.85 ± 0.39	3.46 ± 0.84	-0.39	7.55
	P4: 9.56 ± 0.51	8.56 ± 0.70	-1.01	7.85
Pelvic Obliquity °	P1: 0.32 ± 0.19	0.65 ± 0.66	0.32	47.14**
	P2: -0.69 ± 0.15	0.09 ± 0.27	0.78	-185.30**
	P3: -0.33 ± 0.35	-3.52 ± 0.26	-3.19	-117.42
	P4: 1.15 ± 0.34	1.59 ± 0.19	0.44	22.59**
Pelvic Rotation °	P1: -0.99 ± 1.47	-2.17 ± 2.27	-1.17	-52.52**
	P2: -4.24 ± 1.04	-1.32 ± 0.92	2.92	-74.16**
	P3: -6.95 ± 1.48	-5.88 ± 1.19	1.07	-11.81
	P4: -6.18 ± 0.69	-4.47 ± 0.51	1.71	-22.73**
Hip Flexion/Extension °	P1: 11.64 ± 0.27	9.29 ± 0.34	-2.35	15.87*
	P2: 10.84 ± 0.59	1.27 ± 0.89	-9.57*	111.68**
	P3: 12.81 ± 0.14	11.72 ± 1.09	-1.09	6.28
	P4: 9.15 ± 0.61	9.21 ± 0.82	0.07	0.52
Hip Abduction/Adduction °	P1: 7.51 ± 0.85	8.54 ± 0.40	1.03	9.10
	P2: -0.09 ± 0.57	2.40 ± 0.19	2.49	152.11**
	P3: 1.33 ± 0.53	0.11 ± 0.72	-1.22	120.48**
	P4: -1.16 ± 0.66	-1.22 ± 0.40	-0.06	-3.67
Hip Rotation °	P1: -9.88 ± 0.57	-12.20 ± 0.86	-2.32	-14.83
	P2: 11.29 ± 0.74	11.69 ± 0.95	0.40	2.46
	P3: 0.88 ± 0.45	-0.71 ± 0.55	-1.60	1329.36**
	P4: -16.63 ± 1.42	-12.66 ± 0.77	3.98	-19.20*
Knee Flexion/Extension °	P1: 15.04 ± 0.99	18.73 ± 0.40	3.70	15.48*
	P2: 12.31 ± 1.12	15.19 ± 0.82	2.88	14.81
	P3: 24.43 ± 0.46	20.28 ± 0.54	-4.15	13.13
	P4: 21.84 ± 0.52	22.41 ± 1.04	0.57	1.81
Knee Varus/Valgus °	P1: -5.67 ± 0.17	-6.03 ± 0.93	-0.36	-4.41
	P2: 2.50 ± 0.70	3.92 ± 0.32	1.42	31.18**
	P3: -1.48 ± 0.61	-4.42 ± 0.38	-2.93	-70.28**
	P4: 0.10 ± 0.60	1.44 ± 0.51	1.35	123.45**
Ankle Plantar/Dorsiflexion °	P1: 4.12 ± 0.30	4.81 ± 0.36	0.70	11.01
	P2: 0.40 ± 0.44	1.07 ± 0.51	0.67	64.95**
	P3: 1.68 ± 0.39	3.55 ± 0.53	1.87	50.53**
	P4: 6.13 ± 0.45	0.65 ± 0.33	-5.48*	114.30**
Walking Speed ($m \cdot s^{-1}$)	P1: 0.96 ± 0.05	1.04 ± 0.02	0.08	5.66
	P2: 1.49 ± 0.03	1.48 ± 0.03	-0.01	0.48
	P3: 1.44 ± 0.06	1.46 ± 0.03	0.01	0.49
	P4: 1.29 ± 0.05	1.26 ± 0.03	-0.03	1.66

** signified parameters which have > 5 ° measurement error or > 15 % variability and ** signified parameters which were discarded from the results.*

A3.4. Discussion

The objective of the pilot study was twofold: **1)** to determine protocol feasibility and **2)** determine the intra-rater reliability of marker placement for lower body kinematics during NW.

A3.4.1. Protocol Feasibility

Protocol feasibility was established if: **1)** data collection was completed within two hours and **2)** all walking tasks using three-dimensional analysis had ≤ 10 sample frame gaps for the marker trajectories. The results revealed protocol feasibility was established for all walking tasks except turning. Right and left turning for all participants (except participant 3 right turn) had excessive marker trajectory gaps (> 15). During data collection, a visual inspection of each trial was executed to determine if there were any visible markers gaps and if the trial could be considered valid. The turning task has a technical limitation due to laboratory layout and amount of motion capture cameras available. Camera field of view for the left side of the walkway is blocked by additional equipment (fixed and unmoveable), which meant once the participant turned, the markers were no longer visible for motion capture. Similarly, the right side of the walkway does not have enough cameras to capture the markers once the participant turns as they are reaching the limits of the field of view. Consequently, turning was excluded from the study. Nevertheless, the protocol was completed within one and a half hours and marker trajectories frame gaps were within the acceptable limit (≤ 10) for NW, DT, SON and SOV walking task. Therefore, the protocol was deemed feasible and with a few protocol amendments (A3.4.4. Protocol Amendments) the protocol design was considered standardised.

A3.4.2. Reliability

The pilot study revealed all parameters for within-session variability were within the accepted < 15 % variance for right and left NW, except for participant 4 right NW knee flexion/extension session two (CV% of 16.40). Although there is variability for knee flexion/extension during this session the SD was 2.54 °, which is within the acceptable variability norm (McGinley *et al.*, 2009). As such, all participants were highly consistent within- each session for right and left NW.

The secondary objective of this pilot study was to determine the intra-rater reliability of marker placement during NW. This was established if the kinematic parameter had ≤ 15 % variance and/or ≤ 5 ° measurement error between-sessions. Most parameters had a measurement error < 5 ° and CV% of < 15 %, indicating high consistency for participants between-sessions. Participant 1 revealed low variability in the coronal and transverse plane, with highest variability occurring in the sagittal plane (Table A4.4. and A4.5.). Although, hip flexion/extension and knee flexion/extension for left NW have a CV% > 15, the measurement error was < 4 and therefore within the acceptable limits and considered reliable (McGinley *et al.*, 2009, McGinley *et al.*, 2014). Pelvic tilt and hip flexion/extension (right NW only) both showed between-session measurement error of > 6 ° and CV% of > 42. As such these parameters are not reliable. However, walking speed for this participant was faster during session two (NW right = 1.06 ± 0.04 $m \cdot s^{-1}$ and left = 1.04 ± 0.02 $m \cdot s^{-1}$) compared to session one (NW right = 0.90 ± 0.02 $m \cdot s^{-1}$ and left = 0.96 ± 0.05 $m \cdot s^{-1}$). Kinematic gait patterns such as the sagittal plane have found to vary with changes in walking speed between-sessions (van der Linden *et al.*, 2002, Anderson and Madigan, 2014). Kinematic variability was associated with a true change rather than inconsistent marker placement (McGinley *et al.*, 2009). This potentially explains the difference between-sessions and highlights the importance of real-time walking speed analysis during data collection.

Participant 2 also revealed low variability between-sessions for the coronal and transverse plane, with high variability in the sagittal plane. Again, measurement errors for pelvic tilt and hip flexion/extension were > 6 ° for both right and left NW. Unlike participant 1, this does however reflect marker misplacement. The ASIS markers during session two were too low and this explains the posterior tilt values given (NW right = -3.76 ± 0.95 ° and left = -3.48 ± 0.70), whereas session one pelvic tilt is anterior (> 5 °). Thus, misplacing the ASIS markers not only affects pelvic tilt range of motion but also hip flexion/extension due to PiG model limitations (Appendix Two: Plug-in Gait Marker Model Limitations). Marker misplacement accounts for 75 % of the kinematic error (Gorton *et al.*, 2009) and a measurement error of > 5 ° indicated low reliability of a parameter (McGinley *et al.*, 2009). As such, marker placement of the pelvis requires improvement prior to the study. Post pilot study advice and additional training was sought from a chartered physiotherapist (a falls prevention specialist) regarding accurately identifying pelvis landmarks using various palpating movement techniques.

Participant 3 was the most reliable participant, as all parameters measurement error was within the accepted limits. Although, hip abduction/adduction had a CV% of 28.98 for right NW, the measurement error was -2.40 ° therefore, accepted reliability. In addition, participant 4 had low between-session variability. Hip rotation had a CV% of -15.24 (right NW) and -19.20 (left NW), again measurement error was < 4 ° and so within limits. The only measurement error value outside this limit was ankle plantar-dorsiflexion (left NW -5.48 °) which had a reduce range of motion during session two (0.65 ± 0.33 ° vs. 6.13 ± 0.45 °). During walking, the foot can move inside the shoe causing movement artefacts which results in inaccurate measurements (Stacoff *et al.*, 1991, Stacoff *et al.*, 2000, Bishop *et al.*, 2012, Sinclair *et al.*, 2013a). This difference may be a result of movement artefact as oppose to marker misplacement.

Overall, the pilot study demonstrated reliable marker placement for all planes of motion during NW. This may be attributed to the assessor (well-trained) who applied the markers accurately (except for misplacing the ASIS marker during the re-test for one participant). Also, the same assessor data processed and calibration was performed for every session, with cameras positioned to optimum to minimise measurement error. All these factors assist with achieving reliable data.

A3.4.3. <u>Pilot Study Limitations</u>

The technical limitations (laboratory layout and amount of motion capture cameras) of the Biomechanics laboratory meant the turning task was not feasible and excluded from the main study. To incorporate this task in future work, the Biomechanics laboratory would require an altered layout and a minimum of five additional motion capture cameras, to allow for sufficient field of view for motion capture especially in the horizontal plane of gait walkway. Although this was a pilot study, it is limited by the small sample size which affected the statistical analysis. This meant the calculation of intra-class correlation coefficients to assess reliability was prevented and this was the main reason for selecting CV%. However, from the results section (Table A3.2.-A3.5.) this clearly illustrates the disadvantage of CV%, as it provided some misleading results. If the mean contains positive and negative values or is close to zero, then the CV% will be a high value. Consequently, parameters did have high values especially for the coronal and transverse plane and subsequently had to be discarded from analysis. Furthermore, repetition affect analysis was not calculated due to sample size. Such analysis may have highlighted if participants were improving their performance after each task (e.g. walking speed increased after each walking task) and may cause confounding variables. However, these walking tasks are known everyday tasks and consequently learning effects are unlikely to be controlled. In addition, the age range (59.3 ± 4.4 yrs) was in the youngest age criteria for the study requirement and as such although feasibility was established for the 55-64 years this may not be the case for older adults over the age of 65. However, all participants completed the study within one and half hours and therefore the extra thirty minutes may allow the older participants more time to complete tasks or rest.

A3.4.4. Protocol Amendments

During the pilot study, it was visible force plate targeting was occurring for the NW task, as participants were looking down at the plate (although instructed otherwise) and altering their stride pattern. Walking tasks with force plate contact have found to change participants gait pattern (Martin and Marsh, 1992, Oggero *et al.*, 1997, Ballaz *et al.*, 2013). Therefore, an extended familiarisation period will be implemented to minimise force plate targeting for the main study. Participants will also perform the NW task without force plate contact to familiarise themselves with the walkway in the laboratory. Timing gates (Brower System, Utah, USA) will be positioned in the middle of the walkway to calculate walking speed during data collection. This will allow the researcher to determine if participants were walking slower for the NW task with force plate contact compared to no force plate contact. If participants were slower, then the walking trial would be repeated. In addition, the obstacle clearance tasks currently are kinematic only with no force data; as such, the obstacle will be positioned prior to the force plate to allow landing forces to be calculated. Research identified ground reaction forces (landing forces) during obstacle clearance have age-related differences (Brunt *et al.*, 2005, Buckley *et al.*, 2010). Finally, as the turning task was not standardised due to technical feasibility this task will be excluded from data collection for the main study.

A3.5. Conclusion

In conclusion, the pilot study demonstrated the protocol was feasible for four walking tasks (normal walking, dual task walking, stepping onto and off an obstacle and stepping over an obstacle). The turning task was not feasible due to technical limitations and as a result is excluded from the main study. Marker placement was highly reliable in the coronal and transverse plane. Majority of sagittal plane parameters were reliable and within acceptable limits. However, marker misplacement and walking speed did affect parameters pelvic tilt and hip flexion/extension. Although for the most part marker placement was highly accurate, one participant was affected by misplacement of the pelvis markers which highlights the importance of accurate placement. Post pilot study, additional training was undertaken from a charted physiotherapist specialising in falls prevention to advice on ensuring accurate marker placement. In addition, incorporating real-time walking speed analysis into the main study and normal walking tasks without force plate contact will allow the assessor to monitor targeting and walking speed variability. Consequently, with the additional amendments the protocol is standardised for the study.

Appendix Four: Mini-Mental State Examination (MMSE)

Orientation

Year Month Day Date Time: ____/5

Country County Town Place Room: ____/5

Registration

Examiner names 3 objects (e.g. apple, table, tape): ____/3

Repeat each then, all 3 together Number of tries_____

Attention and Calculation

Subtract 7 from 100: ____/5

Continue 5 times- 93 86 79 72 65

OR: Spell 'WORLD' backwards- DLROW

Recall

Ask for names of 3 objects learned earlier: ____/3

Language

Name a Pen and Watch: ____/2

Repeat 'No ifs, ands, or buts': ___/1

Give a 3 stage command. Score 1 for each: ____/3

E.g. 'Place index finger of right hand on your nose and then on your left ear'

Obey a command on paper: ____/1

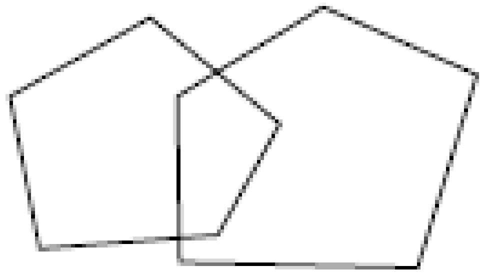
'Close your eyes'

Write a sentence: ____/1

'Subject and a verb'

Copying

Copy a pair of intersecting pentagons: ____/1



__________/30

EAGLeS

Essex Ageing and Gait Longitudinal Study

Questionnaires

The following questionnaires are designed to provide us (The Essex Ageing and Gait Longitudinal Study) with an understanding of you as a person from a social-demographic, health, functionality, physical activity and leg dominance perspective.

Personal Information

Full Name:	
Date of Birth:	
Gender:	
Address:	Post Code:
Contact Number:	Home: Mobile:
Email Address:	
Current Employment Status:	☐ Working ☐ Unemployed (less than 1 year) ☐ Long term unemployed (1 year +) Unable to work due to: ☐ Injury ☐ Disability ☐ Retired
What is your occupation, or your last occupation before unemployment or retirement?	
What industry is or was this in?	
Are you or were you…	☐ An employee ☐ Self-employed ☐ Self-employed with employees
Do you or did you supervise any other employees?	☐ Yes ☐ No
What is your highest educational award?	
Do you have Children?	☐ Yes ☐ No
Do you have Grandchildren?	☐ Yes ☐ No

APPENDIX FIVE

<u>Emergency Contact Details</u> (Please provide 2)

Emergency Contact 1:

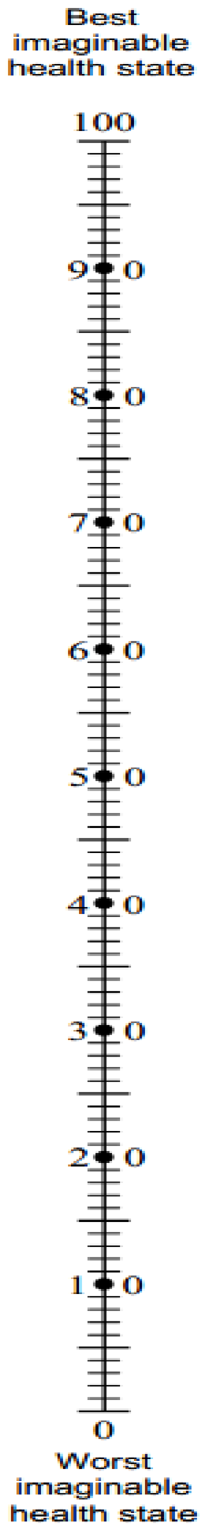
Full Name:	
Contact Number:	Home: Mobile:

Emergency Contact 2:

Full Name:	
Contact Number:	Home: Mobile:

Health Questionnaire

We would like you to indicate on this scale how you perceive your own health today. Please do this by drawing a line on the scale below.



APPENDIX FIVE

1. Do you have a heart condition?

Yes ☐

No ☐

If no, please proceed to Qn. 2a

Which heart condition? Can be multiple, please complete other questions for each type.	In what year were you diagnosed?	Are you prevented in any way from doing any activities because of this heart condition?
Heart Attack ☐		Yes ☐ No ☐ If Yes, what activities: ________________ ________________
Angina ☐		Yes ☐ No ☐ If Yes, what activities: ________________ ________________
Cardiac Arrhythmia ☐		Yes ☐ No ☐ If Yes, what activities: ________________ ________________
Others, Please specify:		Yes ☐ No ☐ If Yes, what activities: ________________ ________________

2a. Do you currently have cancer?

Yes ☐

No ☐

If no, please proceed to Qn. 3a

2b. What type of cancer?

____

2c. In what year were you diagnosed?

____

2d. Are you prevented in any way from doing any activities because of this cancer?

Yes ☐

No ☐

3a. Do you have diabetes?

Yes ☐

No ☐

If no, please proceed to Qn. 4a

3b. Which type of diabetes?

Type 1 ☐

Type 11 ☐

3c. In what year were you diagnosed?

____

3d. Are you prevented in any way from doing any activities because of the diabetes?

Yes ☐

No ☐

4a. Do you have a high blood pressure or are you taking medication to control your blood pressure?

Yes ☐

No ☐

If no, please proceed to Qn. 5a

4b. In what year were you diagnosed?

____

4c. Are you prevented in any way from doing activities because of the high blood pressure?

Yes ☐

No ☐

5a. Do you have high cholesterol?

Yes ☐

No ☐

If no, please proceed to Qn. 6a

5b. In what year were you diagnosed?

____

5c. Are you prevented in any way from doing any activities because of the high cholesterol?

Yes ☐

No ☐

6a. Do you suffer from osteoporosis?

Yes ☐

No ☐

If no, please proceed to Qn. 7a

6b. In what year were you diagnosed?

____

6c. Are you prevented in any way from doing activities because of the osteoporosis?

Yes ☐

No ☐

7a. Do you have arthritis?

Yes ☐

No ☐

If no, please proceed to Qn. 8a

Which type of arthritis? Can be multiple, please complete other questions for each type.	In what year were you diagnosed?	Are you prevented in any way from doing any activities because of this arthritis?
Rheumatoid Arthritis ☐		Yes ☐ No ☐ If Yes, what activities: ____________________ ____________________
Osteoarthritis ☐		Yes ☐ No ☐ If Yes, what activities: ____________________ ____________________
Others, please specify:		Yes ☐ No ☐ If Yes, what activities: ____________________ ____________________

8a. Do you have a respiratory (breathing) condition?

Yes ☐

No ☐

If no, please proceed to Qn. 9a

Which type of respiratory condition? Can be multiple, please complete other questions for each type.	**In what year were you diagnosed?**	**Are you prevented in any way from doing any activities because of this arthritis?**
Asthma ☐		Yes ☐ No ☐ If Yes, what activities: ________________ ________________
Chronic Bronchitis ☐		Yes ☐ No ☐ If Yes, what activities: ________________ ________________
Emphysema ☐		Yes ☐ No ☐ If Yes, what activities: ________________ ________________
Others, please specify:		Yes ☐ No ☐ If Yes, what activities: ________________ ________________

9a. Have you had a stroke?

Yes ☐ How many? _____________

No ☐

If no, please proceed to Qn. 10a

9b. If yes, when was the stroke (if you had more the one stroke please tick the box of when each occurred).

Last 6 months ☐

12 months ☐

1-2 years ☐

3-4 years ☐

4-5 years ☐

Over 6 years ☐

9c. Was that a T.I.A. (mini stroke)?

Yes ☐

No ☐

9d. Are you prevented in any way from doing any activities because of the stroke?

Yes ☐

No ☐

What activities: ____

____

____

10. Do you have any other major medical condition we have not asked about?

Yes ☐

No ☐

If no, please proceed to Qn. 11

Which major medical condition? Please specify below.	In what year were you diagnosed?	Are you prevented in any way from doing any activities because of this medical condition
		Yes ☐ No ☐ If Yes, what activities: __________ __________
		Yes ☐ No ☐ If Yes, what activities: __________ __________
		Yes ☐ No ☐ If Yes, what activities: __________ __________
		Yes ☐ No ☐ If Yes, what activities: __________ __________
		Yes ☐ No ☐ If Yes, what activities: __________ __________
		Yes ☐ No ☐ If Yes, what activities: __________ __________

Medication

Do you take any medication?

☐ Yes

☐ No

If you are not taking any medication, please proceed to Qn. 12a

The next few questions are about medicines. We are interested in any medicine prescribed by a doctor that you have taken or were supposed to take in the last 4 weeks. We are also interested in all other medicines not prescribed by a doctor that you have taken such as laxatives, cough and cold medicines, vitamins, minerals and dietary supplements. Please list down all medications both prescription and non-prescription.

For each medication, complete the details below including name, type, strength and when you take them as per the following examples.

MELOXICAM TABLET 7.5MG Three times a day
CEPHALEXIN CAPSULE 500MG Once a day
HYPROMELLOSE EYE DROPS 10MG/ML Twice a day
TRUAMCINOLONE ACETON CREAM 200MCG/G Once a day

11. Which drug? Please specify below.	What do you take this for?	When do you take this?

Falls and Fractures

This section is about falls you may have had in the past year- including both falls that did not result in an injury as well as those that did.

12a. How many falls have you had in the last 12 months?

____

If 0, please proceed to Qn. 13

12b. How many of these falls were inside your own home?

____

12c. How many of these falls were outside your own home?

____

12d. How many of these falls required medical treatment or limited your activities for more than 2 days?

____

13. Have you broken any bones in the last 12 months?

Yes ☐

No ☐

If no, please proceed to Qn. 14a

13a. Which bones have you broken in the last 12 months?

____

____

____

14a. Have you had any other surgery or operations in the last 12 months?

Yes ☐

No ☐

If no, please proceed to Qn. 15a

14b. How many times have you had surgery in the last 12 months?

____

14c. What was the surgery for? Please list below.

____

____

____

Hearing

15a. Have you ever been prescribed a hearing aid?

Yes ☐

No ☐

If no, please proceed to Qn. 16a

15b. Do you wear a hearing aid nowadays?

Yes, most of the time ☐

Yes, some of the time ☐

No ☐

15c. Were you issued with a hearing aid in the last 12 months?

Yes ☐

No ☐

15d. How much difficulty, if any, do you have with your hearing, even if you are wearing your hearing aid?

None ☐

Slight difficulty ☐

Great difficulty ☐

Vision

16a. Do you suffer from any eye problems (includes wearing lenses or glasses prescribed or non-prescribed)?

Yes ☐

No ☐

If no, please proceed to Qn. 17

Which eye problem? Can be multiple, please complete other questions for each type.		In what year were you diagnosed?	Are you prevented in any way from doing any activities because of this arthritis?
Glaucoma	☐		Yes ☐ No ☐ If Yes, what activities: ____________________ ____________________
Macular Degeneration	☐		Yes ☐ No ☐ If Yes, what activities: ____________________ ____________________
Cataracts	☐		Yes ☐ No ☐ If Yes, what activities: ____________________ ____________________
Others, please specify:			Yes ☐ No ☐ If Yes, what activities: ____________________ ____________________

16b. In the last 12 months, have you had cataract surgery in one or both of your eyes?

Yes- both eyes ☐

Yes- one eye ☐ Which eye: ________________________________

No ☐

If no, please proceed to Qn. 16d

16c. Has the cataract surgery improved your daily living?

Yes ☐

No ☐

16d. Do you ever feel that problems with your vision make it difficult for you to do the things you want to do?

Yes ☐

No ☐

16e. Do you currently wear glasses or contact lenses?

Yes ☐

No ☐

If yes: what is your prescription (if known)?

____

16f. Do you wear eye glasses or contact lenses for?

Distance viewing ☐

Reading ☐

Both ☐

16g. Can you see well enough to recognise letters in a newspaper?

Yes- with glasses or contact lenses ☐

Yes- without glasses or contact lenses ☐

No ☐

16h. Can you see well enough to recognise the letters in a headline?

Yes- with glasses or contact lenses ☐

Yes- without glasses or contact lenses ☐

No ☐

16j. To what extent, if at all, does your vision interfere with your ability to carry out the following activities? Question applies to sight with both eyes, assuming you are wearing glasses or contact lenses if necessary. The questions relate to your visual ability, not your physical ability for each. Please tick where applicable.

Type of Activity	Not Applicable	Not at all	A little	Moderately	A lot
Seeing in the distance	☐	☐	☐	☐	☐
Recognising faces across the street	☐	☐	☐	☐	☐
Watching TV	☐	☐	☐	☐	☐
Seeing in bright light	☐	☐	☐	☐	☐
Seeing in poor light	☐	☐	☐	☐	☐
Appreciating colours	☐	☐	☐	☐	☐
Driving a car/riding a bicycle by day	☐	☐	☐	☐	☐
Driving a car/riding a bicycle at night	☐	☐	☐	☐	☐
Walking inside	☐	☐	☐	☐	☐
Walking outside	☐	☐	☐	☐	☐
Using steps	☐	☐	☐	☐	☐
Crossing the road	☐	☐	☐	☐	☐
Using public transport	☐	☐	☐	☐	☐
Travelling independently	☐	☐	☐	☐	☐
Moving in unfamiliar surroundings	☐	☐	☐	☐	☐
Jobs/ study/ housework	☐	☐	☐	☐	☐
Hobbies/ leisure activities	☐	☐	☐	☐	☐

Smoking and Alcohol

17a. Do you currently smoke cigarettes, pipe or cigars?

Yes ☐

No ☐

If no, please proceed to Qn. 18a

17b. How many cigarettes, cigars or pipes do you usually smoke a day?

____

18a. How often do you have a drink containing alcohol?

Never ☐

Less than monthly ☐

Monthly ☐

Weekly ☐

Few times a week ☐

Weekends only ☐

Daily or almost daily ☐

If never, you do not need to fill in the following questions.

19b. How many units of alcohol do you drink on a typical day of drinking? For example, a small glass of wine is around one and half units of alcohol.

1 or 2 ☐

3 or 4 ☐

5 or 6 ☐

7 to 9 ☐

More than 10 ☐

19c. How often do you have 6 or more drinks on one occasion?

Never ☐

Occasionally ☐

Less than monthly ☐

Monthly ☐

Weekly ☐

Daily or almost daily ☐

Physical Activity

1. Are you able to walk up and down stairs to the first floor of a building without help?

Yes ☐

No ☐

2. Are you able to walk half a mile without help (without walking aids)?

Yes ☐

No ☐

3. How much difficulty, if any, do you have doing the activities listed below?

Please tick where applicable.

Type of Activity	No difficulty at all	A little difficulty	Some difficulty	A lot of difficulty	Unable to do it
Pulling or pushing a large object like a living room chair	☐	☐	☐	☐	☐
Stooping, crouching or kneeling	☐	☐	☐	☐	☐
Lifting or carrying weights over 10 pounds (4kg) like a heavy bag of groceries	☐	☐	☐	☐	☐
Reaching or extending your arms above shoulder level	☐	☐	☐	☐	☐
Writing or handling or fingering small objects	☐	☐	☐	☐	☐

4. Getting up and going to bed

Please put a time in **each** box

	At what time, do you normally get up?	At what time, do you normally go to bed?
On a weekday		
On a weekend day		

5. Getting about

Which form of transport do you use **most often?**
Please tick (✓) one box **ONLY** per line

Distance of journeys	Car	Walk	Public transport	Cycle
Less than 1 mile				
1-5 mile(s)				
More than 5 miles				

6. TV or video viewing

Please tick (✓) on **every** line (average over the last 4 weeks)

Hours of TV or video watched per day	None	Less than 1 hour a day	1 to 2 hours a day	2 to 3 hours a day	3 to 4 hours a day	More than 4 hours a day
On a weekday before 6pm						
On a weekday after 6 pm						
On a weekend before 6 pm						
On a weekend after 6 pm						

7. Stair climbing at home

Please tick (✓) on **every** line (average over the last 4 weeks)

Number of times you climbed up a flight of stairs (approx 10 steps) each day at home	None	1 to 5 times a day	6 to 10 times a day	11 to 15 times a day	16 to 20 times a day	More than 20 times a day
On a weekday before 6pm						
On a weekday after 6 pm						

8. Activities in and around the home

Please tick (✓) on **every** line (average over the last 4 weeks)

Approximate number of hours each week	None	Less than 1 hour a week	1 to 3 hour a week	3 to 6 hours a week	6 to 10 hours a week	10 to 15 hours a week	More than 15 hours a week
Preparing food, cooking and washing up							
Shopping for food and groceries							
Shopping and browsing in shops for other items (e.g. clothes)							
Cleaning the house							
Doing the laundry and ironing							
Gardening							
Caring for family members or friends							
Use a computer							
Play a musical instrument							
Read							

9. Activities outside your own home.

Please tick (✓) on **every** line (average over the last 4 weeks)

Approximate number of hours each week	None	Less than 1 hour a week	1 to 3 hour a week	3 to 6 hours a week	6 to 10 hours a week	10 to 15 hours a week	More than 15 hours a week
Visit with friends or family							
Go to a senior centre							
Do volunteer work							
Attend Church or take part in Church activities							
Attend a club or group meetings							
Attend a concert, movie, lecture or sport event							
Go to bingo							
Dance/ aerobic classes							
Play golf							
Play a racket sport (e.g. tennis)							
Play a sport (do not include racket sport)							
Run or jog							
Walk to do errands (e.g. going to the local shop)							
Walking for exercise (e.g. hiking)							
Riding a bicycle or a stationary cycle?							
Swimming/ water aerobics							
Yoga/ flexibility training							
Weights training							

10. If you do any other type of physical activity which was not previously mentioned please write below, including number of hours in a typical week.

Footedness Questionnaire

Instructions: Answer each of the following questions as best as you can. Think about which foot you would prefer to use to perform each activity and circle accordingly. Try imagining yourself performing the activity and if needed, act it out. There are 5 options to choose from:

1) Right always - you always use your right foot to perform the activity
2) Right usually - you prefer using your right foot to perform the activity but sometimes you use your left foot as well
3) Equally - you use both feet equally often to perform the activity
4) Left always - you always use your left foot to perform the activity
5) Left usually - you prefer using your left foot to perform the activity but sometimes you use your right foot as well

	Which foot would you use to:					
1.	Kick a stationary ball at a target in front of you	**Ra**	**Ru**	**Eq**	**La**	**Lu**
2.	Stand on one foot	**Ra**	**Ru**	**Eq**	**La**	**Lu**
3.	Smooth sand at the beach	**Ra**	**Ru**	**Eq**	**La**	**Lu**
4.	Step up on a chair first	**Ra**	**Ru**	**Eq**	**La**	**Lu**
5.	Stomp on a fast moving bug	**Ra**	**Ru**	**Eq**	**La**	**Lu**
6.	Balance one foot on a railway track	**Ra**	**Ru**	**Eq**	**La**	**Lu**
7.	Pick a marble with your toes	**Ra**	**Ru**	**Eq**	**La**	**Lu**
8.	Hop on one foot	**Ra**	**Ru**	**Eq**	**La**	**Lu**
9.	Help push a shovel into the ground	**Ra**	**Ru**	**Eq**	**La**	**Lu**
10.	Put most of your weight on during relaxed standing	**Ra**	**Ru**	**Eq**	**La**	**Lu**
11.	Is there any reason why you have changed your foot preference for the activities mentioned above?				**YES**	**NO**
12.	Have you been given any training or encouragement to use a particular foot for certain activities?				**YES**	**NO**
13.	If you have answered YES to Qn 11 or 12, please explain:					

This is the end of the survey. Thank you for answering these questions.

Appendix Six: Shapiro-Wilk Test of Normality

A6.1. Chapter Four

Table A6.1.1. Shapiro-Wilk Test of Normality for Chapter Four for whole group analysis.

Parameter	Shapiro-Wilk Test of Normality
Age (yrs)	$W_{140} = 0.971$, $P = 0.005$*
Walking Speed (m·s^{-1})	$W_{140} = 0.986$, $P = 0.160$
Pelvic Tilt	
Range of Motion (RoM)	$W_{140} = 0.992$, $P = 0.602$
Maximum Tilt - Stance	$W_{140} = 0.991$, $P = 0.485$
Minimum Tilt - Stance	$W_{140} = 0.991$, $P = 0.466$
Maximum Tilt - Swing	$W_{140} = 0.993$, $P = 0.718$
Minimum Tilt - Swing	$W_{140} = 0.993$, $P = 0.757$
Pelvic Obliquity	
RoM	$W_{140} = 0.976$, $P = 0.013$*
Maximum Obliquity - Stance	$W_{140} = 0.983$, $P = 0.085$
Minimum Obliquity - Stance	$W_{140} = 0.969$, $P = 0.003$*
Maximum Obliquity - Swing	$W_{140} = 0.970$, $P = 0.004$*
Minimum Obliquity - Swing	$W_{140} = 0.981$, $P = 0.047$*
Pelvic Rotation	
RoM	$W_{140} = 0.995$, $P = 0.934$
Maximum Rotation - Stance	$W_{140} = 0.994$, $P = 0.802$
Minimum Rotation	$W_{140} = 0.991$, $P = 0.536$
Maximum Rotation - Swing	$W_{140} = 0.990$, $P = 0.380$
Hip Flexion/Extension	
RoM	$W_{140} = 0.989$, $P = 0.358$
Flexion - Stance	$W_{140} = 0.987$, $P = 0.213$
Extension	$W_{140} = 0.995$, $P = 0.888$
Flexion - Swing	$W_{140} = 0.989$, $P = 0.350$
Hip Abduction/Adduction	
RoM	$W_{140} = 0.989$, $P = 0.360$
Adduction - Stance	$W_{140} = 0.994$, $P = 0.822$
Adduction - Swing	$W_{140} = 0.994$, $P = 0.868$
Abduction - Swing	$W_{140} = 0.993$, $P = 0.719$
Hip Rotation	
RoM	$W_{140} = 0.980$, $P = 0.037$*
Maximum Rotation - Stance	$W_{140} = 0.971$, $P = 0.005$*
Minimum Rotation - Stance	$W_{140} = 0.975$, $P = 0.012$*
Maximum Rotation - Swing	$W_{140} = 0.982$, $P = 0.065$
Minimum Rotation - Swing	$W_{140} = 0.996$, $P = 0.969$
Knee Flexion/Extension	
RoM	$W_{140} = 0.986$, $P = 0.168$
Flexion at Loading Response (LR)	$W_{140} = 0.988$, $P = 0.262$
Extension at Terminal Stance (TS)	$W_{140} = 0.967$, $P = 0.002$*
Flexion - Swing	$W_{140} = 0.990$, $P = 0.453$

Parameter	**Shapiro-Wilk Test of Normality**
Knee Varus/Valgus	
RoM	$W_{140} = 0.905$, $P = 0.000$*
Maximum Varus - Stance	$W_{140} = 0.983$, $P = 0.077$
Minimum Valgus - Stance	$W_{140} = 0.862$, $P = 0.000$*
Maximum Varus - Swing	$W_{140} = 0.926$, $P = 0.000$*
Minimum Valgus - Swing	$W_{140} = 0.865$, $P = 0.000$*
Ankle Plantar/Dorsiflexion	
RoM	$W_{140} = 0.989$, $P = 0.316$
Plantarflexion (LR)	$W_{140} = 0.992$, $P = 0.650$
Dorsiflexion - Stance	$W_{140} = 0.924$, $P = 0.000$*
Maximum Plantarflexion	$W_{140} = 0.896$, $P = 0.000$*
Moments (Nm/kg)	
Hip Flexion/Extension	
Flexion Moment - Stance	$W_{140} = 0.978$, $P = 0.021$*
Extension Moment	$W_{140} = 0.936$, $P = 0.000$*
Flexion Moment - Swing	$W_{140} = 0.982$, $P = 0.068$
Hip Abduction/Adduction	
Maximum Moment (First Peak)	$W_{140} = 0.981$, $P = 0.055$
Minimum Moment (First Peak)	$W_{140} = 0.893$, $P = 0.000$*
Maximum Moment (Second Peak)	$W_{140} = 0.979$, $P = 0.029$*
Minimum Moment (Second Peak)	$W_{140} = 0.870$, $P = 0.000$*
Knee Flexion/Extension	
Flexion Moment (LR)	$W_{140} = 0.975$, $P = 0.010$*
Extension Moment (TS)	$W_{140} = 0.987$, $P = 0.201$
Flexion Moment Pre-Swing (PSw)	$W_{140} = 0.977$, $P = 0.020$*
Extension Moment - Swing	$W_{140} = 0.979$, $P = 0.027$*
Knee Varus/Valgus	
Varus Moment (First Peak)	$W_{140} = 0.940$, $P = 0.000$*
Valgus Moment (First Peak)	$W_{140} = 0.985$, $P = 0.118$
Varus Moment (Second Peak)	$W_{140} = 0.980$, $P = 0.037$
Valgus Moment (Second Peak)	$W_{140} = 0.992$, $P = 0.583$
Ankle Plantar/Dorsiflexion	
Plantarflexion moment	$W_{140} = 0.851$, $P = 0.000$*
Dorsiflexion moment	$W_{140} = 0.866$, $P = 0.000$*
Powers (Watts/kg)	
Hip Power	
Hip Generation 1	$W_{140} = 0.902$, $P = 0.000$*
Hip Absorption 2	$W_{140} = 0.985$, $P = 0.120$
Hip Generation 3	$W_{140} = 0.979$, $P = 0.031$*
Knee Power	
Knee Generation 0	$W_{140} = 0.924$, $P = 0.000$*
Knee Absorption 1	$W_{140} = 0.941$ $P = 0.000$*
Knee Generation 2	$W_{140} = 0.952$, $P = 0.000$*
Knee Absorption 3	$W_{140} = 0.981$, $P = 0.043$*
Knee Generation 4	$W_{140} = 0.983$, $P = 0.087$
Ankle Power	
Ankle Absorption 1	$W_{140} = 0.965$, $P = 0.001$*
Ankle Generation 2	$W_{140} = 0.986$, $P = 0.168$

* parameter not normally distributed in accordance with the Shapiro-Wilk Test of Normality.

Table A6.1.2. Shapiro-Wilk Test of Normality for Chapter Four for the age groups.

Parameter	**55-64 yrs (n = 63)**	**65-74 yrs (n = 65)**	**≥ 75 yrs (n = 12)**
MMSE	W_{63} = 0.801, P = 0.000*	W_{65} = 0.746, P = 0.000*	W_{12} = 0.774, P = 0.005*
TUG	W_{63} = 0.960, P = 0.041*	W_{65} = 0.914, P = 0.000*	W_{12} = 0.888, P = 0.111
Rhythm			
Cadence (steps/min)	W_{63} = 0.984, P = 0.562	W_{65} = 0.985, P = 0.627	W_{12} = 0.854, P = 0.041*
Step Time (s)	W_{63} = 0.971, P = 0.149	W_{65} = 0.959, P = 0.031*	W_{12} = 0.958, P = 0.761
Stride Time (s)	W_{63} = 0.983, P = 0.530	W_{65} = 0.968, P = 0.094	W_{12} = 0.884, P = 0.099
Single-support Time (s)	W_{63} = 0.974, P = 0.198	W_{65} = 0.961, P = 0.041*	W_{12} = 0.955, P = 0.715
Phases			
Double-support Time (s)	W_{63} = 0.984, P = 0.612	W_{65} = 0.953, P = 0.016*	W_{12} = 0.915, P = 0.248
Foot-off (%)	W_{63} = 0.201, P = 0.000*	W_{65} = 0.954, P = 0.018*	W_{12} = 0.901, P = 0.162
Limp Index (s)	W_{63} = 0.977, P = 0.280	W_{65} = 0.902, P = 0.000*	W_{12} = 0.943, P = 0.537
Opposite Foot Contact (%)	W_{63} = 0.975, P = 0.237	W_{65} = 0.930, P = 0.001*	W_{12} = 0.967, P = 0.876
Opposite Foot-off (%)	W_{63} = 0.979, P = 0.341	W_{65} = 0.952, P = 0.014*	W_{12} = 0.962, P = 0.809
Pace			
Walking Speed (m·s^{-1})	W_{63} = 0.985, P = 0.660	W_{65} = 0.975, P = 0.216	W_{12} = 0.824, P = 0.018*
Step Length (m)	W_{63} = 0.980, P = 0.409	W_{65} = 0.948, P = 0.008*	W_{12} = 0.900, P = 0.159
Stride Length (m)	W_{63} = 0.987, P = 0.764	W_{65} = 0.965, P = 0.063	W_{12} = 0.884, P = 0.099
Base of Support			
Step Width (m)	W_{63} = 0.946, P = 0.008*	W_{65} = 0.951, P = 0.012*	W_{12} = 0.936, P = 0.445
Pelvic Tilt			
Range of Motion (RoM)	W_{63} = 0.981, P = 0.454	W_{65} = 0.986, P = 0.660	W_{12} = 0.959, P = 0.768
Maximum Tilt - Stance	W_{63} = 0.985, P = 0.280	W_{65} = 0.983, P = 0.515	W_{12} = 0.935, P = 0.430
Minimum Tilt - Stance	W_{63} = 0.983, P = 0.522	W_{65} = 0.984, P = 0.564	W_{12} = 0.965, P = 0.848
Maximum Tilt - Swing	W_{63} = 0.984, P = 0.594	W_{65} = 0.986, P = 0.648	W_{12} = 0.962, P = 0.809
Minimum Tilt - Swing	W_{63} = 0.979, P = 0.371	W_{65} = 0.985, P = 0.645	W_{12} = 0.971, P = 0.924

Parameter	**55-64 yrs (n = 63)**	**65-74 yrs (n = 65)**	**≥ 75 yrs (n = 12)**
Pelvic Obliquity			
RoM	W_{63} = 0.928, P = 0.001*	W_{65} = 0.979, P = 0.347	W_{12} = 0.890, P = 0.117
Maximum Obliquity - Stance	W_{63} = 0.973, P = 0.187	W_{65} = 0.975, P = 0.211	W_{12} = 0.924, P = 0.325
Minimum Obliquity - Stance	W_{63} = 0.925, P = 0.001*	W_{65} = 0.972, P = 0.145	W_{12} = 0.932, P = 0.399
Maximum Obliquity - Swing	W_{63} = 0.939, P = 0.004*	W_{65} = 0.951, P = 0.011*	W_{12} = 0.880, P = 0.087
Minimum Obliquity - Swing	W_{63} = 0.954, P = 0.020*	W_{65} = 0.983, P = 0.524	W_{12} = 0.887, P = 0.107
Pelvic Rotation			
RoM	W_{63} = 0.985, P = 0.647	W_{65} = 0.988, P = 0.802	W_{12} = 0.942, P = 0.521
Maximum Rotation - Stance	W_{63} = 0.988, P = 0.776	W_{65} = 0.988, P = 0.779	W_{12} = 0.980, P = 0.984
Minimum Rotation	W_{63} = 0.970, P = 0.123	W_{65} = 0.979, P = 0.344	W_{12} = 0.917, P = 0.260
Maximum Rotation - Swing	W_{63} = 0.985, P = 0.631	W_{65} = 0.977, P = 0.254	W_{12} = 0.902, P = 0.166
Hip Flexion/Extension			
RoM	W_{63} = 0.979, P = 0.363	W_{65} = 0.991, P = 0.925	W_{12} = 0.900, P = 0.160
Flexion - Stance	W_{63} = 0.949, P = 0.011*	W_{65} = 0.983, P = 0.491	W_{12} = 0.869, P = 0.064
Extension	W_{63} = 0.995, P = 0.996	W_{65} = 0.991, P = 0.933	W_{12} = 0.972, P = 0.929
Flexion - Swing	W_{63} = 0.975, P = 0.235	W_{65} = 0.984, P = 0.544	W_{12} = 0.840, P = 0.028*
Hip Abduction/Adduction			
RoM	W_{63} = 0.981, P = 0.447	W_{65} = 0.975, P = 0.199	W_{12} = 0.929, P = 0.375
Adduction - Stance	W_{63} = 0.984, P = 0.603	W_{65} = 0.988, P = 0.765	W_{12} = 0.962, P = 0.807
Adduction - Swing	W_{63} = 0.982, P = 0.480	W_{65} = 0.987, P = 0.726	W_{12} = 0.957, P = 0.747
Abduction - Swing	W_{63} = 0.984, P = 0.602	W_{65} = 0.975, P = 0.213	W_{12} = 0.968, P = 0.892
Hip Rotation			
RoM	W_{63} = 0.965, P = 0.073	W_{65} = 0.978, P = 0.283	W_{12} = 0.980, P = 0.983
Maximum Rotation - Stance	W_{63} = 0.885, P = 0.000*	W_{65} = 0.985, P = 0.595	W_{12} = 0.936, P = 0.443
Minimum Rotation - Stance	W_{63} = 0.971, P = 0.137	W_{65} = 0.954, P = 0.016*	W_{12} = 0.933, P = 0.412
Maximum Rotation - Swing	W_{63} = 0.947, P = 0.009*	W_{65} = 0.995, P = 0.995	W_{12} = 0.973, P = 0.937
Minimum Rotation - Swing	W_{63} = 0.986, P = 0.713	W_{65} = 0.979, P = 0.345	W_{12} = 0.967, P = 0.878

Parameter	**55-64 yrs (n = 63)**	**65-74 yrs (n = 65)**	**≥ 75 yrs (n = 12)**
Knee Flexion/Extension			
RoM	W_{63} = 0.988, P = 0.791	W_{65} = 0.981, P = 0.402	W_{12} = 0.964, P = 0.832
Flexion at Loading Response (LR)	W_{63} = 0.981, P = 0.421	W_{65} = 0.978, P = 0.289	W_{12} = 0.891, P = 0.123
Extension at Terminal Stance (TS)	W_{63} = 0.970, P = 0.120	W_{65} = 0.931, P = 0.001*	W_{12} = 0.917, P = 0.262
Flexion - Swing	W_{63} = 0.986, P = 0.703	W_{65} = 0.979, P = 0.324	W_{12} = 0.860, P = 0.049*
Knee Varus/Valgus			
RoM	W_{63} = 0.912, P = 0.000*	W_{65} = 0.896, P = 0.000*	W_{12} = 0.767, P = 0.004*
Maximum Varus - Stance	W_{63} = 0.978, P = 0.311	W_{65} = 0.961, P = 0.037*	W_{12} = 0.972, P = 0.931
Minimum Valgus - Stance	W_{63} = 0.832, P = 0.000*	W_{65} = 0.890, P = 0.000*	W_{12} = 0.929, P = 0.371
Maximum Varus - Swing	W_{63} = 0.878, P = 0.000*	W_{65} = 0.934, P = 0.002*	W_{12} = 0.973, P = 0.936
Minimum Valgus - Swing	W_{63} = 0.878, P = 0.000*	W_{65} = 0.850, P = 0.000*	W_{12} = 0.806, P = 0.011*
Moments (Nm/kg)			
Hip Flexion/Extension			
Flexion Moment - Stance	W_{63} = 0.978, P = 0.315	W_{65} = 0.966, P = 0.072	W_{12} = 0.880, P = 0.088
Extension Moment	W_{63} = 0.916, P = 0.000*	W_{65} = 0.929, P = 0.001*	W_{12} = 0.908, P = 0.203
Flexion Moment - Swing	W_{63} = 0.993, P = 0.969	W_{65} = 0.925, P = 0.001*	W_{12} = 0.902, P = 0.170
Hip Abduction/Adduction			
Maximum Moment (First Peak)	W_{63} = 0.960, P = 0.040*	W_{65} = 0.988, P = 0.780	W_{12} = 0.955, P = 0.707
Minimum Moment (First Peak)	W_{63} = 0.851, P = 0.000*	W_{65} = 0.877, P = 0.000*	W_{12} = 0.938, P = 0.472
Maximum Moment (Second Peak)	W_{63} = 0.966, P = 0.075	W_{65} = 0.975, P = 0.211	W_{12} = 0.920, P = 0.287
Minimum Moment (Second Peak)	W_{63} = 0.841, P = 0.000*	W_{65} = 0.885, P = 0.000*	W_{12} = 0.896, P = 0.141
Knee Flexion/Extension			
Flexion Moment (LR)	W_{63} = 0.957, P = 0.026*	W_{65} = 0.962, P = 0.045*	W_{12} = 0.923, P = 0.308
Extension Moment (TS)	W_{63} = 0.984, P = 0.590	W_{65} = 0.983, P = 0.512	W_{12} = 0.957, P = 0.742
Flexion Moment Pre-Swing (PSw)	W_{63} = 0.970, P = 0.126	W_{65} = 0.968, P = 0.095	W_{12} = 0.922, P = 0.301
Extension Moment - Swing	W_{63} = 0.967, P = 0.087	W_{65} = 0.978, P = 0.312	W_{12} = 0.926, P = 0.341

Parameter	55-64 yrs (n = 63)	65-74 yrs (n = 65)	≥ 75 yrs (n = 12)
Knee Varus/Valgus			
Varus Moment (First Peak)	W_{63} = 0.934, P = 0.002*	W_{65} = 0.920, P = 0.000*	W_{12} = 0.938, P = 0.468
Valgus Moment (First Peak)	W_{63} = 0.978, P = 0.322	W_{65} = 0.994, P = 0.985	W_{12} = 0.813, P = 0.013*
Varus Moment (Second Peak)	W_{63} = 0.984, P = 0.602	W_{65} = 0.971, P = 0.127	W_{12} = 0.936, P = 0.449
Valgus Moment (Second Peak)	W_{63} = 0.986, P = 0.667	W_{65} = 0.985, P = 0.609	W_{12} = 0.956, P = 0.731
Ankle Plantar/Dorsiflexion			
Plantarflexion moment	W_{63} = 0.981, P = 0.454	W_{65} = 0.776, P = 0.000*	W_{12} = 0.955, P = 0.715
Dorsiflexion moment	W_{63} = 0.841, P = 0.000*	W_{65} = 0.853, P = 0.000*	W_{12} = 0.941, P = 0.515
Ankle Plantar/Dorsiflexion			
RoM	W_{63} = 0.986, P = 0.688	W_{65} = 0.982, P = 0.462	W_{12} = 0.951, P = 0.646
Plantarflexion (LR)	W_{63} = 0.981, P = 0.427	W_{65} = 0.989, P = 0.849	W_{12} = 0.948, P = 0.603
Dorsiflexion - Stance	W_{63} = 0.953, P = 0.018*	W_{65} = 0.853, P = 0.000*	W_{12} = 0.896, P = 0.140
Maximum Plantarflexion	W_{63} = 0.907, P = 0.000*	W_{65} = 0.854, P = 0.000*	W_{12} = 0.926, P = 0.335
Powers (Watts/kg)			
Hip Power			
Hip Generation 1	W_{63} = 0.903, P = 0.000*	W_{65} = 0.907, P = 0.000*	W_{12} = 0.767, P = 0.004*
Hip Absorption 2	W_{63} = 0.969, P = 0.111	W_{65} = 0.991, P = 0.916	W_{12} = 0.908, P = 0.200
Hip Generation 3	W_{63} = 0.961, P = 0.042*	W_{65} = 0.964, P = 0.053	W_{12} = 0.939, P = 0.489
Knee Power			
Knee Generation 0	W_{63} = 0.930, P = 0.002*	W_{65} = .923, P = .001*	W_{12} = 0.920, P = 0.288
Knee Absorption 1	W_{63} = 0.977, P = 0.283	W_{65} = .896, P = .000*	W_{12} = 0.881, P = 0.091
Knee Generation 2	W_{63} = 0.962, P = 0.050	W_{65} = .939, P = .003*	W_{12} = 0.691, P = 0.001*
Knee Absorption 3	W_{63} = 0.972, P = 0.156	W_{65} = .977, P = .253	W_{12} = 0.940, P = 0.503
Knee Generation 4	W_{63} = 0.922, P = 0.001*	W_{65} = .974, P = .177	W_{12} = 0.885, P = 0.100
Ankle Power			
Ankle Absorption 1	W_{63} = 0.992, P = 0.951	W_{65} = 0.909, P = 0.000*	W_{12} = 0.968, P = 0.887
Ankle Generation 2	W_{63} = 0.983, P = 0.516	W_{65} = 0.984, P = 0.583	W_{12} = 0.932, P = 0.407

* parameter not normally distributed in accordance with the Shapiro-Wilk Test of Normality.

A6.2. Chapter Five

Table A6.2. Shapiro-Wilk Test of Normality for Chapter Five.

Parameter	Shapiro-Wilk Test of Normality
Age	$W_{129} = 0.972$, $P = 0.009$*
Normal Walking	
Walking Speed	$W_{129} = 0.987$, $P = 0.284$
Toe-Clearance	
MxT1	$W_{129} = 0.979$, $P = 0.047$*
MTC	$W_{129} = 0.964$, $P = 0.002$*
MxT2	$W_{129} = 0.992$, $P = 0.689$
Ipsilateral Limb	
Hip Extension	$W_{129} = 0.990$, $P = 0.444$
Hip Flexion at MxT1	$W_{129} = 0.994$, $P = 0.831$
Hip Flexion at MTC	$W_{129} = 0.991$, $P = 0.571$
Hip Flexion at MxT2	$W_{129} = 0.988$, $P = 0.297$
Hip Adduction in Stance	$W_{129} = 0.989$, $P = 0.370$
Hip Abduction/Adduction at MxT1	$W_{129} = 0.993$, $P = 0.806$
Hip Abduction/Adduction at MTC	$W_{129} = 0.989$, $P = 0.417$
Hip Abduction/Adduction at MxT2	$W_{129} = 0.994$, $P = 0.838$
Knee Extension at Terminal Stance	$W_{129} = 0.987$, $P = 0.281$
Knee Flexion in Swing	$W_{129} = 0.982$, $P = 0.083$
Knee Flexion at MxT1	$W_{129} = 0.985$, $P = 0.171$
Knee Flexion at MTC	$W_{129} = 0.985$, $P = 0.161$
Knee Flexion at MxT2	$W_{129} = 0.990$, $P = 0.442$
Ankle Dorsiflexion	$W_{129} = 0.971$, $P = 0.008$*
Ankle Plantarflexion	$W_{129} = 0.870$, $P = 0.000$*
Ankle Plantar/Dorsiflexion at MxT1	$W_{129} = 0.966$, $P = 0.002$*
Ankle Plantar/Dorsiflexion at MTC	$W_{129} = 0.852$, $P = 0.000$*
Ankle Plantar/Dorsiflexion at MxT2	$W_{129} = 0.973$, $P = 0.010$*
Contralateral Limb	
Hip Flexion/Extension at MxT1	$W_{129} = 0.991$, $P = 0.553$
Hip Flexion/Extension at MTC	$W_{129} = 0.988$, $P = 0.330$
Hip Flexion/Extension at MxT1	$W_{129} = 0.947$, $P = 0.000$*
Hip Adduction in Stance	$W_{129} = 0.978$, $P = 0.035$*
Hip Abduction/Adduction at MxT1	$W_{129} = 0.973$, $P = 0.012$*
Hip Abduction/Adduction at MTC	$W_{129} = 0.981$, $P = 0.064$
Hip Abduction/Adduction at MxT2	$W_{129} = 0.989$, $P = 0.372$
Knee Flexion at Loading Response	$W_{129} = 0.993$, $P = 0.738$
Knee Flexion at MxT1	$W_{129} = 0.829$, $P = 0.000$*
Knee Flexion at MTC	$W_{129} = 0.794$, $P = 0.000$*
Knee Flexion at MxT2	$W_{129} = 0.957$, $P = 0.000$*

Parameter	Shapiro-Wilk Test of Normality
Ankle Dorsiflexion in Stance	$W_{129} = 0.983$, $P = 0.098$
Ankle Plantar/Dorsiflexion at MxT1	$W_{129} = 0.872$, $P = 0.000$*
Ankle Plantar/Dorsiflexion at MTC	$W_{129} = 0.840$, $P = 0.000$*
Ankle Plantar/Dorsiflexion at MxT2	$W_{129} = 0.772$, $P = 0.000$*
Dual Task Walking	
Walking Speed	$W_{129} = 0.993$, $P = 0.742$
Toe-Clearance	
MxT1	$W_{129} = 0.958$, $P = 0.000$*
MTC	$W_{129} = 0.979$, $P = 0.044$*
MxT2	$W_{129} = 0.990$, $P = 0.437$
Ipsilateral Joint Kinematics	
Hip Extension	$W_{129} = 0.995$, $P = 0.951$
Hip Flexion at MxT1	$W_{129} = 0.994$, $P = 0.898$
Hip Flexion at MTC	$W_{129} = 0.996$, $P = 0.983$
Hip Flexion at MxT2	$W_{129} = 0.991$, $P = 0.560$
Hip Adduction in Stance	$W_{129} = 0.990$, $P = 0.473$
Hip Abduction/Adduction at MxT1	$W_{129} = 0.992$, $P = 0.695$
Hip Abduction/Adduction at MTC	$W_{129} = 0.994$, $P = 0.854$
Hip Abduction/Adduction at MxT2	$W_{129} = 0.993$, $P = 0.815$
Knee Extension at Terminal Stance	$W_{129} = 0.983$, $P = 0.097$
Knee Flexion in Swing	$W_{129} = 0.962$, $P = 0.001$*
Knee Flexion at MxT1	$W_{129} = 0.961$, $P = 0.001$*
Knee Flexion at MTC	$W_{129} = 0.989$, $P = 0.406$
Knee Flexion at MxT2	$W_{129} = 0.993$, $P = 0.791$
Ankle Dorsiflexion	$W_{129} = 0.934$, $P = 0.000$*
Ankle Plantarflexion	$W_{129} = 0.869$, $P = 0.000$*
Ankle Plantar/Dorsiflexion at MxT1	$W_{129} = 0.974$, $P = 0.013$*
Ankle Plantar/Dorsiflexion at MTC	$W_{129} = 0.873$, $P = 0.000$*
Ankle Plantar/Dorsiflexion at MxT2	$W_{129} = 0.805$, $P = 0.000$*
Contralateral Limb	
Hip Flexion/Extension at MxT1	$W_{129} = 0.984$, $P = 0.137$
Hip Flexion/Extension at MTC	$W_{129} = 0.987$, $P = 0.247$
Hip Flexion/Extension at MxT1	$W_{129} = 0.983$, $P = 0.113$
Hip Adduction in Stance	$W_{129} = 0.979$, $P = 0.039$*
Hip Abduction/Adduction at MxT1	$W_{129} = 0.970$, $P = 0.006$*
Hip Abduction/Adduction at MTC	$W_{129} = 0.991$, $P = 0.539$
Hip Abduction/Adduction at MxT2	$W_{129} = 0.993$, $P = 0.728$
Knee Flexion at Loading Response	$W_{129} = 0.991$, $P = 0.557$
Knee Flexion at MxT1	$W_{129} = 0.991$, $P = 0.566$
Knee Flexion at MTC	$W_{129} = 0.982$, $P = 0.092$
Knee Flexion at MxT2	$W_{129} = 0.971$, $P = 0.007$*

Parameter	Shapiro-Wilk Test of Normality
Ankle Dorsiflexion in Stance	$W_{129} = 0.949$, $P = 0.000$*
Ankle Plantar/Dorsiflexion at MxT1	$W_{129} = 0.898$, $P = 0.000$*
Ankle Plantar/Dorsiflexion at MTC	$W_{129} = 0.901$, $P = 0.000$*
Ankle Plantar/Dorsiflexion at MxT2	$W_{129} = 0.795$, $P = 0.000$*
Age Groups	
Normal Walking	
MxT1	
55-64 yrs	$W_{54} = 0.985$, $P = 0.712$
65-74 yrs	$W_{61} = 0.955$, $P = 0.026$*
≥ 75 yrs	$W_{14} = 0.958$, $P = 0.692$
MTC	
55-64 yrs	$W_{54} = 0.929$, $P = 0.003$*
65-74 yrs	$W_{61} = 0.981$, $P = 0.478$
≥ 75 yrs	$W_{14} = 0.986$, $P = 0.997$
MxT2	
55-64 yrs	$W_{54} = 0.972$, $P = 0.246$
65-74 yrs	$W_{61} = 0.978$, $P = 0.351$
≥ 75 yrs	$W_{14} = 0.942$, $P = 0.450$
Walking Speed	
55-64 yrs	$W_{54} = 0.974$, $P = 0.282$
65-74 yrs	$W_{61} = 0.965$, $P = 0.077$
≥ 75 yrs	$W_{14} = 0.903$, $P = 0.127$
Dual Task Walking	
MxT1	
55-64 yrs	$W_{54} = 0.963$, $P = 0.098$
65-74 yrs	$W_{61} = 0.935$, $P = 0.003$*
≥75 yrs	$W_{14} = 0.969$, $P = 0.856$
MTC	
55-64 yrs	$W_{54} = 0.965$, $P = 0.113$
65-74 yrs	$W_{61} = 0.988$, $P = 0.804$
≥ 75 yrs	$W_{14} = 0.943$, $P = 0.461$
MxT2	
55-64 yrs	$W_{54} = 0.955$, $P = 0.043$*
65-74 yrs	$W_{61} = 0.975$, $P = 0.244$
≥ 75 yrs	$W_{14} = 0.847$, $P = 0.020$*
Walking Speed	
55-64 yrs	$W_{54} = 0.983$, $P = 0.634$
65-74 yrs	$W_{61} = 0.986$, $P = 0.726$
≥ 75 yrs	$W_{14} = 0.950$, $P = 0.558$

Parameter	Shapiro-Wilk Test of Normality
Fall History	
Older Adults with Fall History	
Normal Walking	
MxT1	W_{25} = 0.968, P = 0.596
MTC	W_{25} = 0.973, P = 0.719
MxT2	W_{25} = 0.941, P = 0.158
Dual Task Walking	
MxT1	W_{25} = 0.936, P = 0.122
MTC	W_{25} = 0.926, P = 0.069
MxT2	W_{25} = 0.913, P = 0.035*
Older Adults without Fall History	
Normal Walking	
MxT1	W_{104} = 0.974, P = 0.039*
MTC	W_{104} = 0.955, P = 0.001*
MxT2	W_{104} = 0.988, P = 0.459
Dual Task Walking	
MxT1	W_{104} = 0.953, P = 0.001*
MTC	W_{104} = 0.995, P = 0.982
MxT2	W_{104} = 0.993, P = 0.843

* parameter not normally distributed in accordance with the Shapiro-Wilk Test of Normality.

A6.3. <u>Chapter Six</u>

Table A6.3. Shapiro-Wilk Test of Normality for Chapter Six.

Parameter	Shapiro-Wilk Test of Normality
Age	W_{98} = 0.972, P = 0.037*
Hand-Grip	
Dominant	W_{98} = 0.845, P = 0.000*
Non-dominant	W_{98} = 0.913, P = 0.000*
Walking Speed	
NW	W_{98} = 0.979, P = 0.126
DT	W_{98} = 0.984, P = 0.305
SON	W_{98} = 0.985, P = 0.355
SOV	W_{98} = 0.993, P = 0.886
Arm Swing	
Dominant Arm	
NW	W_{98} = 0.991, P = 0.788
DT	W_{98} = 0.487, P = 0.000*
SON	W_{98} = 0.977, P = 0.090
SOV	W_{98} = 0.974, P = 0.053
Non-dominant Arm	
NW	W_{98} = 0.986, P = 0.433
DT	W_{98} = 0.945, P = 0.001*
SON	W_{98} = 0.986, P = 0.373
SOV	W_{98} = 0.983, P = 0.250
Forearm Swing	
Dominant Arm	
NW	W_{98} = 0.964, P = 0.010*
SON	W_{98} = 0.977, P = 0.096
SOV	W_{98} = 0.943, P = 0.000*
Non-dominant Arm	
NW	W_{98} = 0.979, P = 0.137
DT	W_{98} = 0.932, P = 0.000*
SON	W_{98} = 0.993, P = 0.922
SOV	W_{98} = 0.980, P = 0.155

* parameter not normally distributed in accordance with the Shapiro-Wilk Test of Normality.

A6.4. Chapter Seven

Table A6.4. Shapiro-Wilk Test of Normality for Chapter Seven.

Parameter	NW	SON	SOV
Walking Speed ($m \cdot s^{-1}$)	W_{85} = 0.988, P = 0.626	W_{85} = 0.982, P = 0.283	W_{85} = 0.974, P = 0.088
MMSE	W_{85} = 0.779, P = 0.000*	W_{85} = 0.779, P = 0.000*	W_{85} = 0.779, P = 0.000*
Fz Peak Force			
F1	W_{85} = 0.884, P = 0.000*	W_{85} = 0.884, P = 0.000*	W_{85} = 0.955, P = 0.005*
F 2	W_{85} = 0.980, P = 0.219	W_{85} = 0.980, P = 0.219	W_{85} = 0.919, P = 0.000*
F 3	W_{85} = 0.969, P = 0.041*	W_{85} = 0.969, P = 0.041*	W_{85} = 0.965, P = 0.022*
Fy Peak Force			
F 4	W_{85} = 0.987, P = 0.547	W_{85} = 0.987, P = 0.547	W_{85} = 0.988, P = 0.612
F 5	W_{85} = 0.982, P = 0.277	W_{85} = 0.982, P = 0.277	W_{85} = 0.941, P = 0.001*
Fz Impulse	W_{85} = 0.957, P = 0.006*	W_{85} = 0.955, P = 0.005*	W_{85} = 0.976, P = 0.114
Braking Impulse	W_{85} = 0.833, P = 0.000*	W_{85} = 0.620, P = 0.000*	W_{85} = 0.844, P = 0.000*
Propulsive Impulse	W_{85} = 0.790, P = 0.000*	W_{85} = 0.854, P = 0.000*	W_{85} = 0.835, P = 0.000*

Parameter	NW	SON	SOV
Moments (Nm/kg)			
Hip Flexion/Extension			
Flexion Moment - Stance	W_{85} = 0.979, P = 0.194	W_{85} = 0.789, P = 0.000*	W_{85} = 0.995, P = 0.983
Extension Moment	W_{85} = 0.931, P = 0.000*	W_{85} = 0.959, P = 0.010*	W_{85} = 0.992, P = 0.896
Flexion Moment – Swing	W_{85} = 0.975, P = 0.097	W_{85} = 0.782, P = 0.000*	W_{85} = 0.953, P = 0.004*
Hip Abduction/Adduction			
Maximum Moment (First Peak)	W_{85} = 0.963, P = 0.017*	W_{85} = 0.972, P = 0.068	W_{85} = 0.990, P = 0.744
Minimum Moment (First Peak)	W_{85} = 0.980, P = 0.204	W_{85} = 0.983, P = 0.352	W_{85} = 0.927, P = 0.000*
Maximum Moment (Second Peak)	W_{85} = 0.986, P = 0.528	W_{85} = 0.991, P = 0.804	W_{85} = 0.968, P = 0.037*
Minimum Moment (Second Peak)	W_{85} = 0.909, P = 0.000*	W_{85} = 0.957, P = 0.006*	W_{85} = 0.954, P = 0.005*
Knee Flexion/Extension			
Flexion Moment (LR)	W_{85} = 0.957, P = 0.007*	W_{85} = 0.986, P = 0.485	W_{85} = 0.978, P = 0.149
Extension Moment (TS)	W_{85} = 0.966, P = 0.027*	W_{85} = 0.904, P = 0.000*	W_{85} = 0.992, P = 0.873
Flexion Moment Pre-Swing (PSw)	W_{85} = 0.963, P = 0.017*	W_{85} = 0.991, P =0.822	W_{85} = 0.984, P = 0.393
Extension Moment - Swing	W_{85} = 0.969, P = 0.042*	W_{85} = 0.967, P = 0.032*	W_{85} = 0.973, P = 0.074
Knee Varus/Valgus			
Varus Moment (First Peak)	W_{85} = 0.952, P = 0.003*	W_{85} = 0.912, P = 0.000*	W_{85} = 0.959, P = 0.009*
Valgus Moment (First Peak)	W_{85} = 0.992, P = 0.884	W_{85} = 0.821, P = 0.000*	W_{85} = 0.858, P = 0.000*
Varus Moment (Second Peak)	W_{85} = 0.977, P = 0.146	W_{85} = 0.980, P = 0.224	W_{85} = 0.987, P = 0.548
Valgus Moment (Second Peak)	W_{85} = 0.988, P = 0.602	W_{85} = 0.892, P = 0.000*	W_{85} = 0.938, P = 0.335
Ankle Plantar/Dorsiflexion			
Plantarflexion moment	W_{85} = 0.795, P = 0.000*	W_{85} = 0.894, P = 0.000*	W_{85} = 0.938, P = 0.001*
Dorsiflexion moment	W_{85} = 0.832, P = 0.000*	W_{85} = 0.900, P = 0.000*	W_{85} = 0.893, P = 0.000*

Parameter	NW	SON	SOV
Powers (Watts/kg)			
Hip Power			
Hip Generation 1	W_{85} = 0.895, P = 0.000*	W_{85} = 0.801, P = 0.000*	W_{85} = 0.923, P = 0.000*
Hip Absorption 2	W_{85} = 0.974, P = 0.084	W_{85} = 0.917, P = 0.000*	W_{85} = 0.987, P = 0.584
Hip Generation 3	W_{85} = 0.967, P = 0.030*	W_{85} = 0.977, P = 0.144	W_{85} = 0.948, P = 0.002*
Knee Power			
Knee Generation 0	W_{85} = 0.947, P = 0.002*	W_{85} = 0.940, P = 0.001*	W_{85} = 0.682, P = 0.000*
Knee Absorption 1	W_{85} = 0.928, P = 0.000*	W_{85} = 0.911, P = 0.000*	W_{85} = 0.896, P = 0.000*
Knee Generation 2	W_{85} = 0.943, P = 0.001*	W_{85} = 0.984, P = 0.366	W_{85} = 0.858, P = 0.000*
Knee Absorption 3	W_{85} = 0.974, P = 0.082	W_{85} = 0.976, P = 0.109	W_{85} = 0.975, P = 0.094
Knee Generation 4	W_{85} = 0.963, P = 0.016*	W_{85} = 0.983, P = 0.333	W_{85} = 0.966, P = 0.024*
Ankle Power			
Ankle Absorption 0	W_{85} = 0.973, P = 0.075	W_{85} = 0.985, P = 0.447	W_{85} = 0.848, P = 0.000*
Ankle Absorption 1	W_{85} = 0.954, P = 0.005*	W_{85} = 0.990, P = 0.754	W_{85} = 0.926, P = 0.000*
Ankle Generation 2	W_{85} = 0.978, P = 0.159	W_{85} = 0.981, P = 0.257	W_{85} = 0.981, P = 0.256

* parameter not normally distributed in accordance with the Shapiro-Wilk Test of Normality.