University of Strathclyde

Bioengineering Unit

A biomechanical analysis of stair ascent and descent in older adults

by

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Declaration of Authenticity and Author's Rights

This thesis is the result of the author's original research. It has been composed by the author and has not been previously submitted for examination which has led to the award of a degree.

Signed: Date:

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Abbreviations

AP – Anterior posterior

AJC – Ankle joint centre

COM – Centre of mass

COP – Centre of pressure

GRF – Ground reaction force

HJC – Hip joint centre

KJC – Knee joint centre

ML - Mediolateral

ROM – Range of movement

Abstract

The population of the UK is ageing and is set to continue to do so for many years. In order to enable older adults to live independently in their own homes it is essential to understand the challenges of activities of daily living, so that designers can plan suitable environments and rehabilitation professionals can know how best to assist older adults who experience difficulties.

Many older adults experience difficulties negotiating stairs and falls on stairs often lead to hospitalization. To date, little research has been undertaken to explore the biomechanical demands of stair climbing and descent, and most of the literature has focussed on younger adults. In order to provide biomechanical data relevant to an ageing population, this research investigated 84 older adults performing stair ascent and descent. The subjects were divided into three age bands, 60+, 70+ and 80+ in order to assess changes related to increasing older age.

Data were collected using an 8 camera VICON system with a custom built staircase enabling forces to be recorded from 2 Kistler force platforms. A full body biomechanical model was developed to comply with the best practice standards using VICON bodybuilder. Temporal data, joint kinematics and kinetics were produced for a full gait cycle and reported on for each age category. Subjects performed the activity with and without a handrail to explore how handrails may be of assistance.

Adults in the oldest age group were found to have biomechanical changes in both stair ascent and descent. The key findings were a redistribution of joint kinetics, reducing the demands at the ankle joint and increasing the demands on the hip and knee extensors. This strategy optimises muscles groups where there are greater strength reserves in older adults. Use of a handrail improved stability and reduced the demands on the lower limbs.

Chapter 1

Review of the Literature on Ageing and the Effects on Biomechanics

This chapter will present literature relating to the ageing process. The physical and cognitive changes with age will be discussed, and the impact on these on the ability of older adults to undertake activities of daily living. The use of biomechanical analysis in performance of activities of daily living will be presented and the rationale for the current research study.

1.1 Ageing statistics

Throughout the 20th century the proportion of people aged 60 or older has increased in all countries of the world (Bond and Coleman, 1990). This trend is likely to continue as advances in technology lead to improved health and quality of life, the Organisation for Economic Co-operation and Development (OECD) predicts continued increase in life expectancy in all of its 27 member countries until the year 2030 (OECD, 2000). In 2000 the number of adults aged 65 and older in the UK accounted for 15.8% of the total population, by 2005 this became 16% and by 2020 it is predicted to be 19% (OECD, 2009). The increase in the number of very old (over 80) is much sharper, rising from 2% of the population in 1960 to 6.5% in 2030 (OECD, 2000). The effect of population increase across different age bands is shown in table 1.1 and graphically illustrated in figure 1.1

Table 1.1 Actual and projected populations by age, United Kingdom, 1998-2021

Age Group	1998	2001	2006	2011	2016	2021

0-14	11,380	11,294	10,890	10,594	10,541	10,603
15-29	11,580	11,358	11,629	11,936	11,837	11,438
30-44	13,347	13,854	13,538	12,477	11,845	12,118
45-59	10,820	11,247	11,993	12,734	13,579	13,273
60-74	7,783	7,767	8,231	9,300	9,920	10,642
75+	4,237	4,434	4,579	4,731	5,007	5,568
All ages	59,237	59,954	60,860	61,773	62,729	63,642

Reproduced from National Statistics (2000), National Population Projections 1998-based

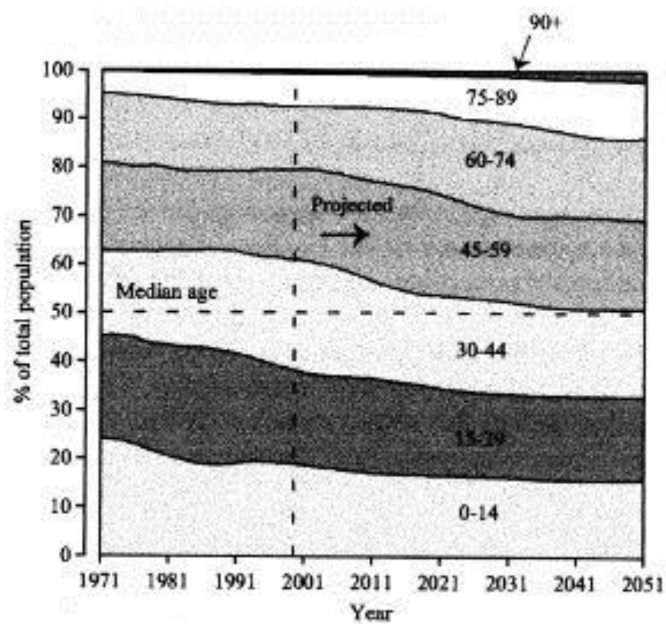


Figure 1.1 Projected age distributions United Kingdom 1971-2051
From National Statistics (2000), National Population Projections 1998-based

The majority of people over the age of 65 remain independent, living in their own home. Information collected for the Scottish Health Statistics (1999) found 18,498 over 65‟s to be resident in nursing homes, which accounted for just 2.4% of this age group. A further 16,677 (2.1% of over 65‟s) were living in residential care homes and 35,927 (4.6%) in sheltered housing. A larger percentage of the population will experience institutional living as a temporary measure and this increases with physical and mental frailty (Bond and Coleman, 1990). At the time of the 1981 census, 9.6% of over 75‟s and

21.6% of over 85‟s were resident in an institution on the night of the census but 40% of these people were not normally resident in that facility (Office of Population, 1993). Statistics produced by Age Concern (AgeConcern, 2000) found the chance of living in a long-stay hospital or care home in 2000 was 0.05% for under 65‟s, 1% for 65-74 year olds, 4.9% for 75-84 year olds and 21.1% for those over the age of 85.

It can therefore be concluded that although the population is ageing, the majority of individuals will remain living in their own home or with families, with only a small percentage of the population requiring nursing home care, mostly at the latter stages of life. As more private dwellings become occupied by older people it is important for designers and those in healthcare to develop an understanding of the specific needs and problems experienced by the older person in order to help maintain independence and quality of life.

1.2 Physical effects of ageing

One of the debates within the field of gerontology is how to differentiate the effects of disease from those caused by the normal ageing process. Ageing is considered to be associated with a decline in physiological effectiveness, which is intrinsic, affecting everyone (Bond and Coleman, 1990). Disease represents a condition in which functions are disturbed in comparison to a normal reference condition (Forbes and Hirdes, 1993), and does not affect all of a population. However there are many diseases that are age related and are seen rarely in young people, such as stroke, arteriosclerosis or dementia and debate exists as to whether these should be seen as part of the ageing process (Goodwin, 1991) or as a separate entity (Shock, 1961).

Whatever the viewpoint of ageing or disability it is widely accepted that physical decrements do become more probable with age (Powell Lawton, 1990), with older subjects having an increased number of impairments. This section of the review will focus specifically on the age related impairments of interest to biomechanics, particularly how physical parameters change with age.

1.2.1 Anthropometry

Anthropometry is the measure of the body in fixed positions and was initiated by the military, in order to optimise the fit of equipment to soldiers. Following the Second World War the emphasis was shifted to the civilian population, and more recently the increase in elderly people resulting in a dominant consumer group has created a need for work in this area (Rogers et al, 1996). The elderly have been recognised as a distinct group anthropometrically and are included separately in commercial computer databases such as ERGOBASE ™ and PeopleSIZE ™ and anthropometric publications such as OLDER ADULTDATA (DTI, 2000).

Height has been widely reported to decrease with age from age 35 onwards as a result of vertebral compression, loss of vertebral disc height, loss of muscle tone and postural changes (OLDER ADULTDATA, 2000). Schultz (1992) reviewed previous literature and reported a 3% decrease in height between subjects aged 18-24 and those aged 65-74. Weight is more variable with age, Vitasalo (1985) reporting lowest body weight in male subjects aged 31-35, highest in those aged 51-55 and intermediate in the elderly aged 7175. In females this trend has also been recorded (OLDER ADULTDATA, 2000), but with maximum weight tending to occur around 10 years later than men and then declining in older age.

One aspect to consider when reviewing anthropometric data is the effect caused by secular trends. An increase in adult stature is continuing in many countries and is considered to be an indicator of the change in nutrition, hygiene and health status of a population (ADULTDATA, 1998). It is therefore only in longitudinal studies that the changes in body size as a result of the ageing process can be determined and it may be found that future elderly populations will increase in size following the secular trend.

1.2.2 Joint range of motion

The range of motion (ROM) of body joints generally diminishes with age (Schultz, 1992) though this does not necessarily lead to functional limitation (Bergstrom et al, 1985). Desrosiers et al. (1995) investigated shoulder joint range of motion in 360 subjects aged 60 and over. They found a statistically significant age related decrease in flexion and abduction in both male and female subjects between the three age groups studies (60-69, 70-79 and 80-94). Fiebert et al (1995) found shoulder range of motion decreased in a linear fashion by decade from the sixth to ninth decade in a group of 102 healthy volunteers aged over 60. The results of these two studies are shown in figure 1.2. There was also a statistically significant decrease in shoulder ROM compared to the standards produced by the American Academy of Orthopaedic Surgeons (AAOS) representative of a normal adult population.

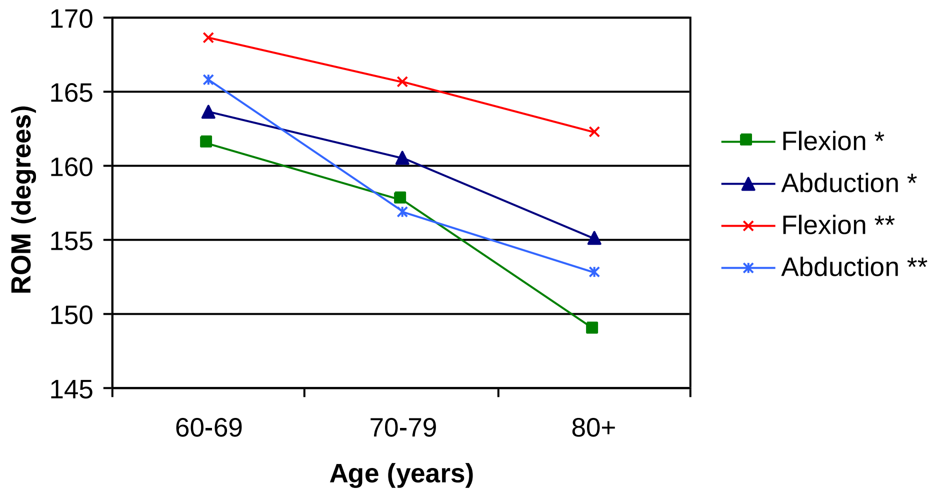


Figure 1.2 Change in shoulder joint range of motion with age * Desrosiers et al. (1995) ** Fiebert et al. (1995)

Steenbekkers (1998) investigated ROM for a wide selection of joints in adults and found a significant age related decrease in mobility for the majority of joints. They found that there were also sex effects on joint mobility. In young subjects, women tend to have greater ROM than men, but in the elderly the pattern of ROM loss is variable with joint and sex.

Bergström et al (1985) measured ROM of the back and peripheral joints in 81 subjects aged 79. They found a prevalence of restriction of range of motion ranging from 17% for the knee joint to 72% for the thoracolumbar spine (table 1.2). For the majority

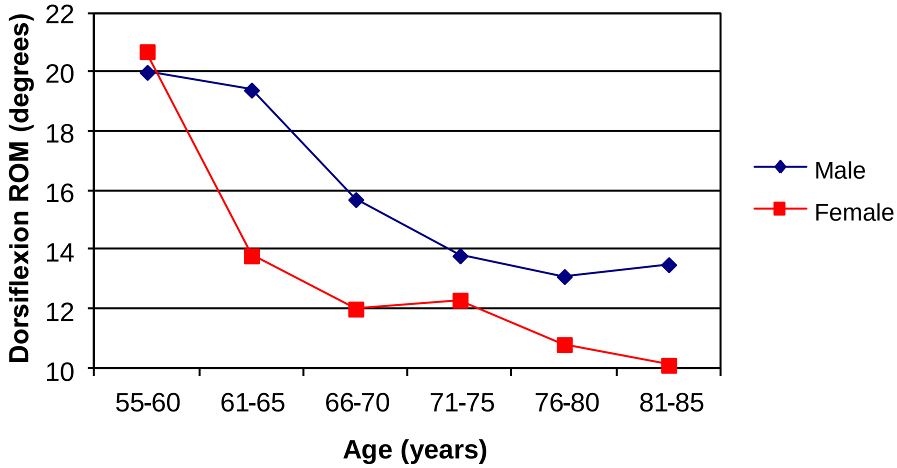
of the subjects this restriction did not result in a joint complaint, the frequency of joint complaints ranged from 3% for the ankle to 32% for the back. The authors concluded that although a considerable proportion of their sample group had some restriction of

ROM compared to younger subjects, that in the majority this was limited and could be ascribed as a stiffness of the locomotor system either due to the change in elastic tissue components or due to age related changes in behaviour and physical activity.

Table 1.2 Prevalence of restricted range of motion (ROM) in 79 year olds
From Bergström et al. (1985)

	Females n=54 (%)	Males n=37 (%)	Total n=89 (%)
Cervical spine	40	43	41
Thoracolumbar spine	63	84	72
Shoulders	31	46	37
Wrists	50	51	51
Fingers	31	19	26
Hips	67	57	63
Knees	21	11	17

Vandervoort et al (1992) investigated the effect of age and sex on mobility of the ankle joint (figure 1.3). ROM of ankle dorsiflexion decreased from 20 to 13.5 degrees from age 55-60 and age 81-85 in males and from 20.7 to 10.1 for the corresponding age groups in females. The study also investigated the resistive torque to passive movement of the ankle, a value that provides an objective measure of the mechanical resistance of connective tissue within the muscle, tendons and joint capsules. They reported that resistive torque showed an overall trend to increase with age and was greater for males than females. This would suggest that there is a decrease in flexibility of connective tissue structures with ageing that may be a factor in loss of joint range of motion. They highlighted the fact that in an elderly subject, a higher proportion of the total dorsiflexor muscle strength will be required to overcome the intrinsic resistance and this may result in muscles becoming ineffective if both weakness and high resistance is present.



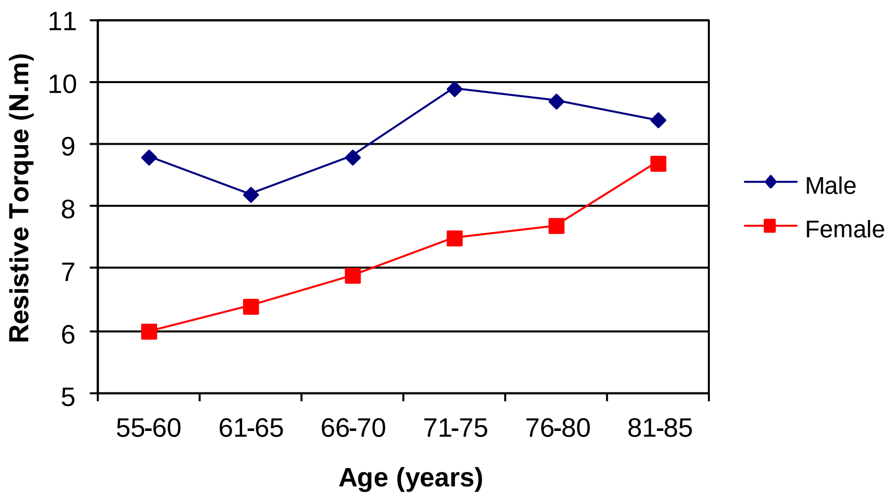


Figure 1.3 Ankle joint dorsiflexion range of motion and passive resisted torque at 10 degrees of dorsiflexion from age 55-60 to age 81-85

From Vandervoort et al. (1992)

The effects of decreased ROM on the ability to perform activities of daily living have not been well studied either in young or old populations (Schultz, 1992). Badley et al (1984) investigated the relationship between ROM impairment and functional disability in subjects with three types of arthritis. Comparing actual ROM with subjective information on ability to perform tasks, they were able to present thresholds of ROM necessary to perform individual activities of daily living (ADL). Loss of knee joint flexion correlated with loss of ability to undertake mobility based activities i.e. walking, getting in and out of a chair, bed or bath and ascending or descending stairs. Loss of hip joint flexion correlated with loss of ability to perform bending tasks and a loss of motion of the joints of the hand correlated with loss of ability to perform dextrous activities such as opening a jar. The authors do not state how much loss of joint movement was present to be considered as a joint impairment.

Jette and Branch (1985) investigated the relationship between impairment and disability in 776 non-institutionalised elderly, determining musculoskeletal impairment by performance of 10 gross body movements which put the major body joints through their complete ROM. They found a relationship between musculoskeletal impairment (or loss of ROM) and physical disability, but concluded that disability is a complex phenomenon influenced by many non-body related factors such as income, age, gender and living situation. Furthermore they felt that due to the many other factors, a musculoskeletal impairment may not result in disability in the elderly person.

Bergstrom et al (1985) followed up from their study of prevalence of joint impairments in 79-year olds by investigating the functional consequences of these impairments. They found restricted knee joint ROM correlated highly with disability to enter public transportation and that restricted hip movements limited climbing stairs. In their study, 80% of the subjects were capable of performing basic personal care and over two thirds walked unaided even though restricted ROM was found in one fifth to two thirds of all subjects (table 1.2). The authors concluded that many elderly people compensate for loss of ROM and that disability directly related to joint range of motion is infrequent even at the age of 79.

1.2.3 Muscle function

It is widely recognised that there is a decrease in muscle function with ageing (Andrews et al, 1996; DTI, 2000; Schultz, 1992; Schultz, 1995; Thompson, 1994). This section will first review the physiological changes that occur with ageing of muscle, then present the reported decline in strength, and assess how this decline is thought to impact on functional activity.

Thompson (1994) presented a review of the literature on all aspects of the effect of ageing on skeletal muscle physiology. Biochemically, reviewing six separate studies, it was concluded that the metabolic capacity of muscle was not adversely affected with age but was more influenced by the activity of subjects, those who participated in more activity having different biochemical characteristics than more sedentary subjects. Muscle atrophy was reported as a typical age related phenomenon, occurring as a result of a decrease in the number of muscle fibres and a decrease in the actual size of individual fibres. This muscle atrophy was found to be greatest in the weight bearing muscles of the lower extremity and less in the upper limbs, affecting type II (fast twitch) muscle fibres more than type I fibres (slow twitch). Force production which declines with ageing was attributed to many factors such as a decrease in the excitation-coupling mechanisms, decrease in cross sectional area, and an inability to recruit all motor units maximally during a contraction due to alterations in the central nervous or cardiorespiratory system. More recent work (Akima et al, 2001), investigating the muscle function in subjects aged 20-84, concluded that muscle strength losses were mainly due to a decline in the mass of the muscle with age, with both male and female subjects demonstrating a correlation between cross sectional area of quadratus femoris and peak torque during knee extension.

The rate of development of muscle torque was reviewed by Schultz (Schultz, 1995) studying the unpublished works of Chen (1993) and Thelen et al (1994). Thelen et al had investigated the rate of development in ankle torque in young and old adults and reported a significant increase in the time taken to reach a fixed force in the elderly subjects. The results implied that it is the rate of torque developments rather that the strength that is often critical to restoration of balance following a trip. This change in torque development in older subjects was felt to be as a result of changes in muscle contraction physiology rather than a decrease in neural processing speed.

Many authors have investigated the actual strengths of different muscle groups with different populations. Muscle actions generally described using the following terminology (Jones and Barker, 1996):

Isotonic: muscle contraction resulting in the movement of a fixed load

Isometric: tension is generated in the muscle but no movement shortening or lengthening occurs.

Isokinetic: muscle action resulting in constant angular velocity of the joint

Concentric: muscle contraction resulting in shortening of the muscle

Eccentric: muscle contraction as lengthening occurs in the muscle

Maximal isometric muscular strength amongst three generations of Finnish men (n=388) has been studied (Vitasalo, 1985). Isometric elbow flexion, knee extension, trunk flexion, trunk extension and grip strength were obtained from specially developed dynamometers. A decline in strength was found from the youngest to the oldest groups for all muscles tested, and on a percentage scale the oldest had the following decline related to the youngest: knee extension (47%), trunk extension (42%), hand grip (42%), trunk flexion (35%) and elbow flexion (35%).

Andrews et al (1996) presented values for isometric muscle force measured using a hand held dynamometer with 156 subjects aged 50-59, 60-69 and 70-79. In tests of eight upper and five lower extremity movements, isometric strength declined with increasing age. Gender and weight were also reported to statistically influence force measurements. Reported forces for lower extremity movements in females using the dominant leg are represented in figure 1.4.

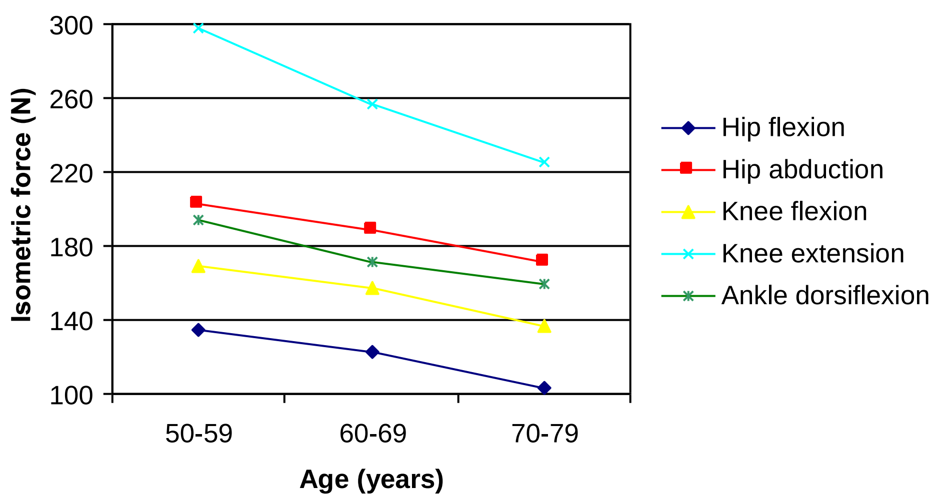


Figure 1.4 Ageing effect on mean maximum isometric muscle contraction for lower limb muscles in women Data from Andrews et. al (1996)

Hand strength in adults was investigated by Mathiowetz et al (1985). Grip and pinch strengths were tested in 638 subjects aged 20 to 94 years. The highest grip strength was reported in the 25-39 age group, with a decline correlating highly with age. Key, grip and palmar pinch was relatively stable for age 20 to 59 followed by a gradual decline with age of low to moderate correlation. The authors suggested that a curvilinear relationship occurs between hand strength and age, with strength peaking between 20 yrs and 50 yrs then decreasing afterwards.

Vandervoort et al(1992) as well and investigating ROM and resistance in musculature around the ankle, also investigated maximal isometric dorsiflexion. They reported a 30% decrease in strength between subjects aged 55-60 and subjects aged 81-85, and reported this to be similar to many other studies.

Isokinetic muscle contraction of the quadriceps was studied by Lindle et al (1997) in 654 women and men aged 20-93 years. Regression analysis revealed significant agerelated reductions in both concentric and eccentric isokinetic peak torques although eccentric muscle work was found to be less affected by age than concentric work. It was also reported that there was a decline in quality of the muscle expressed as the strength per kilogram of fat free muscle mass. This decline was significant for both men and women for concentric activity and in men for eccentric activity. The authors suggested that elderly men and women have differences in mechanical and elastic properties of the muscle enabling elderly women to utilise more elastic energy to assist with eccentric work. No differences were observed in the loss of strength with age between slow and fast velocities. There was a small velocity related difference between men and women, with women showing a greater age-related decrease in fast velocity peak torque than men.

The studies reported above have been cross-sectional in design i.e. data was simultaneously collected from a group of subjects of differing ages. This is generally the easiest and cheapest method of collecting data but as subjects are from different birth cohorts it is not possible to indicate the effect of ageing or rate of change within an individual (Kelly and Kroemer, 1990). Longitudinal studies follow specific subjects over time. Unfortunately these studies are more expensive and time consuming and measurement error can be greater due to the time periods between data collection (Rogers et al, 1996). Subject samples may change over time as people die leading to a survival bias (Desrosiers et al, 1998), but longitudinal studies are the most useful method to provide information about age related changes for any given individual.

Bassey (1998) reported on measurements of grip strength measured longitudinally over an 8-year period, for 350 elderly subjects originally aged 65 or over (figure 1.5). They found grip strength decreased by an average of about 13% in men over the eightyear period, and by 17% in women (both significant at $P<0.0001$). This rate of about 2% decrease in strength per year is similar to the changes reported in the cross-sectional studies reported previously. They reported similar finding to Mathiowetz et al (1985), in that loss of strength was more rapid in the subjects who were aged 75 and older compared to those over 65.

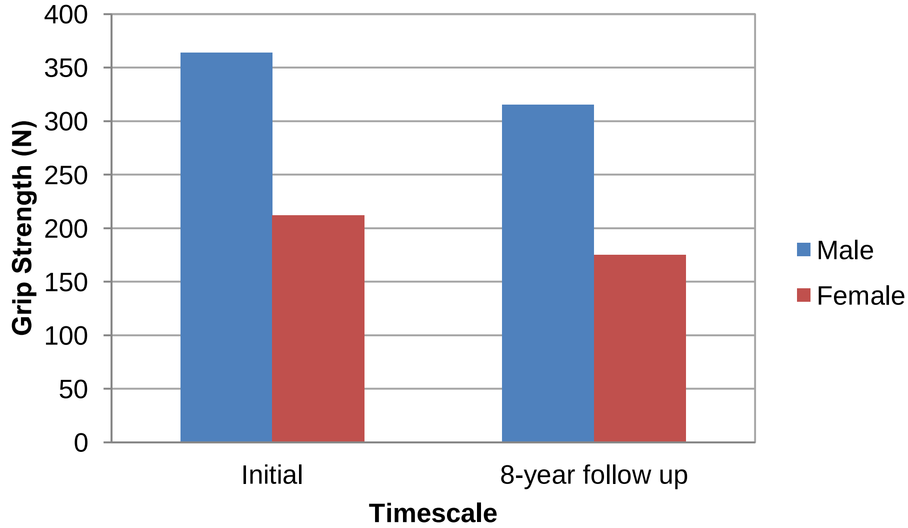


Figure 1.5 Longitudinal change in grip strength in adults aged over 65 Data from Bassey (1998)

The extent to which decline in muscle function leads to mobility impairments is not fully understood (Schultz, 1995). Rantanen et al (1999) studied the association between motor disability, physical activity and muscle strength in a group of 1,002 women aged over 65 who reported difficult in performing criterion tasks. They found that subjects with lower levels of muscle strength reported a greater degree of motor disability and proposed that there is a minimum requirement of strength for certain tasks. As strength decreases to that level, which was not investigated, then difficulty will be reported as the subject is working to maximum capacity. It is difficult to determine what is the causative factor in deterioration of function. Does weakness precede disability, or do the impairments caused by disability lead to decreased activity, therefore resulting in deconditioning and loss of muscle strength? Rantanen et al could not answer this, but the model developed did show a spiralling effect on muscle strength and disability, which the authors proposed could be reversed by increasing activity and manageing pain and chronic disease.

A significant association between physical activity, which is known to correlate with muscle strength, and functional ability was also reported in a five-year longitudinal study (Laukkanen et al, 1998) and in a review of the literature (Rikli and Jones, 1997). Rantanen et al (1994) assessed the strength and functional ability of 287 75-year olds. They found a significant relationship between strength and ability, with subjects who had the greatest strength reporting the least amounts of disability and performing better on step climbing and timed walks.

Schultz (1995) addressed the question of impairments and strength requirements differently by presenting the actual joint torques required to perform several activities of daily living. He reported that the joint torques required to perform activities such as rising from a chair and maintaining balance after a push were generally much lower than the normal strengths reported for older adults, therefore concluding that declines in strength could not fully explain decreased performance in older adults. Gross et al (1998) supported this argument in their study on the effect of muscle strength and movement speed when standing from sitting. Twenty-six elderly females (aged 64-84) and 12 young females were tested for strength and underwent biomechanical analysis of rising from a chair. Although the elderly subjects had only half the muscle strength of the younger subjects, they were still able to rise from a chair without the use of hands. Schultz commented that some older adults are unable to rise from a chair even when they had demonstrated sufficient strength, and concluded that factors other than declines in muscle strength are needed to explain why this occurs.

1.2.4 Proprioception

Proprioception is the awareness, at both a conscious and subconscious level, of the position of the body in space. This awareness results from a combination of sensory inputs from the peripheral proprioceptors in the muscle and joints, along with vision and vestibular information.

Hurley et al (1998) investigated proprioception by assessing the joint position sense (JPS) of the knee joint of young, middle aged and elderly subjects by asking subjects to reproduce a test knee angle. They found the acuity of JPS in the elderly subjects was worse than the young and middle aged and that JPS acuity decreased significantly with age. Petrella et al (1997) also investigated knee joint proprioception but in a standing rather than a seated position, as they felt this better represented normal function. A decline in proprioception, described as an increase in the error of reproducing a test movement, was found to be significant between young and old subjects. The authors also investigated the effect of physical activity on proprioception and found that active older subjects had significantly better proprioception than sedentary older subjects (figure 1.6), indicating that exercise may help to reduce the decline in proprioception with age.

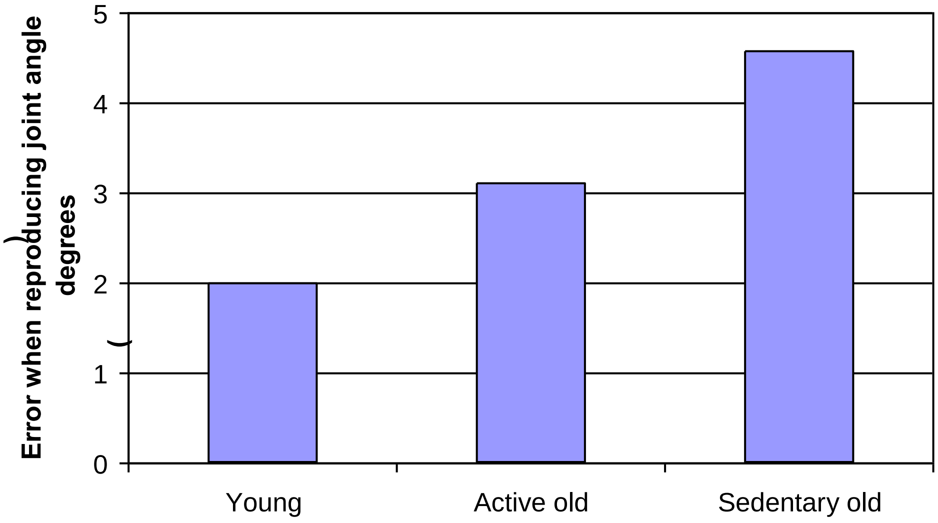


Figure 1.6 Effect of age and activity on the ability to reproduce a test joint angle, a statistically significant decline in proprioception was found with age and activity level.
From Petrella et al. (1997)

Thelen et al. (1998) studied the ability to detect rotation at the ankle in young and old females by having subjects stand with one foot on a servo-controlled platform. The platform induced a plantar or dorsiflexion rotation at a variety of differing speeds and degree of movement. The older subjects had a three to fourfold increase in the amount of movement that was required before they could perceive a change in dorsi or plantarflexion at the ankle, indicating a decline in proprioception.

Hurley et al. (1998) studied the relationship between JPS and functional performance in a series of tests consisting of a timed walk, stairs ascent and stair descent. A decrease in JPS was found to have a significant correlation with decreasing functional performance, and was more strongly related than the effects of decreased muscle strength. This suggests that decreased proprioception may have a detrimental effect on functional performance.

In a review of the literature of sensory function in older people, Wolfson (1997) reported that joint position sense, two-point discrimination and vibratory sense all decrease with increasing age. However these changes were not felt to be the primary causative factor in balance decrements, a view supported by Hurley et al. (1998), who concluded that decreased stability in older adults was likely to be a result of the cumulative effects of deficits in sensory, central processing and motor pathways.

1.2.5 Reaction times

Changes in reaction time, which is the delay that occurs between the onset of a stimulus to the onset of the response, have been well studied. Wolfson, (1997) reported these summarises from the results of previous work:

Reaction times increase rapidly from the sixth decade onwards. The nerve conduction time in afferent and efferent nerves declines only slightly with most of the change in response time due to sensorimotor processing (i.e. the time to identify the stimulus and select the correct motor response). The amount of time required for sensorimotor processing is affected by the complexity of the task. Therefore, the time required for choice reaction-time tasks has a greater increase than simple reaction-time tasks with an increase in age.

Schultz (1995) reviewed work performed by Thelen et al. (1994) who had studied the time taken to develop ankle torque in response to a stimulus. The experiment enabled collection of data on reaction time (time until torque started to develop) and on the time take to reach a certain magnitude of torque. They found the reaction times of old females were only 15 ms longer than young females. However when combined with the time it took from stimulus to develop 60 Nm of plantarflexion torque the difference between young and old was 161 ms. This highlighted that central processing is slower, and the muscles ability to respond in an effective manner is also less with age.

Sakari-Rantala et al. (1998) investigated the association of simple reaction time with mobility in 500 elderly subjects. They found a significant relationship between increased reaction times and decline in basic mobility (timed walk and stair climbing), which in turn was associated with functional limitations.

1.2.6 Balance

Balance in the elderly has been well studied and has universally been reported to decline with age (Rogers et al., 1996; Schultz, 1992; Wolfson, 1997). Hurley et al (1998) investigated postural sway in 20 young, 10 middle aged and 15 old subjects using a custom designed strain gauged „swaymeter‟. Subjects were tested in bipedal stance with their eyes open, bipedal stance with eyes closed and in one-legged stance with eyes open. In bipedal stance with open eyes, postural stability was reported to be similar in the three age groups. However with eyes closed the middle aged and elderly subjects had significantly decreased stability compared to the young subjects, and for one legged stance none of the elderly or middle aged could maintain balance for the test period of 15 seconds, and for a shorter period of 7 seconds the elderly had significantly less stability. The authors considered the increased postural sway in the elderly to be related to an age related decrease in muscle spindle sensitivity and accompanied by deficits in central processing and motor pathways.

Wolfson (1997) and Schultz (1992) suggest decreases in balance occur due to changes any of the component neuromuscular functions. These would include decreased vision, decreased vestibular function, decreased muscle strength or endurance, increased muscle latency, decreased cutaneous or joint proprioception and decreased sensorimotor processing.

Balance deficits in the elderly result in an increase in the risk of falls and fall related injuries (Nevitt, 1997). Lord et al. (1999) investigated lateral postural sway in older people with and without a history of falls. They found that impaired lower limb proprioception, quadriceps strength and reaction time were the best predictors of lateral sway when performing a near tandem stability test. Older subjects demonstrated poorer balance on both tests with eyes open and closed and demonstrated a greater need to take protective steps. Subjects who had a history of falls were found to have decreased proprioception, visual acuity and quadriceps strength and increased lateral sway on testing.

1.2.7 Summary of physical factors associated with ageing

In summary, it can be seen that there are many changes in physical function with ageing. Joint ROM, tissue flexibility, strength and proprioception all decline progressively with age. Reaction times increase as does the time to develop torque by a given muscle group. A combination of these factors leads to a decline in balance and higher risk of falling. The static anthropometry of older subjects is different to younger subjects possibly due to age related changes but also due to secular trends in health and nutrition.

The effect of the above physical factors on functional performance in older subjects has been discussed and overall it can be concluded that decline in function is multifactorial with all of the above physical factors playing a part. It is also recognised that there are other elements that play a role in disability, such a cognitive function, social circumstances and environment and a brief review of the psychological factors associated with ageing will be covered in the next section.

1.3 Psychological factors with ageing

Deterioration in cognitive function with ageing has been strongly associated with disease. Bond and Coleman (1990) report that diseases of the brain account for most cases of intellectual decline in the elderly, particularly the memory failing and clouding of understanding associated with dementia. The average age of onset of Alzheimer‟s type dementia is 75 and the prevalence of all dementias doubles approximately every five years between the age of 65 and 85 (Schaie and Willis, 1996). However there are changes in cognitive functioning with ageing that do appear to be independent of pathology and a useful review is found in Bond and Coleman (1990).

One of the larger studies of adult intellectual functioning has been conducted by the Seattle Longitudinal Study (Schaie, 1994; Schaie, 1996). This study was conducted over a period of 35 years, with 6 cycles of testing at seven year intervals, and in total involved over 5000 subjects. Subjects performed a battery of tests to assess many different components of cognitive function and psychomotor ability and some of these results are demonstrated in figure 1.7.

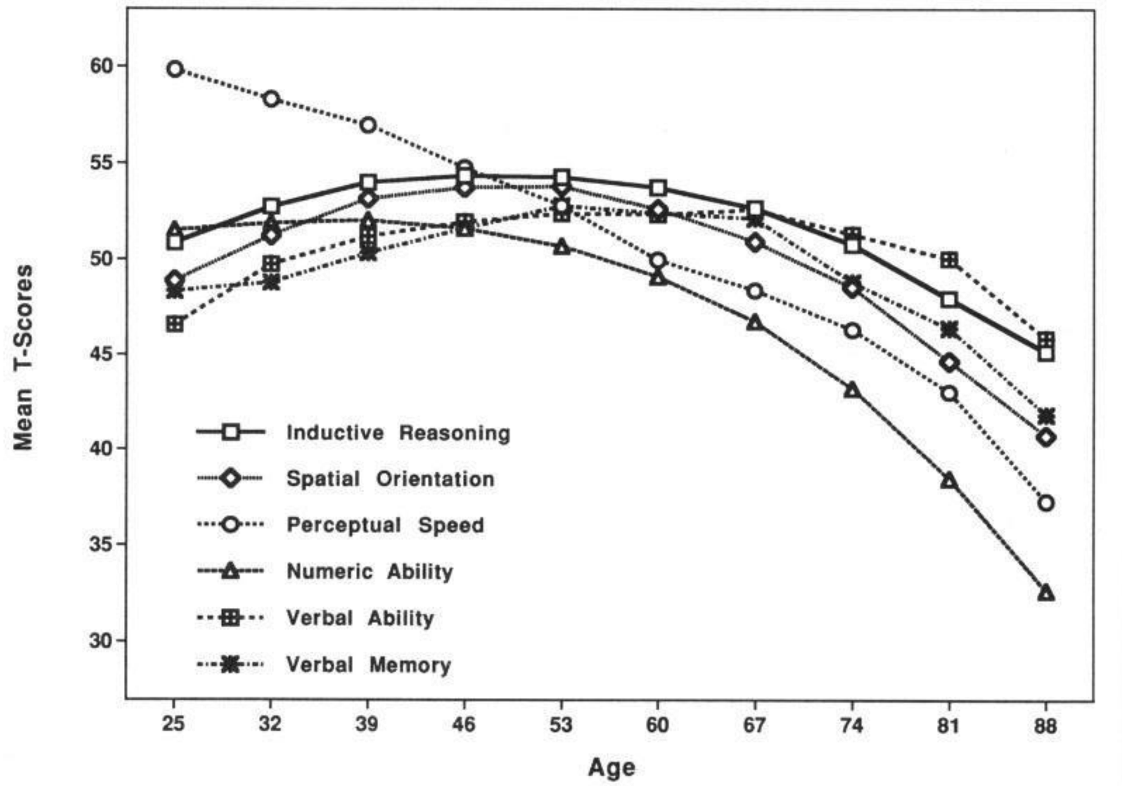


Figure 1.7 Longitudinal changes in various aspects of cognitive functioning with age From Schaie (1996)

For the majority of variables (word fluency, spatial orientation, inductive reasoning, verbal ability and verbal memory), age related changes begin to commence in the early 60"s with a modest decline, followed by a more marked decline after the age of 80. Perceptual speed was found to decline from age 25 onwards and numeric ability begins to decline in the 50"s. A measure of practical intelligence was found to peak by age 60 and a steep decline was not observed prior to age 80. Psychomotor speed and motor cognitive flexibility also followed a similar pattern of being stable until age 60 with moderate decline thereafter.

Useful reviews of the impact of ageing on memory are presented in OLDERADULT DATA (DTI, 2000) and Bond and Coleman (1990). For working, or short term memory, it is suggested that there is no decline with ageing, adults in their 70‟s being able to memorise information for a short time as effectively as younger adults. Studies of longer-term memory have tended to show up differences between young and older subjects and it has been suggested that as people age, there is a decline in the resources available to them to process information, therefore reducing the accuracy of both recognition and recall. Deficits have been reported in both the correct acquisition of the information supplied and the retrieval of that information at a later stage. This will vary with the type of memory being tested, and in some areas such as semantic memory (the knowledge about particular facts about the world), age related declines are rare. It must also be noted that there is a wide variation in reported abilities amongst the elderly population with a number of older adults never experiencing a decline and continuing to perform at a level similar to younger adults.

Impairments in cognitive functions, whether they are as a result of disease or ageing, are known to affect ability in activity of daily living function. Rozzini et al (1993) investigated 549 community dwelling elders aged over 70, to determine whether ADL scales and a physical performance test could detect health status. They found that the health factor that acted as the best predictor of dysfunction in basic and instrumental activities of daily living, was cognitive impairment. They also found that subjects with lower cognitive function performed less well on physical performance tests assessed by an observer. Tinetti (1986) reported mental status to be a significant component contributing to safe mobility as problem solving is required to avoid obstacles and compensate for physical disabilities, and Guralnik et al (1989) state that ADL‟s have a large cognitive component which can never be fully separated from the physical capabilities of the individual.

1.4 Prevalence of ADL impairments

The prevalence of ADL impairments in the elderly gives an impression of how the physical and psychological decrements with age impact on the lives of elderly people. One of the largest studies of ADL impairments was conducted by the National Health Interview Survey in the USA (Dawson et al, 1987), investigating ADL impairments in 16,148 householders aged 50 and over. Clark et al (1990) investigated 244 independently living adults aged 55 to 93, Bergstrom et al (1985) looked at 134 adults aged 79 years, and Gill et al (1999), 1088 community dwelling elders aged 72 or older. Results of these surveys are summarised in table 1.3.

Table 1.3 Prevalence of ADL limitations in the elderly

	Dawson et al. (1987) n > 10,000 aged 65+	Clark et al. (1990) n = 244 aged 55-93	Bergström et al. (1985) n = 134 aged 79	Gill et al. (1999) n = 1088 aged 72+
	Percentage of subjects reporting difficulty with ADLs			
Eating	1.8		4	
Using toilet	4.3		9	
Dressing	6.2	28	11	
Transferring	8.0	28	10	16
Getting outside	9.6			
Bathing	9.8	28		30
Walking	18.7		31	
Housekeeping	23.8	31	31	
Preparing meals	7.1	45		
Grocery shopping	11.1	53		

These studies report a wide variation in the reported inability to perform ADLs, possibly as a consequence of the differing age groups of the subjects in the studies, as it would be expected that older subjects would have a greater degree of impairment. Of the activities performed within the home, housekeeping was reported to be difficult by the most subjects (23.8 - 31 % of subjects), followed by walking (18.7 – 31%), bathing (9.8 – 30%) and preparing meals (7.1 – 45%). Dawson et al (1987) also collected information on the number of subjects reporting difficulty with an ADL who received assistance with that ADL. Of the home management activities subjects tended to receive more help (e.g. housekeeping 81% of subjects receiving help, preparing meals 85% of subjects receiving help) compared to the ADLs associated with personal care (61% of subjects receiving help with bathing and 25% receiving help with walking). This disparity may be as a result of the provision of social services to elderly people or due to the willingness of family and friends to perform household chores rather than personal care.

1.5 Biomechanics research on the effect of ageing on ADLs

From the above sections it can be seen that there has been a large amount of work on physical and psychological effects of ageing, the effect of these impairments on the performance of ADLs and the actual prevalence of difficulties of ADLs. Most work in this area has concluded that the decline in the ability to perform ADLs is multifactorial and cannot be explained by loss of strength, ROM or psychological decline alone. Each activity of daily living will have its own set of demands on the body, some requiring more flexibility, some greater strength and so on. It would therefore seem logical to investigate the biomechanical requirements of ADLs.

There has been relatively little quantitative research on the biomechanics of ADLs and the changes that occur with the ageing process even though the need for this has been identified by many authors (Kelly and Kroemer, 1990; Schultz, 1992; Rogers et al, 1996; Kerrigan et al, 1998). This next section will look at the movements which have undergone some biomechanical studies in older adults.

1.5.1 Gait

Gait has been the most comprehensively studied activity of daily living with an aged population and a useful review of work from the 1960"s onwards has been performed by Prince et al (1997). Early work concentrated on spatial-temporal parameters such as walking speed, stride length, and cadence, with studies overall concluding a decrease in all of these variables. As technology improved kinematic analysis became possible (the study of the movement of body segments) using electrogoniometers or external body markers. Studies overall reported a decrease in ankle dynamic range of motion, decreased knee extension at terminal stance, increase in anterior pelvic tilt with a resulting increase in hip flexion throughout stance.

More recent studies have investigated the kinetic changes with ageing (the study of forces and the energetics of movement). These studies are summarised in table 1.4. All the studies (Winter et al, 1990; Kerrigan et al, 1998; McGibbon and Krebs, 1999; DeVita and Hortobagyi, 2000) concluded that ankle plantarflexion torque during stance through to toe off is reduced in elderly subjects. McGibbon and Krebs (1999) reported this only between young adults and impaired elderly subjects, and not between young and healthy elderly adults, when controlling for centre of gravity velocity. DeVita and Hortobagyi (2000) controlled the speed of walking of subjects to determine whether the reduced torques were neuromuscular adaptations of the elderly or simply as a result of a slower chosen speed of walking. It was found that at the same walking speed elderly subjects had lower ankle plantarflexor torques but generated more propulsive force from the hip extensors. This redistribution of joint torques was interpreted as an alteration in the motor pattern used in gait, suggesting that the underlying neuromuscular components of a motor performance may change with age.

Table 1.4 Studies assessing the effects of ageing on the biomechanics of gait

Study	Subjects	Analysis Method	Age related changes
Winter et al. (1990)	Older adults (n=15), mean age 68 Young adults (n=12), mean age 26	2-dimensional gait analysis from digitised video and force platform data. Linksegment model to calculate joint moments and power. Ten walks at natural cadence.	Reduced walking velocity Reduced stride length Decreased ankle plantarflexion power and increased energy absorption at knee during toe off Decreased ankle dorsifexion ROM at heel contact
Kerrigan et al. (1998)	Older adults (n=31), age 65-84 Young adults (n=31), age 18-36	3-dimensional gait analysis using optoelectronic motion analysis system and two force platforms. Commercial package to calculate kinematics and kinetics. Six walks at normal and six walks at fast speed.	Reduced peak hip extension Increased anterior pelvic tilt Reduced ankle plantarflexion Reduced ankle power generation

McGibbon and Krebs (1999)	Healthy older women (n=16) age 72.5 ± 5.6 Functionally limited older women (n=24) age 73.5 ± 7.2 Healthy young women (n=20) age 27.0 ± 4.2	3-dimensional gait analysis using optoelectronic motion analysis system and two force platforms. Newtonian inverse dynamic algorithm and Lagrangin 5point differentiation algorithm used to compute kinematics and kinetics. Cadence fixed at 120 steps per minute.	Reduced walking velocity Reduced plantarflexion power (this was significantly lower for the functionally limited older adults compared to the healthy young)
DeVita and Hortobagyi (2000)	Older adults (n=12) age 69.0 ± 6.5 Young adults (n=14) age 21.6 ± 2.7	2-dimensional gait analysis from digitised video and force platform data. Linksegment model to calculate joint moments and power. Expressed torque generated at the hip, knee and ankle as a proportion of total support torque Five walks at fixed speed of 1.5 m/s	Reduced step length Increased hip flexion throughout stance Decreased ankle ROM in stance Greater net hip extensor torque throughout stance and push off Reduced net knee and ankle torque at push off

1.5.2 Rising from sitting

In a similar pattern to the study of gait in the elderly, rising from sitting was first investigated using spatial-temporal methods and more recently kinematic and kinetic method have been used. Laporte et al (1999) performed a comprehensive literature review of work in this area, considering the biomechanical requirements of the sit to stand transfer and the age related changes. Recent studies considering the kinematics and kinetics of rising are summarised in table 1.5.

Age related changes in the pattern of movement form sit to stand has been reported by many authors (Alexander et al, 1991; Schultz et al, 1992; Gross et al, 1998; Papa and Cappozzo, 2000). One of the common findings was the increase in trunk rotation and generation of forward momentum in the time period prior to lifting the buttocks from the seat. This strategy has been suggested to increase stability due the centre of mass of the body being located within the base of support as lift off occurs thus reducing the likelihood of falling backwards. Papa and Cappozzo (2000) suggested this also reduced the global muscular effort required however Schultz et al (1992) felt this strategy did not reduce joint torques, indicating that elderly people place greater importance in attaining postural stability than reducing muscle strength requirements. Gross et al. (1998) studied EMG activity in the lower extremity and found an earlier activation in ankle extensors in the elderly group when compared to the young group. They considered this to be another strategy adopted by the elderly to enhance postural stability, as the co-contraction of plantar and dorsiflexors would serve to stiffen the ankle prior to weight transference from the chair to the seat. Joint torques required to rise from sitting were found to be

Table 1.5. Studies assessing the effects of ageing on the biomechanics of sit to stand

Study	Subjects	Analysis Method	Age related changes
Alexander et al. (1991) Schultz et al. (1992)	Healthy older adults, mean age 72 (n=23) Impaired older adults, mean age 84 (n=11) Young adults (n=17)	Instrumented, adjustable height chair. Eight markers tracked with video. 2- dimensional biomechanical model estimating internal loads used to calculate kinetics and centre of mass location.	Increased rotation of upper body segments, thighs and legs prior to lift off Greater anterior displacement of floor reaction location at lift off Reduced knee extension torque in impaired group (otherwise no age related change in torque requirements) Greater upper limb forces used by impaired subjects

VanderLinden et al. (1994)	Older adults (n=8) mean age 68.8	3-dimensional motion analysis using optoelectric system, force plate and surface electromyography. No kinetic calculations made Fast and slow tests with differing knee and ankle ROM.	No significant difference from previously reported data on young adults
Gross et al. (1998)	Older women (n=26) age 70.1 ± 5.8 Young women (n=12) age 24.2 ± 2.4	3-dimensional motion analysis using optoelectric system, EMG and forceplates on floor and on seat. Newtonian inverse dynamics to calculate kinetics.	Decreased vertical force and momentum in fast trials Decreased rate of torque development in fast trials Increased hip flexion before lift off Increased forward momentum generation prior to lift off Earlier activation of ankle extensors causing cocontraction of ankle muscle at lift off
Papa and Cappozzo (2000)	Older adults (n=35) age 65-81 Young adults (n=16) age 22-34	Inverted pendulum model using static joint co-ordinates from photography and force plate data. Fast and slow trials.	Increased trunk flexion and forward momentum prior to lift off. Delayed elevation of centre of mass until centre of mass over base of support Decreased maximum speed

much lower than the actual strength present in the elderly group, suggesting that difficulty in performing this activity may be related to other factors such as fear of falling or decline in balance.

1.5.3 Stair ascent and descent

Ascending and descending the stairs can be an activity that poses many problems for older adults. Studies have found that around 20% of adults aged over 50 reported difficulty climbing stairs, increasing to 45% of individuals aged over 80 (Powell Lawton, 1990; Startzell et al, 2000; Verghese et al 2008). In a study of older adults (mean age 79) attending health centres in the US (Reuben and Siu, 1990), nearly 10% of subjects were unable to climb stairs at all. Stairs can also prove hazardous, with falls on stairs accounting for more than 10% of fatal fall accidents (Startzell et al, 2000), the majority of these fatalities occurring in the over 65 age group. In the UK approximately 20,000 hospitalizations and 900 deaths occur each year due to falls on steps or stairs in adults over 65 (Hamal and Cavanagh, 2004). Around 75% of these falls occur during stair descent (Tinetti et al, 1988) indicating that either the demands of stair descent are greater or that there are more serious consequences resulting from a slip or trip going down the stairs. Verghese et al. (2008) investigated the ability of 310 community dwelling older adults (mean age 79) to ascend and descend stairs. Of these, 140 reported difficulties ascending stairs and 83 reported difficulty descending stairs. The authors looked at other health problems in to determine what factors may influence the ability to climb stairs. They found that difficulty in climbing up stairs was associated with hypertension, arthritis, depressive symptoms and with poor balance, reduced grip strength, and neurologic gait abnormalities. Difficulty climbing down stairs was associated with higher prevalence of falls and fear of falling was a major factor.

Few studies have investigated the effect of ageing on the biomechanics of stair ascent and descent. Studies of younger adults (McFadyen and Winter, 1988; Costigan, 2002; Riener et al, 2002; Nadeau et al, 2003) have found that compared to normal walking there are greater demands on the musculoskeletal system during stair ascent and descent. Larger hip and knee extensor moments are required as a result of propelling the body upwards during stair ascent or braking during descent. Greater joint ROM is also required for stair ascent and descent than normal walking, approximately 80 degrees of knee flexion compared to 64.5 degrees (Rowe et al., 2000). These demands may be close to the maximum joint ROM and strength available in an older adult.

Recent studies that have investigated the effect of ageing on stair ascent and descent are summarised in table 1.6.

Livingston (1996) looked at the kinematic changes between a group of young and older women (n=5 in each group) whilst walking up and down an instrumented staircase of 6 stairs. Three of the stairs contained switch mats which triggered a digital clock when contacted. From this information, the cycle time, stance duration and walking velocity could be determined. They controlled for height between the age groups as they recognised that this has an impact on stair climbing measures such as walking velocity. During stair scent it was found that the older adults were somewhat slower than their younger counterparts with a longer step cycle time and hence slower cadence.

Table 1.6 Studies assessing the effects of ageing on the biomechanics stair climbing and descent

Study	**Subjects**	**Analysis Method**	**Age related changes**
Livingston (1996)	Older adults (n=5) age 67.4 ± 6.5 Young adults (n=5) age 22.8 ± 3.1	Switch mat data on 3 consecutive stairs. Cine camera at 50Hz. Measured cycle time, stance phase duration, cadence and movement velocity.	Slower cadence, longer step cycle time, and decreased velocity during stair climbing. Prolonged stance phase during stair descent
Christina & Cavanagh (2002)	Older adults (n=12) age 73.3 ± 1.9 Young adults (n=12) age 24 ± 3.3	2 force plates mounted in a staircase. Measured stair descent 5 times in each of 2 different illuminance conditions. Measured ground reaction forces and calculated the coefficient of friction required	Less vigorous push off during stair descent. More cautious use of available friction at foot strike and toe off. Faster loading rates on initial contact.

Hortobagyi et al (2003)	Older adults (n=14) age 74 ± 3 Young adults (n=13) age 22 ± 2	2-D gait analysis from digitised video and force platform data. Link-segment model to calculate knee joint angular position and velocity and knee extensor moments. Compared moments to maximal isometric knee extension moment measured with a force plate to calculate relative effort 5 trials of stair ascent and descent	Reduced knee angular velocity at peak torque generation in stair ascent and increased angular velocity at peak torque during stair descent. Reduction in peak knee extensor moment. Relative effort of stair ascent and descent significantly increased.
Stacoff et al. (2005)	Older adults (n=8) age 76.5 ± 4.2 Middle age adults (n=5) age 63.5 ± 2.7 Young adults (n=7) age 33.7 ± 7.9	2 force plates mounted in staircase. 8-10 repetitions at 3 different stair inclinations. Measured ground reaction forces only	Reduced rate of stair ascent leading to reduced ground reaction forces No differences between the middle and old age group.
Reeves et al. (2008[a+b], 2009)	Older adults (n=15) age 74.8 ± 2.8 Young adults (n=17) age 24.6 ± 4.1	3-D motion analysis with a 4 step staircase instrumented with 3 force plates. Inverse dynamics to determine kinetics. EMG of 4 leg muscles. Three trials of ascent and descent with ant without a handrail. Muscle strength measured on an isokinetic dynamometer	Reduced peak knee and ankle moments in older adults during stair ascent. Reduced peak ankle moment during stair descent. Redistribution of joint moments from the ankle to the knee in older adults. Older adults are operating close to their maximum capabilities

During descent the most significant difference was that the older adults displayed much larger stance phase durations that younger adults, which appeared to be the chief compensatory mechanism for maintaining stability during stair descent. This would suggest that the older adults are requiring changes in their gait to improve stability. However the number of subjects in this study was small, with no details about the physical capabilities of the subjects, so it is difficult to relate this data to a wider population.

Stacoff et al (2005) investigated the change in vertical ground reaction forces using force plates embedded in staircases of differing inclinations and three groups of subjects (young (n=7), middle (n=5) and old aged (n=8)). They detected a small but significant increase in the ground reaction forces generated by the younger group in their study and attributed this to the faster rate of stair ascent in this group. They found no significant differences between the groups during stair descent and no differences at all between the middle and old age groups, though the number in each group was low and may not have been sufficient for statistical power. The study did not provide any insight into why stair climbing is an activity that older adults find difficult.

Christina and Cavanagh (2002) also studied ground reaction forces during stair descent but controlled for speed in their study, as they felt some of the differences reported by previous authors related purely to differences in walking speed between the age groups. They found that older adults had a less vigorous push off from the step on stair descent and a greater loading rate when landing on the next step. This could be attributed to a lack of control at touchdown when compared to the young or an increase in joint stiffness – a factor reported by Hortobagyi and DeVita (2000). The purpose of this study had been to investigate the frictional demands of stair descent and the influence of age and illumination. Looking at the relative coefficients of friction during the stair descent, it was found that the older subjects used a more cautious approach reducing the frictional demands of the activity at touch down and toe off.

More information was provided by Hortobagyi et al (2003). They investigated a group of young (n=13) and older adults (n=14), first measuring the maximum joint moments that could be produced during an instrumented leg press. They then performed 2-D movement analysis of the subjects performing stair ascent and descent on a custom built instrumented staircase and calculated the relative effort of the activity by dividing the maximum generated moment by the maximum moment produced during the bench press. During stair ascent it was found that the relative effort for young adults was 54% compared to 78% in older adults. In stair descent the figures were 42% and 88% respectively. These figures were supported by electromyographic (EMG) activity, with the relative muscle activity recorded in the vastus lateralis and biceps femoris of the older subjects being greater than 1.5 times that of the younger subjects. The study concluded that for healthy older adults some of the difficulty performing ADL"s such as stair climbing and descent may be due more to the fact that the person is working at a higher level of relative effort than to the absolute functional demands imposed by the task. The work of Hortobagyi et al (2003) provide some figures for peak knee joint moments in young and older adults. This work was performed using 2-D technology and although the authors have taken precautions to ensure that movement occurred in a plane in line with the cameras, there will be errors introduced if movement occurred out of plane.

Following the data collection for this study, some new work in this research area was published by a group from Manchester Metropolitan University (Reeves et al, 2008[a,b] and Reeves at al, 2009). These authors investigated stair ascent and descent in 15 older adults (mean age 74.8 years) and 17 younger adults (mean age 24.6 years), using a purpose built 4-step staircase instrumented with three force plates. Motion capture was performed using a 3-D optoelectronic system., and EMG signals were obtained from vastus lateralis, biceps femoris, gastrocnemius and tibialis anterior. Similarly to

Hortobagyi et al (2003), the authors wished to determine if older adults were working close to their functional limits and to determine this they measured concentric and eccentric isokinetic muscle strength in the knee and ankle extensors using an isokinetic dynamometer. These authors found that the rate of stair ascent and descent did not differ between the young and old subjects and that there were no changes of the proportion of time spent in stance, differing from the findings of Livingston (1996). When ascending stairs they found that the older adults produced a reduced peak knee extension moment and reduced peak ankle extension moment compared to the younger adults, and during stair descent there was a reduced peak ankle moment in the older group. They compared the peak moments produced during activity to those obtained from the isokinetic dynamometer. Although the older adults had reduced their peak ankle moments during stair ascent, the percentage of their maximum available moment was at 93%, compared to 85% for young adults. The authors suggested the reduction of peak ankle moments was a necessity to prevent the plantarflexors working close to their maximum capabilities. The knee moments for the older adults expressed as a percentage of the maximum concentric knee extensor moments were 42% for stair descent and 75% for stair ascent (corresponding to 30% and 53% in young subjects). The redistribution of joint moments is a strategy that was reported by Hortobagyi et al (2003), and indicates that the demands on the ankle plantarflexors is reduced in the older adults as there is greater reserve in the knee extensors to safely perform the activity. As both these studies have investigated young versus old subjects it is impossible to know when these changes start to be seen in older adults and whether there is a critical age when the reduction in lower limb strength affects stair performance.

The most recent study in this area is by Novak and Brouwer (2011). They investigated 23 young (23.7 ± 3.0 years, range 20-30 years) and 32 older adults (67.0 ± 8.2 years, range 55-83 years) undertaking stair ascent and descent on a custom built 4-step staircase. They used a 3D motion analysis system combined with a forceplate in the second step, and computed sagittal and frontal plane kinetics using a link-segment model and an inverse dynamics approach. The study has the largest sample size of older adults performing stairs to date but the authors included subjects aged 55+ which is younger than similar studies (Reeves et al. 2009), and would have fallen into the classification of middle aged used by Stacoff et al. (2005). Novak and Brouwer (2011) reported similar changes in joint moment profiles to those reported by Reeves et al. (2009), with reduced ankle plantarflexor moments in the older group balanced by an increase in hip and knee extensor moments and also an increase in the hap abductor moments in the older adults.

They reported that the changes in moment contributions were greatest at the times of transition from double to single support in early stance, and single to double support in late stance and less during mid stance. These authors did not measure muscle strength and so could not conclude if these changes were related to weakness, but it does suggest that older adults are needing to change their stair gait pattern, either for safety or due to declining functional ability.

Handrails are commonly recommended for older adults to improve safety and to assist with stair climbing and descent. Only one study to date has investigated the impact of handrail use on the biomechanics of stair climbing. Reeves et al (2008a) studied 11 older adults performing stair ascent and descent using the laboratory detailed above. They performed three trials without a handrail and three trials lightly holding the handrail. They found that the handrail did not effect the timing of events in the gait cycle, or the amount of movement required to perform the activity. Studying the vertical ground reaction forces they concluded that the subjects had not used the handrail to offload the legs but rather to aid balance. The handrail used resulted in a redistribution of joint moments during stir ascent, increasing the peak knee extension moment in the leading leg and reducing the peak ankle extension moment in the trailing leg. The authors felt this strategy was a “safe” strategy as the knee has more available strength capacity than the ankle (Reeves, 2009). During stair descent the opposite was observed with greater ankle moments being produced and reduced knee moments, when using the handrail. The authors reported that this was also a balance control strategy related to the position of the centre of pressure in the leading foot. They felt that more work is required in this area to investigate how less able adults performed this task as their subjects were all healthy, confident stair climbers.

1.5.4 Summary of biomechanical studies on ageing and ADLs

Reviewing the literature it can be seen that elderly people use different movement strategies than young adults in their approach to common activities of daily living. Some of these strategies reduce the required strength and others are made to enhance postural stability, even at the cost of increasing strength requirements. Comprehensive analysis of the tasks of walking and rising from sitting have been performed and have provided useful information on what aspects require targeting in a rehabilitation or preventative programme. Studies of elderly subjects performing more complex tasks so far are limited and often simplistic in their approach. The majority of work has compared older adults to young able adults and provides little knowledge on the effects of the ongoing ageing process.

1.6 The EQUAL project

This research was undertaken as part of a larger project funded by the Engineering and Physical Sciences Council (EPSRC) in the UK under their Extending Quality Life initiative. The project was multidisciplinary, involving the Bioengineering Unit and Psychology Department at the University of Strathclyde, the Product Design Department at Glasgow School of art and the Physiotherapy Department at Queen Margaret‟s University College (QMUC) in Edinburgh. The title of the research project was:

"Integration of biomechanical and psychological parameters of functional performance of older adults into a new computer aided design package for inclusive design"

The aim of the project was to produce a software tool that could be used by designers to predict the ability of an older adult to use a product or negotiate an environment, and hence design new products in a more inclusive manner. In order to provide greater information on the ageing process, as well as enabling designers to select their market group, it was decided to study adults in three age groups, 60‟s 70‟s and 80‟s. The software was to include physical data including maximum joint ROM and maximum isometric joint strength at a range of joint angles. These data were collected by Dr Dinesh Samuel of QMUC and reported in his thesis (Samuel, 2005). Psychological performance of older adults was included from data produced by Dr Lauren Potter (Potter, 2005). The biomechanics inputs for the software form the basis for this thesis.

In order to provide the most useful software package, several activities were proposed to be studied which would enable the software designers at Glasgow School of art the ability to create a model of an older adult capable of performing several tasks. These included normal walking, rising from a chair and sitting down, climbing and descending stairs, navigating doors and lifting objects to different heights. Functional activities of the hand were to include opening jars, lifting objects and manipulating remote controls. All these movements were assessed and provided in details sufficient for the software. The final report of this project is included in appendix 1.1.

1.7 Aims of this thesis

This thesis aims to study in greater depth the biomechanics of a selection of the movements undertaken in the EQUAL project. The area it was felt where detailed study would enhance the already available knowledge base was that of stair climbing and descent. These movements have been studied previously but have not looked at the changes experienced with increasing age and functional restriction. No studies had investigated the impact of using a handrail on the biomechanics on stair performance with large numbers of older adults.

The large number of subjects in this study is unusual in biomechanical research and allows subjects to be divided into genders to determine any differences between the sexes in performance linked to ageing. As there have been reported differences in ROM and strengths between the genders, this may impact on the strategies used to perform the activity. Using the data collected by Dr Samuel it is also possible to determine how much of an older adults joint ROM and strength are required to perform these common activities of daily living and hence provide greater information on why limitations in these activities are seen with increasing age.

In summary the aims are:

1. To provide detailed biomechanical analysis of stair ascent and descent in older adults
2. To determine any age and gender related effects over the three decades from the 60‟s- over 80‟s
3. To investigate whether using a handrail affects the biomechanics of stair ascent and descent
4. To investigate if any changes are related to physical changes such as strength and flexibility

Chapter 2.

A Review of Biomechanical Methods

This chapter presents the methods used in human movement studies to determine kinematics and kinetics. Reviews of the methods of capturing human movement data have been performed previously (Winter, 1990; Cheng, 1996; Furnee, 1997) and will not be detailed here. This chapter focuses on the methods used to process co-ordinate data obtained from an optoelectronic system using passive reflective markers mounted on the subject, coupled with force data from force platforms and strain gauged instrumentation.

2.1 Joint kinematics

Kinematics is the study of body motion, or simply the description of the relative movement between two adjacent bones considered to behave as rigid bodies. In order to fully describe the 3-dimensional motion of a rigid body it is necessary to track the position of three points on it. A set of three orthogonal axes can then be determined for that bone and hence its rotation relative to the orthogonal axes of the laboratory. If the orientations of two bones linked by a joint are known, then it is possible to calculate the angles of rotation between those bones.

Traditionally markers were placed on the skin over anatomically relevant points. More recently it has been acknowledged that markers do not remain stationary with respect to the bone due to interposition of soft tissues, and skin movement artefacts can lead to considerable error in calculated kinematics (Cappozzo et al, 1996; Cappello et al, 1997; Lucchetti et al, 1998). It has therefore been proposed to improve accuracy by placing clusters of markers on a segment in the areas where skin movement is least. If these markers are considered to move rigidly with the bone, then by simple vectorial mathematics, fixed anatomical landmarks can be located if suitable calibration has been performed.

Cappozo et al (1996) proposed what they termed a calibrated anatomical systems technique (CAST) to enable the use of a cluster of markers on a segment. From three markers placed on a segment a technical frame of orthogonal axes can be determined. This technical frame will move with the underlying bone during activity and can be used during the collection of data using a stereophotogrammetric system. However this technical frame does not describe the geometry of the underlying bone and information about the location of anatomical landmarks is required. This information is obtained during a calibration procedure in which the anatomical landmark is identified and the vectorial relationship between it and the technical frame is computed. As the segment is considered to be a rigid body this relationship will be fixed throughout movement so knowledge of the position and orientation of the technical frame can be used to calculate the position of the anatomical landmarks.

2.1.1 Anatomical axis definitions

Anatomical landmarks are used to determine a bone-embedded system of axes (or frame of reference) for each section and mathematical methods are used to compute the relative position of these systems. The choice of how to define an axis system for each body segment is based on several factors (Chadwick, 1999; Fioretti et al., 1997; Cappozzo et al., 1995):

It must be based on recognised anatomical landmarks for repeatability;

It must be based on practical experimental methods;

It must produce results that can be expressed in established anatomical and physiological terminology;

It must be accurate;

This thesis has used the International Society of Biomechanics (ISB) recommendations for axis sets or those of Cappozzo et al. (1995) where possible to aid standardisation. Axis sets for each segment are detailed in Chapter 3. Anatomical landmarks were located by palpation and a pointer technique was used to maximise accuracy in the upper limbs. Lower limb anatomical landmarks were defined using a marker positioned directly above the landmark as it was not practical to have the older adults stand still during a prolonged calibration. Methods such as x-ray computed tomography and bone-embedded markers are reported to have better accuracy than palpation (Chadwick, 1999) but for ethical reasons were not suitable for this study.

Orthogonal axis sets are determined from the position vectors of three known points. Figure 2.1 represents a typical axis system for the pelvis.

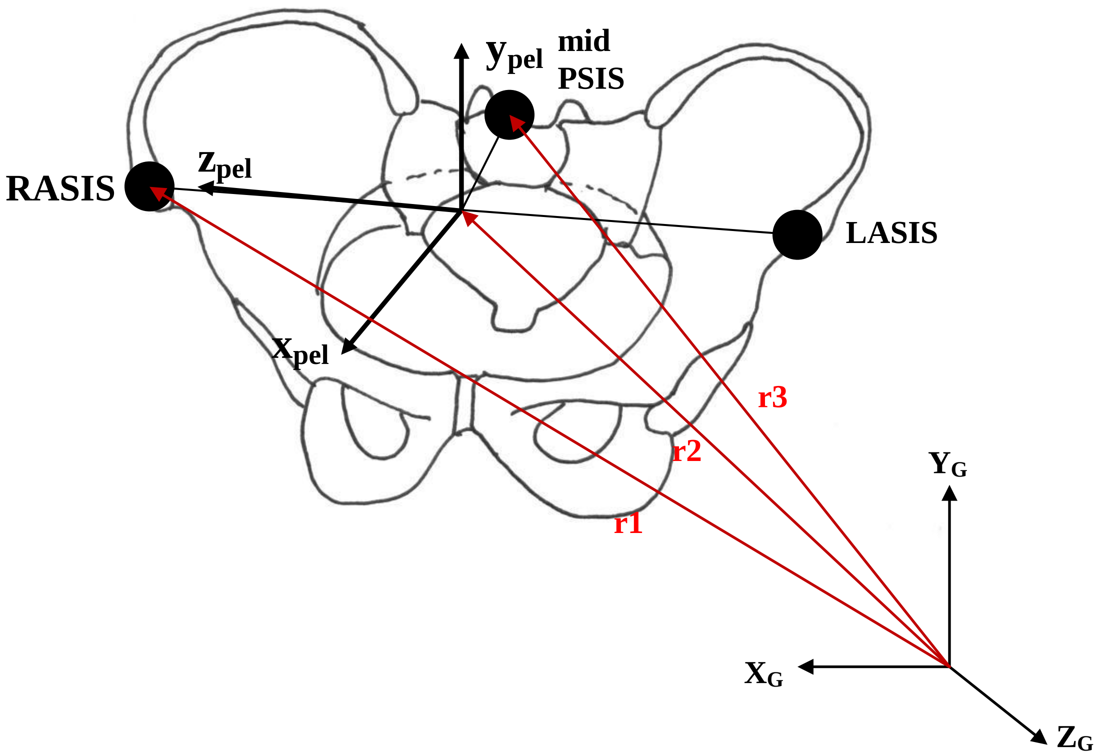


Figure 2.1 Determining three orthogonal axes from position vectors The position of the right anterior superior iliac spine (RASIS) and the midpoint of the front (mid ASIS) and rear (midPSIS) of the pelvis are determined by the position vectors $\mathbf{r}_1$, $\mathbf{r}_2$ and $\mathbf{r}_3$. The z axis of the pelvis $\mathbf{z}_{pel}$ is defined by the unit vector:

$$\mathbf{z}_{pel} = (z_1 i + z_2 j + z_3 k) = (\mathbf{r}_1 - \mathbf{r}_2)/|\mathbf{r}_1 - \mathbf{r}_2| \qquad (2.1)$$

The y axis determined as being orthogonal to the plane containing the three points. It is calculated by first estimating the x axis then using a vector cross product to calculate the orthogonal unit vector.

$$\text{Estimated } \mathbf{x}_{pel} = \mathbf{x}_{est} = (\mathbf{r}_2 - \mathbf{r}_3)/|\mathbf{r}_2 - \mathbf{r}_3| \qquad (2.2)$$

$$\mathbf{y}_{pel} = (y_1 i + y_2 j + y_3 k) = (\mathbf{z} \times \mathbf{x}_{est})/|\mathbf{z} \times \mathbf{x}_{est}| \qquad (2.3)$$

The true x axis is the cross product of the z and y axes

$$\mathbf{x_{pel}} = (x_1 i + x_2 j + x_3 k) = \mathbf{y} \times \mathbf{z} \qquad (2.4)$$

A rotation matrix (${}^{G}\mathbf{R}_{T}$) can be constructed for the bony embedded frame as:

$$({}^{G}\mathbf{R}_{T}) = \begin{bmatrix} x_1 & y_1 & z_1 \\ x_2 & y_2 & z_2 \\ x_3 & y_3 & z_3 \end{bmatrix} \qquad (2.5)$$

2.1.2 Analysis methods

Once the unit vectors of the three axes in the bone embedded system are known with respect to the global system it is then possible to represent the bone‟s position as a translation (**t**) and three dimensional rotation (${}^{G}\mathbf{R}_{B}$) from the origin of the global (laboratory) reference system (figure 2.2):

$$\underline{\mathbf{t}} = [t_1\ t_2\ t_3] \qquad (2.6)$$

$${}^{G}\mathbf{R}_{B} = \begin{bmatrix} i_{wrt}I & j_{wrt}I & k_{wrt}I \\ i_{wrt}J & j_{wrt}J & k_{wrt}J \\ i_{wrt}K & j_{wrt}K & k_{wrt}K \end{bmatrix} \qquad (2.7)$$

where **t** is the position vector of the origin of the bone embedded axis in the global system and $i_{wrt}I$ is the direction cosine of the ith axis of the bone embedded system with respect to the Ith axis of the global system. Any point in the bone embedded system ${}^{B}\mathbf{p}$ can be represented in the global axis system ${}^{G}\mathbf{p}$ by the following:

$${}^{G}\mathbf{p} = ({}^{G}\mathbf{R}_{B}.{}^{B}\mathbf{p}) + \underline{\mathbf{t}} \qquad (2.8)$$

As ${}^{G}\mathbf{R}_{B}$ is an orthogonal matrix of unit vectors ${}^{G}\mathbf{R}_{B}^{-1}$ is the transpose. In order to represent a point in the global axis system as a point in the bone embedded axis system the following calculation is performed:

$${}^{B}\mathbf{p} = {}^{G}\mathbf{R}_{B}^{-1}.({}^{G}\mathbf{p} - \underline{\mathbf{t}}) \qquad (2.9)$$

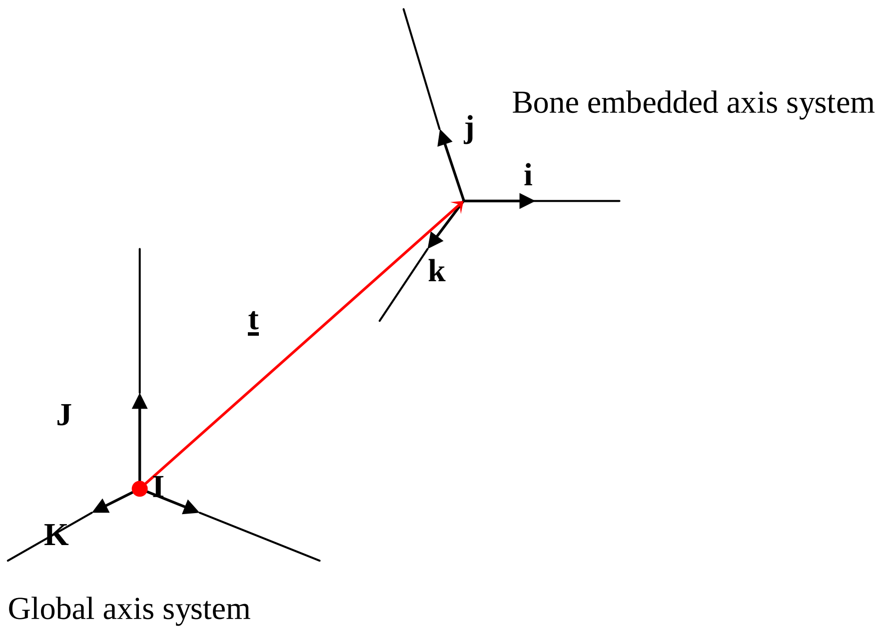


Figure 2.2 Representation of the bone embedded and global axis systems

The above equations relate to the position and orientation of one bone embedded system with respect to the global system. To determine joint kinematics, the relationship between two adjacent bones is needed. For this purpose the proximal bone embedded frame is described as X, Y, Z with unit vectors I, J, K and the distal bone embedded frame as x, y, z with unit vectors i, j, k. The position vector of the origin of the proximal segment in the global frame is $\underline{\mathbf{t}}_p$ and the position vector of the origin of the distal segment is $\underline{\mathbf{t}}_d$ (figure 2.3).

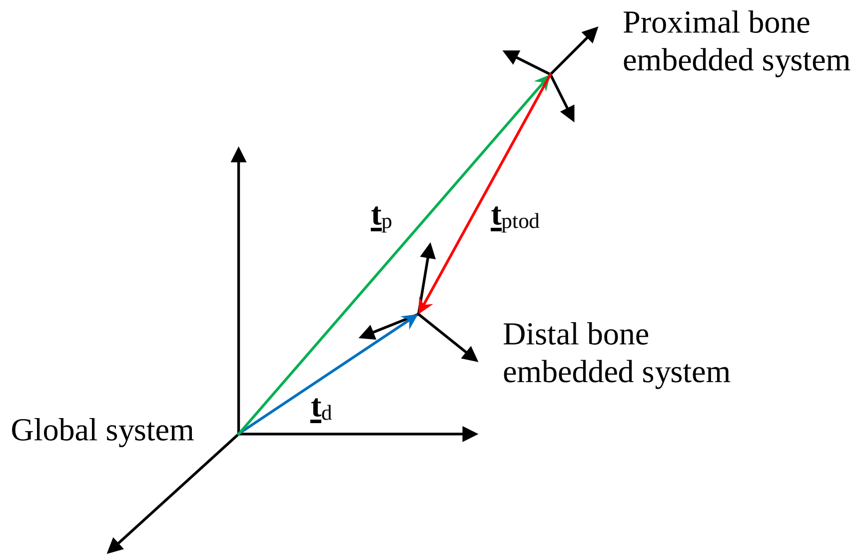


Figure 2.3 Representaion of the global, proximal and distal bone embedded axis systems

Using equation (2.8) and eliminating the global position vector:

$$ {}_p\mathbf{R}_d = {}_G\mathbf{R}_{B(proximal)}^{-1} . {}_G\mathbf{R}_{B(distal)} \quad (2.10) $$

$$ \underline{\mathbf{t}}_{(p\ to\ d)} = {}_G\mathbf{R}_{B(proximal)}^{-1} . (\underline{\mathbf{t}}_d - \underline{\mathbf{t}}_p) \quad (2.11) $$

The orientation of the distal co-ordinate system with respect to the proximal system (${}^p\mathbf{R}_d$) can be expressed as the scalar products of the unit vectors as follows:

$$ {}^p\mathbf{R}_d = \begin{bmatrix} \mathbf{I.i} & \mathbf{I.j} & \mathbf{I.k} \\ \mathbf{J.i} & \mathbf{J.j} & \mathbf{J.k} \\ \mathbf{K.i} & \mathbf{K.j} & \mathbf{K.k} \end{bmatrix} \quad (2.12) $$

Several methods have been proposed to determine the joint kinematics from the direction cosines obtained above, namely Euler and Carden angles, floating axis and screw displacement. A method considering the relative movement of one segment using a two-step method was proposed by Cheng et al (2000).

2.1.3 Euler and Cardan angles

Assuming that the two bony segments are aligned, the distal bone embedded frame can reach the same orientation as the proximal bone embedded frame by undergoing three successive rotations (Cappozo et al, 2005). The rotation matrix of the distal coordinate system relative to the proximal, ${}^{p}\mathbf{R}_{d}$, can be broken down into three basic rotation matrices (Fioretti et al, 1997), parameterised in terms of three component rotations of magnitude (α, β, and γ) around the co-ordinate axes of x, y and z where:

$$\mathbf{R}_x(\alpha) = \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos\alpha & -\sin\alpha \\ 0 & \sin\alpha & \cos\alpha \end{bmatrix} \quad \text{Rotation around x} \qquad (2.13)$$

$$\mathbf{R}_y(\beta) = \begin{bmatrix} \cos\beta & 0 & \sin\beta \\ 0 & 1 & 0 \\ -\sin\beta & 0 & \cos\beta \end{bmatrix} \quad \text{Rotation around y} \qquad (2.14)$$

$$\mathbf{R}_z(\gamma) = \begin{bmatrix} \cos\gamma & -\sin\gamma & 0 \\ \sin\gamma & \cos\gamma & 0 \\ 0 & 0 & 1 \end{bmatrix} \quad \text{Rotation around z} \qquad (2.15)$$

After these basic rotations have occurred, the joint orientation matrix can be determined following set rules (Cappozo et al, 2005). If the rotations were to occur around the distal z axis first, then the around the current distal x axis, and then the current distal y axis the matrix would be:

$$^{p}\mathbf{R}_{d} = \{[\mathbf{R}_{z}(\gamma)\ \mathbf{R}_{x}(\alpha)]\ \mathbf{R}_{y}(\beta)\} \qquad (2.16)$$

To determine the magnitude of the rotations the equation 2.12 and 2.16 can be expanded to: (2.17)

$$\begin{bmatrix} I.i & I.j & I.k \\ J.i & J.j & J.k \\ K.i & K.j & K.k \end{bmatrix} = \begin{bmatrix} \cos\gamma\cos\beta - \sin\gamma\sin\alpha \mathrm{sib}\beta & -\sin\gamma\cos\alpha & \cos\gamma\sin\beta + \sin\gamma\sin\alpha\cos\beta \\ \sin\gamma\cos\beta + \cos\gamma\sin\alpha\sin\beta & \cos\gamma\cos\alpha & \sin\gamma\sin\beta - \cos\gamma\sin\alpha\cos\beta \\ -\cos\alpha\sin\beta & \sin\alpha & \cos\alpha\cos\beta \end{bmatrix}$$

From this set of equations the angles, and hence the three rotations can be determined as:

$$\alpha = \sin^{-1} K.j$$

$$\beta = \sin^{-1}(-K.i/\cos\alpha) \qquad (2.18)$$

$$\gamma = \sin^{-1}(-I.j/\cos\alpha)$$

If the 1st and 3rd rotations were to occur around the same axis the angles are termed Euler, otherwise they are generally termed Cardan or Bryant angles. The three rotations are anatomically considered to correspond to flexion-extension, abduction-adduction and internal-external rotation, and a representation of this for the knee is presented in Fitzsimmons (1995).

The limitation of Euler and Cardan angles is that they are order dependent as matrix multiplication is generally not commutative. This means that the same three rotations must be used in the same order for results to be comparable. Angles have been reported in many different combinations in the literature and this lack of standardisation makes comparisons difficult.

2.1.4 Floating axis technique

The floating axis technique was developed by Grood and Suntay (1983) and uses three non-orthogonal unit base vectors denoted as $\mathbf{e}_1$, $\mathbf{e}_2$ and $\mathbf{e}_3$. Two of the axes, $\mathbf{e}_1$ and $\mathbf{e}_3$ are embedded in the proximal and distal segments respectively and the third, or floating axis, $\mathbf{e}_2$ is the common perpendicular (figure 2.4). This axis is described as floating as it is not fixed in either bone, but moves in relation to both bone embedded axes. The rotations between the two bodies are represented by angles α,β, and γ. Two of these relative rotations (α and γ) can be considered as a spin of each body on its own fixed axis whilst the other body remains stationary. The magnitude of these rotations are measured by the angle formed between the floating axis and a conveniently selected reference axis in each segment. The third relative rotation occurs around the floating axis and is the angle between the two bone embedded axes (β).

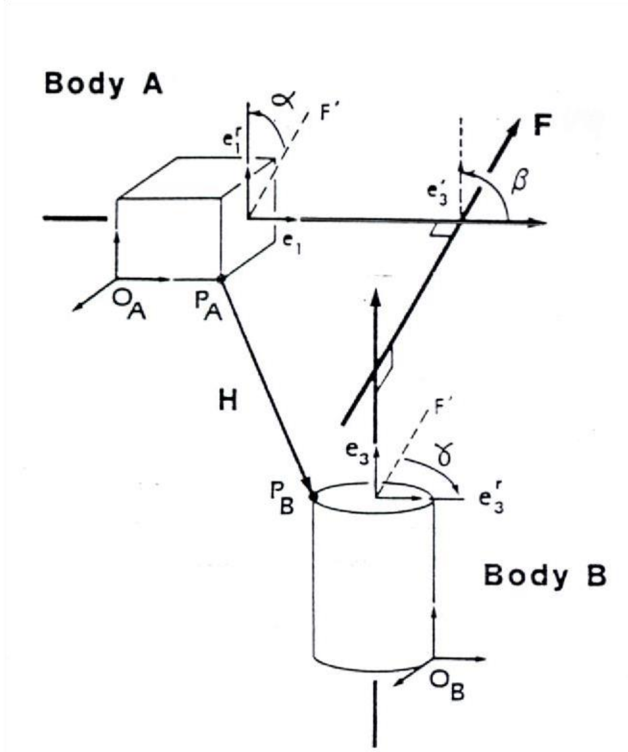


Figure 2.4 The floating axis joint co-ordinate system. Two axes are embedded in the proximal and distal segment with unit base vectors $\mathbf{e_1}$ and $\mathbf{e_3}$. The third axis, F, is the common perpendicular and has unit base vector $\mathbf{e_2}$. (Grood and Suntay 1983)

Grood and Suntay (1983) applied the floating axis technique to knee joint kinematics as shown in figure 2.5. The long axis of the tibia was chosen as $\mathbf{e_3}$ with rotations about it corresponding to knee internal/external rotation. The femoral medial lateral axis, Z, was chosen as $\mathbf{e_1}$ with rotations about it corresponding to knee flexion/extension. Movements about the floating axis therefore corresponded to knee abduction and adduction.

Mathematical analysis detailed in the above paper by Grood and Suntay demonstrates how the angles of rotation can be determined from the direction cosine matrix described in equation 2.12:

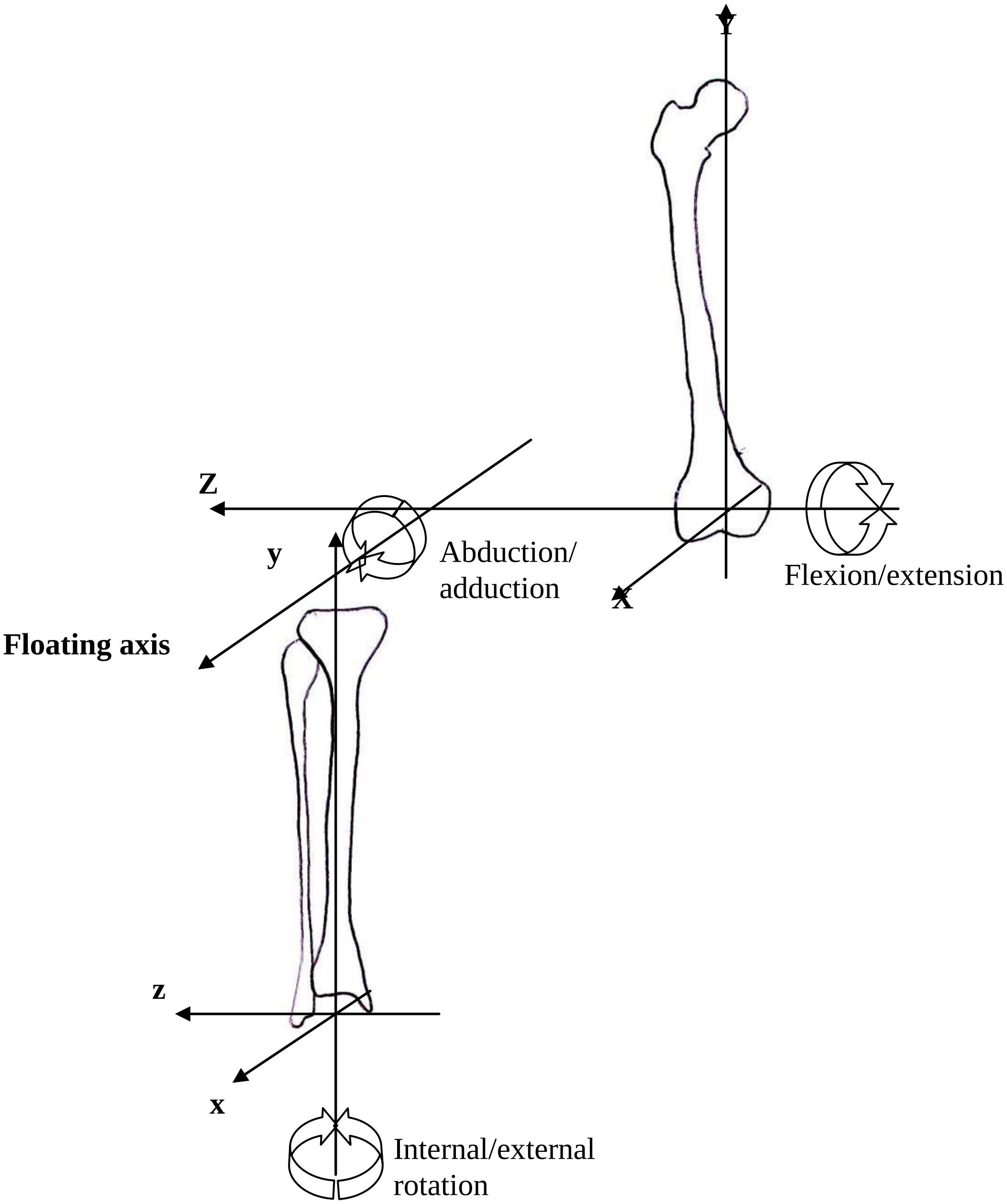


Figure 2.5 Floating axis technique utilised by Grood and Suntay (1983) to determine knee joint angles

I.i I.j I.k cosαcosγ-sinγsinαcosβ sinβsinα –cosαsinγ-
cosγsinαc

J.i J.j J.k = -cosγsinα-cosαcosβsinγ cosαsinβ
sinαsinγ- cosγcosαc

K.i K.j K.k **sinγsinβ** **cosβ** **cosγsinβ**

(2.19) By equating elements of these two matrices the three joint angles can be determined from the direction cosines (Fitzsimmons, 1995):

$$\boldsymbol{\alpha = \tan^{-1} (K.i/K.k)} \qquad (2.20)$$

$$\boldsymbol{\beta = \cos^{-1} K.j} \qquad (2.21)$$

$$\boldsymbol{\gamma = \tan^{-1} (I.j/J.j)} \qquad (2.22)$$

For the example of the knee, flexion was represented by α, abduction by (β-π/2) for the right knee and (π/2-β) for the left knee, and tibial rotation by γ.

The floating axis technique has benefits compared with Euler angles as it is not order dependent. The selection of the reference axis allows measurements to be easily related to anatomical structures and movements to be described in manner easily related to the anatomical and clinical descriptions of joint movements.

2.1.5 Screw displacement axis

This method describes the rotation between two segments as a rotation about an axis followed by a translation along the same axis. This axis will be uniquely defined for the system and the method of calculation is detailed elsewhere (Woltring, 1991; Woltring, 1994) . The disadvantage of this method is that it does not allow simple description of the joint angles in clinical terms (Chadwick, 1999) and the rotational and translational magnitudes are not comparable directly with other methods (Cheng, 1996). For this reason this method will not be detailed further in this thesis.

2.1.6 Two-step rotation method

A two-step rotation method has been proposed, describing the movement of a limb segment from one attitude to another (Cheng, 1996; Cheng et al., 2000). The first rotation is the rotation of the long axis of the limb segment about a specific axis passing through the proximal joint, perpendicular to the long axis, and the second is an axial rotation about the long axis. This method is particularly suited for determining the axial rotation of a segment. The limitation of this method is that it provides the relative pose of the segment to that of its starting position and does not consider the proximal limb segment.

2.1.7 Comparison of kinematic calculation methods

Fioretti et al. (1997) compared Cardanic, floating axis and the helical axis method in computing knee joint angles during gait. They concluded that there was no significant difference in the angles obtained from any of the methods and that one method could not be considered superior to another. More important was the accuracy with which the bone embedded axis system was located.

The floating axis method was chosen to analyse the joint kinematics in this thesis due to the joint angles being produced in a clinically understandable manner. This method also eliminates the need to consider order dependence. In all cases the axis of flexion and extension ($\mathbf{e}_1$) was considered to be the Z axis of the proximal segment, the axis around which internal and external rotation occurs ($\mathbf{e}_3$) was the y axis of the distal segment and add/abduction is described as occurring about the floating axis.

2.2 Kinetics

Kinetics is the study of the forces and moments that occur during an activity. Moments and forces resulting from external loading of a limb segment from the environment are balanced by internal loading from the muscles, ligaments and joint surfaces. There are three types of external loading, namely gravitational loading, inertial loading and contact loading.

Gravitational loading on a body segment is due to the effect of gravitational pull on the mass of the segment. It acts vertically downwards and at the location of the centre of mass of that segment, with its magnitude being the mass of the segment (**m(seg)**) multiplied by the acceleration due to gravitational pull(**g**):

$$\mathbf{F(g)=m(seg) * g} \qquad (2.23)$$

If a body segment of mass **m(seg)** is accelerating with a linear acceleration **a**, the force producing that must be **ma**. D‟Alembert‟s principles consider that a force must act in the opposite direction to obtain equilibrium and this is the inertial force (or loading) (**F(i)**):

$$\mathbf{F(i)=m(seg)*a} \qquad (2.24)$$

If an angular acceleration (**α**) occurs it is opposed by inertial torque (**T**):

$$\mathbf{T = I(seg)\alpha = m(seg)k^2 \alpha} \qquad (2.25)$$

Where **I(seg)** is the segment mass moment of inertia and **k** is the segment radius of gyration.

Contact force is the force between the body and its environment. The contact forces are the reaction of the surface to the gravitational and inertial forces acting on the body in order to maintain equilibrium. The contact forces can be measured by force plates or other suitable strain gauged instrumentation.

2.2.1 Body segment parameters

In order to calculate the magnitude of the external loading it is necessary to know the mass of body segments, the location of the centre of mass in the segment and the radii of gyration. There have been many studies of body segment parameters and a review of these can be found in Pearsall and Reid (1994). Studies have been performed on cadavers, (Chandler et al., 1975; Dempster (1971-1973) from Winter, 1990) and on live subjects using various estimation techniques (Drillis and Contini, 1966; Zatsiorsky and Seluyanov, 1983). Uncertainty exists as to the accuracy of using cadaver studies to represent the living state due to the partially embalmed state of the tissues (Pearsall and Reid, 1994).

Ideally, one would wish to use data for body segment parameters collected from a group of subjects similar to the group being studied (male and female older adults in the current research). No study was found containing information on segment mass and centre of mass location for all of this population and therefore a compromise must be made. The effect of incorrectly estimating body segment parameters will result in some error in kinetic results but this has found to be low at about 1% of subject body weight (Petrella et al., 1997; Bothner et al., 2001).

2.2.2 Joint centres

In order to maximise the accuracy of joint moment calculation and the location of bone-embedded axes, it is necessary to locate the joint centres. This section discusses methods used to determine the joint centres of the hip, knee, ankle, shoulder, elbow and wrist.

2.2.3 Hip joint centre

Several methods have been proposed in the literature to determine the position of the hip joint centre, but most accepted are either a functional or prediction approach. The functional approach (Cappozzo, 1991; Shea et al., 1997; Leardini et al., 1999) estimates the hip joint centre to be at the pivot point of the femur and pelvis and is estimated by calibration movements of the hip. Leardini et al. (1999) reported this method to be most accurate compared to the "gold standard" of roentgen stereophotogrammetric analysis (RSA). Besier et al. (2003) found the functional approach produced more repeatable hip moments in the frontal plane during gait than a prediction approach, although there was no difference in moments in the other planes or joint kinematics. One of the limitations of the functional approach is that 40-45 degrees of hip flexion and abduction are required to perform it accurately. Although the subjects in this study were anticipated to have this range of motion it was felt they may have difficulty balancing to perform this movement. Another potential problem with the prediction method is that it is based on calculating marker velocity. If any error is made in determining marker location, such as measurement "noise", this will give a large amount of variation in the computed locations of the instantaneous centres (Lamoreux, 1996)

Several prediction methods have been proposed (Andriacchi and Stickland, 1983; Bell et al., 1989; Bell et al., 1990; Davis et al., 1991; Seidel et al., 1995) . All these methods estimate the position of the hip joint centre from regression equations based on pelvic dimensions. The methods of Andriacchi and Strickland (1983) and Seidel et al. (1995) require accurate palpation of the pubic symphysis and were not considered for this study due to the difficulty in palpating this landmark. Recent studies (Leardini et al., 1999; Stagni et al., 2000) support the use of Bell"s method in the absence of a functional approach as it has better accuracy in a medio-lateral and anteroposterior direction than the other prediction methods. Misposition of the hip joint centre in the above two directions was found by Stagni et al to have the greatest impact on hip joint moment errors.

This study used the method of Bell et al. (1989) to estimate the location of the hip joint centre based on the positions of the anterior superior iliac spines of the pelvis. In the standard axis system of the pelvis the hip joint centre was located at a point 14% of the inter ASIS distance medial to, 30% of the inter ASIS distance distal to, and 19% of the inter ASIS posterior to the corresponding ASIS.

2.2.4 Knee joint centre

The knee joint exhibits complex joint motion due to rolling and sliding between the joint surfaces during flexion and extension. The axis of rotation is therefore not fixed but moves along a path based on the geometry of the articulating surfaces. The knee has been modelled as a ball and socket joint, a four bar cruciate linkage and a simple hinge joint (see review in (Siegler and Liu, 1997)). To facilitate analysis many authors have simplified the knee joint to a hinge with a fixed point for the joint centre.

The knee joint centre can be defined in the thigh co-ordinate system or the shank coordinate system. In the femoral (thigh-based) axis system, it can be considered it to be at the midpoint of the medial and lateral epicondyles of the femur (Davis et al., 1991; Cappozzo et al., 1995; Stagni et al., 2000). The knee joint centre location in a shank based co-ordinate system was based on anatomical measurements from eight tibiae (Ishai, 1975). This method was used by Fitzsimmons (1995) and was modified by Stansfield (2000) to include anthropometric data from other sources. No reports are available to suggest which method would lead to greatest accuracy. It is suggested that researchers chose which method best suits the study to be undertaken (Siegler and Liu, 1997). Fitzsimmons (1995) proposed that the shank based system may be preferable due to the lower skin movement artefacts in the shank compared to the thigh. However in this study, as a rigid cluster based marker system is utilised for the thigh, these errors would be reduced. A thigh-based system was therefore selected in this study, with the knee joint centre being predicted as the mid-point between the medial and lateral femoral epicondyles in the axis system of the femur.

2.2.5 Ankle joint centre

Modelling of the ankle joint has tended to result in a large simplification to a two hinge or one hinge joint (Lundberg, 1997). Problems arise due to the relatively small segments and the difficulty in defining the talus from the surface. The two hinge model represents the talocrural and talocalcaneal joints separately and the one hinge model considers all joints between the marker segments as one.

Many authors have reported a one hinge model (Ishai, 1975; Davis et al., 1991; Fitzsimmons, 1995; Kirkwood et al., 1999; Stansfield, 2000). Fitzsimmons and Ishai determined the longitudinal axis of the shank from the location of the fibula head, lateral malleolus and tibial tuberosity with a set of equations determined from cadaveric tibiae. They considered the ankle joint to be located at the level of the lateral malleolus on the longitudinal axis of the shank. Stansfield (2000), considered the ankle as just the talocrural joint to allow inclusion of dorsiflexion and plantarflexion. The joint centre was calculated using the anthropometric data from Isman and Inman (1968) (reported in Stansfield (2000)). Davis et al. (1991) considered the ankle joint centre to be half the intermalleoli distance medial to the lateral malleolus. Similarly Kirkwood et al. (1999) considered the ankle centre to be at the midpoint of the medial and lateral malleoli.

This study considered the ankle joint to be a single hinge with the joint centre predicted at the mid-point of the malleoli.

2.2.6 Shoulder joint centre

The glenohumeral (GH) joint is considered as a perfect ball and socket joint with relatively little GH joint translation during active movement (less than 1mm reported by (Graichen et al, 2000)). The shoulder joint centre is considered to be at the geometric centre of the humeral head and as this is not a palpable point it has to be estimated.

Several methods of estimating the location of the GH joint centre from external landmarks have been proposed including calibration movements (Rau et al., 2000;

Stokdijk et al, 2000), regression equations from scapular landmarks (Meskers et al, 1998) and estimation from the position of the acromion (Cheng, 1996; Schmidt et al, 1999). The International Shoulder Group (ISG) and the ISB recommend the regression method of Meskers et al (1998) but this is not possible in this study due to the use of an optical system and the large skin movement errors that would occur if placing markers over the scapula (Van der Helm, 1997). Cheng (1996) and Scmidt et al. (1999) estimated the position of the shoulder joint centre from the acromion process, Schmidt et al. predicting it to lay a distance 7cm inferiorly and Cheng predicting the location of the joint centre for each subject by observation and measurement with callipers. In a cadaver study, reported by Stodijk et al. (2000), the shoulder joint centre lay at 4.32 cm below the acromion angle in the scapular co-ordinate system. Although Schmidt et al described the position in the global system it would appear that the distance of 7 cm would be too large.

Stokdijk et al. (2000) compared three methods of predicting the GH joint centre; that of Meskers et al. (1998), a sphere fitting technique, and an optimal helical axis technique. They found all of the measures to be reproducible although the actual position predicted varied between the methods, with the reliability of the regression method being poorest compared to cadaver studies. They found sphere fitting and the helical axis method had good inter- and intra-rater reliability and recommended the use of the helical axis method due to its shorter calculation time. To the author‟s knowledge no studies

have compared the accuracy of any of the methods in vivo.

This study used the method of Cheng (1996) to determine the glenohumeral joint centre. A pointer calibration method was used to locate the acromion process and the GH joint centre was predicted to lie at a distance 37mm inferior, 14mm lateral and 8mm anterior to it in the co-ordinate system of the trunk.

2.2.7 Elbow joint centre

The ISB committee on the elbow joint (ISB, 1996) describe both radius and ulnar joint coordinate systems when describing the forearm, leading to joint centres for both these articulations. This study did not require such detail at the elbow joint and so the radius and ulnar are considered as one segment. Other studies using this method have estimated the centre of the joint to be located at the mid-point between the medial and lateral epicondyles (Schmidt et al., 1999; Anglin and Wyss, 2000; Rau et al., 2000).

In keeping with previous work this study estimated the elbow joint centre to be located at the mid-point between the medial and lateral epicondyles of the humerus.

2.2.8 Wrist joint centre

The wrist joint centre is generally considered to be at the mid-point of the ulnar and radial syloids (Cheng, 1996; Scmidt et al., 1999; Anglin et al., 2000; Rau et al., 2000), and this definition was used for this study.

2.2.9 Whole body centre of mass calculations

Studying the movement of the COM of the body may highlight differences in movement strategies in older adults. Several methods of calculating the position of the COM have been proposed with varying levels of complexity.

One method is to double integrate the three components of the ground reaction force with respect to time. (Crowe et al., 1993; Shimba, 1994).

$$CM_i = \iint_t \frac{F_i}{m} .dt \qquad (2.26)$$

Where F_i (i = x, y, z) is the i[th] component of the ground reaction force, m is the total body mass and t is time. This method has been suggested by the authors as the gold standard of COM calculation but relies on force plate contact throughout the activity.

Another method, which avoids the difficulties associated with using force plates, is to assume that the COM is a fixed point in the pelvis. Investigators have chosen the sacrum or the mid-point of the pelvis to represent COM (Saini et al., 1998).

A third alternative estimation method is to calculate the position of the COM from full body kinematics, known as a segmental method. The total body COM can be considered as the weighted sum of the individual body segment COM.

$$CM_i = \frac{\sum_j m_j . p_{i,j}}{\sum_j m_j} \qquad (2.27)$$

Where m_j is the mass of segment j, and $p_{i,j}$ is the i[th] component (i = x, y, z) of the position vector of its centre of mass.

It is important when choosing a method to adopt one which is most suitable for the activity being studied. Saini et al. (1998) compared the three above methods, comparing vertical COM displacement during gait. They found no difference between two models basing COM as a fixed point in the pelvis and a segmental method, concluding that a single marker on the sacrum may be all that is required for COM estimation in gait. However, for the segmental method they considered the upper body as a rigid segment attached to the pelvis and therefore did not determine the influence of limb and trunk motion. When they compared the estimation methods to the force plate method they found results differed significantly. Whittle (1997), compared COM determined from the force plate and pelvis estimation method and found that during gait the COM moved within the pelvis due to the influence of the motion of the trunk, head and limbs. Eng and Winter (1993) found a segmental estimation method produced similar results for lateral COM displacement to the force plate method, but that a single marker produced different results. Rabufefetti and Baroni (1999) suggested that the specific model for COM position estimation selected should reflect the specific analysis that is being performed. They developed a generalised method for the assessment of any COM model by comparing the model outputs to the expected ballistic trajectory when airborne and force platform data when on the ground. They compared these absolute assessments of COM trajectory to those calculated by the pelvis method and the weighted body segment method. For a range of activities including jumping, bending and kneeling they found the segmental method to be more accurate.

As the moments in this study were varied and not always in contact with a force plate a segmental method was chosen to represent full body COM.

2.3 Conclusion

Three-dimensional joint angles can be calculated using several methods of which this thesis has selected the floating axis method. Using the knowledge about the orientation of the bones, combined with the positions of the joint centres, centres of mass and the contact loading a free body diagram can be created. From this joint and intersegmental forces and moments can be calculated. The specific method used for this experimental work is detailed in Chapter 3.

Chapter 3

Biomechanics Methodology

Chapter 2 introduced some of the methods used to determine body segment kinematics and kinetics. This chapter describes in detail the methods used in this study to collect and interpret kinematic and kinetic data.

The biomechanics laboratory at the University of Strathclyde is equipped with a Vicon motion analysis system and so this was the method used to collect data. As the output from the laboratory was required in an easily manageable form for the team at the Glasgow School of Art it was decided that Vicon Bodybuilder software would be used to process the marker data and compute joint kinematics and kinetics.

3.1 Subject recruitment

Subjects were recruited from the geographical area around Glasgow, UK over a period of time between November 2002 and May 2004. Initially information leaflets were distributed to older adults‟ organisations in the surrounding area. Advertisements were placed around the University inviting older students (at the Centre for Lifelong Learning) and staff to participate. Presentations were performed at retired adults clubs, to exercise groups for older adults and to Abbeyfield Society group houses for older adults. No financial reward to participate was offered and subjects were required to attend for 2 full days of testing. Transportation and meals were provided.

From the subjects approached, 125 people volunteered to participate. These subjects were contacted by a member of the research team and medical screening was carried out to exclude those with potential medical and health problems. As part of the testing for the EQUAL project was maximal isometric strength testing a number of exclusion criteria were required:

Exclusion Criteria for older adults:

1. History of myocardial infarction within the previous 2 years
2. Unstable angina
3. History of acute myocarditis, acute pericarditis, aortic stenosis, valvular dysfunction, dysarrhythmia within the previous 10 years
4. Pulmonary disease including severe asthma, chronic bronchitis
5. Pulmonary embolus within the last 2 years
6. Cerebrovascular disease including hemiplegia or hemiparesis
7. Systemic disease active within the previous 2 years e.g. Cancer
8. Lower limb fractures sustained within the previous 2 years and upper limb fractures within 6 months
9. Severe arthritis of joints characterised by inability to perform maximal voluntary contractions without pain and presenting with severe limitation of movement
10. Severe Osteoporosis
11. Neurological disorders including Parkinson‟s syndrome, Multiple Sclerosis, Myasthenia gravis, Poliomyelitis
12. Severe hypertension with resting blood pressure of Systolic > 200 mm Hg and Diastolic > 100 mm Hg
13. Severe impairment of balance and coordination

14. Deep vein thrombosis
15. Alzhiemer‟s disease and other Psychotic illnesses (Inability to provide informed consent and inability to follow instructions)

Following screening, 21 subjects were excluded from the study, leaving a pool of 104 potential volunteers. A further 8 subjects were unable to attend due to temporary illness or poor health of their spouse. Of the 96 remaining subjects there were a larger number in the 60-70 age group than the other groups. It was decided to investigate 30 subjects in each age group and subjects were investigated in the order they had volunteered to prevent any selection bias by the investigators.

Ethical approval for the study had been obtained via the medical ethics committee of University of Strathclyde. Subjects were provided with an information sheet and informed consent was signed prior to participation (appendix 3.1).

3.2 Data collection

3.2.1 The Vicon system

Marker position data were collected using the Vicon motion analysis system from Oxford Metrics Ltd. This system consists of eight charge-coupled device video cameras linked to a data station and a PC. Each camera emits pulsed infrared light at 120 Hz from an array of light emitting diodes and detects reflection of this light from any retroreflective marker within its field of view. Any one camera produces a twodimensional image of a given marker and if two or more cameras receive reflections then the three-dimensional coordinates of the marker in the laboratory space can be reconstructed.

The positions of the cameras within the laboratory measurement space are determined by a calibration procedure. A two stage dynamic calibration was used for this experiment. First an L-shaped calibration frame was placed over the far right force platform when looking along the laboratory. This had four 25mm fixed markers arranged to allow calculation of the origin of the laboratory and the three orthogonal axes of the laboratory system. The second stage used a wand with two 50 mm markers mounted at a known separation. Data were captured as an operator waved the wand throughout the volume required. Then using data from all of the cameras the computer computes the location and orientation of the cameras in relationship to the laboratory. A calibration residual was produced for each camera following the second stage. This is the average distance by which each direction-measurement from the camera concerned deviates from the location of the markers used in the calibration. The calibration residual was below
1mm for all trials.

3.2.2 Marker positions

The markers used were spheres of 14mm diameter covered in retroreflective tape. A total of 52 markers were used and the placement of these is shown in figure 3.1 and on a subject in figure 3.2. The positions of the markers were chosen to minimise skin movement whist also considering the nature of the movement being performed.

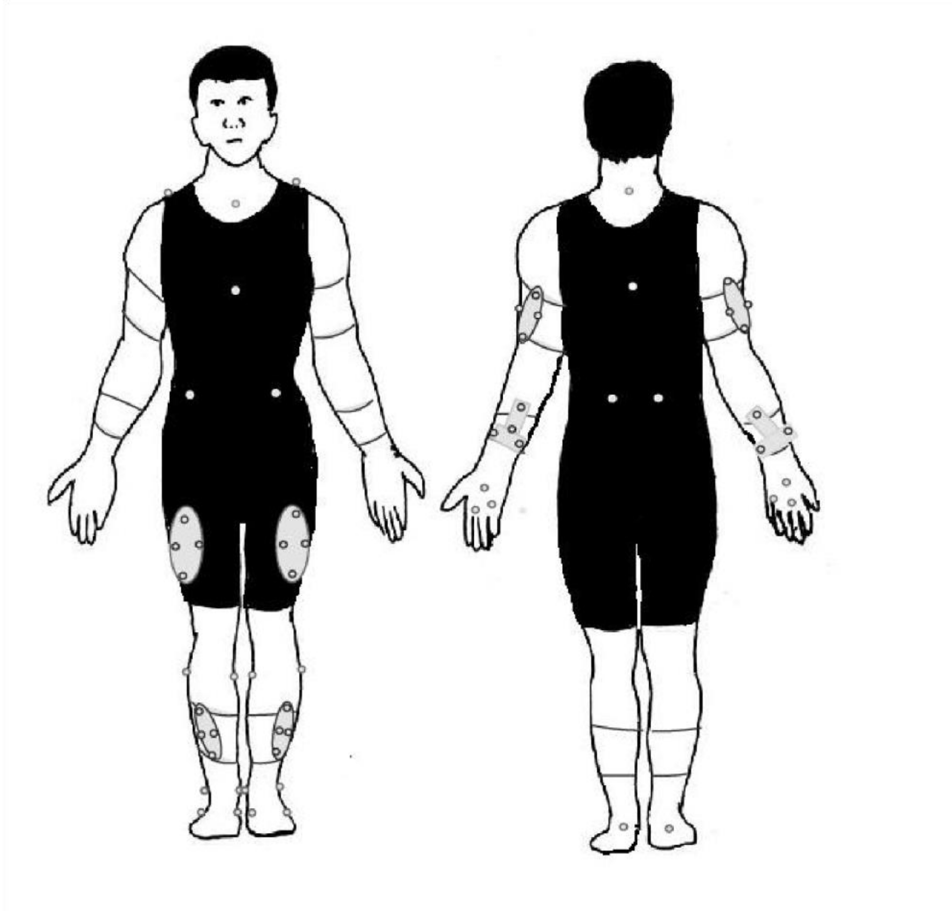


Figure 3.1 Marker placement

Individual markers on: ASIS and PSIS right and left pelvis (4 markers), 3 on the foot in a plane parallel to the floor, 7^{th} cervical vertebrae, 8^{th} thoracic vertebrae, jugular notch, xiphoid process, base of 3^{rd} metacarpal, 3^{rd} metacarpal head, 5^{th} metacarpal head. Technical clusters of 4 markers on upper and lower arm, thigh and shin secured with neoprene bands

<u>Anatomical calibration points</u>:
Lateral epicondyle of femur
Medial epicondyle of femur
Medial malleolus
Lateral malleolus
Acromion process
Lateral epicondyle of humerus*
Medial epicondyle of humerus*
Ulnar styloid *
Radial styloid*

* pointer calibration used

Figure 3.2 Markers placed on a subject.

Knee markers have been removed post calibration

Rigid clusters were used on the trunk, arms, forearms and thighs and were attached to the subject with adjustable neoprene cuffs. The neoprene provided a thin layer of cushioning between the rigid plastic mount, which had been heat moulded to follow the anatomical shape, and also prevented slipping of the cuff due resistance between it and the skin. Using the cuffs facilitated marker mounting on subjects and the use of sufficiently wide cuffs is believed to reduce soft tissue movements. The use of rigid clusters eliminated the need for optimisation procedures to remove deformation in the cluster caused by movement of markers in relation to one another. Clusters were placed as distal as possible on the relevant section to try to better represent the axial rotation of that segment.

Four markers were used on each cluster in a non-linear configuration. Other authors have used clusters containing four (Cappello et al., 1997; Lucchetti et al., 1998) or more markers (Alexander and Andriacchi, 2001) and used least squares methods to determine the centre of the cluster. Cappozzo et al. (1997) performed a study on optimal surfacemounted cluster design and felt that four markers per cluster provided a good practical solution. Manal et al. (2000) found that a rigid cluster of three markers was optimal for estimating tibial rotations, when comparing 11 possible designs. Work done on optimal marker configurations by Lucchetti (1995) and reported by Chadwick (1999) found that four markers on the upper limb segments lead to difficulties with broken trajectories and marker crossover when reconstructing data. Experimental work to determine optimal camera positions found that the clusters of four markers used in this study could be tracked without problems with crossover.

3.2.3 Contact loading

Three Kistler force plates, one model 9281B and two model 9261A, were used to measure contact forces between the ground and the subject. Sampling frequency was 1080 Hz and synchronised with the Vicon motion capture system. The three force plates were positioned in the laboratory co-ordinates system as shown in figure 3.3.

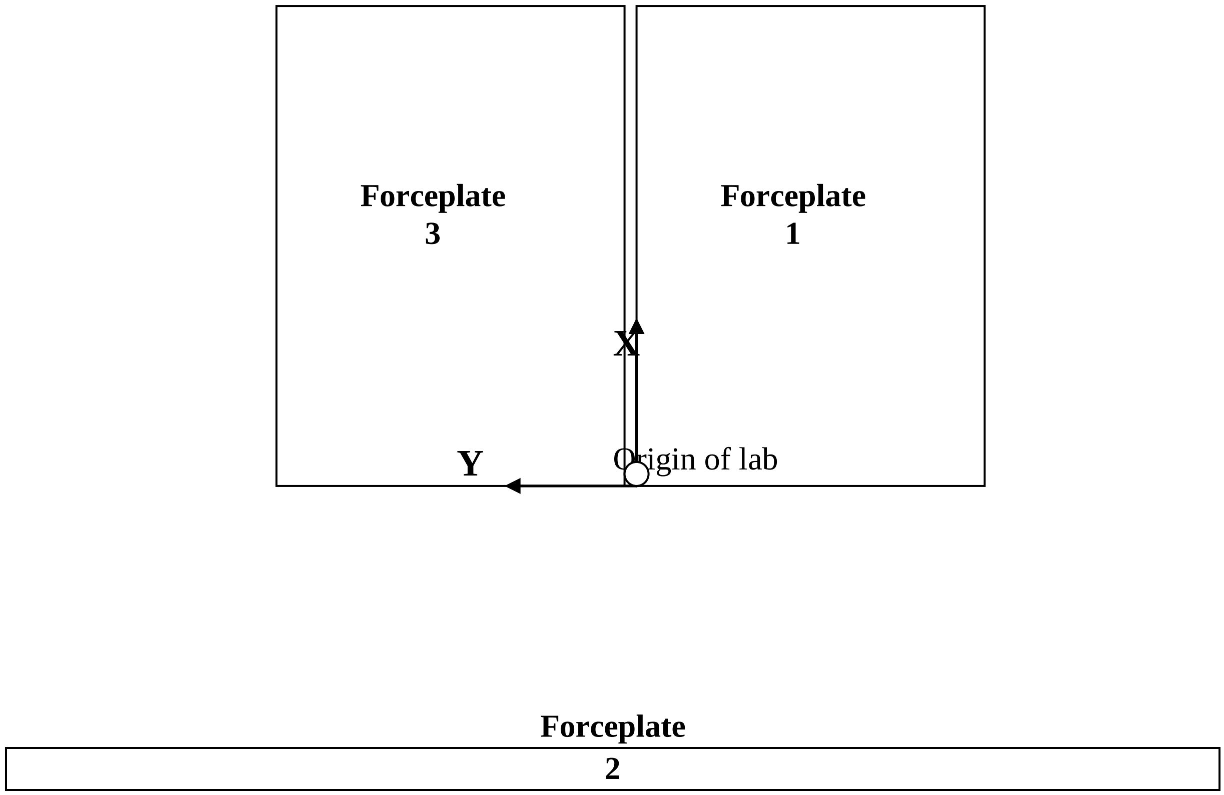


Figure 3.3 Force plate arrangement in the laboratory co-ordinate system

Data obtained from the force plates are the three components of force (Fx, Fy and Fz) and the three components of the moment (Mx, My and Mz) acting at the origin of the force plate. The origins of the force plates (in mm) in the global system were:

Force plate 1 = (300, -54, -200)

Forceplate 2 = (300, -37, 206)

Forceplate 3= (-306, -37, 18)

3.2.4 Staircase

A custom made staircase had been constructed in the Bioengineering Unit for a previous project (Fitzsimmons, 1995). A photograph of this staircase can be seen in figure 3.4 and the dimensions are provided in figure 3.5. It consisted of four steps with a platform at the top. The whole structure was enclosed with handrails to minimize the risk of falls. The step depth was 280mm and the rise was 185mm. This was in line with the Building Regulations of 2000, which advises that in public places the rise should be no more than 190mm and the depth should be greater than 250mm. In a private dwellings the maximum rise can be 220mm and the depth a minimum of 220mm. The handrail was at a height of 900mm above the step height, which was in line with building regulations that advise a height of 900-1000mm.

A section second step was securely bolted to one of the Kistler forceplates by means of four bolts, one in each corner of the surface of the forceplate. This mechanism had been tested and considered to provide good accuracy by Fitzsimmons (1995). The section of the step attached to the forceplate was half of the width of the staircase. This enabled the study of different types of stair gait, either both feet on one step or the reciprocal one over one method.

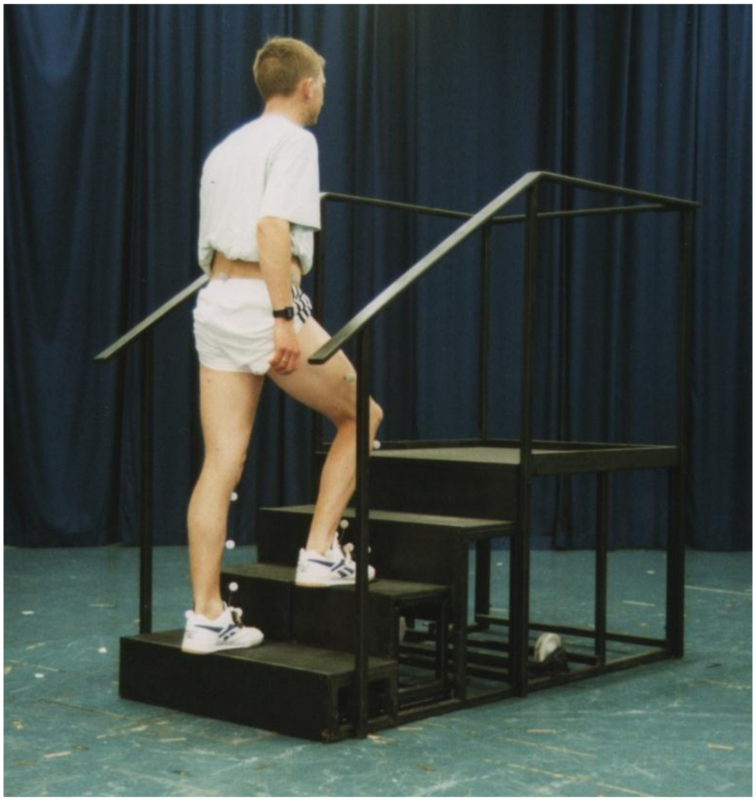


Figure 3.4 Staircase used for stair climbing task

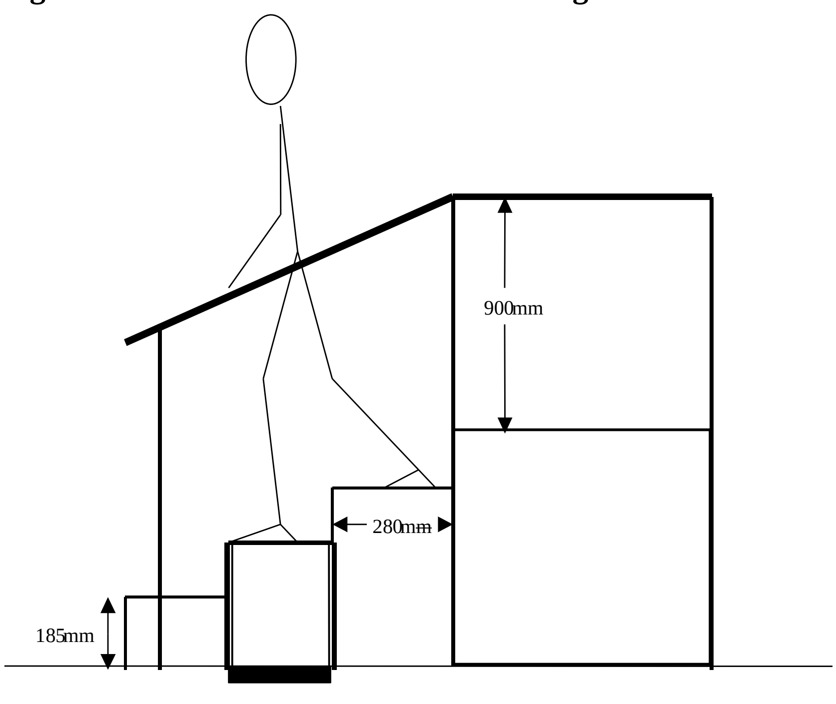


Figure 3.5 Dimensions of staircase

3.2.5 Subject calibration

The subjects wore lycra bodysuits and the markers were placed on them as detailed earlier. Lycra shorts have been found to reduce some soft tissue movement and so reduce errors caused by marker movement (Hazelwood et al., 1997). Static trials were used to identify bony landmarks using the wand technique or additional markers. These markers were removed during dynamic activities. Information on subject height and bodyweight were used to create a calibration file for that individual subject and enable kinematic calculations.

3.2.6 EQUAL testing protocol

Subjects attended the Bioengineering Unit at the University of Strathclyde on two separate occasions normally within a week of each other in order to complete all aspects of the EQUAL project testing (physical examination, psychology, hand biomechanics and full body biomechanics). The session normally lasted around 2 hours and prior to testing subjects were well rested.

On arrival to the laboratory subjects were asked to change into the lycra bodysuit. For modesty some men preferred to wear running shorts on top which still allowed visualisation of the ASIS‟s. Subjects had markers placed on them as detailed in section 3.2.2 secured using toupee tape. Clusters of markers were attached to the thigh, shank, upper and lower arms using custom made neoprene cuffs secured with Velcro. Once all markers were attached, the subject was requested to walk briskly around the laboratory to ensure that they were comfortable with the markers in place, and that no markers were in a position where they may become dislodged. Static calibration trials were then collected using the pointer technique to identify anatomical landmarks.

Subjects then performed the following activities in the following order whilst data were captured using the Vicon motion analysis system:

1. Normal walking – sufficient trials to obtain foot contact on the forceplate three times with both left and right feet
2. Sit to stand and stand to sit three times with each leg on a forceplate, first using hands on the arm rest and then without the arm rest if possible
3. Walking and opening a door, walking through it and closing it – three trials of pulling the door towards and three of pushing it away.
4. Lifting a can to a high shelf three times and a low shelf three times.
5. Ascending and descending stairs sufficient times to get three good foot contacts on the forceplate (normally 3 attempts) first with a handrail and then three attempts with no handrail if possible.

Subjects were given as much rest as they felt they required during testing, normally at least 10 minutes between different activities and were provided with refreshments. Trials were assessed for missing marker trajectories and repeated if necessary. Although the laboratory session was long it was not felt that subjects suffered from fatigue as all activities were performed to the subjects‟ capability level.

3.2.7 Stair testing protocol

Subjects were asked initially to walk up and down the staircase twice to familiarise themselves with the equipment. The investigator measured a run up to the stairs which allowed the subject to take three steps before reaching the stair and then take the first step up with the left foot leading. This meant the right foot would make contact with the forceplate on the second step, which was located in the right hand side of the staircase. The run up allowed subjects to get into a rhythm of walking which was felt to be more natural than commencing stair climbing form a static position. At the top of the staircase, subjects were asked to turn and then pause to ensure they had got their balance. They then descended the stairs with their right leg leading in order that the left foot made a clean contact with the forceplate step. For the practice attempts subjects were instructed to hold onto one of the handrails.

Data were collected from three attempts of stair ascent and descent using the handrails and then three attempts of stair ascent and descent without the use of the handrails if the subject felt they were capable of doing this. A rest was allowed between each attempt whilst the investigator ensured that the data captured was of good quality (i.e all markers were visible for the majority of the trial). For the trials where the handrails were used the subjects were instructed to use both handrails. They were not given advice on how much weight to put through the rails and were advised to use the rails as they would do at home.

3.2.8 Checking trajectories

Following each trial it was possible to playback the activity on the computer screen to check marker visualisation. Trials where markers were obscured or missed at points during the activity were repeated until a satisfactory set of marker trajectories were obtained.

3.2.9 Smoothing

Position data of the markers from the Vicon system were smoothed and small gaps interpolated using a quintic spline routine (Woltring, 1986). A quintic spline was selected for smoothing as it has previously been reported to provide optimal smoothing of displacement data when compared to other methods (Giakis and Balzopoulos, 1997), and it enabled rapid computing within the Vicon software. Quintic splines have also demonstrated less end point error in acceleration data than Butterworth filters and Fourier series (Vint and Hinrichs, 1996). By observation of the smoothed data a MeanSquare Error (MSE) of $2mm^2$ was selected as it appeared to have best fit to the data whilst eliminating high frequency noise.

3.3 Kinematic and kinetic calculations

3.3.1 Vicon Bodybuilder software

Vicon Bodybuilder software uses BodyLanguage, a programming language designed specifically for biomechanical modelling. The software enables the operator to create their own biomechanical model but is scripted so that many of the basic calculations are performed in the background. A Bodybuilder model was created specifically for this thesis by the author and the programming code for this is presented in appendix 3.2.

The programme takes the following steps:

1. Develops a technical coordinate system for the marker clusters
2. Calculates the position of bony landmarks in the technical coordinate system using information from a calibration procedure
3. Develops the anatomical axis system
4. Computes the joint angles using the floating axis method
5. Computes the joint moments
6. Computes the whole body centre of mass

3.3.2 Technical coordinate systems

A technical coordinate system of right-handed orthogonal axes was defined for each segment using three of the four markers. This is illustrated for the thigh segment in figure 3.6.

The software hides the computations but follows the method described below.

Let the three markers be labelled 1,2 and 3 with position vectors in the global system of $\mathbf{r}_1$, $\mathbf{r}_2$ and $\mathbf{r}_3$. The three axes can then be defined by three unit vectors as follows:

$$\mathbf{z} = (z_1 i + z_2 j + z_3 k) = (\mathbf{r}_1 - \mathbf{r}_2)/|\mathbf{r}_1 - \mathbf{r}_2| \quad (3.1)$$

Estimated $\mathbf{y} = \mathbf{y}_{est} = (\mathbf{r}_1 - \mathbf{r}_3)/|\mathbf{r}_1 - \mathbf{r}_3|$ (3.2)

$$\mathbf{x} = (x_1 i + x_2 j + x_3 k) = (\mathbf{y}_{est} \times \mathbf{z})/|\mathbf{y}_{est} \times \mathbf{z}| \quad (3.3)$$

$$\mathbf{y} = (y_1 i + y_2 j + y_3 k) = \mathbf{z} \times \mathbf{x} \quad (3.4)$$

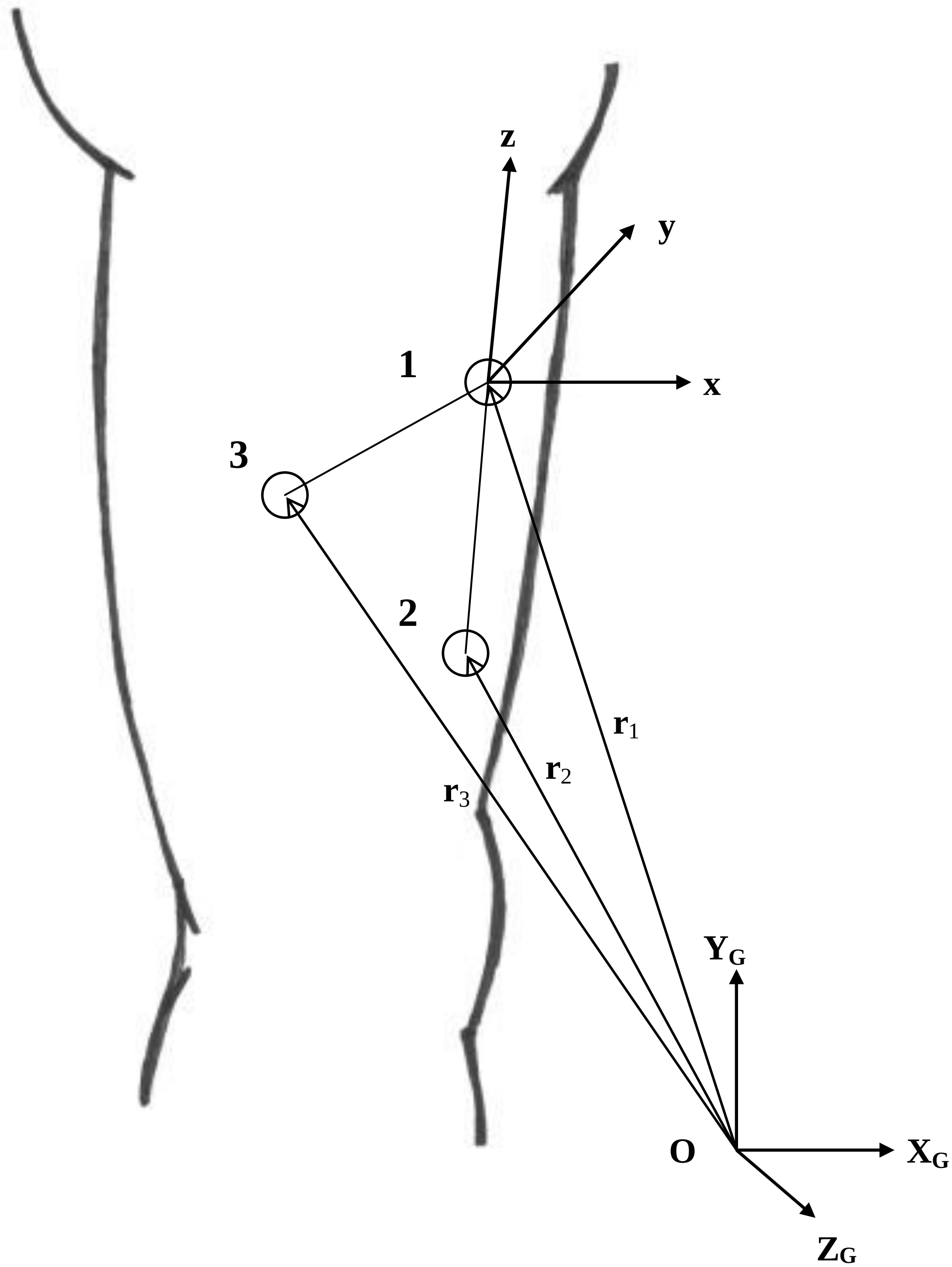


Figure 3.6 Definition of a technical axis system for the thigh segment

A rotation matrix (${}^{G}\mathbf{R}_{T}$) can be constructed as described in section 2.1.1.

$$ {}^{G}\mathbf{R}_{T} = \begin{bmatrix} x_1 & y_1 & z_1 \\ x_2 & y_2 & z_2 \\ x_3 & y_3 & z_3 \end{bmatrix} \qquad (3.5)$$

The origin of the technical system is taken as the position of marker 1, $\mathbf{r}_1$.

The position of any point in the global system (${}^{G}\mathbf{p}$) can be expressed in the technical system as:

$$ {}^{T}\mathbf{p} = {}^{G}\mathbf{R}_{T}^{-1}.({}^{G}\mathbf{p} - \mathbf{r}_1) \qquad (3.6) $$

And the reverse as:

$$ {}^{G}\mathbf{p} = ({}^{G}\mathbf{R}_{T}.{}^{T}\mathbf{p}) + \mathbf{r}_1 \qquad (3.7) $$

3.3.3 Anatomical point calibration

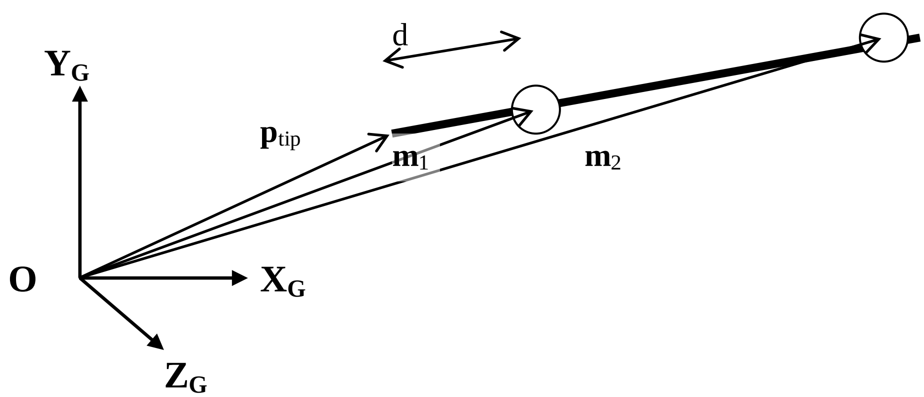


Figure 3.7 Pointer calibration technique

In order to determine the position and orientation of the bone embedded anatomical axis system it was necessary to calibrate the positions of anatomical landmarks in relation to the technical axis system. A pointer with two fixed markers of known distance was held against the anatomical landmark while data were captured for a few frames. The position vector of the pointer tip ($\mathbf{p}_{tip}$) can be determined in the global system from the position of the two markers $\mathbf{m}_1$and $\mathbf{m}_2$ as follows (see figure 3.7):

$$\mathbf{p}_{tip} = [\ (\mathbf{m}_1 - \mathbf{m}_2)/|\mathbf{m}_1 - \mathbf{m}_2|\ ] * d + \mathbf{m}_1 \qquad (3.8)$$

where d is the distance between the marker at $\mathbf{m}_1$ and the pointer tip.

The pointer tip location in the technical axis system and hence the anatomical landmark location can be determined using equation 3.6. During the dynamic trial the position of the anatomical landmarks will remain fixed in the technical system as they are based in the same rigid body. From knowing the position and orientation of the technical axis system throughout the dynamic trial equation 3.7 can be used to determine the position of the anatomical landmarks in the global system and then to define the anatomical axis systems.

3.3.4 Anatomical axis systems

The axis systems used for the lower limb were those suggested by Cappozzo (1996)

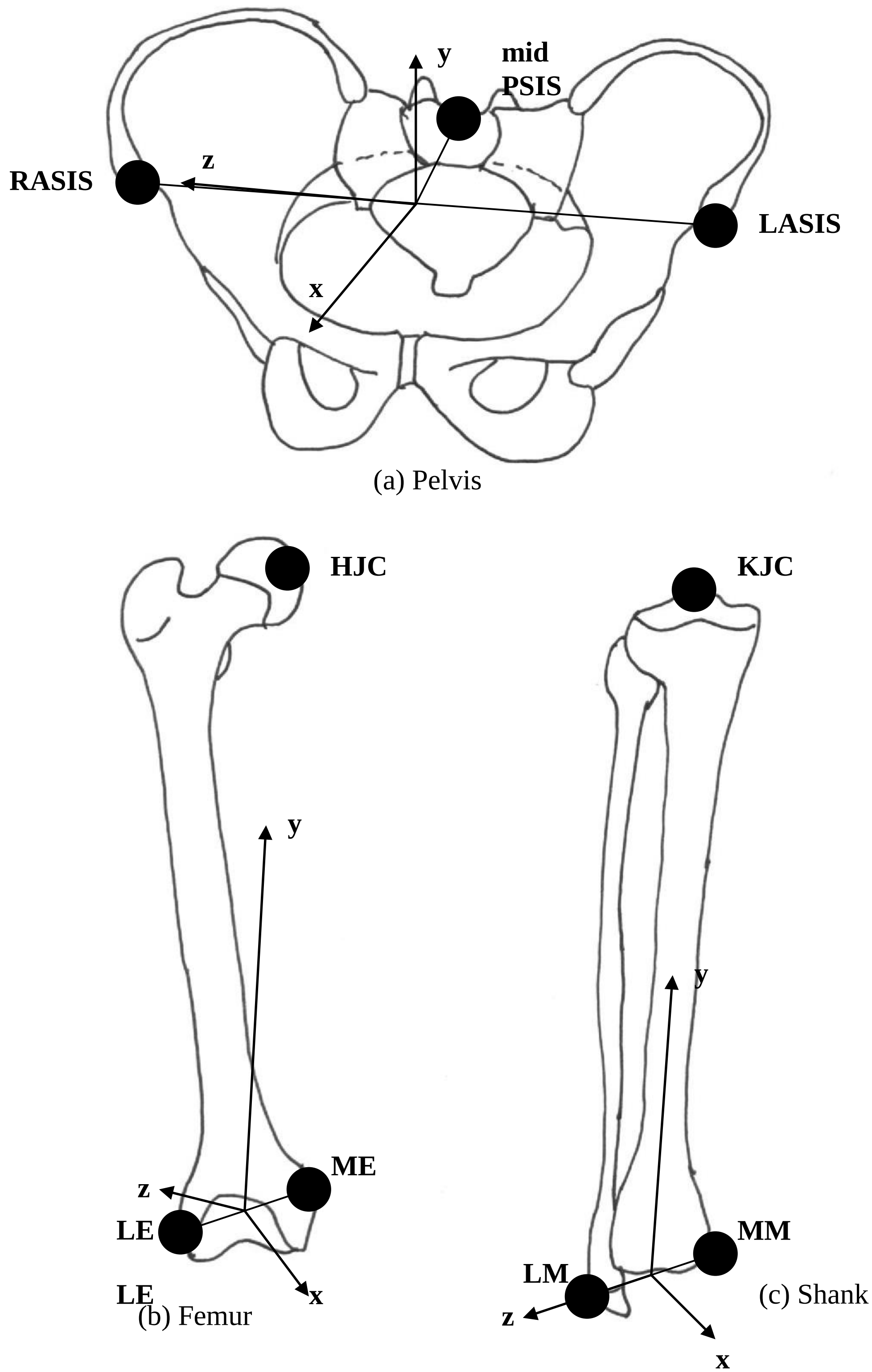


Figure 3.8 Axis systems for the pelvis and lower limb

3.3.4.1 Pelvis anatomical axis system

The points used to set up the pelvis anatomical co-ordinate system were the right and left anterior superior iliac spines (RASIS and LASIS) and the mid-point between the two posterior superior iliac spines (PSIS"s) as follows:

O_p: The origin is at the mid-point between the RASIS and LASIS

Zp: The z axis is defined as passing through the RASIS and LASIS from left to right, i.e. laterally of the right side and medially on the left side

Yp: The y axis runs perpendicular to the plane defined by the ASIS"s and the mid-PSIS"s, pointing superiorly

X_p: The x axis lies in the quasi transverse plane defined by the ASIS"s and the midpoint of the PSIS"s with its positive direction anteriorly

See figure 3.8(a).

3.3.4.2 Femoral anatomical axis system

The points used to set up the femoral axis system are the medial and lateral epicondyles (ME and LE) of the femur and the hip joint centre (HJC)

O_f: The origin is at the mid-point between the LE and ME

Y_f: The y axis is defined as passing through the origin and the HJC pointing superiorly.

Z_f: The z axis lies in the quasi frontal plane defined by the y axis, LE and ME. It is positive from left to right, i.e laterally in the right leg and medially for the left leg

X_f : The x axis runs perpendicular to the plane defined by the LE, ME and HJC with its positive direction anteriorly

See figure 3.8(b)

3.3.4.3 Shank anatomical axis system

The points required to set up the axis system of the shank are the lateral and medial malleoli (LM and MM) and the knee joint centre (KJC). This system puts constraints on the knee joint due to the KJC being defined by the femoral technical frame.

O_s: The origin is at the mid-point between the LM and MM

Y_s: The y axis is defined as running from the mid-point of the LM and MM to the KJC with its positive direction proximally

Z_s: The z axis lies in the quasi-frontal plane and can be defined from the malleoli. It is positive laterally for the right leg and medially for the left leg

X_s : The x axis is orthogonal of the yz plane with its positive direction anteriorly

See figure 3.8(c)

3.3.4.4 Foot anatomical axis system

During the activities the subjects wore shoes for their safety so the marker system used was to represent the shoe. Markers were placed on the heel and the medial and lateral shoe in the same plane parallel to the sole of the shoe.

O_{fo}: The origin is located at the midpoint of the two forefoot markers

Y_{fo}: The y axis is defined as passing through the origin and the heel marker with its positive direction proximal

X_{fo}: The x axis runs perpendicular to the plane defined by the three foot markers.

Z_{fo}: The z axis is perpendicular to Y_{fo} and X_{fo}

3.3.4.5 Trunk anatomical axis system

The points required to set up the axis system for the trunk are from the International Shoulder Group recommendations (van der Helm, 1997) and are the incisura jugularis (IJ – middle of the jugular notch), processus xiphoidius (PX, most caudal point of the xyphoid process), 7th cervical vertebrae (C7) and the 8th thoracic vertebrae (T8).

O_t: The origin is at the mid point between the IJ and C7

Y_t: The y axis is defined as passing through the mid-point of PX and T8 and the mid-point of IJ and C7 pointing superiorly and almost vertical in the initial position

Z_t : The x axis runs perpendicular to the plane defined by IJ, C7 and the midpoint of PX and T8 with its positive direction to the right

X_t: Perpendicular to Z_t and Y_t and pointing anteriorly

3.3.4.6 Humeral anatomical axis system

The points required to define the humeral axis system are the glenohumeral joint centre (GH) and the medial and lateral epicondyles of the humerus (MEp and LEp).

O_{hu}: The origin is the GH joint centre

Y_{hu}: The y axis is defined as passing through the mid-point of MEp and LEp (elbow joint centre) to the GH joint centre, positive direction proximal

X_{hu} : The x axis runs perpendicular to Y_{hu} and a vector (LEp-MEp) with its positive direction pointing forwards

Z_{hu}: The z axis is orthogonal to the yx plane with its direction pointing to the right.

See figure 3.9(a)

3.3.4.7 Forearm anatomical axis system

For this study the forearm was considered as one segment rather than the two-segment model of the radius and ulna proposed by Chadwick (1999). The method of Schmidt et al. (1999) and Cheng (1996) was used. The points required to define the axis system for the forearm are the elbow joint centre, the radial styloid (RS) and ulnar styloid (US).

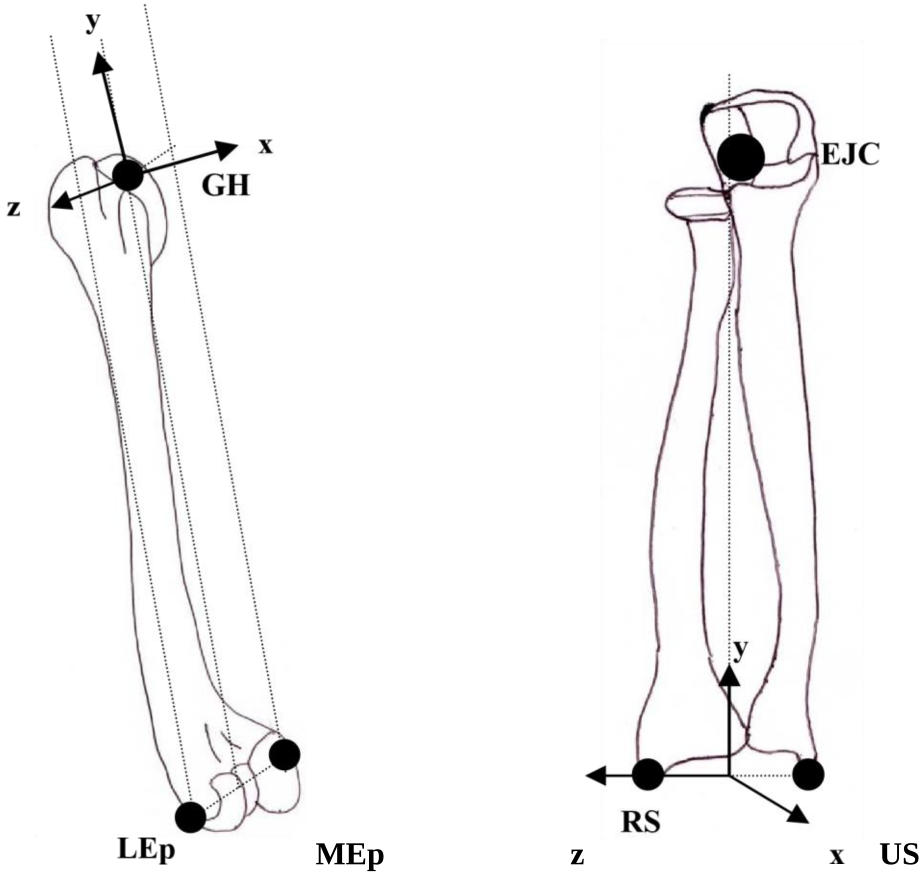
O_{fa}: The origin is the wrist joint centre (mid-point of radial and ulnar styloids)

Y_{fa}: The y axis is defined as passing through the wrist joint centre and the elbow joint centre, positive direction proximal

X_{fa} : The x axis runs perpendicular to Y_{fa} and a vector (RS-US) with it"s positive direction pointing forwards

Z_{fa}: The z axis is orthogonal to the yx plane with its positive direction pointing to the right

See figure 3.9(b)



(a) (b)

Figure 3.9 Axis systems for the upper limb

3.3.5 Calculating joint angles

A macro was used to calculate the joint angles using a floating axis method for each joint following the method described by Cole et al. (1993). The axis around which flexion/extension was calculated was the Z-axis of the proximal segment. Internal/external rotation was considered to be about the Y-axis of the distal segment and abd/adduction around the floating axis.

3.3.6 Calculating joint moments

Joint moments were calculated within the Bodybuilder software. The programming language is hidden to the user but Vicon provides an explanation of the method of calculation. This is detailed in appendix 3.3.

In summary:

1. The forces due to acceleration and moments of inertia are calculated from the position of the centre of mass of the segment (as given in the anthropometric table) in the current frame and the frame before and after.
2. The reactions resulting from the effect of gravity acting on the COM of the segment are then added.
3. If the foot is in contact with a forceplate then the reaction from this is also included.

The programmer must input the body segment parameters selected and instruct the software on how the segments of the body model fit together.

3.3.7 Body segment parameters

Dempster"s body segment parameters were used in this study (Winter, 1990). These parameters were determined from 8 elderly male cadavers.

Table 3.1 Dempster's body segment parameters

Segment	Segment mass/total mass	Centre of mass from distal joint/segment length	Radius of gyration/segment length
Hand	0.0060	0.494	0.297
Forearm	0.0160	0.570	0.303
Upper Arm	0.0280	0.564	0.322

Foot	0.0145*	0.500	0.475
Shank	0.0465	0.567	0.302
Thigh	0.1000	0.567	0.323

*as the shoes were kept on this parameter was changed to 0.0195 in the Bodybuilder program to account for the increased mass.

3.3.8 Whole body centre of mass calculation

The centre of mass (COM) of the whole body was calculated as the weighted sum of the COM of every segment of the body.

$$CM_i = \frac{\sum_j m_j . p_{i,j}}{\sum_j m_j}$$

Where m_j is the mass of segment j, and $p_{i,j}$ is the i^{th} component (i = x, y, z) of the position vector of its centre of mass.

The model had a total of 12 segments:

Pelvis, head and trunk, and upper arm, lower arm and hand, thigh, shank and foot bilaterally.

3.4 Processing and exporting data

Following data capture the individual markers can be visualised within the Vicon software and a moving image of the markers is reconstructed. The markers can then be labelled in order that Bodybuilder can perform the required calculations. At times, markers were obstructed from view by limbs or the environment. If a marker was part of a cluster of four markers then the position of the missing marker could be determined from knowledge of the geometry of the cluster. For other small gaps when the direction of movement was not changing it was possible to anticipate the position of the marker using a cubic spline interpolation. The number of frames that interpolation could be performed over depended on the movement but was not normally above 10 frames. As laboratory experience improved, an ideal set of camera positions reduced the number of missing markers.

Processed data from Bodybuilder was exported into Excel spreadsheets. Custom written software in Matlab was used to further analyse data and enable graphing and statistical analysis.

3.5 Summary

This chapter has detailed how the Vicon system was used to collect data and then perform kinematic and kinetic calculations on it. A Bodybuilder model was developed specifically for this project and aimed at the subject group of older adults. The model was designed with ISB standards in mind to facilitate meaningful sharing of the data.

Chapter 4

Results

4.1 Introduction

This Chapter presents the results for stair climbing and descent in adults aged over sixty participating in the EQUAL project. The biomechanical data is presented to demonstrate the demands of stair ascent and descent and to investigate age and gender related changes. Strength and ROM data collected by Samuel (2005) are used to determine whether any changes are related to differences in physical strength and flexibility.

4.2 Subject characteristics

In total 84 subjects were studied. It was not possible to recruit 30 subjects aged over 80 in the time available, so the final age breakdown of the subjects is shown in table 4.1

Age group	**60’s**		**70’s**		**80’s**	
Sex	Male	Female	Male	Female	Male	Female
Number	15	15	15	15	13	11

Table 4.1 Number of males and females in the three age cohorts

Subject characteristics are presented in table 4.2

Ideally the mean age in the 70‟s and 80‟s would have been around 75 and 85. The fact that the mean is towards the lower end of these groups represents the difficulty recruiting the oldest subjects.

Age group	**60’s**		**70’s**		**80’s**	
Sex	Male	Female	Male	Female	Male	Female
Mean Age (years)	65.7 ± 3.0	65.2 ± 2.9	73.6 ± 3.2	73.5 ± 2.8	81.9 ± 1.9	83.1 ± 2.8
Height (m)	1.73 ±.08	1.62 ±.08	1.73 ±.06	1.58 ±.06	1.72 ±.09	1.57 ±.06
Body Mass (kg)	77.1 ± 12.5	72.8 ±14.5	75.5 ±8.6	69.1 ±13.9	81.9 ±16.9	63.0 ±9.8
BMI (kg/m^2)	25.9 ± 3.2	27.5 ± 5.6	25.6 ± 2.5	27.8 ± 4.5	27.5 ±3.6	25.6 ± 3.9

Table 4.2 Subject characteristics

A two-way analysis of variance (ANOVA) was performed to examine the differences in height and body mass based on gender and age. A significant difference in height and body mass was observed between the men and women ($p < 0.05$), the men being taller and heavier. However, no significant difference was observed in height and body mass with increasing age. An ANOVA was performed on BMI and there was no significant difference in BMI with age or between genders.

4.3 Data processing

Markers trajectories were labelled and smoothed using Vicon Workstation and kinematic and kinetic data obtained using the Bodybuilder code detailed in Chapter 3. The stick figure generated by Bodybuilder was replayed to ensure that a correct representation had been made (i.e. no marker labelling errors) and to determine when significant events occurred. The events recorded were the foot strike and toe off for both feet for a full step cycle including the period when one foot was on the forceplate. From these events it was possible to determine the time spent on each step, the step rate, the proportion of stance and swing and the periods spent in double support (both feet in contact with a step).

Data were imported into Matlab and processed using custom written code detailed in appendix 5.1. In order to compare trials between subjects, each complete step cycle (from foot strike on the forceplate to foot strike on the next step) was normalised to 100 data points. For each subject a set of graphs were plotted for stair ascent and descent using the handrail and without the handrail (figure 4.1 + figure 4.2).

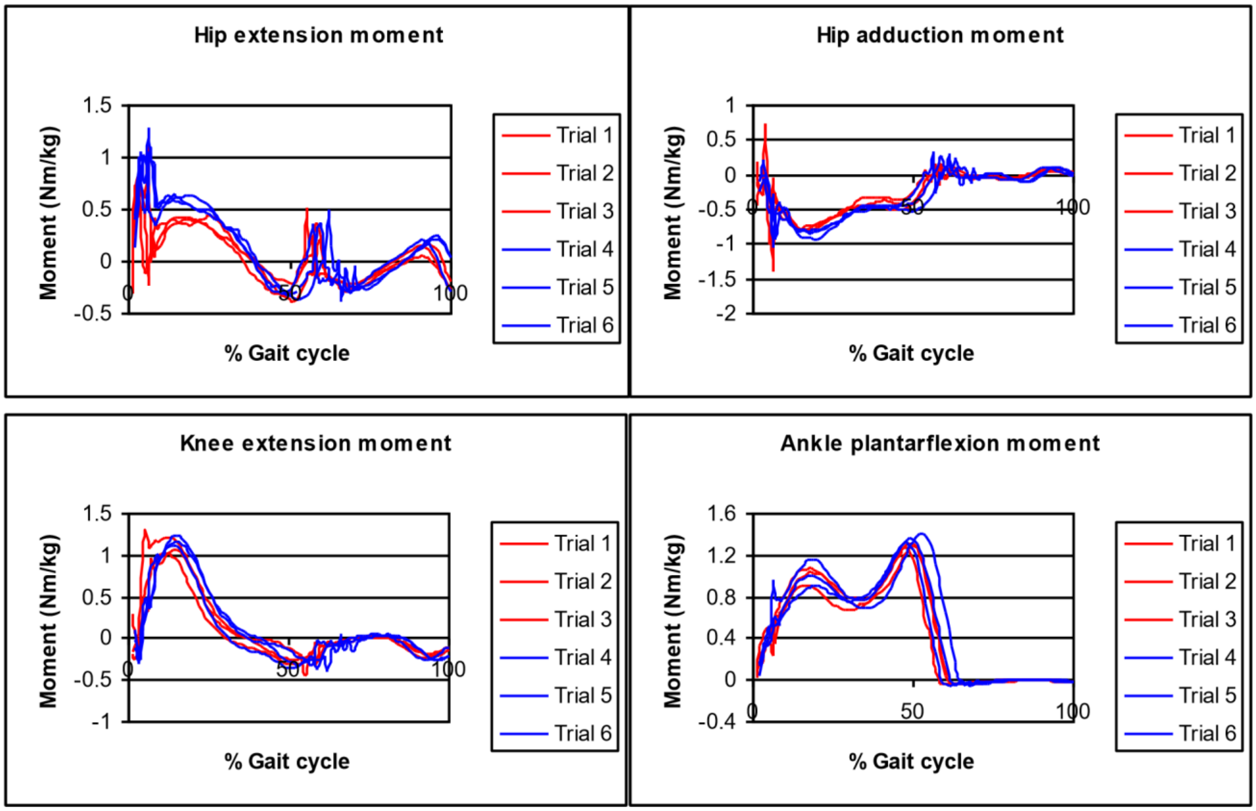
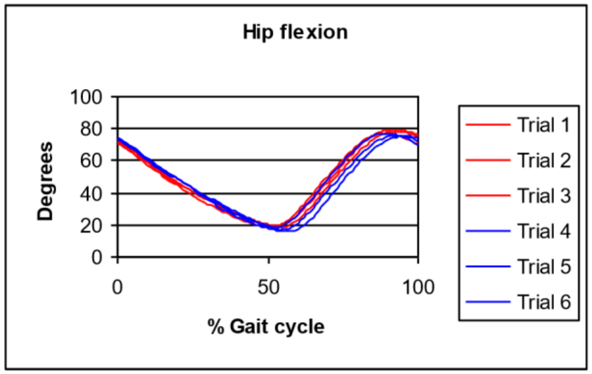
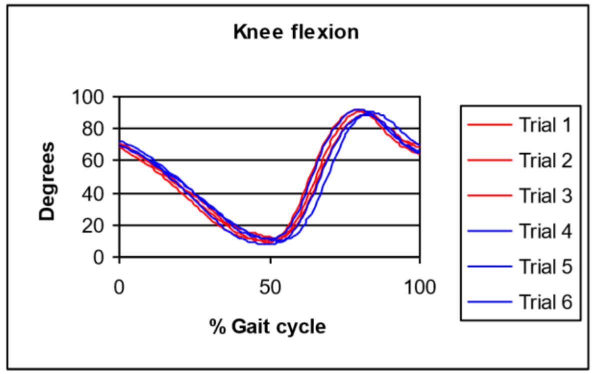
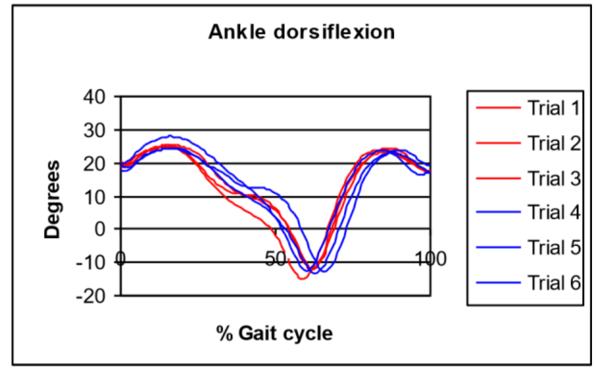
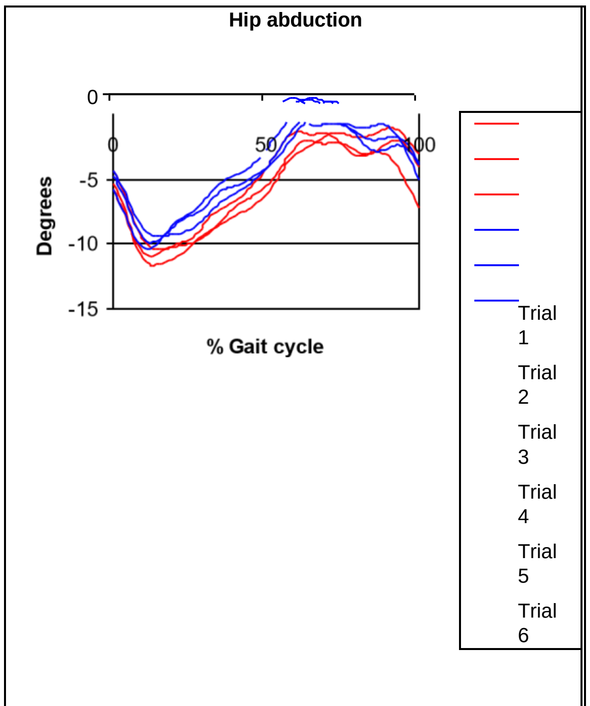


Figure 4.2 Kinetic results for one step cycle of stair ascent (foot strike to foot strike) for a male subject in the 60+ age group. Trials 1-3 (red) are using handrail Trials 4-6 (blue) are without handrail









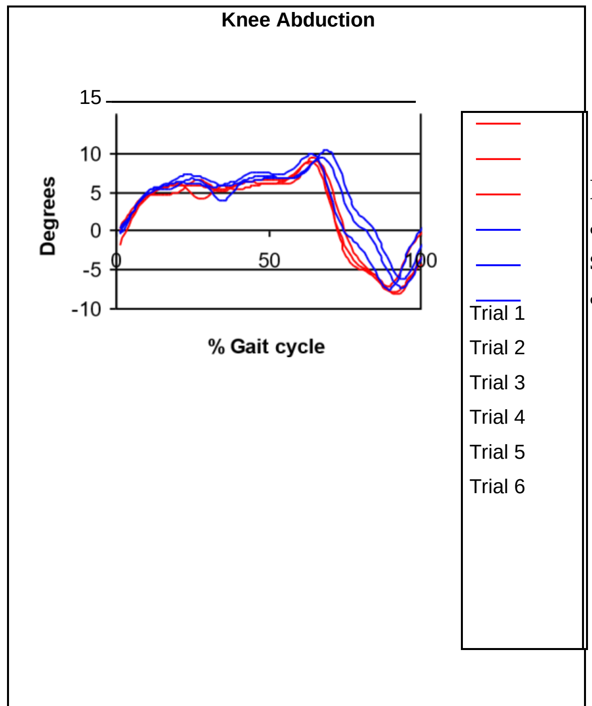
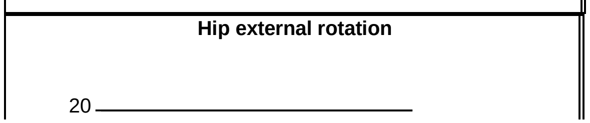
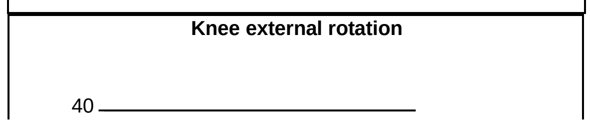


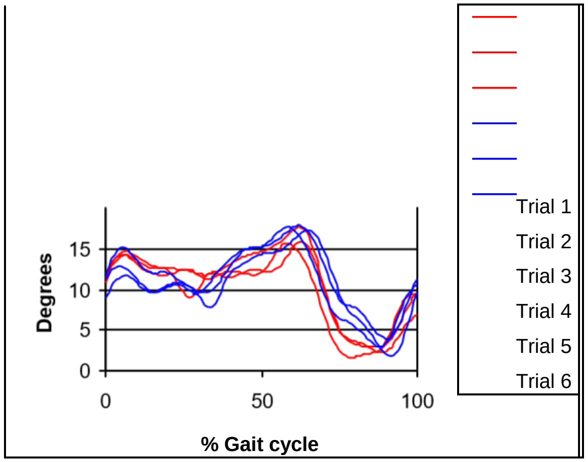
Figure 4.1

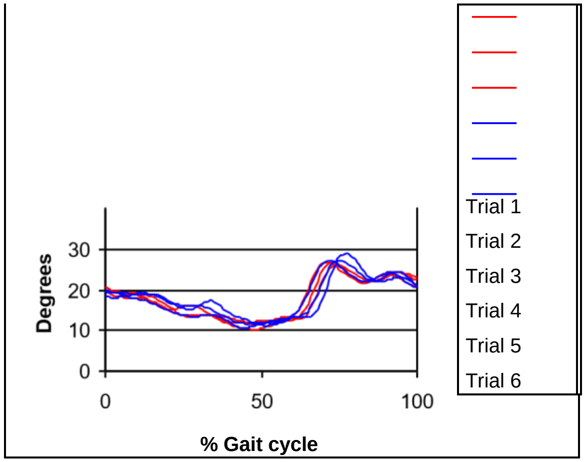
Kinematic results for one step cycle of stair ascent (foot strike to foot strike) for a male subject in the 60+ age group . Trials 1-3 (red) are using handrail

Trials 4-6 (blue) are without handrail)

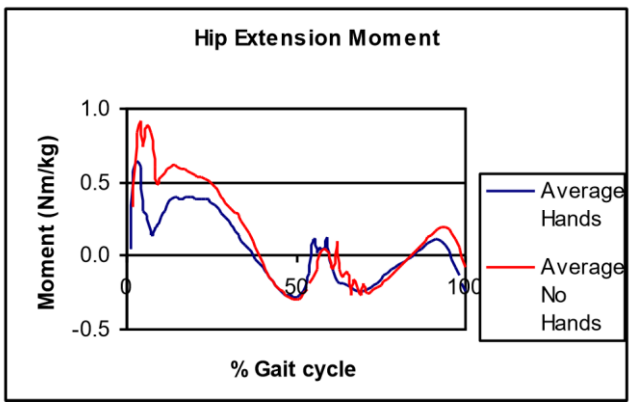


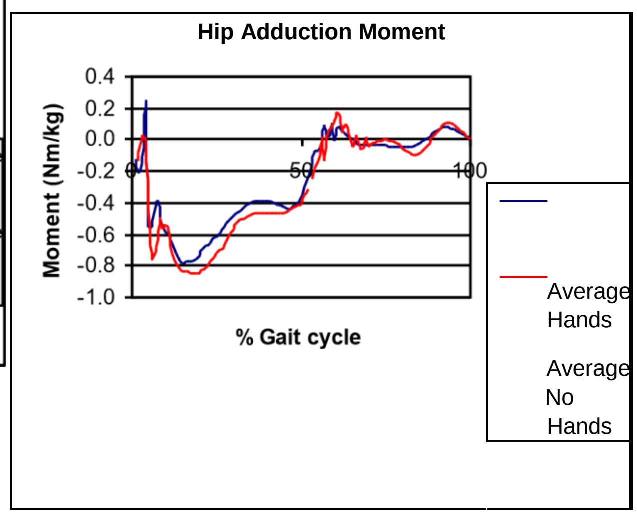



Trial 1
Trial 2
Trial 3
Trial 4
Trial 5
Trial 6
Degrees
15
10
5
0
0
50
100
% Gait cycle


Trial 1
Trial 2
Trial 3
Trial 4
Trial 5
Trial 6
Degrees
30
20
10
0
0
50
100
% Gait cycle

For this subject it can be seen that there is good repeatability over the six trials with the three attempts using the handrail producing a very similar set of curves. The curves without handrail use are also very repeatable for both kinematic and kinetic data. This was the case for the majority of trials for the 84 participating subjects. It was felt that due to this repeatability an average curve could be produced for each subject to represent the attempts with and without use of a handrail to facilitate comparison between subjects. On occasions it was clear from the graph that the subject has performed the stepping activity in a very different way, for example during a near trip or hesitant episode. These individual trials were removed prior to producing the average plots. A set of average plots for the subject above is provided in figures 4.3 and 4.4





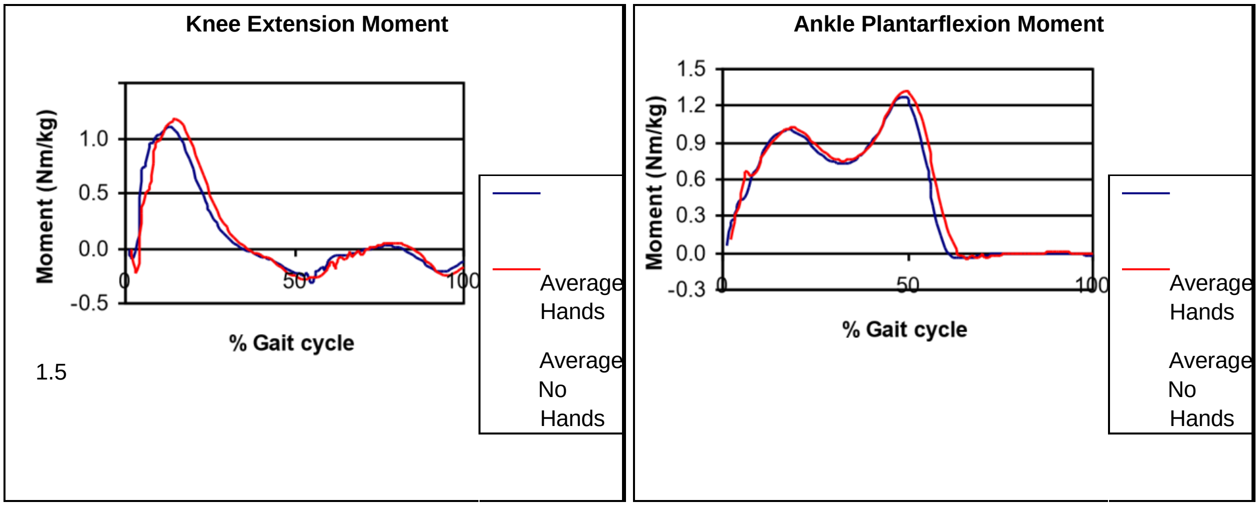
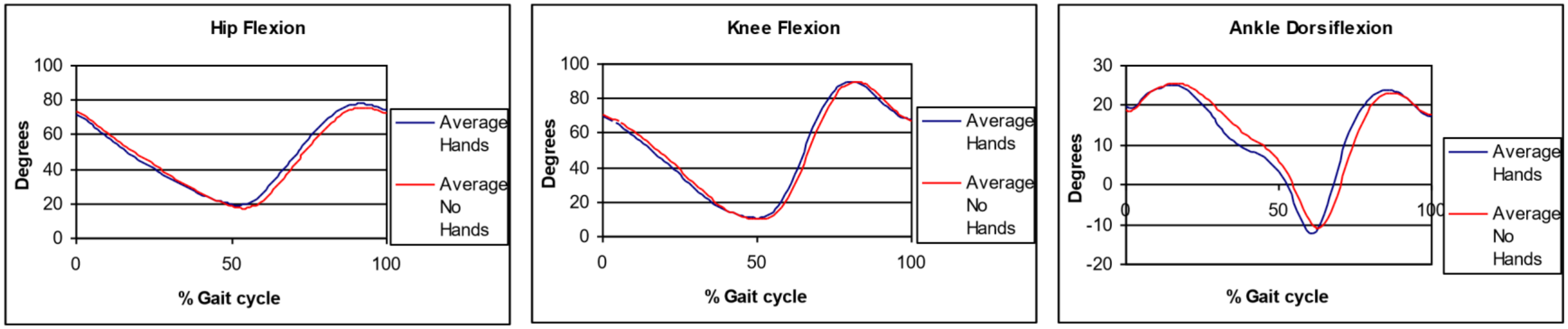
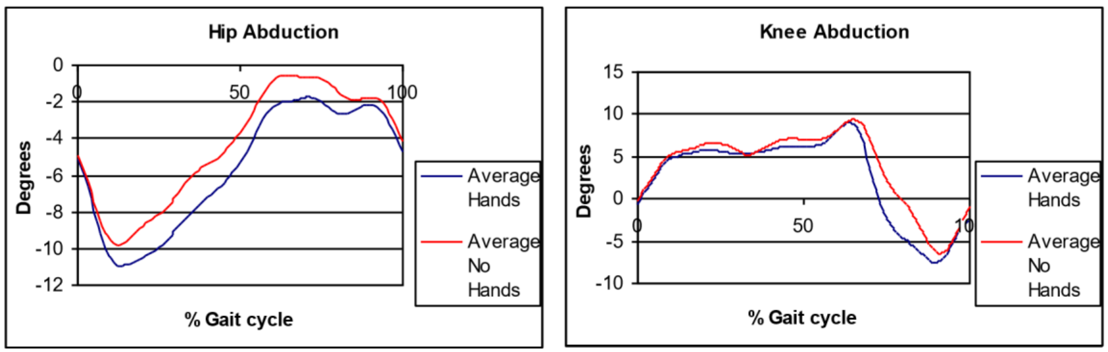


Figure 4.4 Average kinetic results for three step cycle of stair ascent (foot strike to foot strike) for a male subject in the 60+ age group.





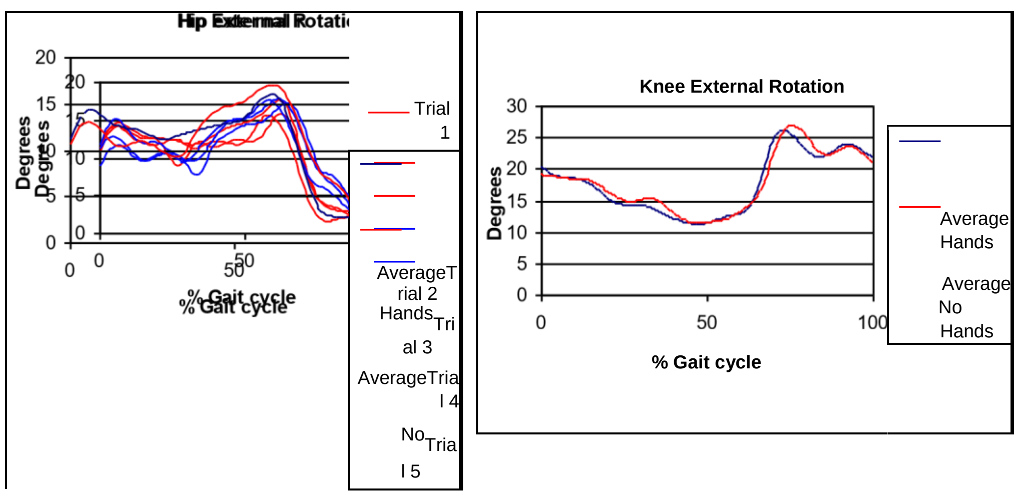


Figure 4.3

Average kinematic results for three step cycles of stair ascent (foot strike to foot strike) for a male subject in the 60+ age group


Hands
Trial 6

The Matlab programme also exported data for further statistical analysis. The normalised step cycle was used and all available trials for each subject (i.e. normally 3 attempts with hand and three without hands). Data exported were:

Maximum and minimum joint moments for each of the reported joints

Maximum and minimum joint angles for each of the reported joints

Maximum and minimum COM velocity

The percentage of the step cycle at which these maximum and minimum events occurred

The statistical programme SPSS was used to study the data and investigate any age related changes.

4.4. Ability to perform task

Of the 84 subjects assessed, all were able to ascend and descend the stairs using a handrail using the normal stair gait pattern of one foot on one step then one foot on the next step. However, without use of the handrail, 5 subjects were unable to ascend the stairs and 7 were unable to descend the stairs. A further subject only felt able to ascend and descend the stairs once without the handrail, as she felt too anxious to repeat the task. The age distribution of those unable to complete the task is shown in table 4.3.

Broken down it can be seen that 20% of the females in their 70"s were unable to ascend or descend stairs without the handrail, 18% of the females in their 80"s could not ascend and 36% could not descend without the handrail and just 1 (or 8%) of the men in their 80"s could not ascend or descend the steps (figure 4.5). As it was found that the problems with stair climbing were more prevalent in the female subjects it was decided that results would compare any biomechanical differences between the genders in addition to the three age groups.

Age group	60's		70's		80's	
Sex	Male	Female	Male	Female	Male	Female
Number unable to ascend stairs without handrail				3	1	2*
Number unable to descend stairs without handrail				3	1	4*

Table 4.3 Number of subjects unable to perform task (* subject only managed 1)

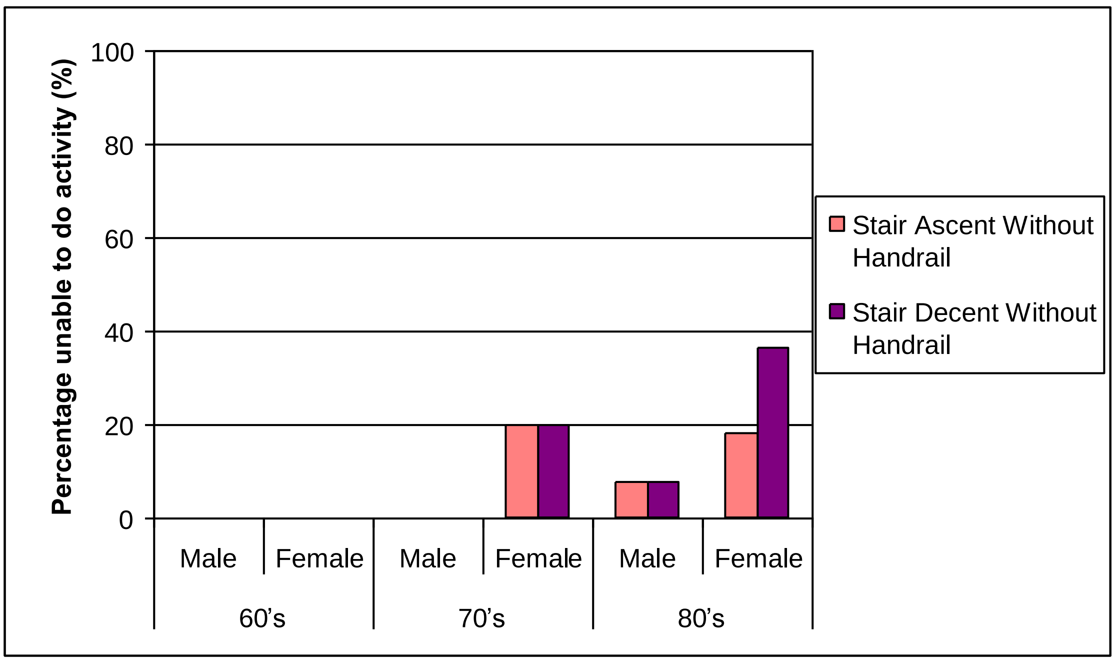


Figure 4.5 Percentage of subjects in each age group unable to use stairs without a handrail

<u>4.5 Stair ascent</u>

4.5.1 Temporal Data

The following parameters were studied:

- Time taken for one cycle (from right foot strike to next right foot strike)

- Rate of stair ascent (steps/min)
- Stance phase (%)
- Swing Phase (%)
- Total double support time (%) (i.e time with both feet on the staircase)

The average of the three trials was used for each subject. Data were assessed for normality using SPSS. All parameters were normally distributed when studied as a group or divided into age and gender categories. The variance of the data was similar between groups and therefore it was concluded that parametric statistical tests could be performed.

The effects of gender were determined using an independent samples t-test. There were no significant differences in any of the parameters between males and females. Therefore it was considered appropriate to consider the age groups as a whole (male and female) for further statistical analysis.

4.5.2 Temporal data using handrail

The temporal data for subjects ascending the stairs using a handrail are presented in table 4.4 and figure 4.6. A one-way ANOVA was performed to determine age effects, and significant age related differences were found for all parameters. On post hoc testing these differences were found to be between the 60‟s and 80‟s and 70‟s and 80‟s. No difference was found between subjects in their 60‟s and 70‟s.

Age Group	60's	70's	80's
Time per step cycle (s)	1.40 (± 0.23)	1.46 (±0.19)	1.69 (±0.27)*
Rate of stair ascent (steps/min)	87.6 (±13.0)	83.6 (±10.4)	72.8 (±11.6)*
Stance phase (% of cycle)	59.6 (±2.46)	60.1 (±2.82)	62.0 (±2.62)*
Swing phase (% of cycle)	40.4 (±2.46)	39.9 (±2.82)	38.0 (±2.62)*
Double support (% of cycle)	24.1 (±3.38)	25.3 (±3.37)	28.0 (±3.72)*

Table 4.4 Temporal data for stair ascent using a handrail. * indicates significant difference at p<0.05 level

Subjects in their 80‟s were significantly slower than the subjects in their 60‟s and 70‟s, spending slightly longer in each stance phase. A longer proportion of the step cycle was spent in double support in the 80‟s age group than the other groups.

4.5.3 Temporal data without a handrail

The temporal data for subjects ascending the stairs without a handrail are presented in table 4.5 and figure 4.6.

Age Group	60's	70's	80's
Time per step cycle (s)	1.35 (±0.2)	1.39 (±0.18)	1.59 (±0.22)*
Rate of stair ascent (steps/min)	90.6 (±13.0)	88.0 (±12.0)	76.8 (±10.4)*
Stance phase (% of cycle)	60.2 (±2.7)	60.1 (±2.8)	63.2 (±2.6)*
Swing phase (% of cycle)	39.8 (±2.7)	39.9 (±2.8)	37.8 (±2.6)*
Double support (% of cycle)	24.3 (±3.4)	25.4 (±4.1)	28.4 (±2.5)*

Table 4.5 Temporal data for stair ascent without a handrail. * indicates significant difference at p<0.05 level

As with the use of the handrail, subjects in their 80‟s were significantly slower and spent a greater proportion of time in stance and double support.

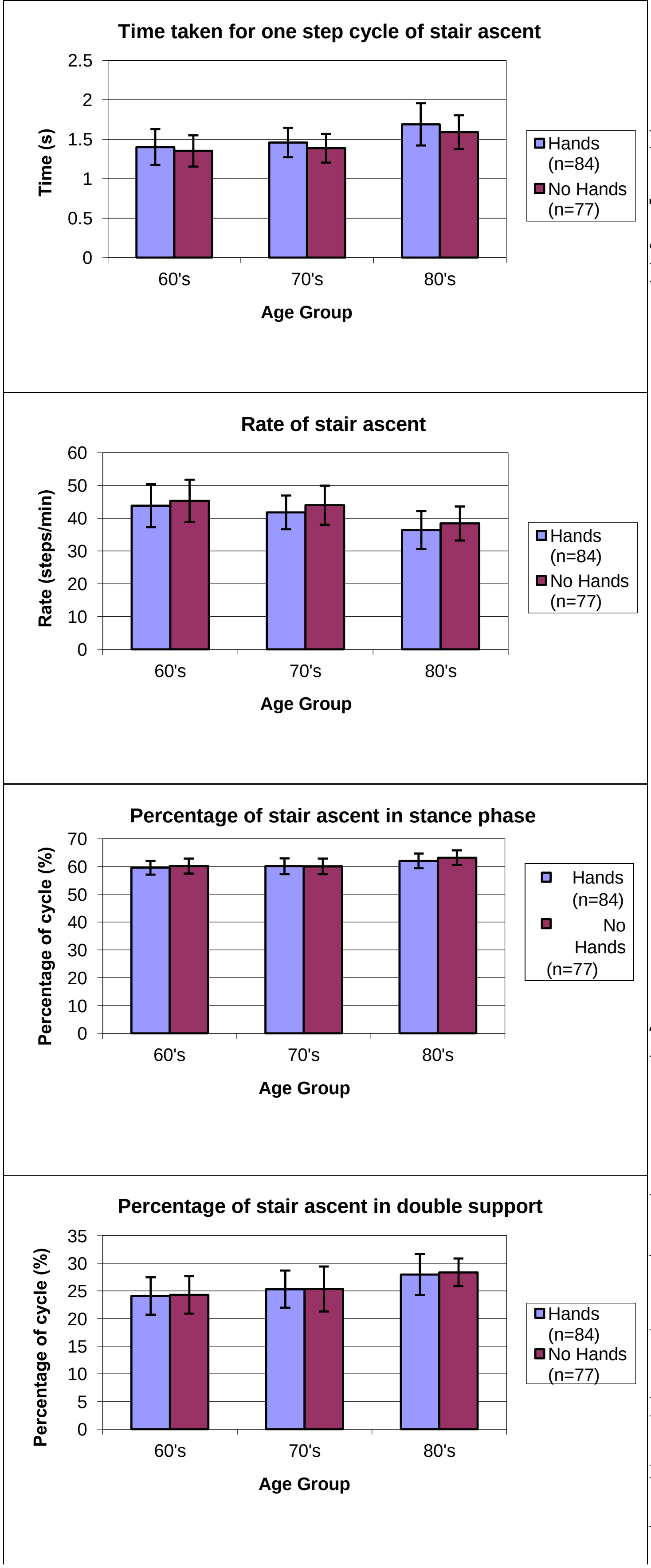


Figure 4.6

Temporal data for stair ascent with and without a handrail (mean ± 1 s.d.)

4.5.4 Effect of hand rail on temporal data during stair ascent

A 3-way ANOVA was used to determine the interaction between, age, sex and handrail use for each of the temporal parameters studied. There were no significant interactions between age and handrail use and sex and handrail use. Therefore the effects of handrail use are not influenced by age or gender.

An independent t-test was used to investigate the change in temporal parameters related to using a handrail. The time taken for each step was significantly less when subjects did not use a handrail ($p<0.05$) and therefore the rate of stair ascent was faster. There was no change in the proportion of time spent in swing or stance phase when using a handrail to ascend stairs.

4.5.5 Kinematics of stair ascent using a handrail

4.5.5.1 Sagittal plane kinematics

Sagittal plane kinematics are shown in figure 4.7 for the subjects in the three age groups. As there were nearly even numbers of males and females in the groups the genders have been combined for the purpose of age group comparison as this created greater statistical power. Some gender related differences in kinematics were found and these will be discussed later. All data were tested for normality prior to statistical analysis. All peak kinematic data were normally distributed apart from maximum hip abduction.

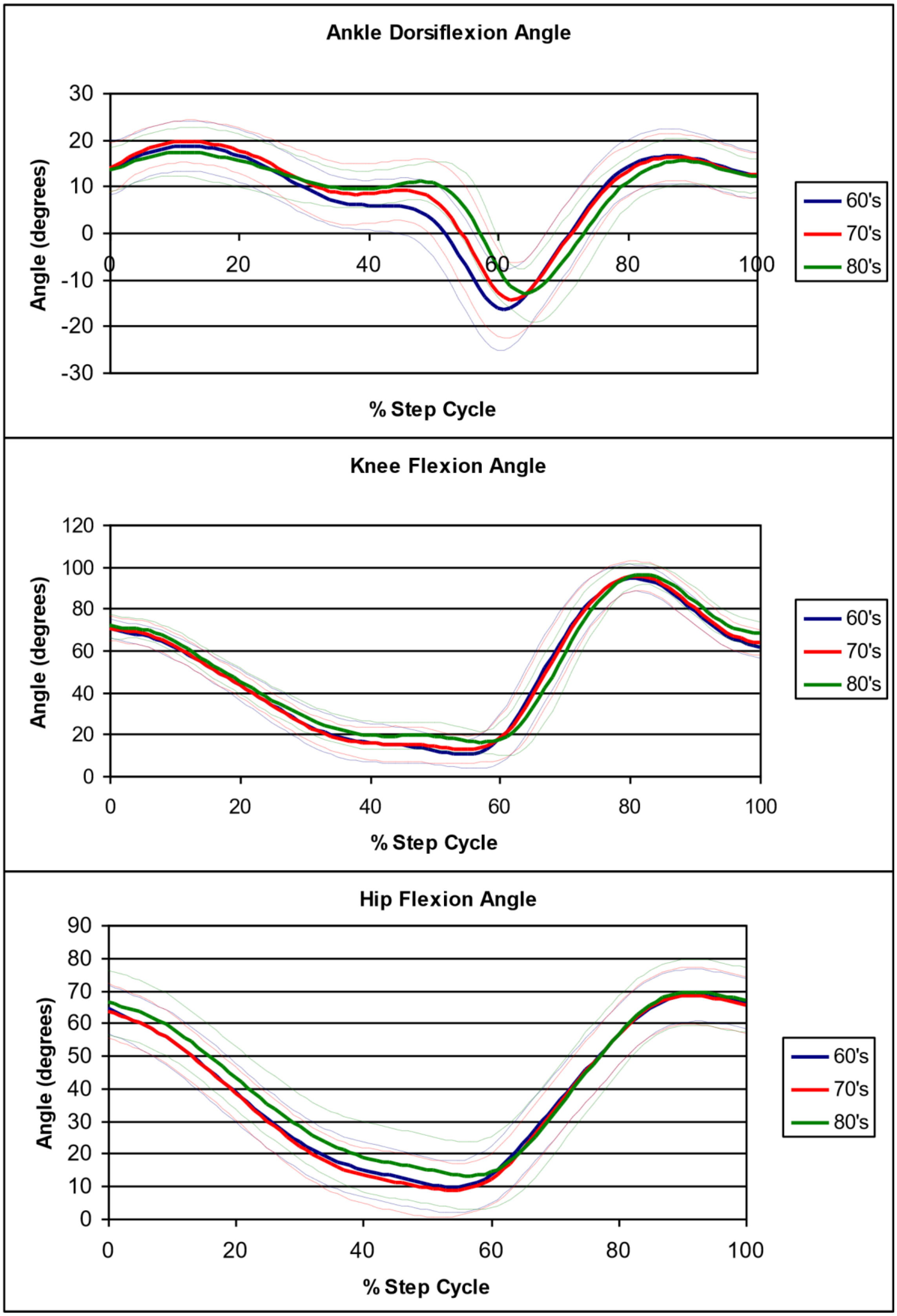


Figure 4.7 Sagittal plane kinematics for stair ascent using hands. Foot strikes are at 0 and 100% of the step cycle.

Ankle joint

At the ankle joint subjects contacted the step with the ankle dorsiflexed an average of approximately 15 degrees. As weight was taken over that foot, ankle dorsiflexion increased slightly to a maximum of around 20 degrees. As weight transferred to the opposite foot the ankle became less dorsiflexed and then progressed into plantarflexion of around 18 degrees prior to toe off. During swing phase the ankle returned to a dorsiflexed position prior to foot strike. Peak angles for ankle dorsflexion and plantarflexion are shown in table 4.6.

		Age Group		
Angle	**Sex**	**60's**	**70's**	**80's**
Maximum ankle dorsiflexion (degrees)	Male	21.0 (±4.6)	22.4 (±3.5)	19.9 (±4.8)
	Female	20.1 (±5.3)	19.1 (±4.4)	19.2 (±4.2)
	Combined	20.6 (±4.9)	20.7 (±4.3)	19.6 (±4.5)
Maximum ankle plantarflexion (degrees)	Male	17.9 (±6.6)	17.5 (±6.0)	15.1 (±7.0)
	Female	20.9 (±9.1)	19.3 (±7.6)	18.8 (±4.2)
	Combined	19.4 (±7.9)	18.4 (±6.8)	16.8 (±6.1)

Table 4.6 Maximum ankle plantar and dorsiflexion angles during stair ascent with hands

Using a 2-way ANOVA it was found that there were no statistically significant differences in peak ankle dorsiflexion or plantarflexion between the age groups or the sexes during stair ascent. From the plots in figure 4.7 it can be seen that there is a delay in the onset of plantarflexion with increasing age, which would reflect the increase in stance phase observed in the oldest group.

Knee joint

At foot strike on the step the knee was flexed by approximately 70 degrees. During stance phase the knee gradually extended to a point where it was approximately 10 degrees from full extension. At toe off the knee then rapidly flexed to its maximum of approximately 95 degrees before extending slightly prior to the foot contacting the next step. Maximum and minimum knee flexion angles are presented in table 4.7

		Age Group

Angle	Sex	60's	70's	80's
Maximum knee flexion (degrees)	Male	95.4 (±6.8)	95.8 (±7.4)	97.5 (±5.5)
	Female	96.9 (±6.7)	98.9 (±5.2)	98.6 (±6.4)
	Combined	96.1 (±6.7)	97.4 (±6.5)	98.0 (±5.8)
Minimum knee flexion (degrees)	Male	10.7 (±6.1)	8.6 (±6.2)	14.2 (±7.0)
	Female	5.1 (±6.3)	9.0 (±5.2)	9.9 (±3.3)
	Combined	7.9 (±6.1)	8.8 (±5.6)	12.3 (±5.9)

Table 4.7 Maximum and minimum knee angles during stair ascent with hands

A 2-way ANOVA found there to be no age or gender effects for maximum knee flexion. For knee extension women were found to have a significantly more extended knee during stance phase than there male counterparts ($p<0.05$). This may be explained by the women being shorter than the men and therefore needing to use their leg length more to progress to the next step. As age increased there was a trend towards a more flexed knee position during stance phase. The difference between the 60 and 80 year olds was significant ($p<0.05$) indicating that the older subjects are not extending their knees as much throughout stance phase, specifically during in later stance.

Hip joint

At initial contact the hip is in a flexed position of approximately 65 degrees. During stance phase the hip extends to approximately 10-15 degrees from full extension and then flexes to its maximum of just under 70 degrees just prior to the next foot contact. Maximum and minimum angles are presented in Table 4.8

		Age Group		
Angle	**Sex**	**60's**	**70's**	**80's**
Maximum hip flexion (degrees)	Male	70.3 (±7.4)	68.8 (±7.2)	72.9 (±10.9)
	Female	67.5 (±8.7)	69.2 (±10.2)	66.7 (±7.1)
	Combined	68.9 (±8.1)	68.8 (±8.7)	70.1 (±9.7)
Minimum hip flexion	Male	12.7 (±6.7)	8.24 (±7.13)	15.78 (±11.05)
	Female	4.7 (±7.4)	6.4 (±9.5)	6.9 (±7.2)

(degrees)	Combined	8.7 (±8.1)	7.4 (±8.3)	11.7 (±10.3)

Table 4.8 Maximum and minimum hip angles during stair ascent with hands

A 2-way ANOVA found there to be no significant difference between age groups or gender for the maximum amount of hip flexion used during stair ascent. Women were found to use a significantly more extended hip position than men at terminal stance ($p<0.05$). This may be explained again by height differences between the groups. There were no significant age related changes in minimum hip flexion, but there was a trend for the oldest age group to have a more flexed hip position throughout stance phase (figure 4.7).

4.5.5.2 Coronal and transverse plane kinematics

Coronal and transverse plane kinematics are presented in figures 4.8 and 4.9. The way the foot was modelled did not allow for calculation of ankle angles in these planes. Coronal and transverse plane kinematics are known to be sensitive to errors in the location of the flexion/extension axis of the joint and should be interpreted with degree of caution (Della Croce, 2005).

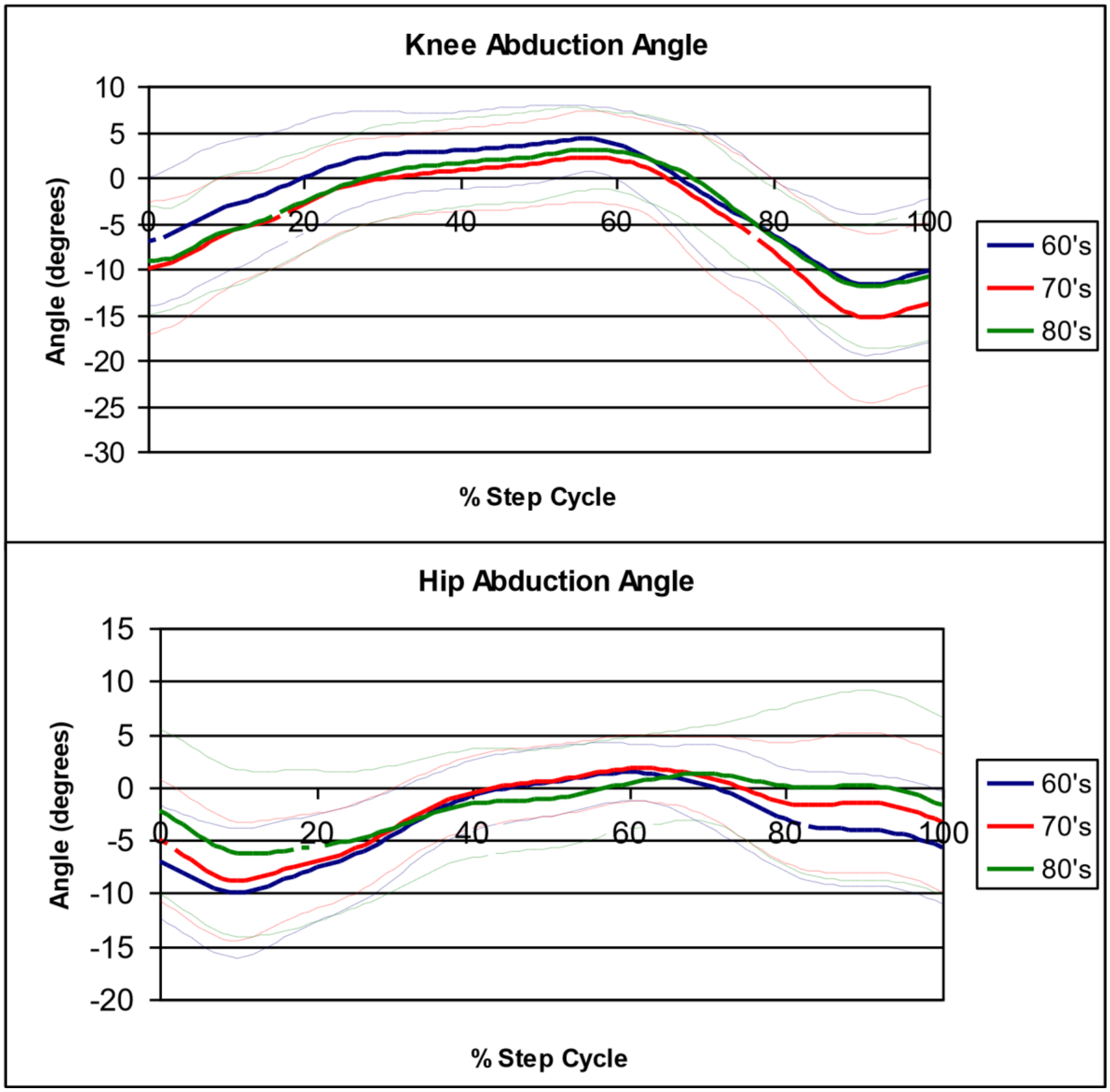


Figure 4.8 Coronal plane kinematics for stair ascent using hands. Foot strikes are at 0 and 100% of the step cycle.

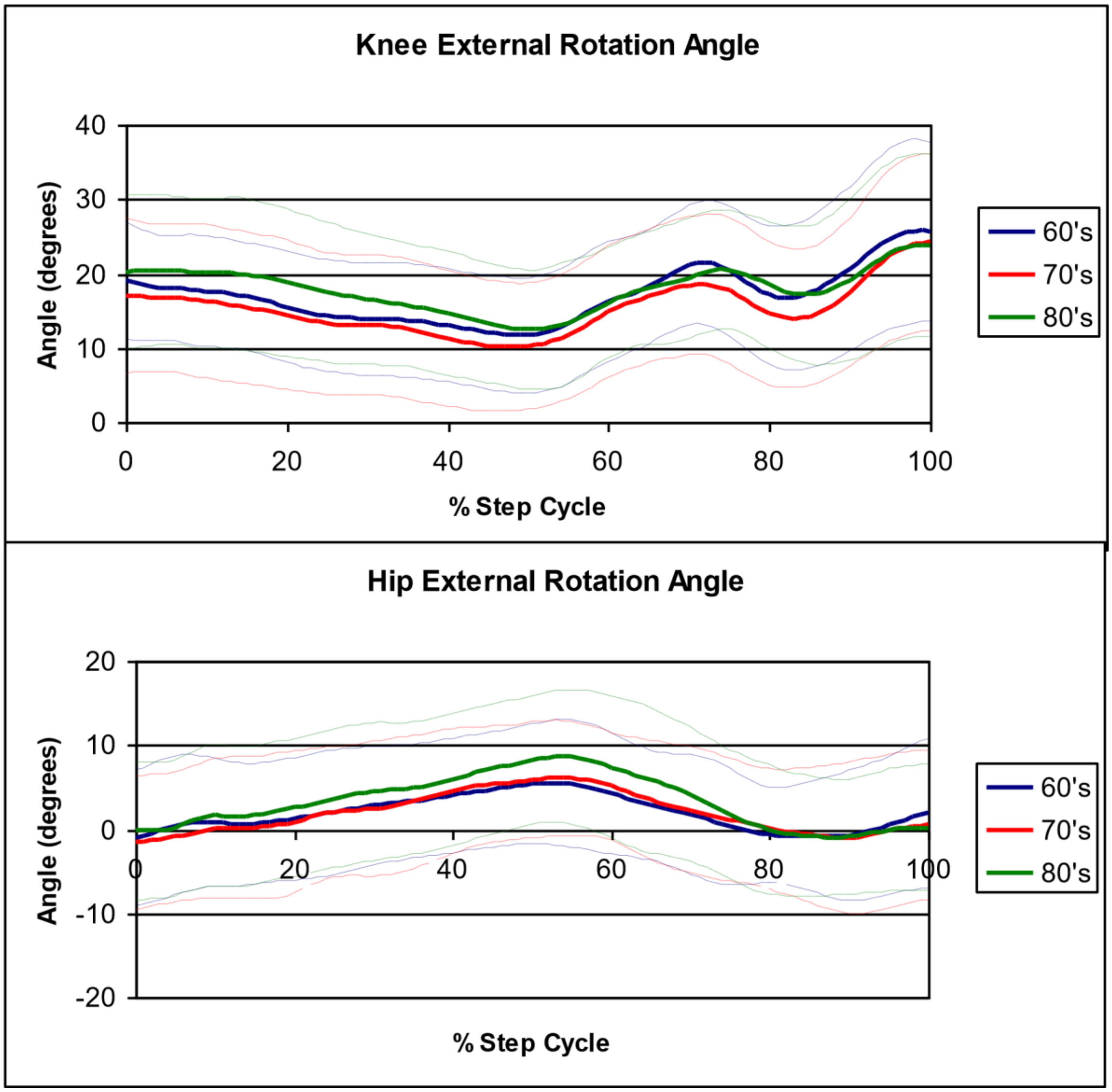


Figure 4.9 Transverse plane kinematics for stair ascent using hands. Foot strikes are at 0 and 100% of the step cycle.

<u>Knee joint</u>

The knee angle initially appeared to be slightly adducted at foot contact progressing to a few degrees of abduction during stance. During the swing phase the knee again appears more adducted. The degree of adduction corresponds to the amount of knee flexion at any point, the more flexion the greater the knee adduction. This indicates that there may be some error with the alignment of the knee flexion/extension axis. Regarding rotation, the knee appeared to be externally rotated throughout the whole movement with around 15 degrees of movement seen between stance and swing. This external rotation reflects normal tibial torsion and the degree of movement again may be affected by knee axis alignment.

A 2-way ANOVA found there to be no age effects in maximum knee add/abduction or internal/external rotation during stair ascent.

Hip joint

At foot strike the hip is adducted approximately 5 degrees. During stance the hip becomes abducted by a few degrees and then adducts again prior to initial contact. Overall there is less than 15 degrees of hip excursion in the coronal plane. The hip remains in a neutral or slightly externally rotated position during the whole step cycle. There is less than 10 degrees of movement into external rotation as the opposite foot is progressed onto the next step.

Statistical testing found there to be no age related changes in maximum hip excursion during stair ascent.

4.5.6 Effect of handrail on kinematics during stair ascent

When the kinematic plots were produced for the 3 age groups without use of the handrail they were found to look almost identical to those using a handrail shown in figures 4.7-4.9. To assess statistically, a paired t-test was used on the parametric data. Without the use of a handrail there was found to be a significant ($p<0.05$) increase in the maximum amount of hip and knee flexion used and a decrease in the amount of ankle plantarflexion (figure 4.10).

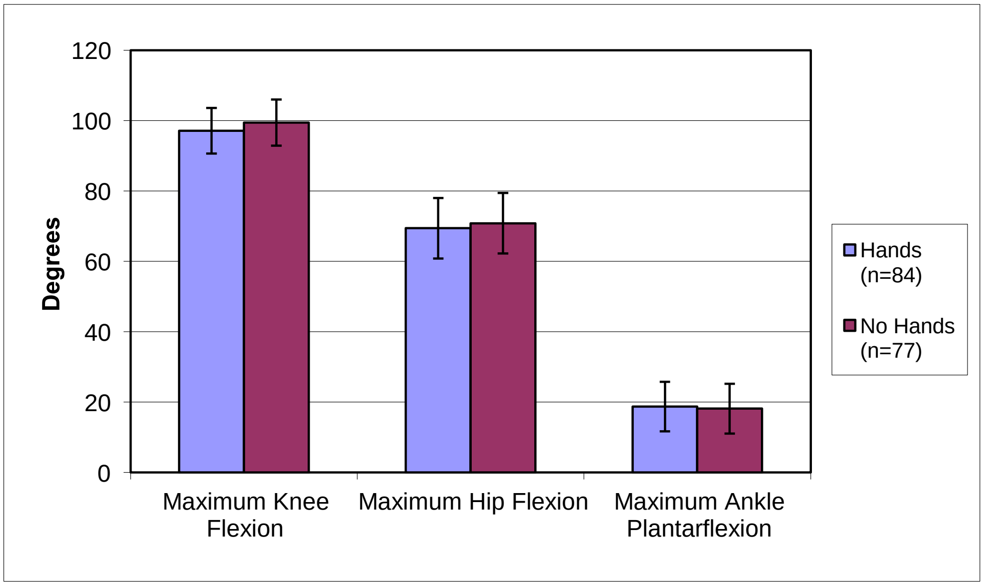


Figure 4.10 Effect of handrail use on joint kinematics

These differences, although significant statistically, were small. For hip flexion there was a 1.5 degree increase, for knee flexion a 2.3 degree increase and for ankle plantarflexion a 1 degree decrease.

4.5.7 Kinetics of stair ascent using a handrail

The plots of the internal moments (i.e. produced by the subject to balance the external moments) are presented in figures 4.11 and 4.14. This thesis will concentrate on the moments balanced predominantly by muscular control, hip extension and abduction, knee extension, and ankle plantarflexion.

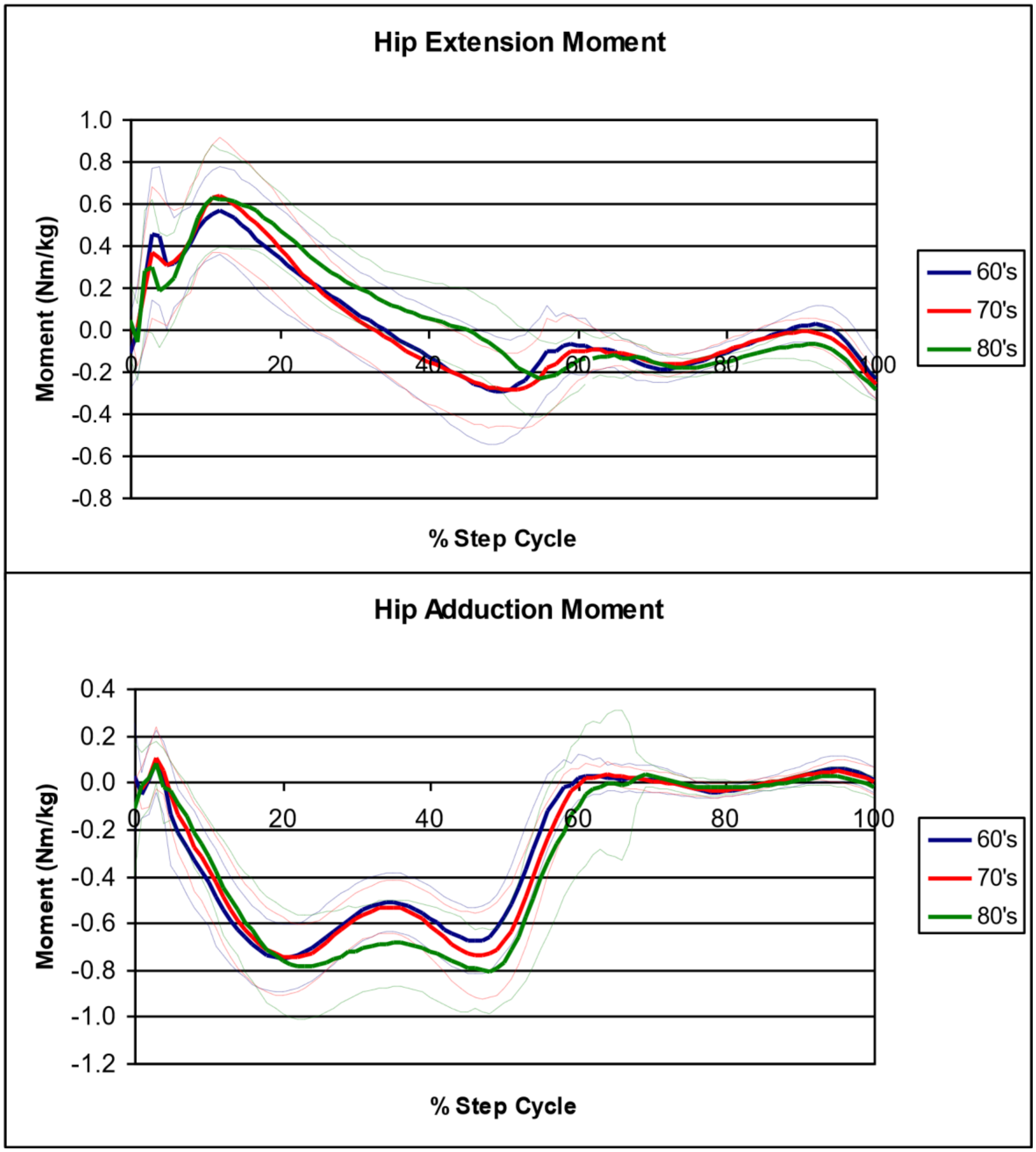


Figure 4.11 Moments produced around the hip joint during stair ascent using hands. Foot strikes are at 0 and 100% of the step cycle.

<u>Hip moments</u>

At initial contact there is an external moment tending to flex and abduct the hip. Throughout stance a hip abduction moment is produced which has a pattern of two peaks. The first peak is during the phase of single support on that limb then the trough occurs as weight is transferred onto the next step and then a further peak is generated prior to toe off. These peaks are similar in magnitude. During swing phase there is very little hip abd/adductor moment. The hip extensor moment is greatest during the first single support phase then progresses to a hip flexor moment during weight transference to the next foot. The hip flexor moment peaks just at the initiation of swing phase and a small flexor moment continues until the next foot strike. Subjects in their 80‟s tended to have a greater hip extension moment in later stance than those in their 60‟s and 70‟s.

Peak hip moments for each individual trial were entered into SPSS and analysed further. Hip extension moments were normally distributed and statistically analysed using 2-way ANOVA. Hip abduction moments were not normally distributed and therefore the non-paramaetric Kruskal Wallis test was used. Mean results for each age group and gender are presented in figure 4.12 and table 4.9. There was a trend for the maximum hip extension moment to decrease with age, however this was not statistically significant. There was no statistical difference between the groups for peak hip abduction moment.

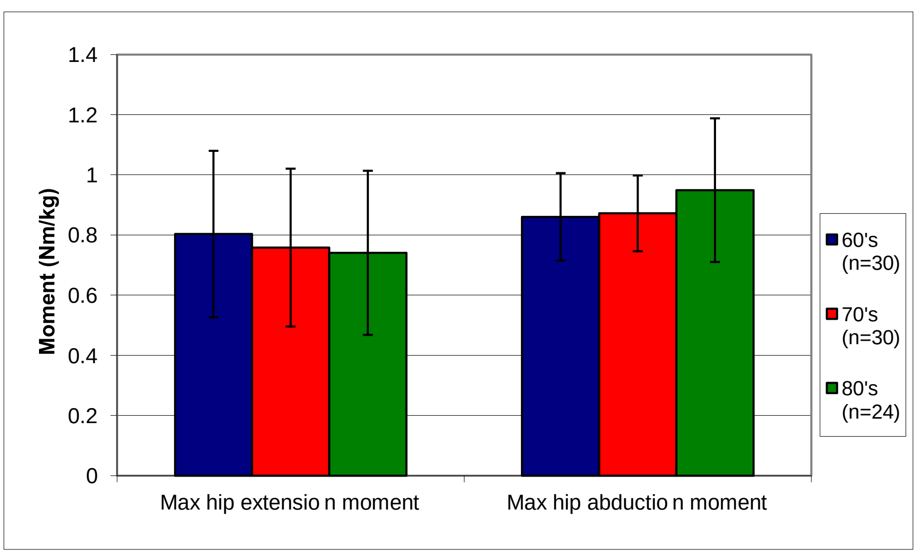


Figure 4.12 Maximum hip moments during stair ascent with hands. (Mean ± 1 s.d.)

		Age Group		
Moment	**Sex**	**60's**	**70's**	**80's**
Maximum hip extension moment (Nm/kg)	Male	0.85 (±0.26)	0.83 (±0.27)	0.83 (±0.31)
	Female	0.75 (±0.29)	0.68 (±0.24)	0.63 (±0.19)
	Combined	0.80 (±0.28)	0.76 (±0.26)	0.74 (±0.27)
Maximum hip abduction moment (Nm/kg)	Male	0.83 (±0.11)	0.86 (±0.11)	0.97 (±0.29)
	Female	0.89 (±0.14)	0.89 (±0.14)	0.92 (±0.16)
	Combined	0.86 (±0.13)	0.87 (±0.13)	0.95 (±0.24)

Table 4.9 Maximum hip moments during stair ascent with hands

Women did have a statistically reduced peak hip extension moment compared to the men ($p<0.05$) in all of the age groups (figure 4.13).

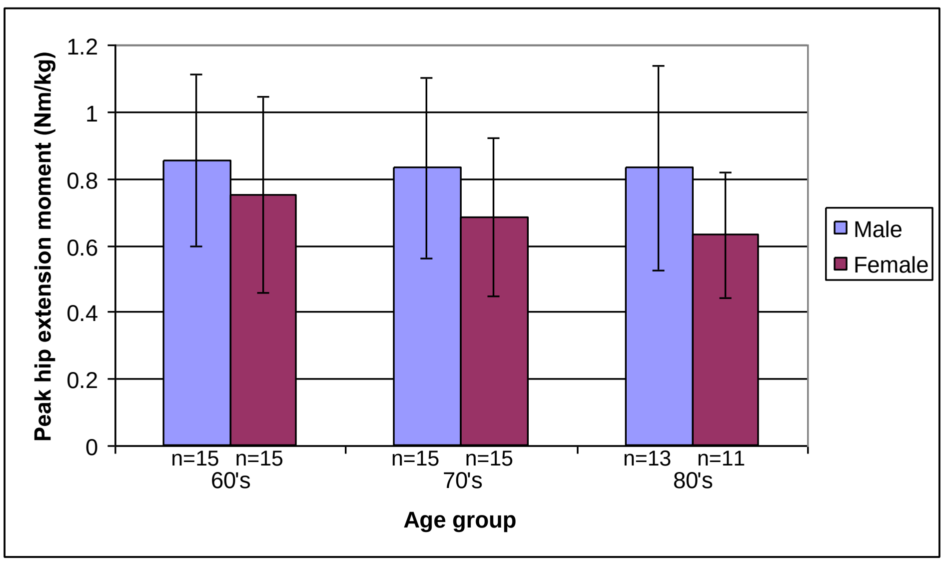


Figure 4.13 Maximum hip moments for males and females during stair ascent. (Mean ± 1 s.d.)

<u>Knee and ankle moments</u>

Plots of the knee extensor and ankle plantarflexor internal moments are presented in Figure 4.14 for the three age groups.

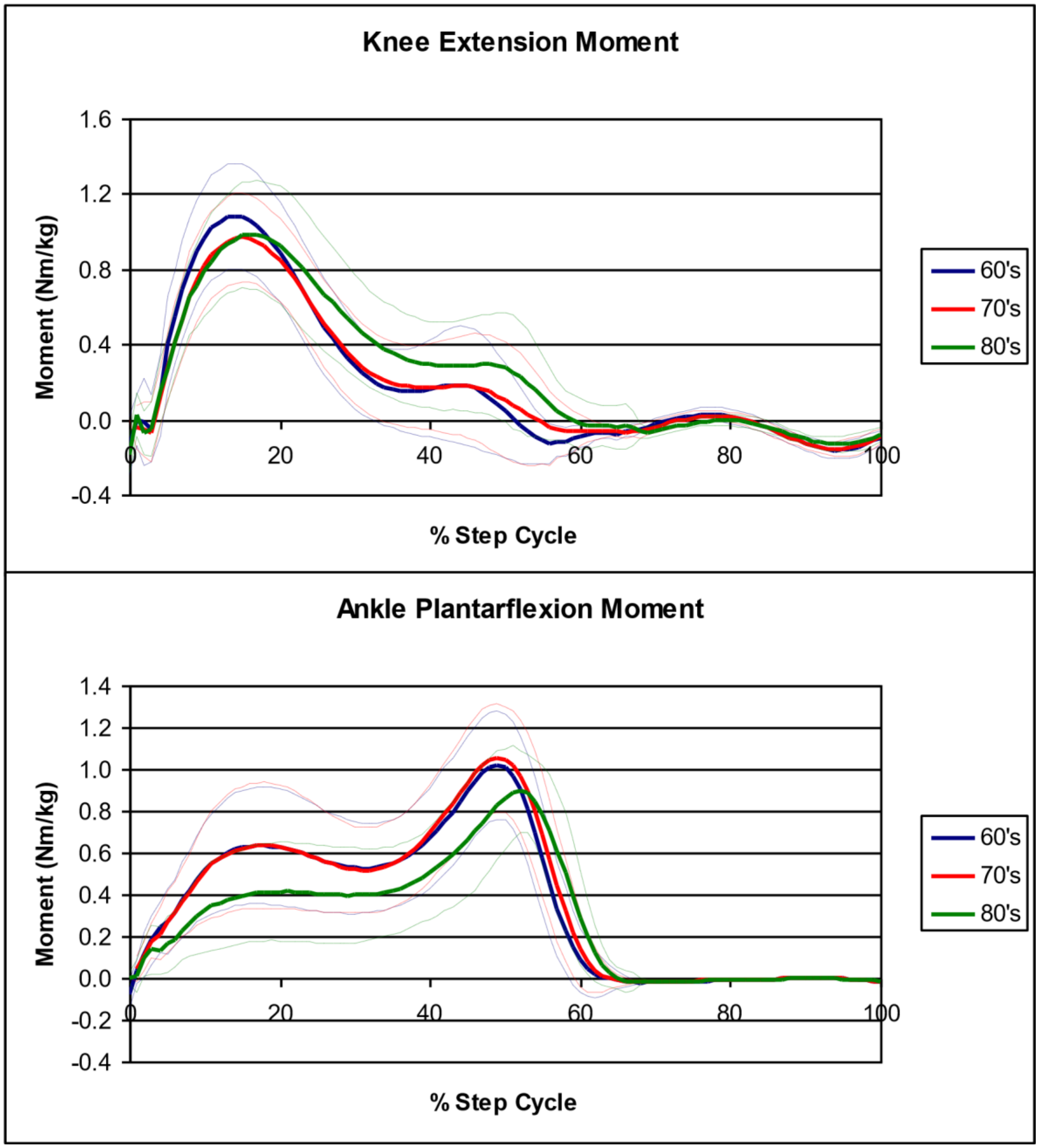


Figure 4.14 Moments produced around the knee and ankle joints during stair ascent using hands. Foot strikes are at 0 and 100% of the step cycle.

Following foot contact with the step, a knee extensor moment is produced which peaks in early stance as the knee gains its most extended position. An extensor moment is maintained throughout stance with a second smaller peak prior to foot off. During swing phase, knee moments are minimal but a slight flexor moment is produced prior to the next foot contact. Adults in their 80‟s had a slightly higher knee extension moment throughout the later part of stance than those in their 60‟s and 70‟s.

At the ankle, a plantarflexion moment is produced following foot contact and is present throughout stance phase. There is an initial peak in early stance at a similar point in time to the peak knee moment and then a second much larger peak just prior to foot off, indicating that the plantarflexors are responsible for much of the propulsion from the step. Adults in their 80‟s did not have such a clear 2 peak pattern as those in their 60‟s and 70‟s and tended to generate an ankle plantarflexion moment more steadily throughout stance phase, and to a reduced level, than the younger subjects. It appears that these subjects were using hip and knee extensors more to compensate for this.

The peak moments for each subject were normally distributed and hence investigated further using parametric statistical test. Mean results for each age group and gender are presented in figure 4.15 and table 4.10.

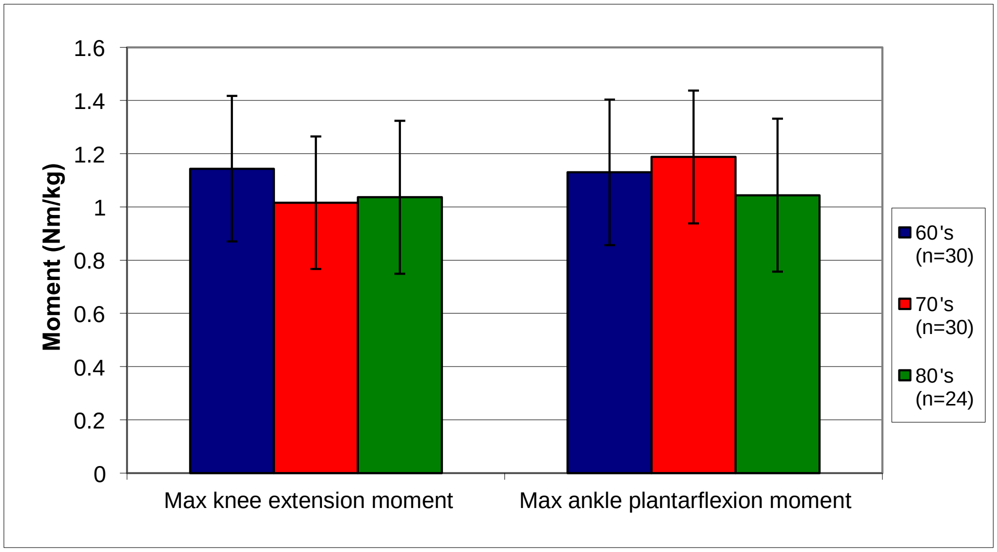


Figure 4.15 Maximum knee and ankle moments during stair ascent with hands

		Age Group		
Moment	**Sex**	**60's**	**70's**	**80's**
Maximum knee extension moment (Nm/kg)	Male	1.33 (±0.18)	1.15 (±0.27)	1.05 (±0.30)
	Female	0.96 (±0.22)	0.88 (±0.18)	1.01 (±0.29)
	Combined	1.14 (±0.27)	1.01 (±0.24)	1.04 (±0.29)
Maximum ankle plantarflexion moment (Nm/kg)	Male	1.16 (±0.14)	1.27 (±0.11)	1.03 (±0.14)
	Female	1.10 (±0.18)	1.11 (±0.16)	1.06 (±0.16)
	Combined	1.13 (±0.17)	1.19 (±0.16)	1.04 (±0.15)

Table 4.10 Maximum knee and ankle moments during stair ascent with hands

Statistically, adults in their 80‟s had a reduced peak plantarflexion moment compared to those in their 70‟s ($p<0.05$). Men had a statistically reduced peak knee extension moment in each increasing age group but this was not seen in women. Women however had a significantly decreased peak knee extension moment when compared to men for each age group. There were no effects of gender on peak ankle dorsiflexion moment.

4.5.8 Effect of handrail on kinetics during stair ascent

Plots of the moments at the hip, knee and ankle were produced for subjects ascending the stairs without use of the handrail, and a comparison of moments with and without a handrail is shown in figure 4.16. There was very little difference between the subjects in their 60‟s and 70‟s and the 70 year olds have been removed from this plot to aid clarity. It can be seen that in early stance there is an increase in hip and knee extensor moments when the handrail was not used. This increase is slightly more apparent in the older subjects. Use of the handrail does not seem to have any effect on hip abduction moments generated though stance for the 60 year olds, but there is a small increase in the peak hip abduction moment during the phase where there is only one leg on the step in the 80 year old group. The ankle plantarflexion moment in early stance seems similar in both conditions in both age groups. However, prior to toe off there is an increase in peak ankle plantarflexion moment generated by both age groups when not using the handrail.

A paired t-test was used to investigate the effect of handrail use on the peak ankle, knee and hip moments as all were normally distributed. Further analysis used 2-way ANOVA to determine if changes were related to age as well as handrail use.

There was a significant increase ($p<0.05$) in peak extension moments at the ankle, knee and hip with the absence of a handrail (table 4.11 and figure 4.17).

Maximum Moment (Nm/kg)	**Handrail used**	**No Handrail**
Hip extension	0.77 (±0.27)	0.85 (±0.31)
Knee extension	1.08 (±0.27)	1.14 (±0.25)
Ankle plantarflexion	1.13 (±0.17)	1.21 (±0.17)

Table 4.11 Handrail effects on maximum moments during stair ascent

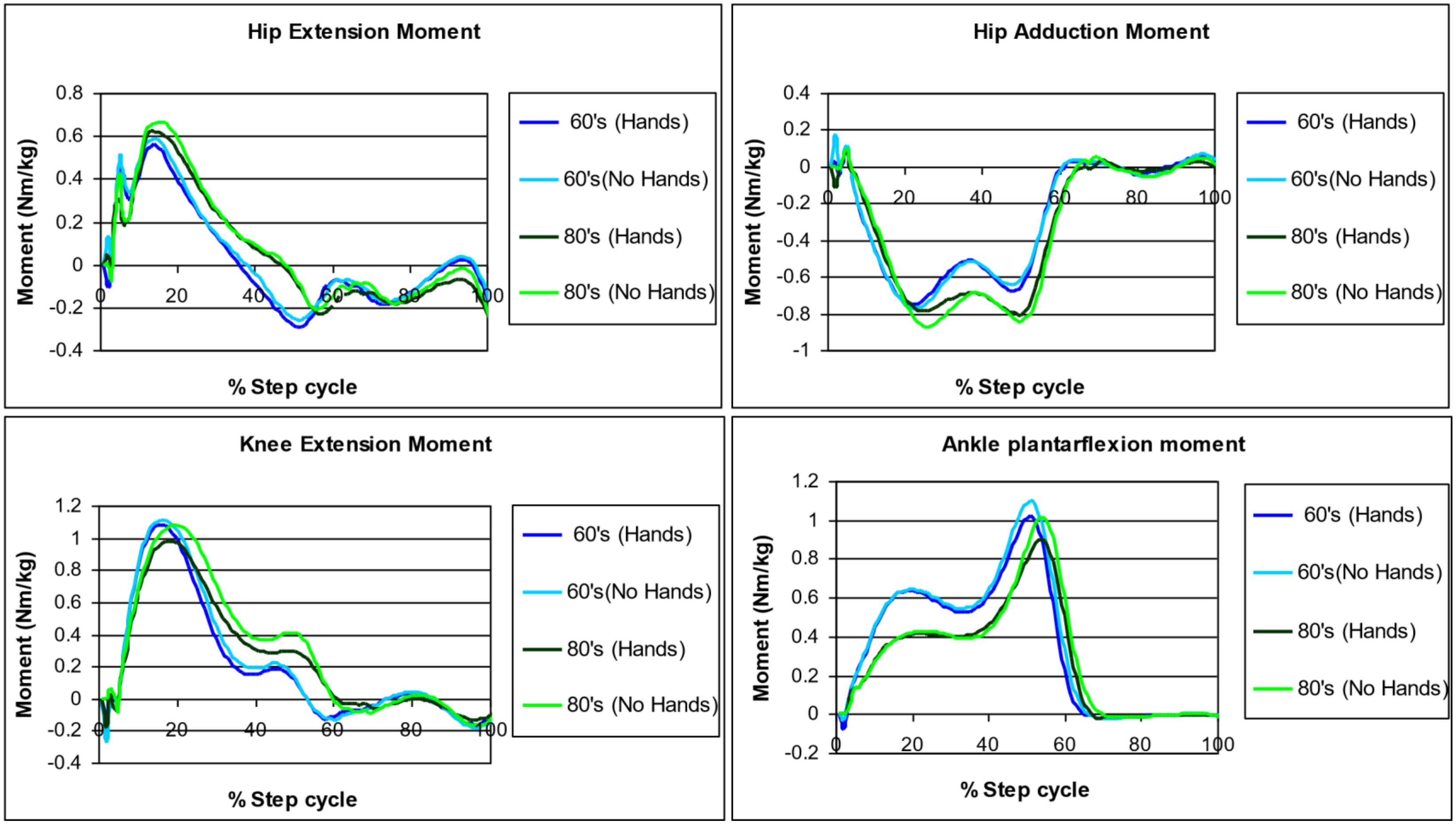


Figure 4.16 Moments at the hip, knee and ankle during stair ascent with and without a handrail

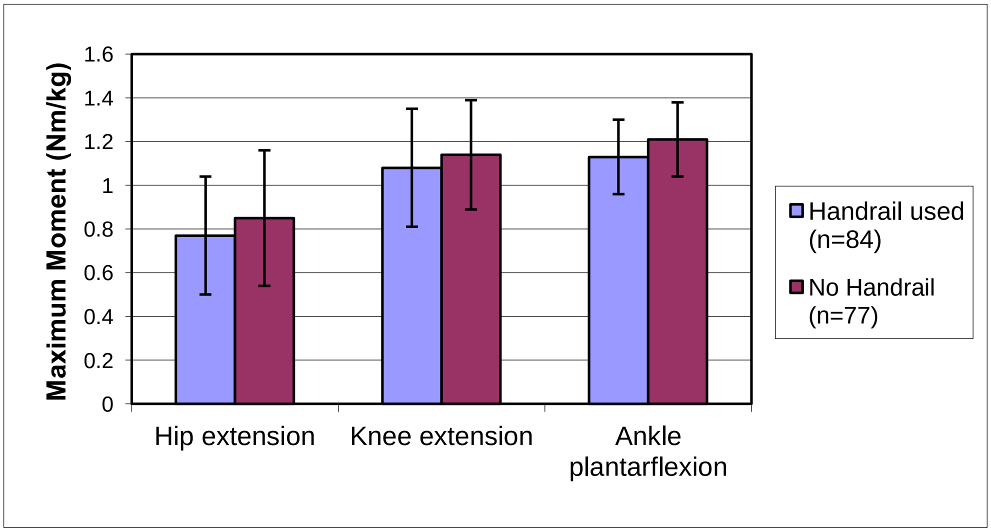


Figure 4.17 Handrail effects on maximum moments during stair ascent

The increase in moments was an average of 0.08 Nm/kg at the hip, 0.06 Nm/kg at the knee and 0.08 Nm/kg at the ankle. For an average 70 kg male this equates to an extra moment of 5.6 Nm at the hip, 4.2 Nm at the knee and 5.6 Nm at the ankle.

The ANOVA found no interaction between age, gender and handrail use, the increase in peak moments being uniform for all ages when the handrail was not used.

4.5.9 COM movement and velocity during stair ascent with a handrail

The position of the COM of the body relative to the laboratory was determined in the three orthogonal axes. As the subjects were walking forwards and up a fixed staircase, the COM was seen to progress forwards and upwards. The degree of excursion in the anterior/posterior and vertical directions represented the size of the staircase and did not demonstrate any changes due to ageing. The excursion of the COM in the mediolateral (ML) direction represents the sway of the COM from side to side whilst progressing up the staircase. The mediolateral (ML) excursion of the COM during one step cycle is shown in table 4.12.

		Age Group		
COM excursion	**Sex**	**60's**	**70's**	**80's**

Mediolateral COM Excursion(mm)	Male	50.7(±16.4)	54.1 (±13.1)	61.9 (±23.8)
	Female	47.7(±11.2)	49.2(±12.2)	44.0(±13.2)
	Combined	49.2(±13.9)	51.7(±12.6)	53.3(±21.1)

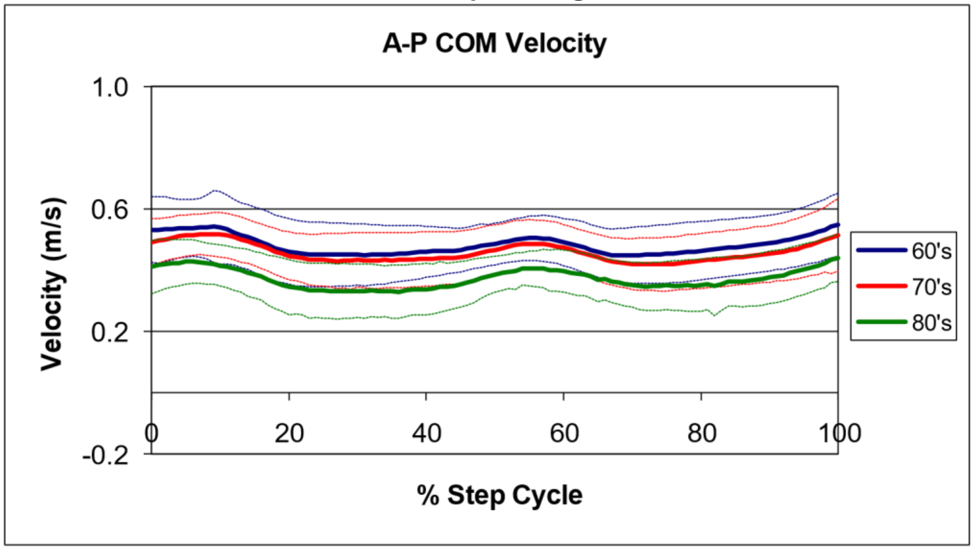
Table 4.12 COM excursion during stair ascent with a handrail

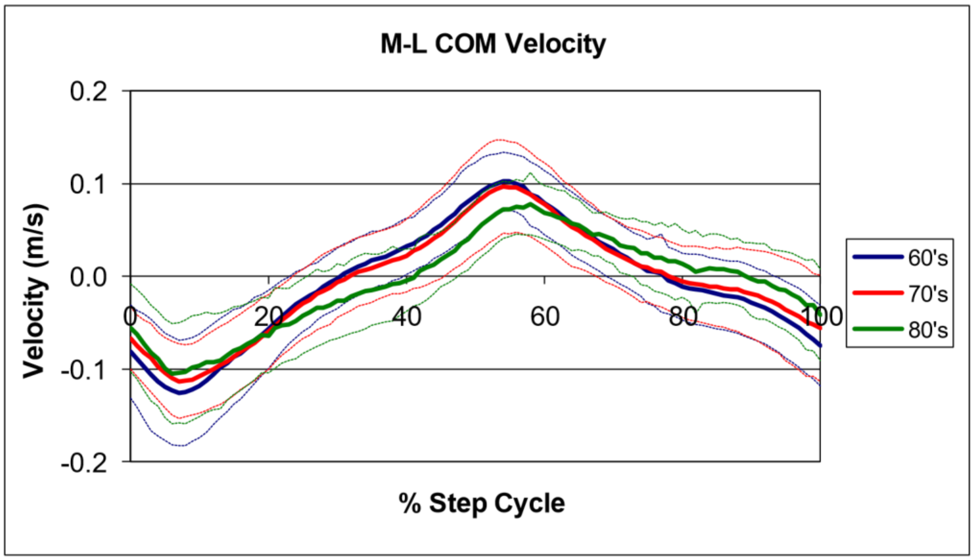
2-way ANOVA demonstrated there was no significant change in the amount of ML COM excursion with increasing age. It had been anticipated that older subjects, due to balance deficits may not be able to control COM excursion as well as the younger subjects, but did this does not appear the case in this group. Male subjects had a significantly ($p<0.05$) larger excursion of the COM than their female counterparts.

The velocity of the centre of mass in three orthogonal directions was studied to determine if and how older adults adopted strategies to reduce COM motion. Plots of COM velocity for a whole gait cycle are shown in figure 4.18 and the maximum velocities used are shown in table 4.13.

		Age Group		
Maximum COM velocity	**Sex**	**60's**	**70's**	**80's**
A-P (m/s)	Male	0.59(±0.07)	0.59(±0.09)	0.49(±0.08)
	Female	0.61(±0.12)	0.55(±0.10)	0.48(±0.05)
	Combined	0.60(±0.10)	0.57(±0.10)	0.49(±0.07)
M-L (m/s)	Male	0.14(±0.04)	0.12(±0.06)	0.15(±0.04)
	Female	0.12(±0.05)	0.12(±0.03)	0.10(±0.03)
	Combined	0.13(±0.05)	0.12(±0.04)	0.12(±0.05)
Vertical (m/s)	Male	0.55(±0.07)	0.56(±0.06)	0.50(±0.06)
	Female	0.54(±0.08)	0.53(±0.04)	0.49(±0.06)
	Combined	0.55(±0.07)	0.55(±0.05)	0.49(±0.06)

Table 4.13 Maximum COM velocity during stair ascent with a handrail





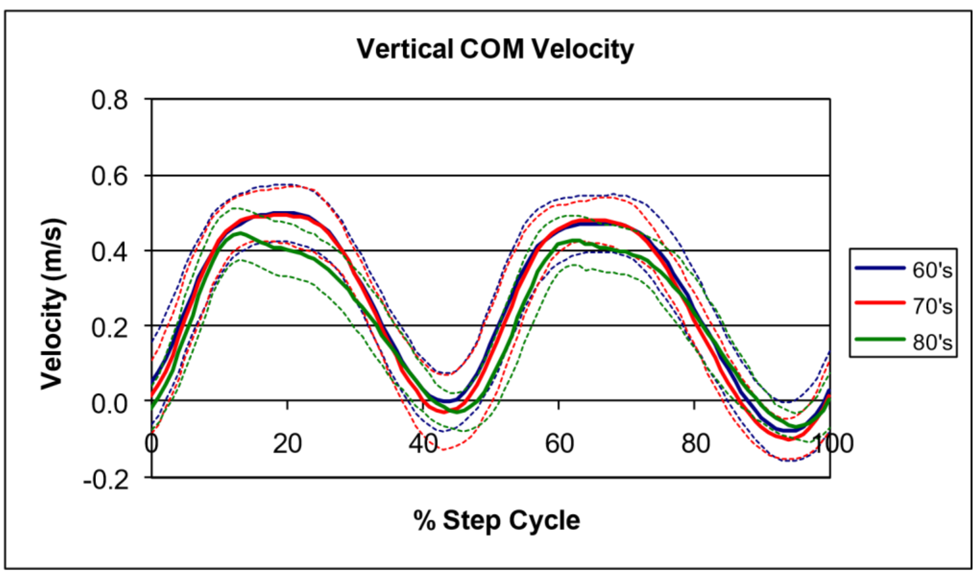


Figure 4.18 COM velocity during stair ascent. Foot strikes are at 0 and 100% of cycle

In the A-P direction the velocity of the COM was in the forward direction and varied only slightly, with a small increase in velocity after foot contact on each step. 2-way ANOVA revealed no gender differences but found there to be a significant age related decrease in A-P velocity in the subjects in their 80‟s compared to those in their 60‟s and 70‟s.

In the M-L direction the COM increases in speed towards the weight bearing foot following initial contact on the step. This is followed by a period of slowing down and then as the opposite foot contacts the step COM velocity increases towards that side. ANOVA revealed no age related changes in M-L velocity but as with M-L COM excursion there was a gender effect with females having a lower peak velocity than males.

In the vertical direction there was a 2-peaked pattern for COM velocity. These two peaks correspond to the time following foot contact when the body is being progressed upwards towards the next step due to extension of the hip and knee. This corresponds to the phases of single support. The subjects in their 80‟s had reduced velocity of the COM in the vertical direction compared to the other groups (ANOVA $p<0.05$). They also did not maintain the same COM velocity through single support, demonstrating an earlier deceleration compared to the younger groups.

In section 4.3.2 it was reported that the speed of stair ascent was similar in the 60‟s and 70‟s and decreased in the 80 year old group. The COM velocity has a similar pattern and may be explained simply by the changes in the speed of stair climbing.

4.5.10 Effect of handrail use on COM movement

The mean excursion of the COM in a mediolateral direction with and without handrail use whilst ascending the stairs is shown in Table 4.14.

	Mediolateral COM excursion (mm)	
Age Group	**Handrail used**	**No Handrail**
60‟s	49.2 (±13.9)	62.6(±17.3)
70‟s	51.6(±12.6)	61.2(±14.8)
80‟s	53.3(±21.1)	81.9(±35.6)
Mean overall	51.2(±15.7)	67.3(±24.4)

Table 4.14 COM excursion during stair ascent with and without a handrail

A 2-way ANOVA was performed to investigate the effects of age and handrail use on the mediolateral COM movement during stair ascent. There was a significant increase in the amount of excursion of the COM when the handrail was not used ($p<0.05$), from an average of 51.2mm to 67.3mm. There was also a significant increase in the amount of excursion demonstrated in the 80+ group compared to the younger groups when not using a handrail as shown in figure 4.19. There was interaction between handrail use and age in the ANOVA, suggesting that the oldest group are more affected by the loss of the handrail.

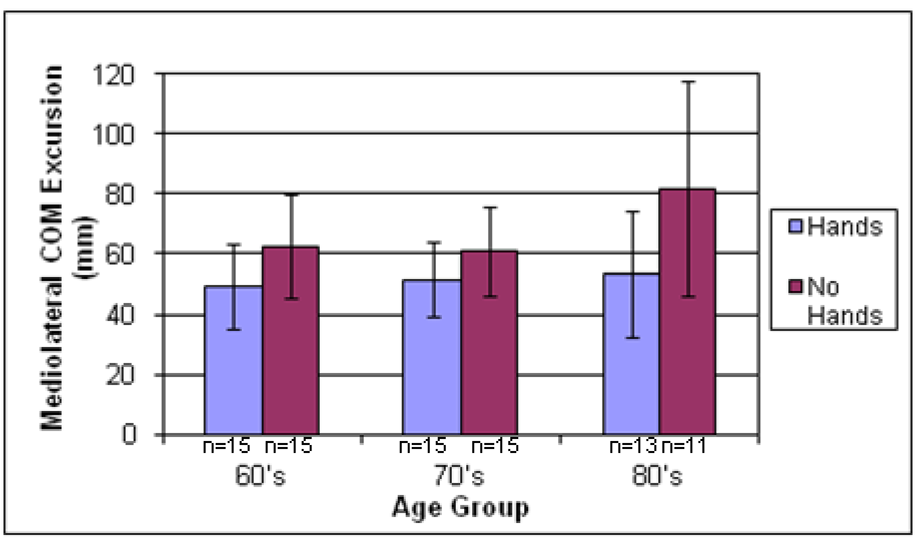


Figure 4.19 Handrail effects on mediolateral COM excursion during stair ascent

4.5.11 Summary for stair ascent

Ascending the stairs with a handrail was an activity all subjects were able to perform. The rate of stair ascent was significantly slower in the 80-year-old group, with an increase in the time spent in stance and double support. The decline in velocity related to a decline in velocity of the COM in the A-P and vertical direction in the oldest age group. Demands on joint range of motion in the hips, knees and ankles were similar between the three ages with a maximum hip flexion of around 70 degrees, knee flexion of 98 degrees, ankle dorsiflexion of 21 degrees and plantarflexion of 19 degrees. These are much larger joint ranges than would be required for level walking (typically 30 degrees of hip flexion, 45 degrees of knee flexion and 10 degrees of dorsiflexion (Whittle, 1996)). Subjects in their 80‟s adopted a more flexed position in the lower limb during the stance phase of gait, with a non-significant change at the ankle and hip and a significant increase in knee flexion. Some changes in joint kinetics were observed between the age groups. The oldest age group had a reduced ankle plantarflexion moment throughout stance, which seems to be compensated by increasing hip and knee extensor moments in later stance phase compared to the younger subjects. There were significantly lower peak ankle plantarflexion moments in the 80+ group but no significant difference in the peak knee or hip extensor moment produced during the activity.

Ascending stairs without a handrail was found to be a more difficult activity, with 5 subjects unable to perform this at all, all of whom were in the older two age groups. Subjects ascended steps more quickly without the handrail but spent the same proportion of time in swing and stance. There was little effect on joint kinematics when ascending stairs without use of hands but there was a significant increase in the joint moments required to extend the hip, knee and ankle. This increase in joint moments required was present in all age groups but was larger for subjects in the 80-year-old group. Without the handrail, the excursion of the COM side to side was much higher, especially in the 80+ age group.

These findings are discussed in Chapter 5.

4.6 Stair descent

4.6.1 Temporal data

Data were tested for normality and all data were normally distributed as a whole and when divided into gender and age groups. 2-way ANOVA was used to assess the following parameters as with stair ascent:

- Time taken for one cycle
- Rate of stair descent
- Stance phase (%)
- Swing Phase (%)
- Total double support time (%) (i.e time with both feet on the staircase)

4.6.2 Temporal data using a handrail

Gender differences were found for stair descent using a handrail. Male subjects spent statistically more time in stance phase than female subjects and had an increased period of double support ($p<0.05$). The increase in double support accounted for 2% of the complete gait cycle.

Unlike stair ascent where there were age related effects on all temporal parameters, there was only an age related change in the rate of stair descent (or time per step cycle).

This change was observed between the 60‟s and 80‟s and 70‟s and 80‟s but no difference was found between subjects in the 60‟s and 70‟s. There were no age effects on the proportion of time spent in stance and swing, and correspondingly double support. Temporal data are presented in table 4.15 and figure 4.20.

Age Group	60's	70's	80's
Time per step cycle (s)	1.27 (± 0.26)	1.40 (±0.27)	1.69 (±0.35)*
Rate of stair descent (steps/min)	97.8 (±17.4)	88.6 (±16.8)	74.2 (±15.6)*
Stance phase (% of cycle)	59.4 (±2.6)	59.1 (±3.7)	60.1 (±3.2)
Swing phase (% of cycle)	40.4 (±2.6)	39.9 (±3.7)	38.0 (±3.2)
Double support (% of cycle)	19.2 (±2.8)	19.7 (±3.8)	19.6 (±5.5)

Table 4.15 Temporal data for stair descent using a handrail. * indicates significant difference at p<0.05 level

4.6.3 Temporal data without a handrail

There were no effects on any parameter related to gender when stair descent was performed using a handrail. There were significant age related effects between the 60‟s and 80‟s and 70‟s and 80‟s for the time taken for each step cycle and hence the rate of stair descent, with the oldest subjects being slower. There was a slight, but significant, increase in the amount of time spent in stance for the 80‟s compared to the 60‟s but no change in the double support time across the age groups. It may be that by increasing the amount of time in stance the oldest group are increasing stability to cope with the loss of the handrail.

Data are presented in table 4.16 and in figure 4.20.

Age Group	60's	70's	80's
Time per step cycle (s)	1.20 (± 0.23)	1.24 (±0.24)	1.53 (±0.29)*
Rate of stair descent (steps/min)	103.8 (±19.4)	99.2 (±16.6)	81.2 (±15.6)*
Stance phase (% of cycle)	60.0 (±2.4)	60.5 (±3.6)	61.6 (±3.3)*

Swing phase (% of cycle)	40.0 (±2.4)	39.5 (±3.6)	38.4 (±3.3)*
Double support (% of cycle)	19.3 (±3.1)	20.5 (±4.5)	21.4 (±6.0)

Table 4.16 Temporal data for stair descent without using a handrail.

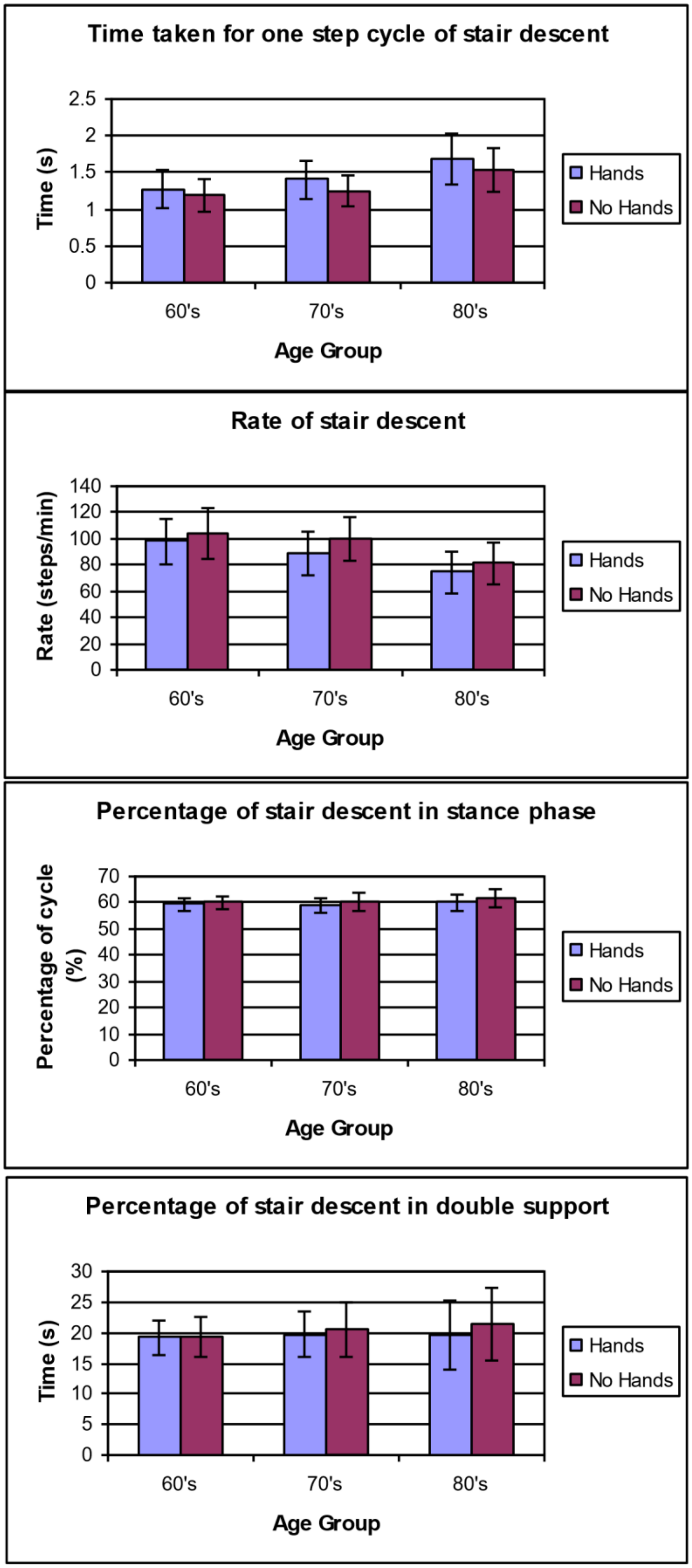


Figure 4.20

Temporal data for stair descent with and without a handrail

4.6.4 Effect of hand rail on temporal data during stair descent

The time take on each step and the rate of stair descent were both significantly ($p<0.05$) affected by the use of a handrail. Subjects went faster down the stairs spending less time on each step when no handrail was present. The proportion of time spent in stance and double support was also affected be the use of the handrail. Without the handrail, subjects spent more time in stance and in double support. It would therefore appear that although subjects are descending the stairs faster, that they are trying to adopt a more stable method of stair descent.

4.6.5 Kinematics of stair descent using a handrail

4.6.5.1 Sagittal plane kinematics

Sagittal plane kinematics are shown in figure 4.21 for the subjects in the three age groups. All data were tested for normality prior to statistical analysis. All peak kinematic data were normally distributed.

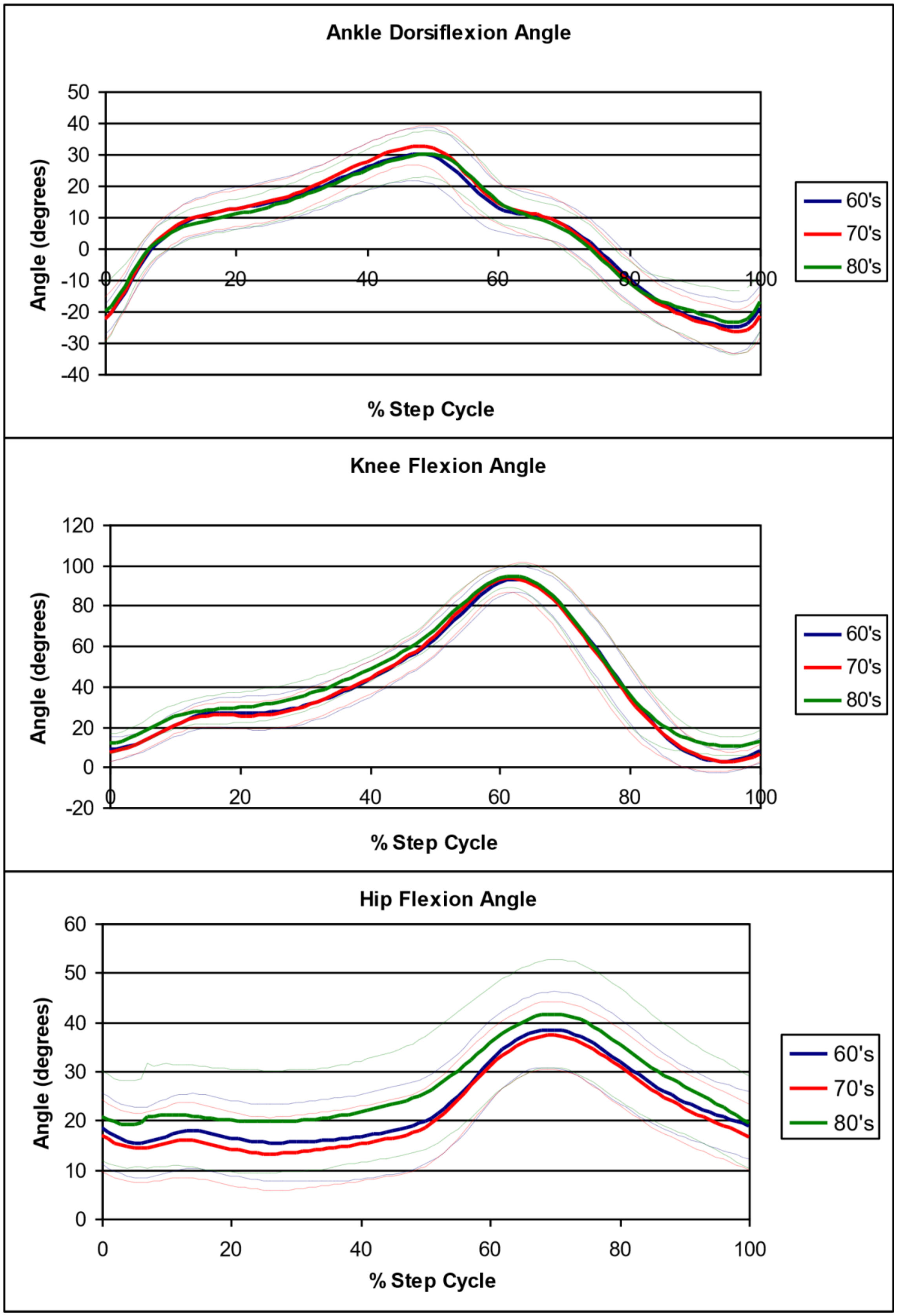


Figure 4.21 Sagittal plane kinematics for stair descent using hands. Foot strikes are at 0 and 100% of the step cycle.

Ankle joint

At the ankle joint subjects contacted the step with just over of 20 degrees plantarflexion. As weight was transferred over the foot the ankle became increasingly dorsiflexed and peaked at around 33 degrees of dorsiflexion towards the end of stance. Approaching toe off the ankle became less dorsiflexed, and then becoming plantarflexed as it was lowered to the next step prior to initial contact. Peak angles for ankle dorsiflexion and plantarflexion are shown in Table 4.16

		Age Group		
Angle	**Sex**	**60's**	**70's**	**80's**
Maximum ankle dorsiflexion (degrees)	Male	29.9(±2.8)	36.4(±1.2)	31.6(±1.4)
	Female	35.2(±1.5)	32.3(±1.5)	28.7(±2.9)
	Combined	32.5(±1.6)	34.6(±1.0)	30.4(±1.4)
Maximum ankle plantarflexion (degrees)	Male	27.1(±1.2)	27.4(±1.1)	24.4(±2.7)
	Female	27.8(±1.9)	29.1(±1.5)	29.3(±2.5)
	Combined	27.4(±1.1)	28.2(±0.9)	26.4(±1.9)

Table 4.16 Maximum ankle plantar and dorsiflexion angles during stair descent using hands

Using a 2-way anova there were found to be no significant age or gender effects on peak ankle angles during stair descent.

Knee joint

As the descending foot contacted the step the knee was in a nearly fully extended position. Knee flexion occurred fairly gradually throughout stance phase peaking just prior to toe off at around 95 degrees. This flexing of the knee enables the opposite limb to be progressed down to the next step. During swing phase the knee was quickly extended in preparation for the next foot contact as the descending leg was lowered to the step. Peak knee angles are shown in table 4.17

		Age Group		
Angle	**Sex**	**60's**	**70's**	**80's**
Maximum knee	Male	94.7(±1.2)	92.4(±1.15)	95.7(±1.7)

flexion (degrees)	Female	96.9(±1.4)	99.5(±2.1)	98.9(±1.7)
	Combined	95.7(±0.9)	95.5(±1.3)	97.0(±1.3)
Minimum knee flexion (degrees)	Male	4.2(±1.0)	3.3(±1.4)	11.7(±1.0)
	Female	0.4(±1.0)	1.6(±1.4)	5.9(±1.3)
	Combined	2.3(±0.8)	2.6(1.0)	9.3(±1.0)

Table 4.17 Maximum and minimum knee angles during stair descent using hands

A 2-way ANOVA found there to be no statistically significant age related differences in the amount of knee flexion used during stair descent. Female subjects were found to use significantly more knee flexion than there male counterparts ($p<0.05$). This may be explained by the fact that the women were shorter than the men and therefore needed to flex the trailing knee more to enable foot contact for descending leg. The amount of extra knee flexion was 4 degrees, female subjects using an average of 98 degrees compared to 94 degrees in the male subjects. For knee extension there were statistically significant ($p<0.05$) age and gender related changes. Subjects in their 80‟s differed from the subjects in their 60‟s and 70‟s by not extending there knees as fully in preparation for stance and during early stance. Female subjects had a more extended knee position at initial contact and early stance than their male colleagues. This difference again was around 4 degrees meaning that overall the female subjects use an extra 8 degrees of knee movement during stair descent than the males. Whether these differences are related to the female subjects having better knee flexibility than the males of whether these are purely effects of height will be explored later.

Hip joint

Compared to stair climbing there is a lot less sagittal plane movement at the hip during stair descent. At initial foot contact the hip is slightly flexed (15-20º). This flexion remains fairly constant during stance phase and then the hip flexes to progress the swinging limb to the next step. Peak hip flexion is in early swing and then the hip extends in preparation for the next foot contact on the step below. Peak hip angles are presented in table 4.18.

		Age Group		
Angle	**Sex**	**60's**	**70's**	**80's**
Maximum hip flexion (degrees)	Male	40.5(±2.2)	35.1(±1.9)	45.1(±3.4)
	Female	39.6(±1.8)	40.8(±2.1)	42.3(±3.1)
	Combined	40.0(±1.4)	37.6(±1.5)	44.0(±2.4)
Minimum hip flexion (degrees)	Male	15.5(±1.9)	12.4(±1.8)	20.9(±3.0)
	Female	11.1(±1.6)	12.8(±2.1)	12.6(±2.1)
	Combined	13.3(±1.3)	12.6(±1.4)	14.2(±0.9)

Table 4.18 Maximum and minimum hip angles during stair descent using hands

2-way ANOVA found no age or gender related effects for the maximum amount of hip flexion used during stair ascent, although there did appear that there was a trend for 80 year old subjects to have a more flexed hip posture throughout the activity. Female subjects had a significantly ($p<0.05$) reduced amount of hip flexion in stance phase on average using 4 degrees more hip extension than there male counterparts. Men in their 80‟s were found to remain more flexed in stance than those in their 60‟s and 70‟s ($p<0.05$).

4.6.5.2 Coronal and transverse plane kinematics

Coronal and transverse plane kinematics for stair descent are shown in figures 4.22 and 4.23.

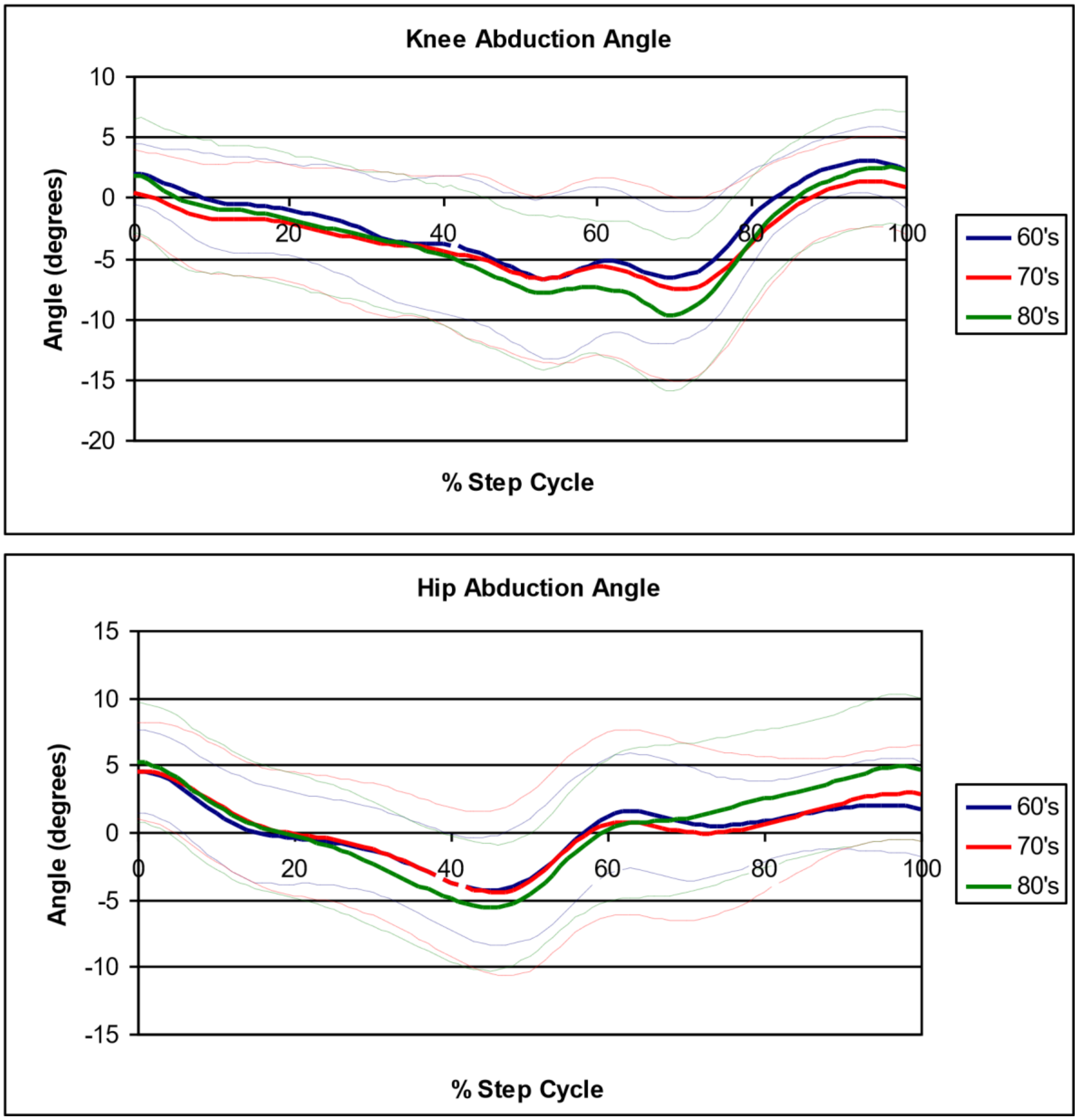


Figure 4.22 Coronal plane kinematics for stair descent using hands. Foot strikes are at 0 and 100% of the step cycle.

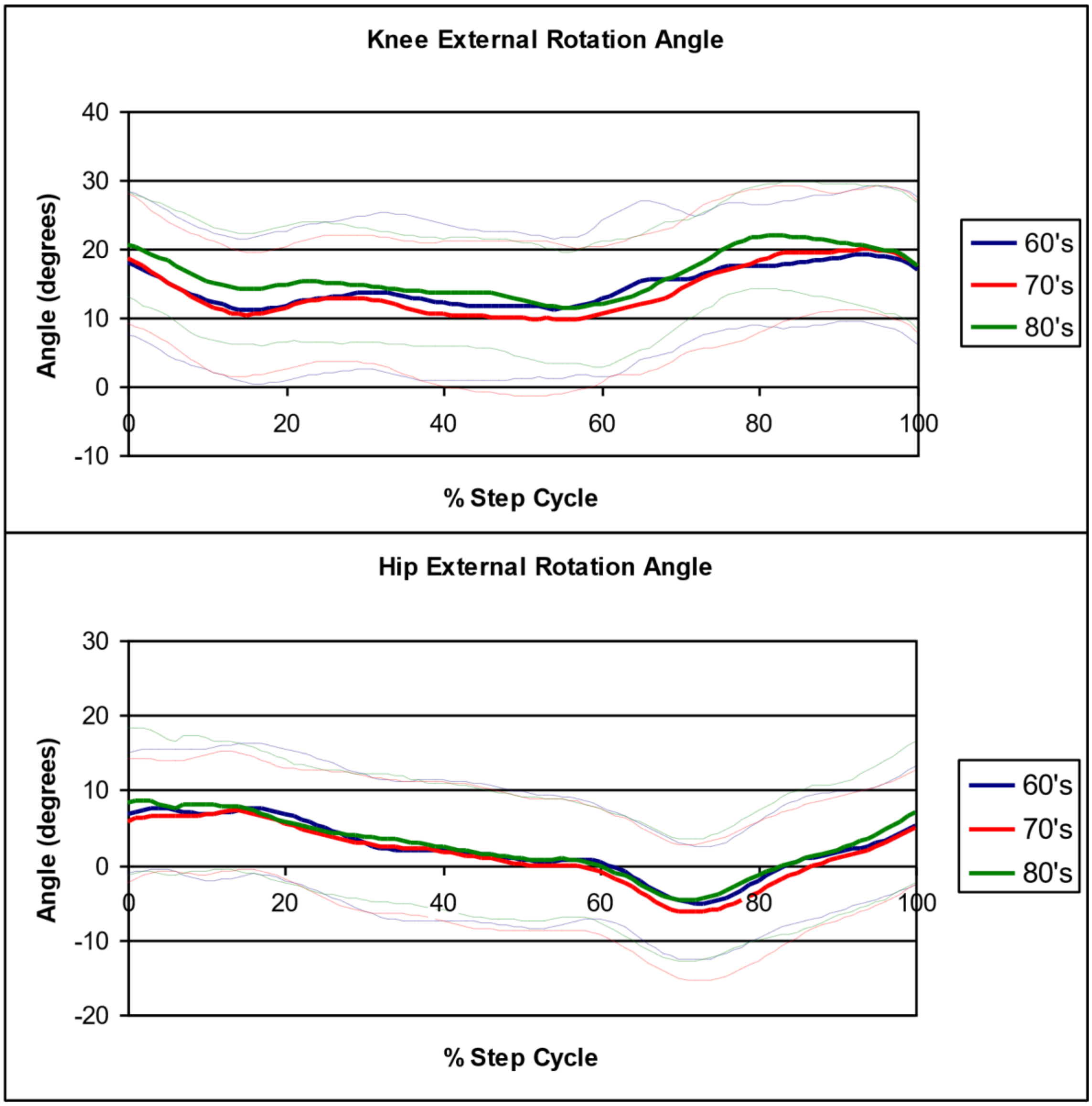


Figure 4.23 Transverse plane kinematics for stair descent using hands. Foot strikes are at 0 and 100% of the step cycle.

Knee joint

At initial contact the knee is in a neutral position regarding ab/adduction and externally rotated (tibial torsion). As the knee flexes through stance there is a corresponding increase in knee adduction. This may indicate some error with alignment of the knee axis resulting in some knee flexion being detected as knee adduction. Knee rotation is fairly static throughout stair descent and would appear to represent normal tibial torsion. 2-way ANOVA found there were no significant age or gender related differences in the amount of joint excursion.

Hip joint

At initial contact the hip is slightly abducted. As the opposite leg progresses past and then lowers to the next step the hip becomes adducted and reaches peak adduction late stance. During swing phase the hip is slightly abducted and remains so until the next foot contact. Overall the hip only moves about 10 degrees in the coronal plane.

The hip is slightly externally rotated at initial foot contact and then rotates to neutral and then slightly internally rotated as the pelvis and opposite leg progress forwards. During swing phase the hip rotates slightly towards external rotation as the leg progresses to the next step. Total hip internal/external rotation is around 15 degrees.

2-way ANOVA found there to be no significant differences between the 3 age groups, or between males and females in relation to sagittal and coronal plane peak hip angles.

4.6.6 Effect of handrail on kinematics of stair descent

When the kinematic plots were produced for stair descent without a handrail the movement patterns were identical to stair descent using a handrail. Peak angles were within 1 degree of those found when using the handrail so there were no significant effects of use of the handrail. The inability of some subjects to descend the stairs without using the handrail would not appear to be related to an increased demand on joint ROM.

4.6.7 Kinetics of stair descent using a handrail

Plots of the internal moments generated during stair descent are presented in figures 4.24 and 4.26. As with stair ascent this thesis will concentrate on the moments balanced predominantly by muscular control i.e. hip extension and abduction, knee extension and ankle plantarflexion.

Hip moments

Hip moments in the sagittal plane were not very large throughout stance or swing phase during stair descent. For the majority of the time a small moment was produced by the hip flexors balancing a small external extension moment when the COM passes behind the hip joint. As the upper body progresses forwards approaching toe off less of a hip flexor moment was produced, then there is a small flexor moment during initial swing phase. In the coronal plane, there is an immediate increase in the hip abduction moment produced following foot contact and then a hip abduction moment is produced throughout stance. This moment represents the hip abductors working eccentrically as the pelvis tilts allowing the swing leg to progress to the next step.

Studying the kinematic plots, it can be seen that subjects in their 80‟s produced significantly lower hip flexor moments during stance phase, occasionally producing a hip extension moment. This may be as a result of the more flexed hip posture adopted during stance for these subjects reported in section 4.6.7.

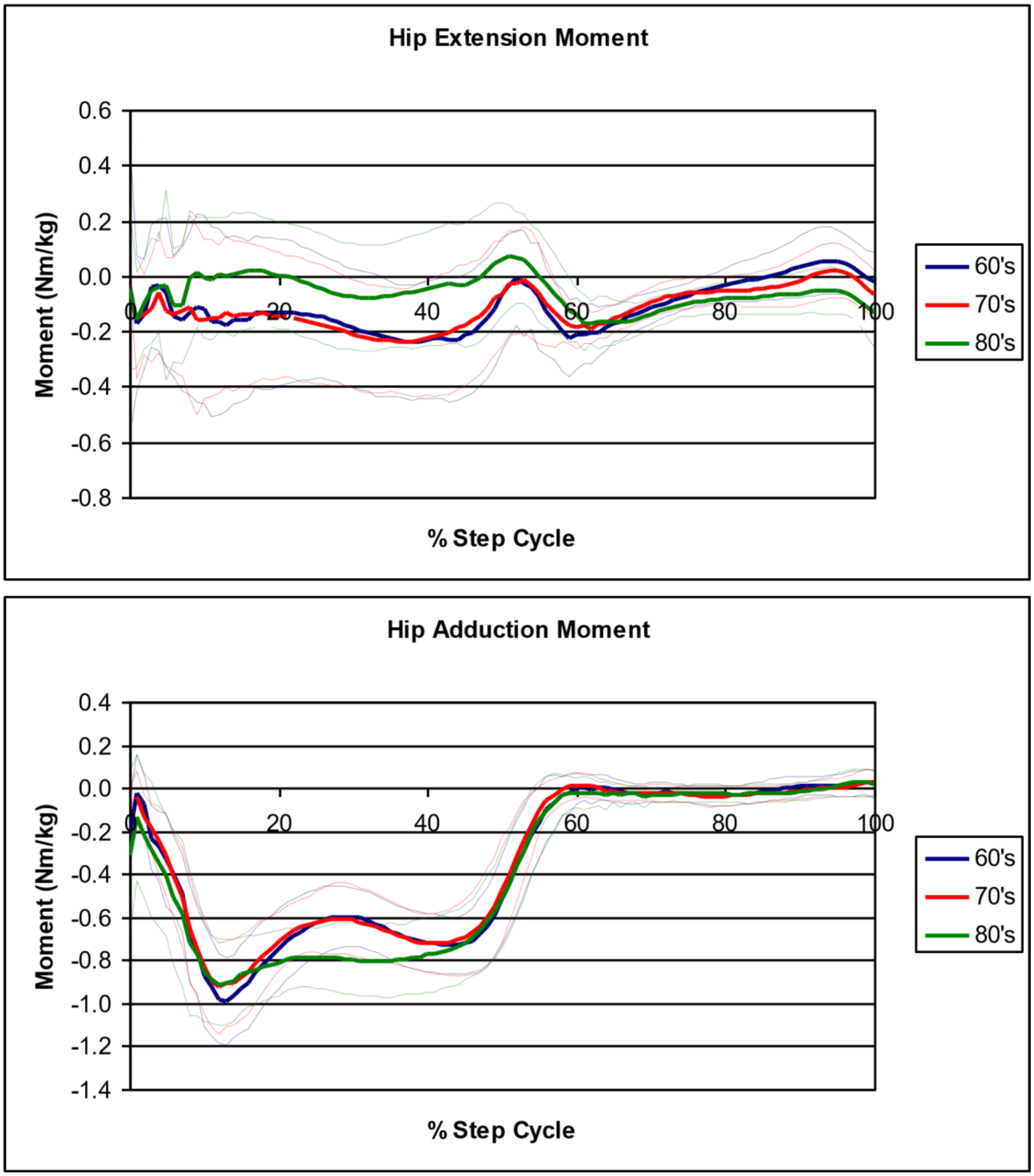


Figure 4.24 Moments produced around the hip joint during stair descent using hands. Foot strikes are at 0 and 100% of the step cycle.

Peak hip moments for each individual trial were entered into SPSS and analysed further. Hip moments were not normally distributed and were therefore analysed using a Kruskall Wallis test to determine age related changes. The maximum hip extension and abduction moments generated are shown in figure 4.25 and table 4.19.

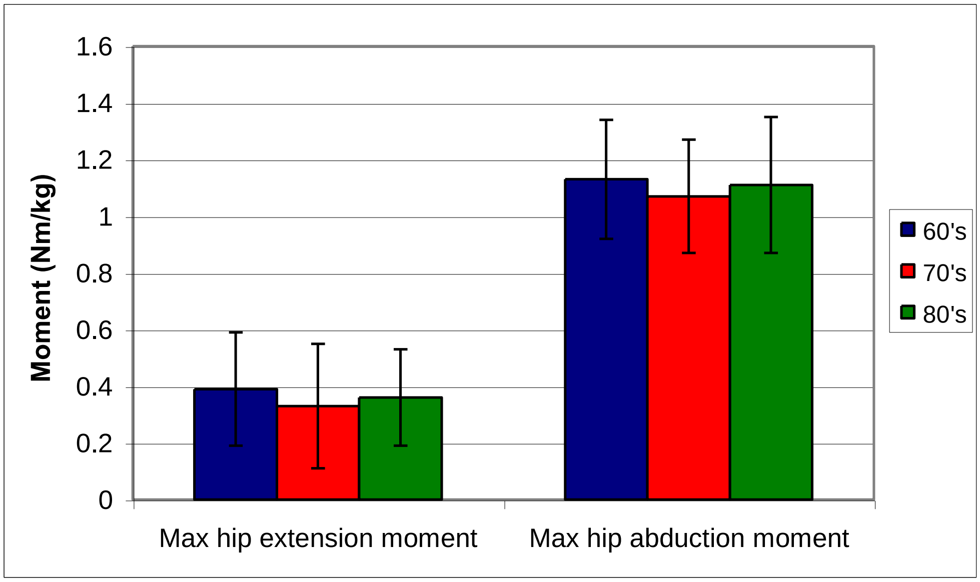


Figure 4.25 Peak hip moments during stair descent using a handrail

		Age Group		
Moment	**Sex**	**60's**	**70's**	**80's**
Maximum hip extension moment (Nm/kg)	Male	0.39 (±0.22)	0.31 (±0.22)	0.40 (±0.19)
	Female	0.39 (±0.19)	0.34 (±0.23)	0.30 (±0.11)
	Combined	0.39 (±0.20)	0.33 (±0.22)	0.36 (±0.17)
Maximum hip abduction moment (Nm/kg)	Male	1.10 (±0.19)	1.01 (±0.18)	1.08 (±0.23)
	Female	1.17 (±0.24)	1.12 (±0.21)	1.13 (±0.25)
	Combined	1.13 (±0.21)	1.07 (±0.20)	1.11 (±0.24)

Table 4.19 Maximum hip moments during stair descent with hands

There were no significant differences in the peak hip moments produced between the age groups and no differences between the genders.

Knee and ankle moments

Plots of the knee extensor and ankle plantarflexor internal moments are presented in figure 4.26 for the three age groups.

Following foot contact with the step a knee extensor moment is produced. A small peak occurs in early stance, corresponding to the time when the opposite leg will be leaving the step i.e. the start of single support phase. There is then a difference between the subjects in their 80‟s and those in the 60‟s and 70‟s. The younger groups have a decrease in the knee extension moment produced during mid-stance, increasing again to a second, larger, peak prior to toe off from the step. The knee extensor moment corresponds to the knee becoming increasingly flexed and therefore must be produced by the quadriceps muscle groups working eccentrically (lengthening) to balance the external moment tending to flex the knee. The subjects in their 80‟s do not have the same decrease in knee extensor moment during mid-stance, but continue to increase the knee

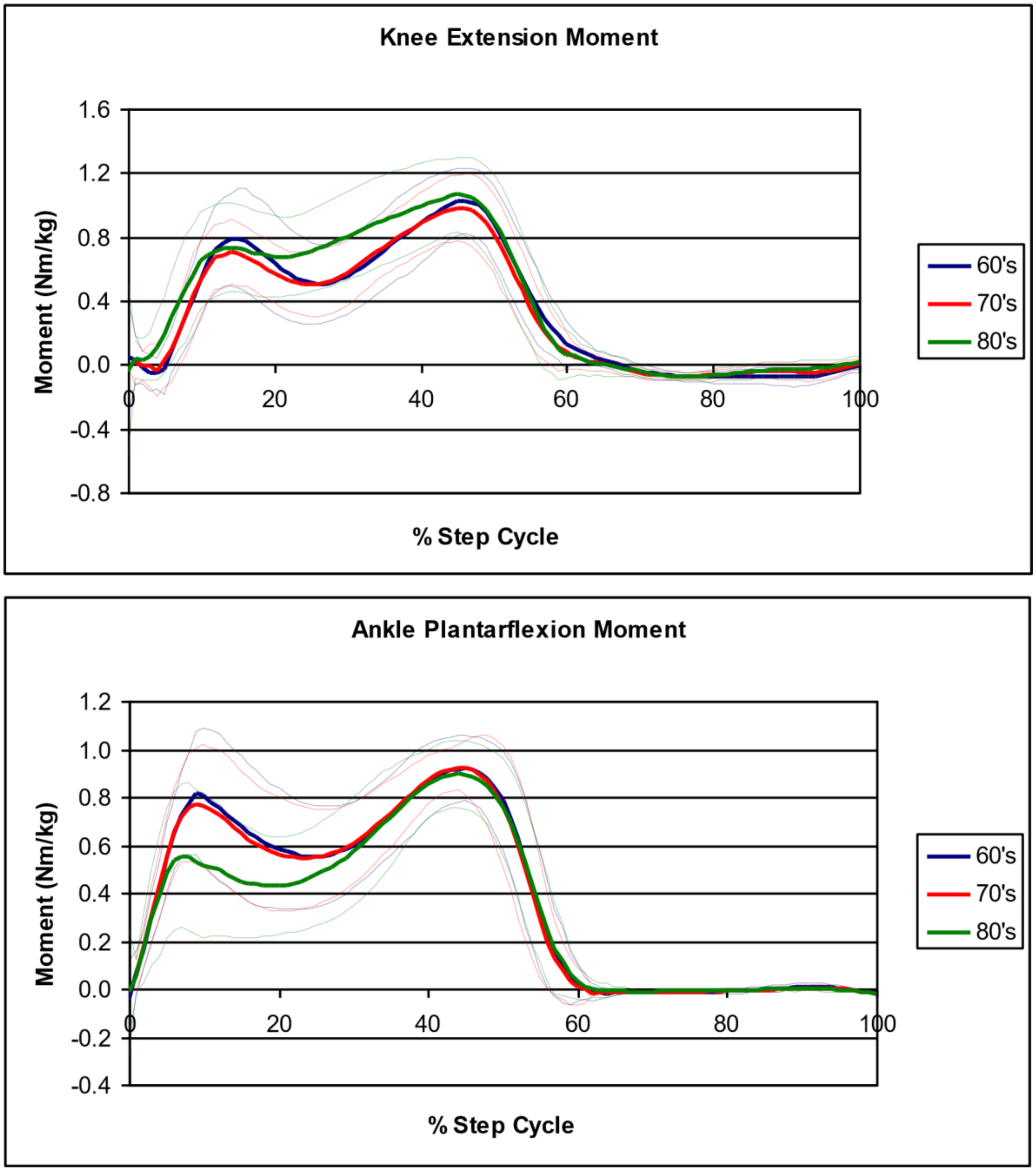


Figure 4.26 Moments produced around the knee and ankle joint during stair descent using hands. Foot strikes are at 0 and 100% of the step cycle. extension moment throughout stance, peaking at the point prior to toe off. Therefore the oldest group are producing a larger knee extension moment throughout stance than the younger groups. This increase in knee extensor moment may in part be due to the slightly more flexed knee position following initial contact but may be part of a more complex balancing of moments which will be considered in Chapter 5.

At the ankle, a plantarflexion moment is produced following foot contact and is present throughout stance phase. There is an initial peak in early stance at a similar point in time to the peak knee moment and then a second larger peak just prior to foot off.

Adults in their 80‟s had a reduced plantarflexion moment early in stance compared to the

younger groups but were able to produce the same peak moment prior to toe off.

The peak knee and ankle moments for each subject were not normally distributed and hence investigated further using non parametric statistical test. Mean result for each age group and gender are presented in figure 4.27 and table 4.20.

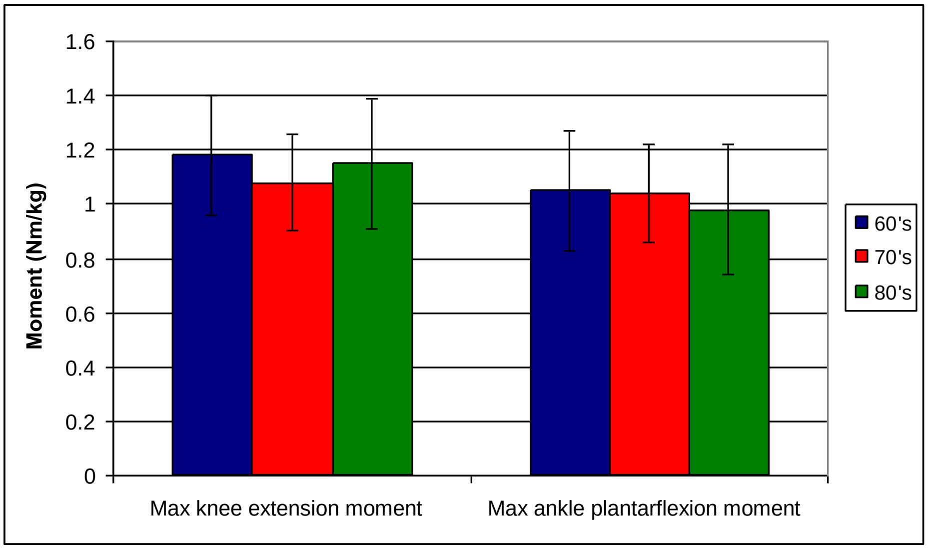


Figure 4.27 Maximum knee and ankle moments during stair descent with hands

		Age Group		
Moment	**Sex**	**60's**	**70's**	**80's**
Maximum knee extension moment (Nm/kg)	Male	1.23 (±0.21)	1.15 (±0.21)	1.23 (±0.27)
	Female	1.12 (±0.22)	1.01 (±0.12)	1.05 (±0.15)
	Combined	1.18 (±0.22)	1.08 (±0.18)	1.15 (±0.24)
Maximum ankle plantarflexion moment (Nm/kg)	Male	1.04 (±0.18)	1.08 (±0.12)	0.96 (±0.15)
	Female	1.06 (±0.14)	0.99 (±0.11)	0.99 (±0.14)
	Combined	1.05 (±0.16)	1.04 (±0.12)	0.98 (±0.14)

Table 4.20 Maximum knee and ankle moments during stair descent with hands

Statistically, there was no significant difference in the peak knee or ankle extension moment generated during stair descent between the three age groups. This may indicate theta there is a minimum requirement for the moment generated during stance phase to successfully descend the stairs. From the plots of kinetic data it could be seen that the older subjects did not develop a plantarflexion moment as rapidly in early stance. This resulted in a reduced first peak for the plantarflexion moment. The value of the first plantarflexion peak was obtained from the kinetic data and entered into SPSS for analysis. ANOVA determined there to be a reduced first peak in the 80 year old group compared to the younger groups but at a significance level of $p<0.1$.

4.6.8 Effect of handrail on kinetics during stair descent

Plots of the moments at the hip, knee and ankle were produced for subjects descending the stairs without use of the handrail, and a comparison of moments with and without a handrail is shown in figure 4.28. There was very little difference between the subjects in their 60‟s and 70‟s and the 70 year olds have been removed from this plot to aid clarity.

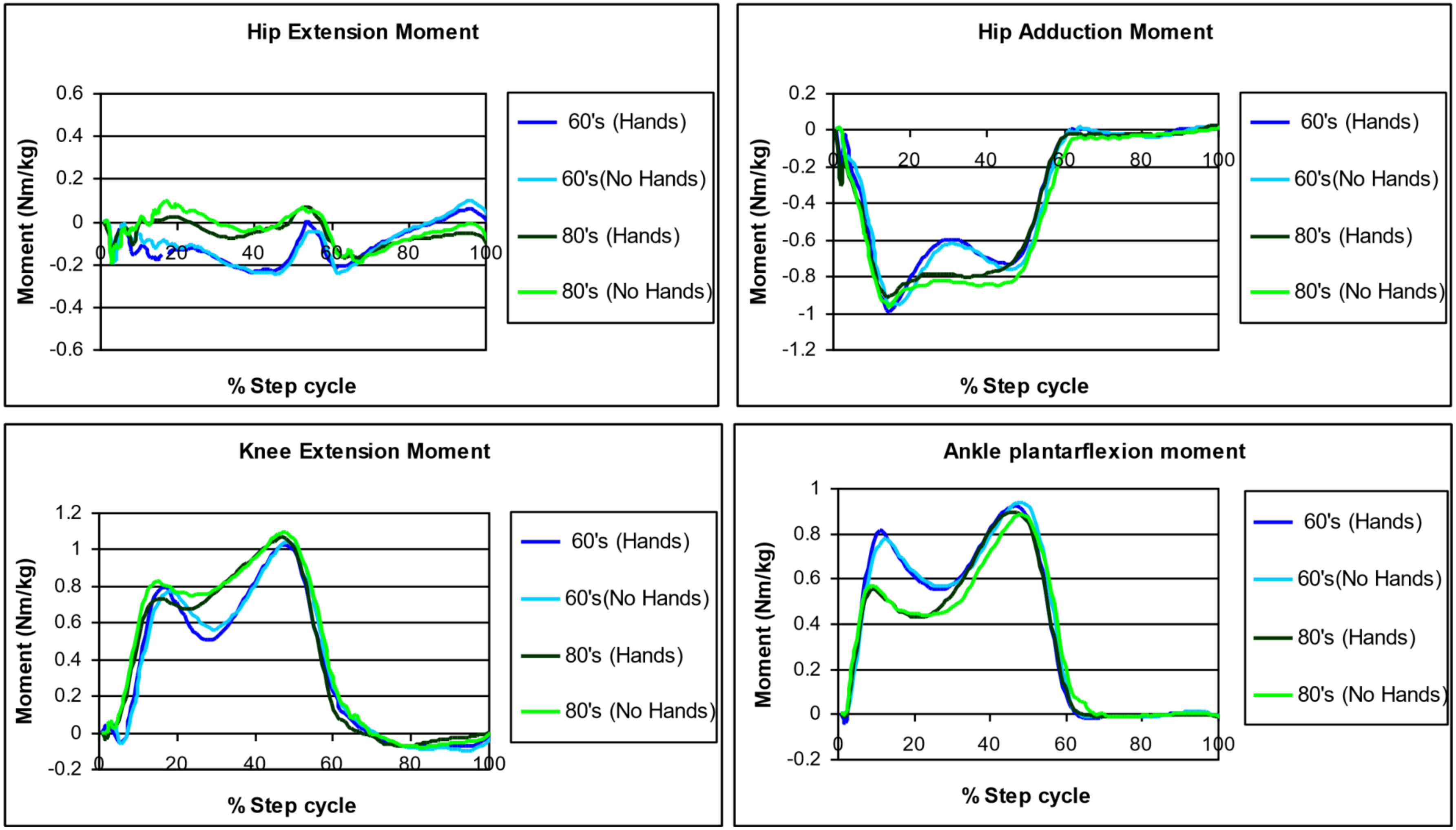


Figure 4.28 Moments at the hip, knee and ankle during stair descent with and without a handrail

The most noticeable changes to the kinematic plots when not using the handrail are seen in the hip and knee extensor moments in the first half of stance phase. Subjects in their 80‟s produce a larger hip extensor moment and knee extensor moment during early stance. This is seen to a lesser degree in the younger subjects. In both age groups there is an increase in hip abduction moment throughout stance phase. The first peak ankle plantarflexion moment is unchanged in the older subjects and slightly reduced in the younger, though the overall peak moment appears to be similar or slightly increased prior to toe off. Overall it would appear from studying the kinetic plots that the extra work involved in descending the stairs without a handrail is distributed mainly at the hip and knee joints and to a lesser degree at the ankle joint.

A paired t-test (or sign test for non-parametric data) was used to investigate the effect of handrail use on the peak ankle, knee and hip moments. There was a significant increase ($p<0.05$) in peak extension moments at the ankle, knee and hip with the absence of a handrail, and a significant increase in peak hip abduction moment (table 4.21 and figure 4.29).

Maximum Moment (Nm/kg)	**Handrail used**	**No Handrail**
Hip extension	0.36 (±0.20)	0.44 (±0.23)
Knee extension	1.14 (±0.22)	1.18 (±0.24)
Ankle plantarflexion	1.02 (±0.14)	1.06 (±0.16)
Hip abduction	1.10 (±0.21)	1.17 (±0.23)

Table 4.21 Handrail effects on maximum moments during stair descent

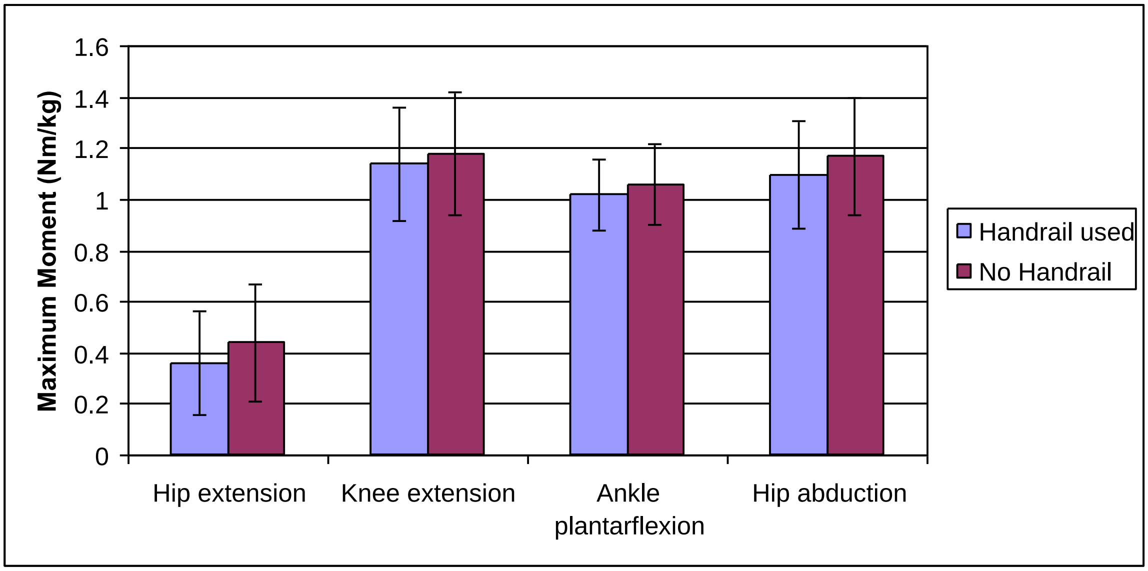


Figure 4.29 Handrail effects on maximum moments during stair descent

The increase in moments were an average of 0.08 Nm/kg for hip extension, 0.07 for hip abduction, 0.04 Nm/kg for knee extension and 0.04 Nm/kg for ankle plantarflexion. For an average 70 kg male this equates to an extra moment of 5.8 Nm for hip extension, 5 Nm for hip abduction, 2.9 Nm for knee extension and 2.9 Nm for ankle plantarflexion.

4.6.9 COM movement and velocity during stair descent with a handrail

As the subjects were walking forwards and down a fixed staircase, the COM was seen to progress forwards and downwards. The degree of excursion in the anterior/posterior and vertical directions represented the size of the staircase and did not demonstrate any changes due to ageing. The excursion of the COM in the mediolateral (ML) direction during one step cycle is shown in table 4.22

		Age Group		
COM excursion	**Sex**	**60’s**	**70’s**	**80’s**
Mediolateral COM Excursion(mm)	Male	60.8(±16.7)	56.2 (±14.2)	75.2 (±30.4)
	Female	58.1(±20.2)	49.2(±19.2)	60.6(±11.0)
	Combined	59.4(±18.3)	52.7(±16.9)	68.5(±24.3)

Table 4.22 COM excursion during stair descent with a handrail

2-way ANOVA found that there was a significant increase in the excursion of the COM in the ML direction with age ($p<0.05$). Post-hoc testing revealed this to be between the 60‟s and 80‟s and between the 70‟s and 80‟s but not between the 60‟s and 70‟s. There was no significant change in COM excursion related to gender.

The velocity of the centre of mass in three orthogonal directions was studied to determine if and how older adults adopted strategies to reduce COM motion. Plots of COM velocity for a whole gait cycle are shown in figure 4.30 and the maximum velocities used are shown in table 4.23.

		Age Group		
Maximum COM velocity	**Sex**	**60’s**	**70’s**	**80’s**
A-P (m/s)	Male	-0.95(±0.20)	-0.84(±0.18)	-0.68(±0.20)
	Female	-1.02(±0.23)	-0.80(±0.21)	-0.65(±0.26)
	Combined	-0.99(±0.22)	-0.82(±0.19)	-0.67(±0.22)
M-L (m/s)	Male	0.20(±0.04)	0.19(±0.05)	0.21(±0.06)
	Female	0.21(±0.07)	0.16(±0.05)	0.19(±0.05)
	Combined	0.21(±0.06)	0.17(±0.05)	0.20(±0.05)
Vertical (m/s)	Male	-0.63 (±0.08)	-0.60(±0.08)	-0.55(±0.09)
	Female	-0.64(±0.10)	-0.57(±0.11)	-0.53(±0.07)
	Combined	-0.63(±0.09)	-0.58(±0.10)	-0.54(±0.08)

Table 4.23 Maximum COM velocity during stair descent with a handrail

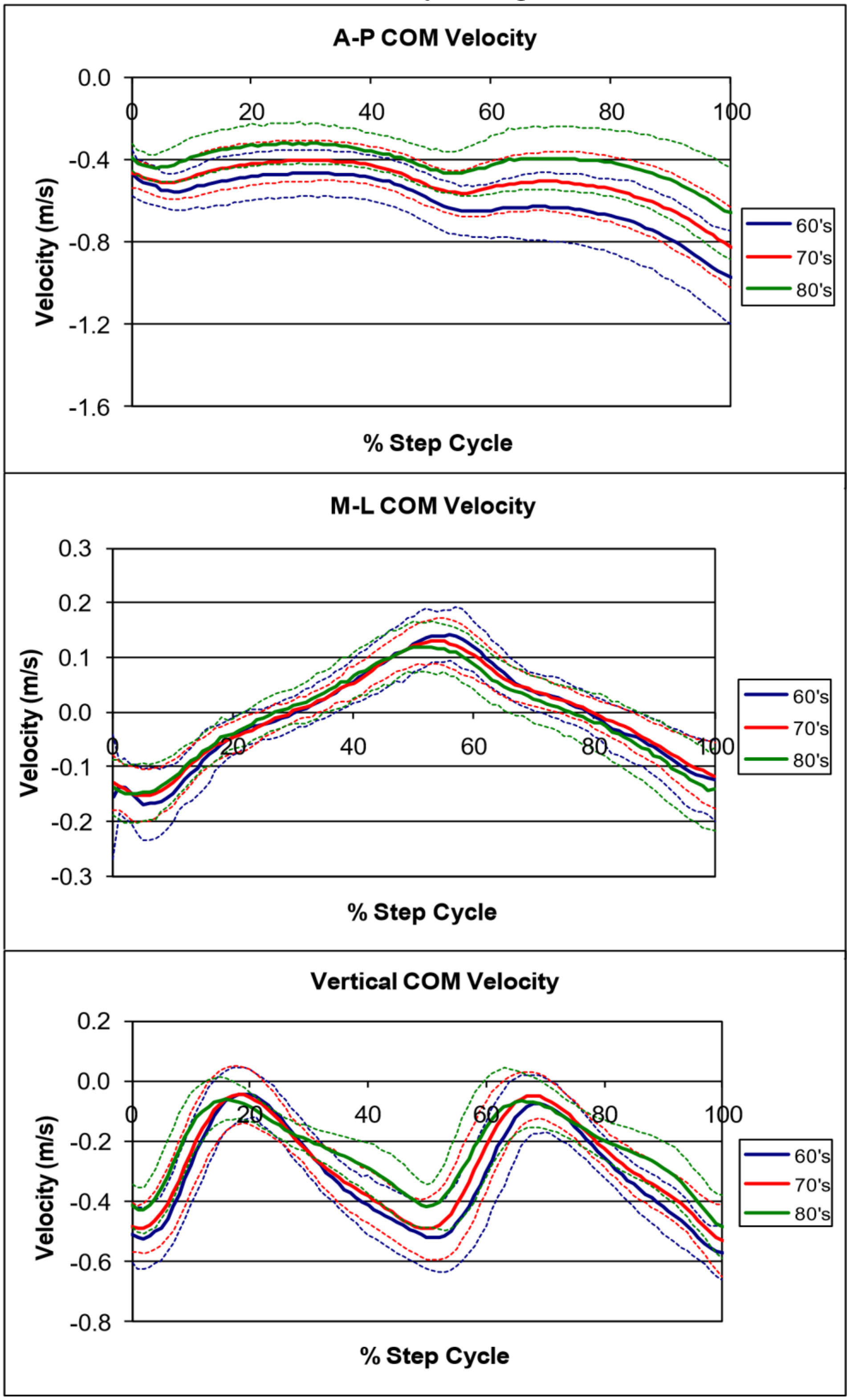


Figure 4.30 COM velocity during stair descent. Foot strikes are at 0 and 100% of cycle

In the A-P direction the velocity of the COM was negative as the orientation of the stairs in the laboratory was with the horizontal axis from positive to negative in the horizontal direction of progression for stair ascent. The COM velocity increased as the subjects continued down the stairs, possibly implying that the subjects increased their speed overall as they descended the stairs. There were peak areas of COM velocity during early stance. A 2-way ANOVA found there to be no gender related effects on AP velocity but there were age related effects. Post hoc analysis found there to be a significant decrease in the AP COM velocity with each decade ($p<0.05$).

In the M-L direction the COM increases in speed towards the weight bearing foot following initial contact on the step. This is followed by a period of slowing down and then as the opposite foot contacts the step COM velocity increases towards that side. ANOVA revealed no significant age or gender related changes in M-L velocity.

In the vertical direction the COM velocity is negative as the z axis of the lab was positive vertically upwards. Following initial contact the vertical velocity of the COM reduces during stance then increases as the opposite limb is lowered to the next step. This corresponds to the phases of single support. 2-way ANOVA found that there were no gender effects but there was a decrease in the maximum vertical COM velocity with age. This effect was significant between the 60‟s and the 80‟s groups only.

In section 4.6.2 it was reported that the speed of stair ascent was similar in the 60‟s and 70‟s and decreased in the 80 year old group. The COM velocity has a similar pattern and may be explained simply by the changes in the speed of stair climbing.

4.6.10 Effect of handrail use on COM movement

The mean excursion of the COM in a mediolateral direction with and without handrail use whilst descending the stairs is shown in Table 4.24.

	Mediolateral COM excursion (mm)	
Age Group	**Handrail used**	**No Handrail**
60‘‘s	59.4 (±18.3)	63.3(±19.5)
70‘‘s	52.7(±16.9)	61.0(±16.6)
80‘‘s	68.5(±24.3)	93.7(±32.9)
Mean overall	59.6(±20.6)	70.4(±26.5)

Table 4.24 COM excursion during stair descent with and without a handrail

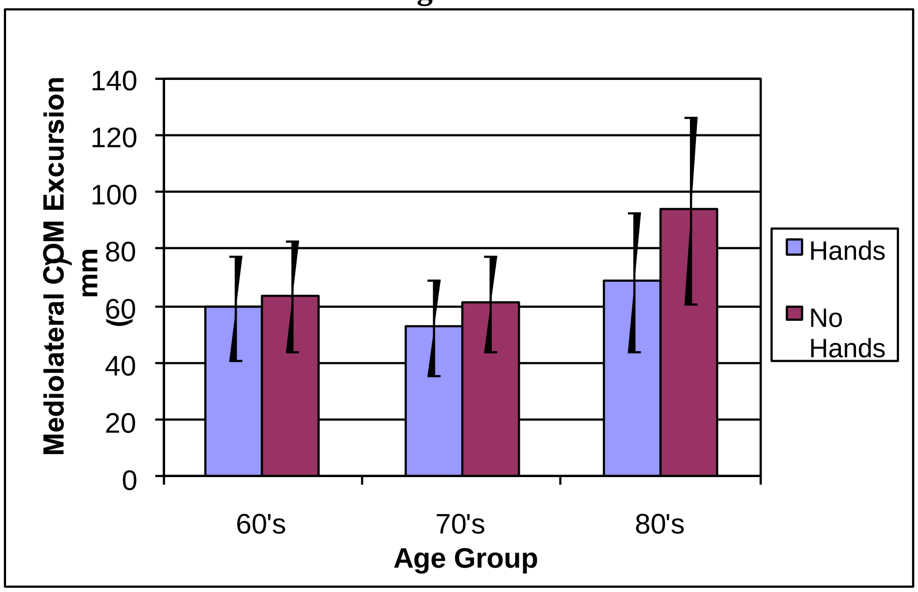


Figure 4.31 Handrail effects on mediolateral COM excursion during stair descent

A 2-way ANOVA was performed to investigate the effects of age and handrail use on the mediolateral COM movement during stair descent, as it was felt that problems with control of the COM excursion may impact performance. There was a significant increase in the amount of excursion of the COM when the handrail was not used ($p<0.05$), from an average of 59.6mm to 70.4mm. There was also a significant increase in the amount of excursion demonstrated in the 80+ group compared to the younger groups when not using a handrail as shown in figure 4.31. There was interaction between handrail use and age in the ANOVA, suggesting that the oldest group are more affected by the loss of the handrail.

4.6.11 Summary of stair descent

All subjects were able to descend the stairs whilst using a handrail. The rate of stair descent was significantly slower in the 80-year-old group but there was no change found in the time spent in stance or double support. The decline in velocity related to a decline in velocity of the COM in the A-P and vertical direction in the oldest age group.

Joint range of motion in the hips, knees and ankles were similar between the three ages with a maximum hip flexion of around 45 degrees, knee flexion of 99 degrees, ankle dorsiflexion of 36 degrees and plantarflexion of 29 degrees. The demands are different to those of stair climbing, but also much higher than in normal walking and this is discussed in Chapter 5.

Some changes in joint kinetics were observed between the age groups. The oldest age group had a reduced ankle plantarflexion moment in early to mid-stance, which seems to be compensated by increasing hip and knee extensor moments during this phase of gait.

Descending stairs without a handrail was found to be a more challenging activity, and seven subjects were unable to perform this at all, all of whom were in the older two age groups. Subjects descended steps more quickly without the handrail but adopted their gait pattern to significantly increase the proportion of time spent in stance phase and in double support. There was little effect on joint kinematics when descending stairs without use of hands but there was a significant increase in the joint moments required to extend the hip, knee and ankle. This increase in joint moments required was present in all age groups but was largest for subjects in the 80-year-old group. The COM excursion in a mediolateral direction was significantly higher when not using a handrail and more noticeably again in the oldest age group.

4.7 Comparison of stair ascent and descent

Table 4.25 summarises the temporal, kinematic and kinetic demands of stair ascent and descent without a handrail to enable comparison of the two activities. This is the mean across all age groups (n=77 subjects for stair ascent, n=75 subjects for stair descent).

Participants were slower on average during stair ascent than stair descent. Greater ankle ROM was required for stair descent than ascent, but at the knee and hip greater ROM was required for stair ascent. Hip and ankle extensor moments were higher for stair ascent compared to stair descent, whereas hip abduction moments were greatest during descent.

As more of the participants had difficulty with stair descent than ascent these findings are discussed in Chapter 5.

		Stair Ascent	**Stair Descent**
Temporal	Time per step cycle (s)	1.43 (±0.22)	1.30 (±0.28)
	Rate (steps/min)	86.2 (±13.2)	96.6 (±19.8)

	Stance phase (% of cycle)	60.9 (±3.0)	60.7 (±3.1)
	Swing phase (% of cycle)	39.1 (±3.0)	39.3 (±3.1)
	Double support (% of cycle)	25.7 (±3.8)	20.4 (±4.5)
Kinematic	Maximum ankle dorsiflexion (degrees)	20.5 (±4.0)	33.1 (±6.5)
	Maximum ankle plantarflexion (degrees)	18.1 (±7.1)	27.4 (±6.4)
	Maximum knee flexion (degrees)	99.4 (±6.6)	96.0 (±5.8)
	Maximum hip flexion (degrees)	70.8 (±8.6)	40.2 (±8.8)
Kinetic	Maximum ankle plantarflexion moment (Nm/kg)	1.21 (±0.17)	1.06 (±0.16)
	Maximum knee extension moment (Nm/kg)	1.14 (±0.25)	1.18 (±0.24)
	Maximum hip extension moment (Nm/kg)	0.85 (±0.31)	0.44 (±0.23)
	Maximum hip abduction moment (Nm/kg)	0.91 (±0.14)	1.17 (±0.23)

Table 4.25 Comparison of mean values for all ages during stair ascent and descent

4.8 Comparison with physical assessment

As part of the EQUAL project, a physical assessment was performed by a physiotherapist, and reported in his PhD Thesis (Samuel, 2005). A measurement of joint ROM was performed at the hips (flexion, extension and abduction), knees (flexion and extension) and wrists (flexion and extension). A custom built isometric dynamometer was constructed in the Bioengineering Unit at the University of Strathclyde and Samuel

(2005) undertook assessment of the maximum isometric muscle strength in the following muscle groups and positions:

- Knee extension strength at 20°, 60° and 90° of knee flexion
- Knee flexor strength at 20°, 60° and 90° of knee flexion
- Hip extensor strength at 0°, 30° and 45° of hip flexion
- Hip flexor strength at 0°, 30° and 45° of hip flexion
- Hip abduction strength at 0° of hip flexion

This section presents the maximum strength data obtained by Samuel (2005) and compares it to the findings from the biomechanical analysis of stair ascent and descent. Samuel reported the produced moments in Nm and did not normalise these results to body mass as had been done in the biomechanical analysis. In order to allow comparison the biomechanical data has been multiplied by the mean body mass for each age group to produce a moment in Nm. Maximum isometric knee extensor strength was found to be produced in the testing position of 60° of knee flexion and maximum isometric hip extensor strength was found to be produced at 45° of hip flexion (Samuel,2005). These testing positions will be used in this comparison, along with hip abduction at 0° of hip flexion.

The peak isometric knee extensor moments produced on the dynamometer are compared with the peak knee extensor moments during stair ascent and descent in table 4.26.

Gender	**Age Group**	**Maximum isometric knee extensor moment (Nm) (Samuel 2005)**	**Peak knee extensor moment during ascent (Nm)**	**Peak knee extensor moment during descent (Nm)**
Males	60‟s	100.3	102.5	94.8

	70"s	88.4	86.8	86.8
	80"s	75.7	85.9	100.7
Females	60"s	54.1	69.9	81.5
	70"s	54.5	60.8	69.7
	80"s	47.7	63.6	66.2

Table 4.26 Isometric knee extensor strength compared with peak moments produced

Samuel (2005) found that there was an age related decline in the maximum knee extensor moment produced but that this decrease was not statistically significant. Female subjects had a significantly reduced knee extensor moments compared to the male subjects. For nearly all cases the peak knee extensor moment produced during the activity is higher than the peak isometric knee extensor moment produced on the dynamometer. Potential reasons for this will be discussed in Chapter 5.

Peak hip extensor moments from Samuel (2005) are compared with those during activity in table 4.27.

Gender	Age Group	Maximum isometric hip extensor moment (Nm) (Samuel 2005)	Peak hip extensor moment during ascent (Nm)	Peak hip extensor moment during descent (Nm)
Males	60"s	100.3	65.5	30.1
	70"s	88.4	62.7	23.4
	80"s	75.7	67.9	32.8
Females	60"s	54.1	54.6	28.4
	70"s	54.5	47.0	23.4
	80"s	47.7	39.7	18.9

Table 4.27 Isometric hip extensor strength compared with peak moments produced

Samuel (2005) reported a decline in hip extensor strength with each increasing decade but this was not statistically significant. There was a significant difference between the genders with female subjects having approximately 53-63% less strength than the male subjects. During stair ascent the peak hip extensor moments produced are lower than the maximum isometric moment for male subjects. The female subjects in their 60‟s have peak hip extensor moments greater than the isometric moment produced.

Comparison of the peak isometric hip abduction moment obtained from the dynamometer is compared the hip abdution moment during stair ascent and descent in table 4.28. The hip adductor moments produced during activity were higher than the maximum hip abduction produced isometrically on the dynamometer.

Gender	Age Group	Maximum isometric hip abductor moment (Nm) (Samuel 2005)	Peak hip abductor moment during ascent (Nm)	Peak hip abductor moment during descent (Nm)
Males	60‟s	61.1	64.0	84.8
	70‟s	53.1	64.9	76.2
	80‟s	45.0	79.4	88.5
Females	60‟s	33.8	64.8	85.1
	70‟s	31.0	61.5	77.3
	80‟s	24.4	58.0	71.2

Table 4.28 Isometric hip abduction strength compared with peak moments produced

Joint ROM measured at the hip and knee by Samuel (2005) is presented in table 4.29.

		Joint ROM (degrees)				
Gender	**Age Group**	**Knee extension**	**Knee flexion**	**Hip extension**	**Hip flexion**	**Hip abduction**
Males	60‟s	0.1	128.0	15.8	109.0	40.6
	70‟s	0.0	124.3	13.9	104.7	36.2

	80‟s	-0.3	117.4	13.0	97.4	29.5
Females	60‟s	1.0	119.9	12.3	108.0	35.8
	70‟s	-0.2	120.5	12.7	107.2	38.8
	80‟s	0.2	119.2	12.5	104.4	31.7

Table 4.29 Joint ROM at the hip and knee measured by Samuel (2005)

The amount of joint ROM used during stair ascent and descent at the hip and knee has been presented in Tables 4.7, 4.8, 4.17 and 4.18. The older adults in this study had around 20-30^{o} more flexion at the knee than was required for stair ascent and descent and over 30 degrees more hip flexion than was needed. Participants had very close to full knee extension on physical assessment, which was more than what was necessary for stair ascent and descent. Hip extension and abduction were much greater than what was required for both stair ascent and descent.

In summary, the kinematic demands of stair ascent and descent appear to be well within the joint ROM available to the older adults in this study. The kinetic demands of stair ascent and descent are higher in many cases than the peak isometric joint moments measured by Samuel (2005). These findings are discussed in Chapter 5.

Chapter 5

Discussion

This chapter discusses the results found from the laboratory work. It investigates further some of the age related changes presented in Chapter 4 and explains why these may occur, considering the kinetics and kinematics. The results are compared with previous studies on stair ascent and descent in both young and older adults. The differences between stair ascent and descent are explored to see if there are biomechanical factors that relate to older adults experiencing reluctance to descend stairs in this study. The implications of the findings of this study are discussed in the context of how health professionals dealing with older adults approach rehabilitation to assist with improving both stair climbing and descent. Finally, the limitations of this study and the areas in which further research would be beneficial are presented.

5.1 Ability to ascend and descend stairs

This study investigated healthy adults aged over 60. In order to reduce the risk of injury to participants and ensure that there was a minimal risk of falls, subjects were screened prior to participation and excluded if they had health problems that may put them at risk. This procedure is similar to many previous biomechanical studies (Livington, 1996; Christina and Cavanagh, 2002; Hortogabayi, 2003; Reeves et al., 2008[a+b]) and was considered a necessary safety requirement for this study. By controlling for health problems it also allows more reliably exploration of some of the biomechanical changes resulting from ageing, rather than the effects of diseases that tend to be age related. The subjects who participated could be considered to be healthier and more able than a typical older adult in that age range. Samuel (2005), as part of the EQUAL project, asked participants in this study to complete a questionnaire SF-36, which is widely used to assess health outcomes (Ware and Sherborne, 1992). One section of this questionnaire assesses physical function providing a score out of a maximum of 100. Participants in their 60"s averaged 87.9, those in their 70"s averaged 86.7 and in their 80"s averaged 67.5. A large study of nearly 10,000 community dwelling older adults (Walters et al, 2001) reported physical functioning scores from the SF-36 as

61.9 for those in their 60"s, 55.9 for those in their 70"s and 36.6 for those in their 80"s. The subjects participating in the EQUAL project has physical functioning scores on average at least 20 points higher than their peers, and in the older age groups they were even more capable that their average peers. This shall be taken into consideration when looking at the results.

Of the 84 subjects assessed, all were able to ascend and descend the stairs using a handrail using the normal stair gait pattern of one foot on one step then one foot on the next step. However, even in this very healthy and active group, without use of the handrail, 5 subjects felt unable to ascend the stairs and 7 were unable to descend the stairs. The details of these subjects are presented in Section 4.4. It was found that 20% of the females in their 70"s were unable to ascend or descend stairs without the handrail, 18% of the females in their 80‟s could not ascend and 36% could not descend without the handrail and just 1 (or 8%) of the men in their 80‟s could not ascend or descend the steps. These findings relate well to those of Reeves et al. (2008b), who asked their older participants who were confident on the stairs if they used the handrail when walking up or down the stairs. Of the eleven participants, six rarely or never used the handrail when ascending the stairs but only two rarely or never used the handrail when descending the stairs. Accidents during stair descent are more common than during ascent at a ratio of three to one (Startzell et al, 2000), so the sense of needing to use the handrail for stair descent may relate to issues of confidence as well as the changing biomechanical demands presented in section 5.5.

The ability of all the participants to ascend and descend stairs concurs with previous findings in older adults. In their study of 310 community dwelling older adults Verghese et al. (2008) found that 140 reported difficulties ascending stairs and 83 reported difficulty descending stairs. Although these subjects had difficulties they were able to perform the activity and other studies have found many older adults continue to climb stairs as frequently as younger adults as it is a necessary part of them maintaining independence (Startzell et al, 2000).

5.2 Age related changes in stair ascent

5.2.1 Temporal changes

In the current study it was observed that during stair ascent (either with or without a handrail), subjects in the oldest age group had a significantly reduced step rate, and spent an increased amount of time in stance phase and double support. Table 5.1 compares these finding to those of previous studies of stair ascent without using a handrail.

Table 5.1 Comparison of temporal data for stair ascent

	Current study			Costigan et al. (2002)	Nadeau et al. (2003)	Reeves et al. (2009)		Novac and Brouwer (2011)	
Age	**60's**	**70's**	**80's**	**Mean 24.6**	**41-70**	**Mean 24.6**	**Mean 73.4**	**Mean 23.7**	**Mean 67.0**
Number	30	30	24	35	11	17	15	23	32
Time per step cycle (s)	1.35 (±0.2)	1.39 (±0.18)	1.59 (±0.22)	1.49 (±0.16)	1.30 (±0.18)				
Rate of stair ascent (steps/min)	90.6 (±13.0)	88.0 (±12.0)	76.8 (±10.4)	88 (±10)	93.6 (±12.8)	98 (±13)	92 (±10)	102.5 (±8.9)	94.8 (±13.0)
Stance phase (% of cycle)	60.2 (±2.7)	60.1 (±2.8)	63.2 (±2.6)	68 (±0.3)	60.3 (±1.1)	63 (±2)	64 (±3)		
Swing phase (% of cycle)	39.8 (±2.7)	39.9 (±2.8)	37.8 (±2.6)	32 (±0.3)	39.7 (±1.1)	37 (±2)	36 (±3)		
Double support (% of cycle)	24.3 (±3.4)	25.4 (±4.1)	28.4 (±2.5)		24.9 (±2.1)	27 (±2)	29 (±3)		

The results of this study demonstrate that these subjects were performing the task of stair climbing at a slightly slower rate that previous studies looking at older adults (Nadeau et al, 2003; Reeves et al, 2009; Novak & Brouwer, 2011). Costigan et al.(2002) reported slower step rates in their group of younger adults, and Riener et al. (2002) reported a similar cycle duration to this study (1.40±0.10 s) with healthy young male subjects. Some of this variation between studies may be explained by the laboratory set up. Costigan et al. (2002) used a stair rise of 200mm, Riener et al. (2002), Nadeau et al. (2003) and Reeves et al. (2009) used 170mm, Novak and Brouwer (2011) used 150mm, and the current study used 185mm. A deeper step will require greater joint excursion and subjects may need longer to achieve this. The step height in the current study was selected in line with the Building Regulations of 2000, which advises that in public places the rise should be no more than 190mm and in private dwellings no more than 220mm. The number of steps in the staircase may also have affected the speed of progression. Costigan et al. (2002) only had 2 steps for subjects to negotiate which may have resulted in the need to slow down towards the end of the trial. Nadeau et al. (2003), Reeves et al. (2009), Novak and Brouwer (2011) used a staircase with 4 steps and an upper platform, similar to the current study, which may enable subjects to adopt a more natural stair climbing pattern, and hence be a little faster. Recent work by Cluff and Robertson (2011), investigating stair descent found that a minimum of five stairs were required to reach a steady state of stair descent. Similar work has not been undertaken for stair ascent at this time but it is certainly a factor that requires consideration.

Many studies investigating gait have found a decline in gait velocity with increasing age (Prince et al, 1997; Winter et al, 1990, McGibbon and Krebs, 1999) when walking on the flat and this study indicates that a similar occurrence happens when walking up stairs. Subjects in their 80‟s in this study were significantly slower than subjects in their 60‟s and 70‟s when walking up the stairs. This decrease in velocity may be related to a need to reduce energy expenditure during stair climbing or as a result of the decrease in the ability of the muscle to contract at the required velocity, both issues known to change with age as discussed in Chapter 1. It could be that the older subjects were slower as they were trying to decrease the power required by the muscle, or as a result of decreased confidence. Reeves et al. (2009) reported a decrease in cadence in their older adult group (mean age 73.4) compared to their young adults (mean age 24.6), however this was not significant at the $p<0.05$ level. Brouwer and Novak (2011) reported a significant decrease in cadence in their older subjects (mean age 67.0) compared to the younger subjects (mean age 23.7). The current study had a larger sample size than Reeves et al. (2009) (see table 4.1) and it may have been that Reeves et al.(2009) did not have sufficient subject numbers to provide statistical power.

At this time there are no other known studies that have looked at the effects of increasing old age on the speed of stair performance. In this study the changes in cadence were only observed in the oldest group which may suggest that healthy adults in their

60‟s and 70‟s are able to walk at a speed comparable to younger subjects, and that it is only with increasing age that adaptations in speed are required. Alternatively, there may be some slowing in stair climbing with age and these changes are accelerated as people enter their 80‟s

The subjects in this study adopted a more stable approach to stair ascent in the oldest age group by increasing the amount of time spent in stance and in double support. In the only comparable study investigating stair ascent, Reeves et al. (2008) did not find a significant difference between their young and older group. The increase in double support in the current study was only seen in the 80+ age group and it may be that some of the strategies to make the gait pattern more stable were not required in the younger subjects. Winter et al. (1990) observed changes in the amount of time spent in stance in level walking in a group of fifteen fit elderly subjects (mean age=68) when compared to younger subjects. They reported an increase in stance phase duration from 62.3% in the young to 65.5% in the elderly subjects. The authors summarised that this adaptation related to the need for a "safer and less destabilizing gait pattern" in the older subjects. In this study the increase in the stance phase was approximately 3% from the 60‟s to the 80‟s. It would appear that the oldest group are adopting not only a slower method of stair ascent, but one that is more stable and therefore potentially safer.

5.2.2. Kinematic changes

The kinematics of stair ascent were similar between the three age groups with no significant differences between the peak angles required to perform the activity, apart from knee extension where the 80+ group did not extend their knees as fully during stance phase. The amount of hip flexion used was around 70 degrees, knee flexion was 98 degrees, ankle dorsiflexion was 21 degrees and plantarflexion, 19 degrees. The kinematic plots compare well with those of McFayden and Winter (1998) and Riener et al. (2002) who were investigating young adults. Table 5.2 compares kinematic data from previous studies to the present trial.

	Current study			**Nadeau et al. (2003)**	**Reeves et al. (2009)**	
Age	**60's**	**70's**	**80's**	**Age 41-70**	**Mean 24.6**	**Mean 73.4**
Number of subjects	n=30	n=30	n=24	n=11	n=17	n=15
Maximum Hip Flexion (degrees)	68.9 (±8.1)	68.8 (±8.7)	70.1 (±9.7)	60.1 (±5.6)		
Minimum Hip Flexion (degrees)	8.7 (±8.1)	7.4 (±8.3)	11.7 (±10.3)	4.7 (±6.5)		
Maximum Knee Flexion (degrees)	96.1 (±6.7)	97.4 (±6.5)	98.0 (±5.8)	93.1 (±3.1)	94.2 (±7.8)	95.6 (±4.8)
Minimum Knee Flexion (degrees)	7.9 (±6.1)	8.8 (±5.6)	12.3 (±5.9)	10.0 (±2.7)	14.5 (±4.7)	16 (±5.6)
Maximum Ankle Plantarflexion (degrees)	19.4 (±7.9)	18.4 (±6.8)	16.8 (±6.1)	9.4 (±6.0)	9.4 (±9.5)	10.6 (±4.8)
Maximum Ankle Dorsiflexion (degrees)	20.6 (±4.9)	20.7 (±4.3)	19.6 (±4.5)	29.8 (±2.8)	23.4 (±4.9)	21.5 (±3.9)

Table 5.2 Peak angles during stair ascent

5.2.3 Hip and knee kinematics

Comparing the results of this study to those of Nadeau et al. (2003) and Reeves et al. (2008) there is an increase in the amount of hip and knee flexion used during stair ascent in the current study. Some of this difference will be due to the current study using a step height that was 15mm larger than the one used by these authors and also that the subjects in the current study were on average slightly shorter. The majority of the compensation is in the hip with an increase in hip flexion of approximately 10 degrees and a corresponding difference in knee flexion of 3-4 degrees.

Subjects in the current study did not extend their hips at terminal stance as much as younger subjects in the study by Nadeau et al. (2003). This may indicate a trend to a more flexed hip gait in the older subjects. At this time there are no other studies which compare hip kinematics in stair ascent between young and elderly subjects. This increase in hip flexion was observed in this study by the subjects adopting a position of increased anterior pelvic tilt and associated trunk forward leaning. This has been reported in gait by Kerrigan et al. (1998) in their study of 31 young and 31 elderly subjects, who found the older adults had approximately 6.5° loss of hip extension at terminal stance and an associated increase in pelvic tilt. Stair ascent does not use the same amount of hip extension as gait, typically the hip remains flexed to approximately 8° in stair climbing compare to extending to 15-20° in gait. The reason for the more flexed hip position therefore, does not seem to relate to a loss of hip extension ROM as all subjects were able to walk unaided on the level. Samuel (2005) recorded joint ROM of the subjects in the EQUAL project and reported that mean hip extension was 13.4°, so the subjects were operating well within their available ROM. Kerrigan et al. (1998) hypothesised that the increased pelvic tilt during gait in older adults may be due to subtle hip flexion contractures that cannot be detected by standard clinical testing. It seems that this would be unlikely in this case as subjects had approximately 20 degrees more extension than was measured in the static position and were able to utilise some additional range during gait, which does not suggest a fixed degree of contracture.

Another theory proposed by DeVita and Hortobaygi (2000) is that older subjects adopt a more flexed hip position in order to assist the hip muscle to generate greater hip moment due to a stretch being placed on the muscle. This theory may well be relevant here as a redistribution of joint moments was observed in the oldest age group, who adopted the most flexed position, but who also had an increase in hip extensor moment through stance. Muscle force production is greatest in the mid-range of the muscle and is less in inner range (Sherwood, 2008). By coming away from the inner range of the hip extensors subjects may be facilitating better strength generation. This is reinforced by the findings of Samuel (2005) who found that in these subjects the peak hip extensor moments were greatest isometrically when tested in a position of 45° flexion, then 30° and were lowest when tested at 0°.

An alternative suggestion for the more flexed hip position may be that the subjects are having to spend a greater time observing foot placement on the step due to deterioration in confidence or sensory feedback. In order to look at the step the subjects may be flexing the neck and trunk which would result in the need for a more flexed hip position.

As the vision of these participants was not studied, it is not possible to draw further conclusions, but it is an area that would be of interest for future research.

5.2.4 Ankle kinematics

Another difference in kinematics from between this study and the previous studies in Table 5.1 is the amount of ankle dorsiflexion and plantarflexion used. Peak dorsiflexion and plantarflexion differs between studies, with subjects in the current study having a more plantarflexed foot position at toe off and reduced peak dorsiflexion. However the total amount of excursion at the ankle is relatively similar between studies, being approximately 40° in subjects in the current study and Nadeau et al. (2003) and 33° in Reeves et al. (2009). The differences between studies may be due to differences in protocol for marker placement on the shoes, and sufficient detail is not available in the previous studies. The marker arrangement in this study was such that the markers were placed along the shoe parallel to how the foot was positioned in the shoe. Therefore if a subject was to have a higher heel then the foot would appear slightly plantarflexed in normal standing. The participants in this study wore their own shoes and some of the participants did have small heels on the shoe. To determine ankle movement optimally it would be necessary to be barefoot on the stairs but this poses problems with older adults in terms of safety and comfort.

It is important to consider that some of the variation between studies may be due to differences in the biomechanical methods used to determine the kinematics. Nadeau et al. (2003) do not specify how they determined the joint centres, but do state that they used a Cardanic x-y-z sequence to determine angles such that the local x,y and z axes corresponded respectively to add/adduction, rotation and flexion/extension. This differs from the current study where a floating axis technique was used. Reeves et al. (2009) used the "plug in gait model" produced by VICON to compute the kinematics and kinetics. The Newington - Gage model (Davis et al., 1991) is used to define the positions of the hip joint centre in the pelvis segment, which is different from the current study where the method of Bell et al. (1989) was used. The "plug in gait model" uses a Cardanic y-x-z sequence, which involves rotating around the flex/extension axis first, then the abd/adduction axis then rotation. It is impossible to know the exact differences between the Cardanic method and the floating axis method and the changes in kinematics due to slightly different locations of joint centres. Fioretti et al (1997) compared the floating axis method to Cardanic rotations computed at the knee during gait and did not find a significant difference in the angles produced. Stagni et al. (2000) investigated the effects of hip joint centre mislocation on hip angles during gait and reported changes in knee and hip joint angle were negligible (1-2°) with a change in hip location of 30mm. It is therefore conceivable that there may be small changes due to different methods but these are likely to be of the order of a couple of degrees only.

5.2.5 Summary of kinematic changes

Although there may be small differences in peak angles there is overall consistency in the kinematic plots from this and previous studies suggesting that there is a minimum joint excursion required for normal reciprocal stair climbing. The exact requirements will depend on the geometry of the staircase and the size of the person performing the activity, with greater ROM required for steeper stairs or shorter people. The degree of joint excursion is considerable higher than that of normal gait, which is approximately 30 degrees of hip flexion, 45 degrees of knee flexion and 10 degrees of dorsiflexion (Whittle, 1996; Winter et al, 1990). This would explain why patients with joint limitations such as arthritis, especially in the knee, experience difficulty with stair ascent (Verghese et al, 2008). Knowledge of the joint requirements for normal stair climbing is of use to health professionals rehabilitating older adults experiencing stair climbing difficulties. If sufficient range of movement is not available then alternative strategies for stair ascent may be necessary.

5.2.6 Kinetic changes

The plots for the joint kinetics at the hip, knee and ankles are shown in figure 4.11 and figure 4.14. Subjects in their 60‟s and 70‟s had very similar joint kinetic plots and no statistical changes could be found between them. Differences were only observed with the oldest age group who had the following changes compared to the younger groups:

1. A decrease in plantarflexor moment throughout stance phase with an associated decreased peak plantarflexor moment at toe off.
2. An increase in hip and knee extensor moment, and hip abduction moments, in mid to late stance, though peak moments were not statistically different.

Before exploring some of the reasons for these changes it is useful to compare the kinetic data to the previous literature. Comparisons of the kinetic findings of this study and those previously reported in the literature are shown in table 5.3.

Table 5.3 Peak joint moments during stair ascent without a handrail

	Current study			Nadeau et al. (2003)	Reeves et al. (2009)		Novak & Brouwer (2011)	
Age	**60's**	**70's**	**80's**	**Age 41-70**	**Mean 24.6**	**Mean 73.4**	**Mean 23.7**	**Mean 67.0**
Number of subjects	n=30	n=30	n=24	n=11	n=17	n=15	n=23	n=32
Maximum hip extensor moment (Nm/kg)	0.86 (±0.32)	0.87 (±0.36)	0.83 (±0.25)	0.53 (±0.17)			0.56 (±0.19)	0.55 (±0.18)
Maximum hip abduction moment (Nm/kg)	0.86 (±0.12)	0.89 (±0.11)	0.99 (±0.18)	0.99 (±0.15)			0.62 (±0.16)	0.72 (±0.18)
Maximum knee extensor moment (Nm/kg)	1.19 (±0.25)	1.08 (±0.24)	1.14 (±0.26)	0.98 (±0.18)	1.19 (±0.24)	0.89 (±0.22)	1.06 (±0.20)	0.99 (±0.19)
Maximum plantarflexion moment (Nm/kg)	1.22 (±0.17)	1.26 (±0.19)	1.14 (±0.13)	1.17 (±0.14)	1.48 (±0.27)	1.24 (±0.21)	1.31 (±0.16)	1.19 (±0.11)

5.2.7 Hip kinetics

The mean peak hip extensor moments generated in this study were higher by approximately 0.3 Nm/kg than those reported previously by Nadeau et al. (2003) and Novak and Brouwer (2011). The peak hip extensor moment occurs at the time when the weight is being taken over the lead foot and progression is being made from double to single stance at approximately 15% of the way through the gait cycle (figure 4.11). At this time the hip of the lead leg is flexed and is extending in order to lengthen the leg sufficiently to enable the trail leg to step up. If the hip is more flexed at the point of single stance, then a greater hip extending moment will be required to balance the external moments, and to generate sufficient potential energy to progress to the next step. The step height in this study was 15mm higher than Nadeau et al. (2003) and 35mm higher than Novak and Brouwer (2011). In section 5.2.2 it was discussed that the subjects in this study used more hip flexion than knee flexion to accommodate the increased step height and it may be that this reason is behind the increase in peak hip extensor moments in the current study.

Differences in the biomechanical models used may also account for some of the variability in hip extensor moments. All the studies used a 3-D motion capture system, with integrated force plates and adopted an inverse dynamics approach to determine the kinetics. Nadeau et al. (2003) do not provide sufficient information to be able to determine how they located the hip joint centre. Novak and Brouwer (2011) state that “the hip joint centre was located at one quarter of the distance between the greater trochanters from the left or right trochanter”. The position of the greater trochanters were determined using a pointer calibration system and would be related to a cluster of markers attached to the thigh, as in the CAST system described by Cappozo et al (1996), and used in this study for other lower limb anatomical points. The method of defining the hip joint centre from the greater trochanters is similar to the method of Tylkowski et al. (1982) cited in Bell et al. (1989), however it is normally used in addition with pelvic markers to improve the accuracy in the frontal plane. The current study used the regression method of Bell et al. (1989) to determine HJC location which is likely to have put the HJC in a slightly different location to the other studies, though it is difficult to know by how much (Leardini et al. 1999). Variation in the location of the HJC will have an impact on the kinetic results reported at the hip though the extent for stair climbing has not been investigated. Stagni et al. (2000) reported that during gait, the flexionextension moments are most affected by any mislocation of the HJC especially if the mislocation is in the anterior-posterior plane.

The peak hip extensor moments measured in this study and others are similar to the peak hip extensor moments required for normal gait. Perry and Burnfield (2010) report moments of 0.84 Nm/kg at initial contact during adult gait and similarly Kerrigan et al. (1998) report peak hip extension moment as 0.86 Nm/kg in gait at a comfortable speed in healthy older adults (mean age 72.7 years). This would suggest that stair climbing does not place additional functional demand on the hip extensors compared to gait.

In the coronal plane a hip abduction moment was produced throughout the majority of the stance phase and had a pattern of two small peaks (figure 4.11). This pattern is similarly reported in the work of Costigan et al. (2002), Nadeau et al. (2003) and Novak and Brouwer (2011). In younger subjects (Nadeau et al. (2003), the 60+ and 70+ group in the current study and the young subjects in Novak and Brouwer (2011)), the pattern is that the first peak of hip abduction is the highest, followed by a second lower peak hip abduction moment. However in older subjects Novak and Brouwer (2011) reported that the second peak hip abduction moment was significantly higher than that of younger subjects and the same trend is observed in the current study as can be seen on the second peak in figure 4.11.

The peak hip abduction moments reported in table 5.3 shows less variability than the peak hip extension moments and range from the lowest of 0.62 Nm/kg in younger subjects (Novak and Brouwer, 2011) to 0.99 Nm/kg (Nadeau et al., 2003 and the 80+ group in the current study). Some of the variability may again relate to differences in step height and modelling methods as discussed earlier in this chapter. There is limited published data on the peak hip abduction moments required for gait in older adults. Watt et al. (2010) reported a mean peak hip abduction moment of 0.64 Nm/kg.m in their group of 18 older adults (mean age 70.3 years). The mean height of subjects in this study was 1.66m which would correspond to a peak hip abduction moment of 1.06 Nm/kg. The peak hip abduction moments produced during stair climbing are less that this figure reported for gait, suggesting that inability to produce hip moments are probably not a factor that prevent older adults for performing stairs if they are able to walk on the flat.

5.2.8 Knee and ankle kinetics

The plots of the knee and ankle kinetics from the current study (figure 4.14) follow a similar pattern to those reported previously in the literature (Nadeau et al., 2003; Reeves et al., 2009;Novak and Brouwer, 2011). The knee extensor moment peaks at the start of single stance on the step, demonstrating the knee extensors role in bringing up the body by extending the flexed knee. The peak knee moments from each study are shown in table 5.3 and range from 0.89 Nm/kg (Reeves et al., 2009 (older adults)) to 1.19 Nm/kg (Reeves et al., 2009 (young adults) and current study 60+). The results in this study sit at the higher end or the range. The most likely reason for this (as discussed in Section 5.2.7) is that the step height was slightly larger in this study resulting in an extra few degrees of knee flexion in early stance. The modelling of the knee joint centre is similar where stated and so there are less likely to be difference in knee kinematics due to the variation in location of the knee joint centre.

The two studies that have compared knee kinetics in young and older subjects (Reeves et al. 2009; Novak and Brouwer 2011) demonstrated that there was a second, smaller peak in the knee extensor plot close to push off from the step. This pattern was observed in all age groups in the current study but the second peak moment was higher in the older age group. It appears consistent in the studies that with increasing age there is a different strategy in later stance that will be discussed in Section 5.2.9

The demands on the knee extensors for stair ascent reported here are significantly higher than that of normal gait where peak extensor moments range from 0.44 Nm/kg

(Kerrigan et al., 1998) to 0.52 Nm/kg (Perry and Burnfield, 2010). This requirement to produce almost double the knee joint moment may be one of the limiting factors in older adults‟ ability to ascend stairs.

The ankle moments follow an opposite pattern to the knee moments. There are two peaks on the kinetic plots (figure 4.14). The first occurs during the first phase of single stance, and corresponds with the time when the leg is extending in order to assist the opposite leg to progress upwards. The second peak corresponds to just prior to toe off and indicates that the ankle plantarflexors assist in raising the body to the next step. This pattern is repeatable across the studies included in table 4.14.

The peak plantarflexor moments in studies of older adults were very comparable, ranging from 1.14 Nm/kg in the 80+ group in the current study to 1.26 Nm/kg in the 60+ group. Reeves et al. (2009) and Novak and Brouwer (2011) both reported a statistically significant decline in peak ankle joint moment in the older adults compared to younger adults. In the current study there was a significant decline in ankle plantarflexion moment in the 80+ group when compared to adults in their 60‟s and 70‟s. This suggests that there is an ongoing decline in the peak ankle plantarflexion moment produced during stair ascent with increasing age.

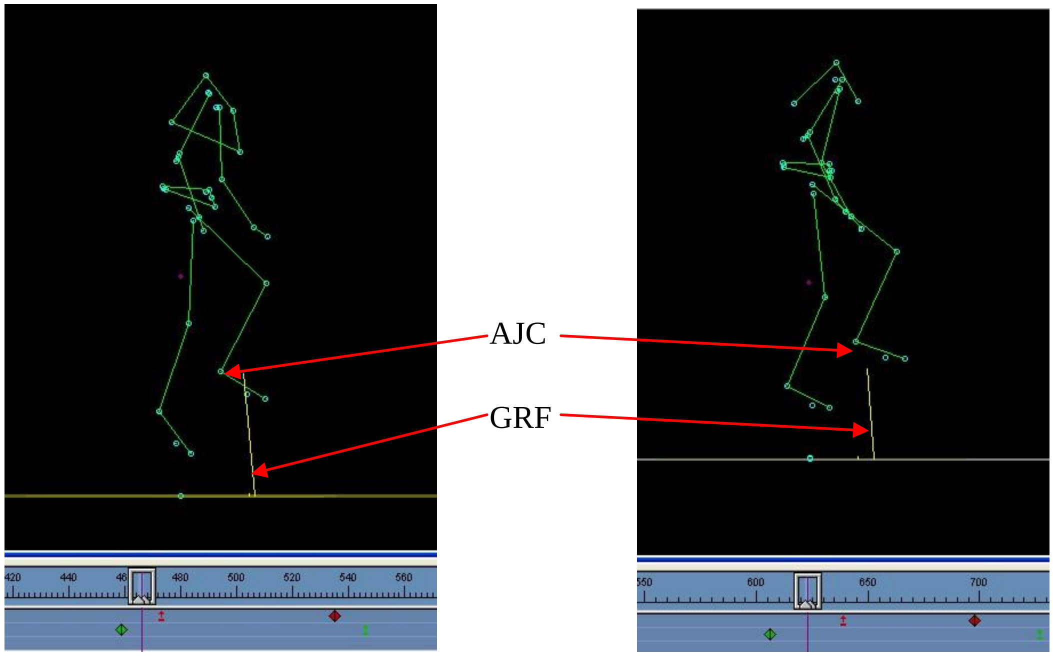
Reeves et al. (2009) used isokinetic dynamometry to measure the peak concentric plantarflexor moment that could be produced at a range of angular velocities. They reported that the older group (mean age 74.8 years) produced a peak ankle plantarflexion moment of 1.5 Nm/kg which was significantly less than that of the young group (mean age 24.6) whose mean peak moment was 1.9 Nm/kg. Reduced strength in older adults is well documented (see Section 1.2.3) and in the ankle it has been observed that there is a linear decline in maximum isometric muscle torque produced from the age of 60 onwards (Vandervoort and McComas, 1986). Older adults will be operating closer to their maximal peak plantarflexion moment during stair ascent, which may account for the need to adopt strategies to decrease the plantarflexion moments used.

5.2.9 Age related redistribution of joint moments during stair ascent

Summarising the previous sections it appears that the older adults in this study are using a different kinetic strategy for stair ascent than younger adults. This finding is similar to that of Reeves et al. (2009) who investigated moment distribution across the knee and ankle, and Novak and Brouwer (2011), who investigated all three lower limb joints.

The most observable change in the kinetics is the decreased planatarflexion moment throughout the stance phase observed in the 80+ subjects compared to the younger subjects. In the first period of double support there does not appear to be significant compensation for this at the hip or knee. If the support moment (the sum of the lower limb moments) at this time was plotted, there would be a reduction in support moment in the lead leg in older subjects. This corresponds to the findings of Novak and Brouwer (2011). The decrease in lead leg support moment appears to be being compensated for by an increased support moment in the trail leg at the corresponding time. In this study the older subjects produced greater peak hip extensor and knee extensor moments in later stance which would produce a greater support moment in the trail leg along with an increase in hip abductor moment. This would imply that there is greater pushing up from the step below than pulling up from the step above during this first phase of double support.

Reeves et al. (2009) explored the mechanisms by which the ankle joint moment was lower in the older adults in their study. They investigated the ground reaction forces and found that these were similar between young and older subjects. However, the distance between the centre of pressure (COP) and the ankle joint centre was smaller in the older subjects, reducing the moment arm length and hence the external moment produced at the ankle. This effect was not formally measured in this study but it is noticeable when looking at the VICON files from the older subject that the ground reaction force passed much closer to the ankle joint (figure 5.1). This would support the theory that the older adults may be intentionally shifting the COP to reduce ankle moment demands during stair climbing.



60+ male

80+ male

Figure 5.1 Relationship between the ground reaction force (GRF) and ankle joint centre (AJC) in a 60+ and 80+ male ascending the stairs with no handrail

Throughout mid to late stance there is a redistribution of joint moments in the oldest group in this study. The decreased plantarflexion moment in later stance is associated with an increase in hip and knee extensor moments and hip abduction moments. It has been discussed earlier that the ankle plantarflexors may be operating close to the available limits in older adults, and also that knee moments are much higher than what is required for normal gait. However stair climbing does not place significantly higher demand on the hip muscles than level walking. Some of this redistribution may be to enable the hip muscles, which have greater strength reserves to have a greater role. The older adults adopted a slightly more flexed hip posture in later stance which will assist the hip extensor muscles to operate in a better part of the length-tension curve.

Although the demands on the knee extensors were already high, the oldest adults adopted a strategy that involved greater work from the knee extensors during late stance. All the subjects in this study had a second smaller peak in the knee extensor moment shortly prior to toe off (figure 4.14). This was more marked in the oldest subjects and is associated kinematically with a reduction in the knee extension in this phase. This pattern was observed in the older subjects studied by Reeves et al. (2009) but was not seen in the younger subjects. These authors hypothesized that the elderly subjects are utilising gastrocnemius as a two joint muscle to transfer energy from the knee to the ankle to enhance the plantarflexion moment. This results in a slowing of knee extension at this phase in gait, resulting in the knee being slightly more flexed at terminal stance. With the knee slightly more flexed the quadriceps muscles will be able to operate at a more optimal length and therefore be able to produce a greater knee extensor moment at this time.

In stair climbing the hip abductor muscles control the lateral pelvic tilt in order that the swing leg can adequately clear the intermediate step (Nadeau et al., 2003). In this study it may be that the oldest subjects are augmenting the hip abductor moments to ensure adequate clearance but also to improve stability in the frontal plane and hence safety. It had been anticipated that older subjects may have an increased movement of the COM in the frontal plane due to poorer control but this was not the case for stair climbing in these subjects.

The second peak of hip abduction is observed just prior to toe off and shows a trend to increase with each increasing decade. This pattern was similarly observed by Novak and Brouwer (2011) between their young and old adults. This peak occurs during the second phase of double support at which time the lead leg has been found to have a reduced support moment. The increase in hip abductor moment in the trail leg may be as a result of altered loading between the two legs, with the trail leg being more instrumental in pushing the body up the stairs than the lead leg is by pulling.

The redistribution of joint moments for stair climbing presented here is similar to those reported during gait (Kerrigan et al.,1998; DeVita and Hortobayi, 2000). This may indicate that there is a neuromuscular adaptation with ageing that is changing the motor pattern associated with gait and stair climbing. This pattern may favour muscle groups that have greatest reserve in the elderly.

5.3 Age related changes in stair descent

5.3.1 Temporal changes

In this study it was observed that subjects in their 80‟s descended stairs significantly slower than subjects in their 60‟s and 70‟s. There was a slight but significant increase in the amount of time spent in stance by the 80 year olds when descending the stairs with no hand rail compared to the younger groups.

The results of the current study are compared to the findings in the literature in table 5.4. The subjects in their 60‟s in the current study are descending stairs at a similar rate as the older subjects studied by Novac and Brouwer (2011), and have a similar cadence to the younger subjects observed by Reiner et al. (2002). In section 5.2.1 it was noted that step height seemed to influence the rate of stair ascent. This does not seem to be as

Table 5.4 Comparison of temporal data for stair descent

	Current study			Reiner et al. (2002)	Reeves et al. (2008[b])		Novac and Brouwer (2011)	
Age	**60’s**	**70’s**	**80’s**	**Mean 28.8**	**Mean24.6**	**Mean 74.9**	**Mean 23.7**	**Mean 67.0**
Number	30	30	24	10	17	15	24	33
Time per step cycle (s)	1.20 (± 0.23)	1.24 (±0.24)	1.53 (±0.29)	1.19 (±0.1)				
Rate of stair descent (steps/min)	103.8 (±19.4)	99.2 (±16.6)	81.2 (±15.6)				110.6 (±10.2)	103.7 (±15.6)
Stance phase (% of cycle)	60.0 (±2.4)	60.5 (±3.6)	61.6 (±3.3)	61.2 (±2.3)	61	63		
Swing phase (% of cycle)	40.0 (±2.4)	39.5 (±3.6)	38.4 (±3.3)	38.8 (±2.3)	39	37		
Double support (% of cycle)	19.3 (±3.1)	20.5 (±4.5)	21.4 (±6.0)	22.4 (±4.6)	25	26		

apparent in stair descent as there is consistency across studies even though step height ranged from 150mm to 185mm. This finding is supported by Reiner et al. (2002) who investigated stair ascent and descent using three different step heights. They found that step cycle time increased with increasing inclination for stair ascent but remained the same during stair descent. These differences between ascent and descent may relate to the difference in the transfer of energy in the two events, which has been stated by McFadyen and Winter (1993). During stair ascent there is a need to transfer muscle energy into potential energy to progress the body upwards. In descent a controlled lowering occurs as potential energy is dissipated by the body. An increase in step height will increase the energy required to be produced and dissipated. It would appear that the body can adapt to increase the rate of dissipation whereas increased muscle energy production takes a greater time.

There appears to be relative consistency between the studies in the amount of time spent in stance. Novac and Brouwer (2011) reported their stance time in seconds, making the results difficult to compare, but they did analyse statistically the difference between their groups, finding that the older group spent more time in stance than the younger group. Reeves et al. (2008[b]), also observed a longer stance phase in the older subjects but did not report on whether this was statistically significant. It can be concluded that there does appear to be a trend for older adults to increase stance phase duration during stair descent and that this continues with increasing age as seen in the current study. This strategy will assist in providing stability during stair descent and potentially make stair descent safer.

5.3.2 Kinematic changes

The kinematic plots for stair descent shown in figures 4.21 and 4.22 demonstrated a very repeatable pattern across the age groups. The shape of the plots compare well to those reported previously in the literature (McFadyen and Winter,1993; Riener et al. 2002; Reeves et al., 2008[b]). The peak angles used during stair descent in the young and old have only previously been reported by Reeves et al. (2008[b]), and these are compared to the current study in table 5.5.

	Current study			Reeves et al. (2008[b])	
Age	**60's**	**70's**	**80's**	**Mean 24.6**	**Mean 73.4**
Number of subjects	n=30	n=30	n=24	n=17	n=15
Maximum Hip Flexion (degrees)	40.0 (±1.4)	37.6 (±1.5)	44.0 (±2.4)		
Minimum Hip Flexion (degrees)	13.3 (±1.3)	12.6 (±1.4)	14.2 (±0.9)		
Maximum Knee Flexion (degrees)	95.7 (±0.9)	95.5 (±1.3)	97.0 (±1.3)	91.8 (±5.9)	90.2 (±6.4)
Minimum Knee Flexion (degrees)	2.3 (±0.8)	2.6 (1.0)	9.3 (±1.0)	12.7 (±4.2)	12.0 (±4.9)
Maximum Ankle Plantarflexion (degrees)	27.4 (±1.1)	28.2 (±0.9)	26.4 (±1.9)	20.6 (±5.4)	22.5 (±4.9)
Maximum Ankle Dorsiflexion (degrees)	32.5 (±1.6)	34.6 (±1.0)	30.4 (±1.4)	33.2 (±4.0)	33.9 (±5.7)

Table 5.5 Peak angles during stair descent without a handrail

Subjects in the current study used a greater ROM at the knee and ankle than in Reeves et al. (2008[b]). As was discussed in section 5.2.2 this may relate to the increased step height in the current study, the slightly shorter participants or variations in the biomechanical modelling. The findings of both studies show that there is a very large joint excursion required at the knee and ankle and less so at the hip.

Reeves et al. (2008[b]) did not report any statistically significant differences between their young and older group for joint kinematics. In the current study there were no significant changes in ankle ROM. However, at the knee the oldest subjects did not gain as much extension in terminal swing and the knee remained slightly more flexed throughout stance (see figure 4.21) and the oldest subjects had greater hip flexion throughout stance and swing phase. This increase in hip flexion was discussed earlier (section 5.2.2) and appears to be in part due to more anterior tilt at the pelvis and a greater forward lean in the trunk. As discussed previously it seems unlikely that this is due to a flexion contracture at the hip as all subjects had much greater ROM than is required for stair descent. There is some possibility that this position may assist visualising foot placement on the stairs and therefore is occurring as a safety mechanism, or alternatively the flexed position may be a strategy to alter the lower limb biomechanics during stair descent.

The increase in hip and knee flexion at initial contact may be used by the oldest subjects to assist in the dissipation of energy at this time. In younger subjects during loading response the main strategy to absorb energy is the rapid motion of the ankle from plantarflexion to dorsiflexion combined with a peak in plantarflexor moment (Perry and Burnfield, 2010). In the older subjects, the increase in hip and knee flexion will enable a larger extensor moment to be produced at both joints at this time possibly redistributing some energy absorption to the hip and knee, where there are likely to be greater reserves.

The ROM requirements at the knee and ankle during stair descent may be one of the factors why older adults experience difficulty with this activity. Ankle dorsiflexion is significantly higher than is required for normal gait. Winter et al. (1990) report peak ankle dorsiflexion of approximately 10° for gait in healthy older adults. In this study an extra 20° was required for stair descent which is close to the actual ROM available at the ankle joint in healthy older adults, a maximum dorsiflexion range of 32° being reported by Reeves et al. (2008[b]). Knee flexion is also much higher than is required for gait and whilst healthy older adults should have sufficient ROM, those with degenerative joint conditions may not have the required ROM to descend stairs in a reciprocal manner. The demands on knee and ankle ROM will be greater with a greater step height and will also be higher for shorter people.

5.3.3 Kinetic changes

In this study changes in the kinetics of stair descent were observed in the subjects aged 80+ when compared to those in their 60‟s and 70‟s (figures 4.24 and 4.26). These findings can be summarised as such:

1. A decrease in the ankle plantarflexion moment produced in early to mid-stance phase.
2. A reduction in the hip flexion moment throughout stance, occasionally becoming a hip extension moment.
3. An increase in knee extensor moment from early stance to the second peak.
4. A higher hip abduction moment through mid-stance.

Although the moments produced in early to mid-stance showed a redistribution of moments, the peak moments produced at the hip, knee and ankle throughout stance were not different. This demonstrates that the older subjects in this study were capable of producing the same moments as the younger subjects and implies that it is not simply due to weakness that the ankle moment is reduced in early stance.

The kinetic plots in this study are highly comparable with those produced by Novac and Brouwer (2011), the changes observed between the young and old participants in that study being replicated between the 60 and 80 year olds in the current study. To date this is the only other study reporting hip, knee and ankle kinetics for stair descent, comparing young and older adults. Reeves et al. (2008[b]) investigated knee and ankle moments and found slightly different kinetic plots. They observed a similar reduction in the first peak ankle plantarflexor moment in older adults, but also found a decrease in the second peak ankle plantarflexor moments and did not observe the increase in knee extensor moment throughout stance as was seen here and by Novac and Brouwer (2011). Some of these differences in studies may be due to differences in the subjects tested. Reeves et al. (2008[b]) investigated 15 older adults but do not state if they were male or female. Novac and Brouwer (2011) studied 32 older adults of whom 19 were female, which was a closer sample size and demographic to the current study. In this study it was found that the female subjects used greater hip and knee ROM during stair descent and had a significantly reduced peak knee extensor moment. It is therefore important to consider the gender differences between studies. A summary of the finding of the three studies is presented in table 5.6.

Table 5.6 Peak joint moments during stair descent without a handrail

	Current study			Reeves et al. (2008[b])		Novak & Brouwer (2011)	
Age	**60's**	**70's**	**80's**	**Mean 24.6**	**Mean 74.8**	**Mean 23.7**	**Mean 67.0**
Number of subjects	n=30	n=30	n=24	n=17	n=15	n=23	n=32
Maximum hip extensor moment (Nm/kg)	0.42 (±0.19)	0.43 (±0.29)	0.49 (±0.18)			0.23 (±0.19)	0.23 (±0.19)
Maximum hip abduction moment (Nm/kg)	1.17 (±0.25)	1.15 (±0.21)	1.21 (±0.21)			0.75 (±0.16)	0.76 (±0.17)
Maximum knee extensor moment (Nm/kg)	1.21 (±0.23)	1.13 (±0.22)	1.22 (±0.29)	0.91 (±0.29)	0.83 (±0.17)	1.11 (±0.16)	1.19 (±0.17)
Maximum plantarflexion moment (Nm/kg)	1.09 (±0.20)	1.08 (±0.14)	1.01 (±0.11)	1.32 (±0.34)	1.03 (±0.14)	1.07 (±0.17)	1.02 (±0.12)

Ankle plantarflexor moments are comparable across the three studies, ranging from 1.01Nm/kg in older adults (current study) to 1.32Nm/kg in younger adults (Reeves et al. 2008[b]). Knee extensor moments were highest in the current study compared to the other two. As discussed in section 5.2.3. the potential causes for this may be the difference in step height, the current study having the largest steps, and biomechanical modelling methods. The hip extensor and abductor moments were much higher than previously reported. Some of this variability may be due to the step height and modelling methods, however this does not explain the fact that the hip extension moment kinetic plot (figure 4.24) does not suggest such a high peak moment. Exploring the raw data provides some clarity as to how the hip extensor moment is higher than seems obvious from the kinetic plot. Figure 5.2 shows the sagittal plane kinetics for a single subject:

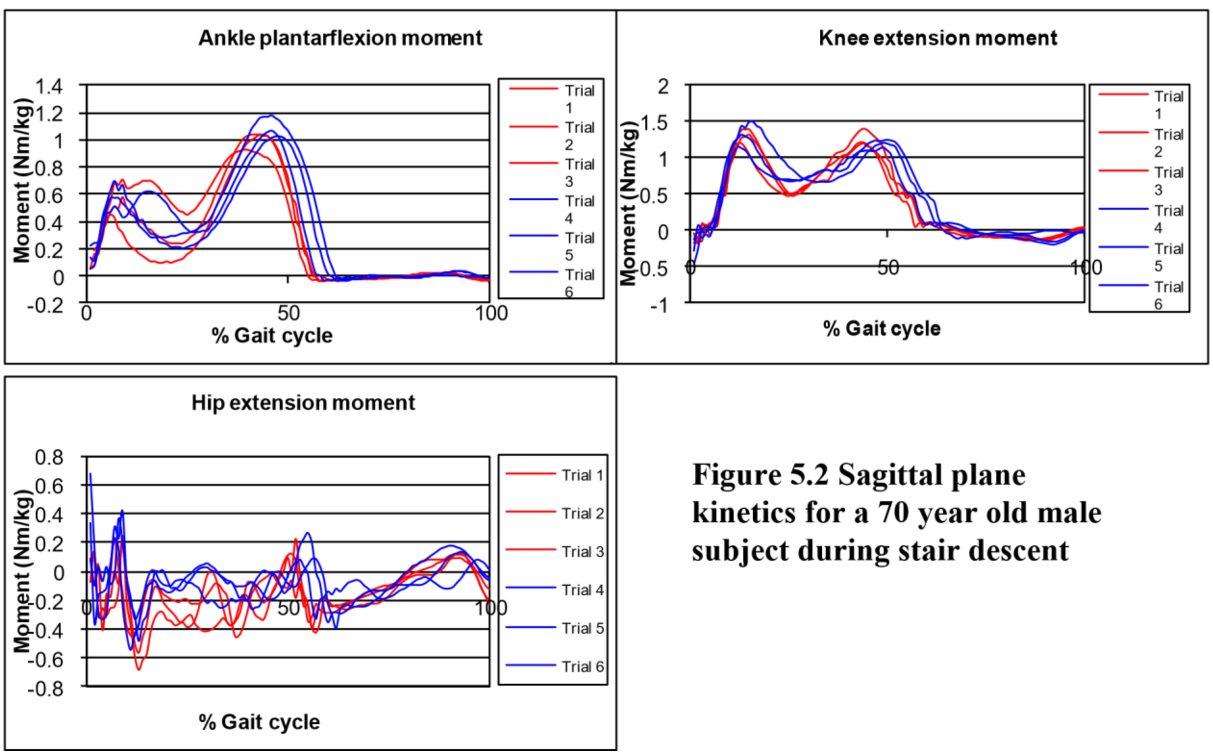


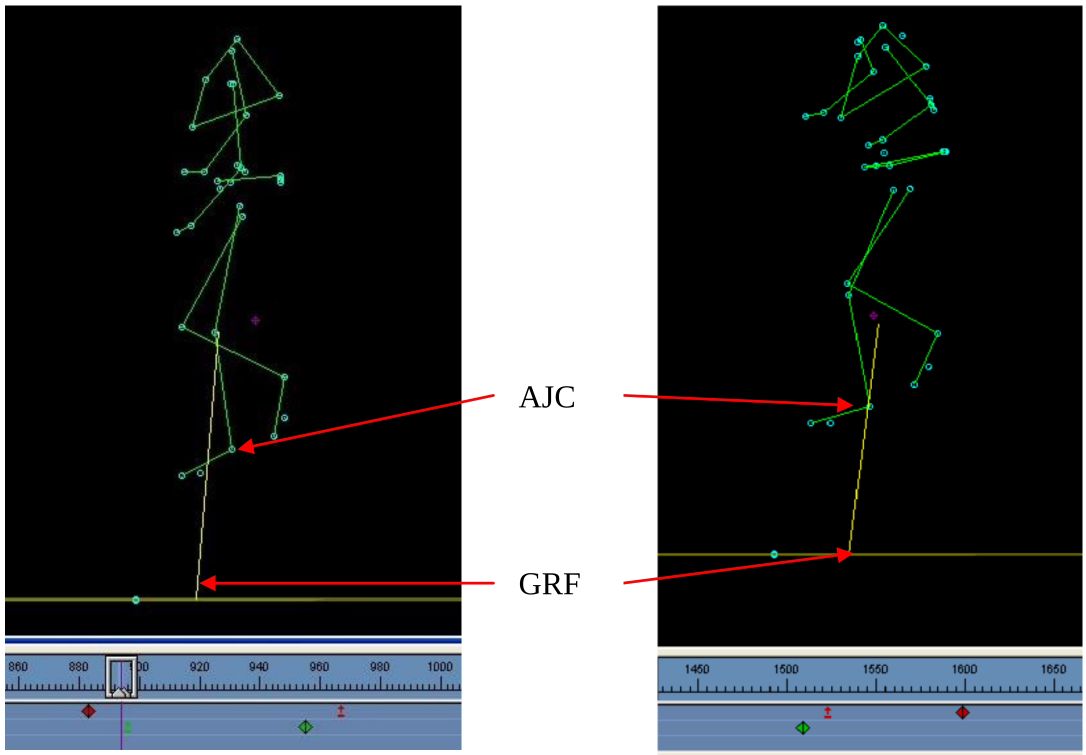
Figure 5.2 Sagittal plane kinetics for a 70 year old male subject during stair descent

For this subject there are some parts of the hip moment profiles that are very repeatable, such as the peak following initial contact. The profile for this individual demonstrates that hip moments are generally low in magnitude throughout stance, but the shape of the plots are irregular with several peaks and troughs throughout stance phase. Between trials there is variability in the location of the peaks and troughs and so when the plots are averaged there will be a smoothing out of the data. The hip moments showed the greatest variability between subjects with the position of the peaks and troughs occurring at different locations. By producing ensemble plots for the age groups the output curve may appear much smoother as negative spikes for one subject will be averaged with positive for another. The peak joint moment reported in table 5.6 was calculated by averaging the peaks from each individual trial. This will result in a higher value that will be observed on the averaged moment plots.

Intra and inter-subject variability in hip moments during stair has been investigated previously. McFadyen and Winter (1998) and Novak and Brouwer (2011) have both reported that the intra-subject variability at the hip is higher than at the ankle and knee. Variability between subjects, as measured by the coefficient of variance (CV), was four times higher at the hip than at the ankle and knee (Novak and Brouwer, 2011). With such high variability the practice of creating ensemble average plots may not be appropriate due to the overall plot shape not necessarily representing any of the individual plots. This factor should be taken into consideration when viewing the kinetic plots in this study and the literature.

5.3.4 Age related redistribution of joint moments during stair descent

Figure 5.3 shows the VICON model output for two subjects (one age 60+ and one aged 80+) during stair descent, just prior to the period of single leg support.



60+ male

80+ male

Figure 5.3 Relationship between the ground reaction force (GRF) and ankle joint centre (AJC) in a 60+ and 80+ male descending the stairs

Although this is a simplified view of the positions of the joint centres, it demonstrates the more flexed posture of the older subject at this pointing the gait cycle. As was observed during stair ascent the older subject has adopted a strategy that positions the GRF closer to the ankle joint centre in the sagittal plane. This strategy certainly will contribute to the reduced plantarflexion moment occurring in the first half of the stance phase and the increase in the knee extensor moment.

Novac and Brouwer (2011) reported very similar kinetic plots to the current study but also included support moments. They found that the older adults had a reduced support moment during the weight acceptance phase on the lead leg but that support moments in later stance were higher in the older subjects. The moment profiles (figures 4.24.and 4.26) in this study conclude with this, as the reduced ankle plantarflexor moment in early stance is not totally compensated for at the hip and knee. This would result in a reduced total support moment in early stance. In late stance the hip flexor moment is reduced in the 80+ subjects with a slightly higher knee extensor moment and comparable ankle plantarflexor moment. This would result in a higher total support moment in late stance in the oldest subjects. This behaviour could represent redistribution in the energy absorption between the lead and the trail leg. The older subjects may be reducing the loading on the lead leg by maintaining more weight on the trail leg. This theory however is not supported by studies into the ground reaction forces produced by young and older adults during stair descent (Christina and Cavanagh, 2002; Reeves et el. 2008[b]). Both groups found that GRFs were equal in young and older adults in the early part of gait when walking at a similar speed.

The older subjects in this study, and also that of Novac and Brouwer (2011), were significantly slower during stair descent than younger subjects. This would reduce the moments required during loading response and may well explain the overall reduction in the total support moment at this time. In the current study the rate of stair descent was not controlled as it was felt that it was important to investigate any changes that occur as a result of increasing age. In future studies it would be useful to control for speed to determine the proportion of biomechanical changes related to changes in gait speed.

In section 5.9 it was discussed that older adults may reduce the peak plantarflexion moment produced in stair ascent as they may be close to their maximum plantarflexor moment production. The same theory cannot be applied to stair descent as the plantarflexor moment peak in late stance is equal across the age groups. The older adults therefore are able to produce the same plantarflexor moment as younger subjects, but for some reason they do not do so in early stance. The plantarflexor moment in early stance is produced by eccentric muscle action in the gastrocnemius and soleus muscles. Muscle power, i.e. the rate of generating a moment, is known to be reduced with age (Thelen et al. 1996). The older subjects may be reducing the ankle moment as they cannot recruit gastrocnemius and soleus at a fast enough rate, but have sufficient knee power to compensate.

From this study it is apparent that the oldest adults adopted a different posture during stair descent and also they redistributed joint moments especially in early stance. It is impossible to tell which event leads to the other as it could be that loss of ROM, dynamic contractures and fear of falling lead to a more flexed posture, which in turn alters the ground reaction force and COP, thus altering moments. Alternatively it may be that, due to reduced power in the ankle muscles, a strategy has to be adopted which reduces the ankle moment and it is that which leads to the more flexed limb position.
The subjects in this study did not appear to have ROM losses when measure statically (Samuel, 2005) but this does not mean that they did not have full active ROM in all limb positions. Future work should investigate active ROM and joint powers around all the lower limb joint, in order to address this question.

5.4 Functional demand of stair ascent and descent

As part of the EQUAL project, work was undertaken by Dinesh Samuel (Samuel, 2005) to measure the functional ability of the participants, in order that the functional demand of activities of daily living could be evaluated. The strength at the hip and knee were reported in Section 4.8 and it appeared that the knee moments used during activity were generally higher than those measured on the dynamometer.

As the biomechanical results in this study have been found to be comparable at the ankle and knee to previous studies, this does not appear to make sense as the older adults had sufficient strength to perform the activity. A reason for the dynamometer moments recorded by Samuel (2005) being lower than was observed functionally will be the type of muscle activity that was being recorded. The dynamometer built for the EQUAL project recorded isometric joint moments i.e. the maximum moment produced at any fixed angle as the muscle attempts to shorten. Functionally, during stair ascent, the hip and knee extensors are working concentrically with both the hip and knee progressing from flexion to extension. The most comparable method to evaluate this would be concentric isokinetic testing. For stair descent the muscles are working eccentrically and therefore eccentric isokinetic testing would be most comparable.

Two studies have investigated the functional demands of stair ascent and descent comparing them to isokinetic joint moment (Reeves et al., 2008[b]; Reeves et al.,2009). These authors reported maximum concentric and eccentric knee extensor moments of 92.1 Nm and 149.4 Nm respectively in their group of older adults. During stair ascent the peak knee moment used was 75% of this and during descent the peak knee moment used was 42% of that available. They also recorded both concentric and eccentric isokinetic ankle moments. During stair ascent their subjects used 95% of the maximum ankle moments and during descent this was 77%.

Comparing the 70 year olds in the current study to those investigated by Reeves et al. (2008[b],2009), it would seem that these adults are not exceeding the peak isokinetic moments that would be expected at their age, however they are requiring a high percentage of the available strength to perform the activity. In future studies for the evaluation of functional performance, it would seem that isokinetic dynamometry would be the most appropriate method.

In conclusion, the functional demands of stair ascent and descent are high, especially for the ankle muscles and this could certainly be a factor that hinders older adults from performing this task as strength declines with increasing age. The strength demands of stair ascent and descent are increased with increasing step height so it is essential that designers consider this when planning environments that older adults need to access.

5.4 The effects of using a handrail

At this time there is very limited data on the effects of using a handrail on the biomechanics of stair ascent or descent. One of the aims of this study was to explore the effects of handrail use on stair gait to fill this knowledge gap. At this time the only other comparable study is Reeves et al. (2008[a]), who investigated stair ascent and descent with and without light handrail use in a group of thirteen older adults (mean age 74.9 years). This thesis is novel in investigating if handrail effects change with increasing older age.

5.4.1 Stair ascent

In the current study six of the older adults did not feel able to attempt the task of stair ascent without use of a handrail, all of whom were capable of performing the task with a handrail. Of these subjects three were in their 70‟s and three in their 80‟s. Five of these subjects were female, and only one was male. The biomechanical plots for these subjects were studied and were not found to be unusual for the age group i.e. the plots were not much lower than their peers or demonstrating altered strategy. The peak moments produced were within 1 s.d. of the gender specific age group normal in all but one case, who walked much slower than the average, suggesting that these subjects were not performing stair ascent with a handrail differently from their peers. Of note is that the male subject was the weakest subject in both the hip and knee isometric muscle tests and that the female subjects were all much weaker than their age equivalent males. Decrease in strength may therefore be a factor that influences the ability to ascend stairs without a handrail.

Investigating the subjects who could ascend stairs with and without a handrail it was found that without the handrail the rate of stair ascent was faster. There was no difference noted in the time spent in stance or double support. This was contrary to what was expected which was that older adults would adopt a slower and safer gait when walking with no handrails, and is different to the findings of Reeves et al. (2008[a]), who reported no significant change in cadence between the two conditions. This difference between the studies does not appear to be due to differences in the staircase as both studies used a handrail height of 0.9 m and there was only 15 mm difference in riser height. There may, however, be some influence of the experimental design. In the current study the participants completed three trials of stair ascent and descent using a handrail first. This was done in order that the investigators could ensure that the older adults would be safe to attempt stair ascent and descent without a handrail. There may have been a familiarisation of the task which could have resulted in subjects being more confident and therefore faster when then asked to attempt without a handrail. Reeves et al. (2008[a]) asked subjects to do the “without hands” task first followed by “with the rails” which may have resulted in the subjects being faster in the trials with the handrail due to improving knowledge of the task. In future studies it would be useful to consider in the design of the experiment to reduce any order effects.

If the change in cadence was not entirely due to an order effect then other explanations should be explored. It was apparent from the laboratory sessions that the stair ascent and descent without a handrail were the tasks during which the subjects felt least comfortable. There may have been a component of wanting to get it over and done with as soon as possible that lead to a rushing of the task. Another point is that using a handrail required the participants to concentrate both on hand and foot placement and that may have resulted in a more deliberate gait which in turn resulted in the decrease in cadence.

Regarding the joint kinematics it was found that there was a small but significant increase in the peak hip and knee flexion used when not using a handrail. This increase was 1.5° at the hip and 2.3° at the knee and occurs during swing phase as the leg is flexed to enable clearance of the step and placement on the next step up. The combination of both an increased hip and knee flexion may be a safety mechanism as it would result in improved clearance of the swing leg. This could lessen the chance of a trip caused by inadequate foot clearance, which would obviously be more serious with no handrail support. This small change, however, may be purely as a result of the increased cadence and care should be made in interpreting it further.

In all of the age groups in the current study it was found that there was an increase in the peak extensor moments at the hip, knee and ankle when ascending stairs without a handrail. Looking at the kinetic plots in figure 4.16, it can be observed that the hip and knee extensor moments are higher throughout stance phase when no handrail is used, and the peak ankle plantarflexion moment is increased just prior to toe off. The increase in joint moment was 0.06 Nm/kg at the knee and 0.08Nm/kg and the ankle and hip. This increase will in part be related to the increased cadence of stair ascent without the handrail. The amount of increase, especially at the ankle may be one of the factors that makes stair ascent without a handrail difficult for older adults as it will take the ankle moment close to its functional limit.

Reeves et al. (2008[a]) reported an altered strategy of stair ascent in their subjects with a redistribution of the joint moments when no handrail was used. They found that the knee extensor moment was lower during stair ascent with no handrails and ankle moments were higher. This strategy would place greater demand on the ankle plantarflexors and does appear counterintuitive as the ankle is operating much closer to its functional demand than the knee joint. However, Reeves et al. (2008[a]) suggested that the older adults adopted this strategy as increasing the knee extensor moment during unaided stair ascent may present a challenge to balance, and therefore the subjects were avoiding this. This joint redistribution was not observed in the current study and the older subjects were able to increase the knee moment without an apparent detriment to their balance. In fact the 80+ subjects had the greatest increase in knee extensor moment when not using a rail, which would not be expected if this affected balance. The strategy in the current study involved increasing both hip and knee extensor moments during the pull up phase and the ankle and knee extensor moments during the push off phase when no handrail was used.

Handrail use is associated with assisting balance and improving safety (Startzell et al. 2000). This study measured the total excursion of the whole body COM in a mediolateral direction for each step cycle, theorising that a greater COM excursion results in a greater challenge to balance. Previous work by Chou et al. (2003) reported that balance impaired older adults demonstrated greater excursion of the COM when clearing obstacles than young adults, and other studies have demonstrated a loss of M-L stability in older adults during gait (Schrager et al., 2008). On average the COM excursion of subjects was 16 mm less when using handrails to ascend the stairs, and this effect increased with increasing age. By reducing the COM excursion it would be expected that there is a reduced risk of falls related to the COM being separated from the centre of pressure (COP) under the foot/feet. The use of handrails therefore results in a less challenging gait regarding balance for these older adults.

In summary when ascending the stairs, use of a handrail resulted in a slower, more balanced gait, with reduced joint moment demands at the hip, knee and ankle. Handrails may enable older adults who are operating close to their functional limits to perform the task of stair ascent when they might not otherwise be able to do so.

5.4.2 Stair descent

Eight of the subjects who were able to descend stairs using a handrail were not able to do so without the handrail. Of these, five were in the oldest age group and seven were female. Six of these subjects were the same subjects who could not ascend the stairs without using hands, and some of the factors for their difficulties were discussed in section 5.4.1.

Similarly to stair ascent, the participants in this study were significantly faster during stair descent when not using a handrail. This will place greater demands on the lower limbs to absorb the energy during stance phase and was not an expected finding of this study. As stair descent can be considered a controlled fall, some of this increased speed may be due to the older adults not being able to effectively slow down the body weight without the use of hands. The handrail may be being used to steady the upper body and enable better control. Reeves et al. (2008[a]) also reported a small increase in cadence though this was not significant at the 99% level. Although the subjects in the current study were faster without a handrail they did adopt a more stable gait with a greater percentage of time being spent in stance and double support. This has not been reported previously. As falls are more frequent during stair descent (Startzell et al. 2000), a more stable strategy is an appropriate response when not using a handrail.

The kinematic demands of stair descent were not affected by use of the handrail as no significant differences were found between the two conditions. This corresponds well with the results from Reeves et al. (2008[a]) who also found no change in ankle and knee joint excursion and only a 2° change in hip flexion. This implies that the lower limbs are able to maintain similar stiffness at initial contact with and without the handrail.

The kinetic responses to handrail use are best divided into two phases. In the first half of the stance phase, there is an increase in the hip and knee extensor moments when subjects are not using the handrail. In late stance the ankle and knee extensor moments are increased in addition to the hip abduction moment. Overall this results in the peak moments at the hip, knee and ankle all being significantly increased when descending stairs without the handrail. For the ankle and knee extensors the increase is 0.04 Nm/kg, the hip extensors 0.08 Nm/kg and the hip abductors 0.07 Nm/kg. This increase in moments without a handrail will be partially contributed to by the increase in cadence. It may also be that subjects using the handrail were taking weight through them. As the handrails were not instrumented it is impossible to say how much load is being taken, and with one force plate a full representation of vertical ground reaction forces cannot be made to determine if these were reduced.

A redistribution of joint moments was reported by Reeves et al. (2008[a]) during stair descent. They found that compared to using a handrail, subjects had an increased knee joint moment and a decreased ankle joint moment than when descending the stairs unaided. On initial inspection they reported this as counterintuitive as the peak ankle joint moments during unaided stair descent will be close to its functional limit whereas the knee will have reserve. They explained this finding as a balance strategy, and as a result of the older subjects maintaining the foot flatter on the stairs for longer, providing an improved base of support. This affected the position of the GRF in relation to the ankle joint centre resulting in the higher moment. A redistribution of joint moments was not observed in the current study in any of the age groups relating to use of the handrail. This study used data from the 76 older adults who could complete both conditions. This is a much larger sample size than Reeves et al. (2008[a]) and may account for some of this difference.

As was found for stair ascent the M-L COM excursion was significantly lower when using a handrail than when descending unaided. The difference of 11.4 mm indicated that the handrail had a significant effect on stability of the body in the frontal plane during stair descent. Reeves et al. (2008[a]) investigated the separation of the COM and COP during stair descent, considering that a smaller COP-COM separation indicates improved balance control. They found that there was an increased separation of the COP-COM during stance phase when descending stairs unaided, indicating that handrails do assist with balance control. This effect on balance control may well be the reason that many older people do not feel safe descending stairs unaided.

In summary, use of a handrail resulted in a slower gait with reduced demands on all lower limb joints and assists with stability in the frontal plane. The subjects in this study were healthy older people but some were unable to perform the activity without a handrail for support. The handrail will offer assistance with balance control, but for more frail older adults the handrail will be required to enable the arms to produce some of the work required for stair ascent in order that the lower limbs can be offloaded, to enable muscles to operate within safe limits.

5.5 Limitations of study

This study investigated 84 healthy older adults who were willing to participate in the study. As discussed earlier, these participants were much more able than their peers and therefore the biomechanical data here cannot be extrapolated to all older adults but to those who match the demographic investigated. The number of participants was high for a biomechanical study, and did provide adequate statistical power to observe differences between groups. However, this sample was still relatively small and may still not cover the full range of results which may be obtained in these age groups, especially in the oldest group.

The staircase used in the study was designed to fit in the lab space available at the time. Having four steps and an upper platform was felt sufficient to get a natural gait style but the recent work of Cluff and Robertson (2011) suggest that at least 5 steps are required in order that steady state gait is obtained. This produces challenges for laboratories undertaking this kind of work, in respects both to space and safety. The consequences of a fall on longer staircases is more serious and the use of safety harness may be indicated as was used by Christina and Cavanagh (2002). The current staircase only enabled recording of forces from a single forceplate, making it impossible to investigate asymmetry in gait or to investigate any changes in the behaviour of the COP.

Errors in the biomechanical data can be introduced from several sources, namely instrumental errors, soft tissue movement artefacts and errors in the location of anatomical points. The equipment in this study was regularly calibrated to ensure it was as accurate as possible and calibration residuals for the VICON system were lower than 1mm – meaning that the accuracy of locating markers was to within a millimetre. The pointer system of calibration was the CAST method described by Cappozo et al (1996) as this avoids the need to place markers over anatomical landmarks where large amounts of skin movement occur. The points were identified with the subject in quiet standing and then were related to the technical frame of the associated cluster of marker on the limb. Although this has been found to be an improvement over placing markers on landmarks there will still be some alteration of the relationship between the anatomical landmarks and the technical frame as the leg moves to the extremes of movement due to tissue deformation and skin sliding. These errors lead to mislocation of a landmark by as much as 15mm in the example of the greater trochanter (Cappozo et al.,1996). Leardini et al. (2005) propose some strategies to manage this such as multiple calibrations with different limb positions and global optimisation. Software developments will be required to enable these practices to become routine and practical in a gait laboratory.

Anatomical landmark location was difficult in some subjects in this study due to overlying soft tissues, especially around the pelvis. A single person performed all the marker placement with subjects to prevent inter-tester variability being a problem.

Overall, it is essential to consider the errors in data in this study, as in any other biomechanical study. The extent of these errors may affect the kinematic and kinetic data produced and the anticipated degree of error has been discussed in the appropriate sections.

Chapter 6:

Conclusions and Recommendations

6.1 Summary of findings

This study has provided a comprehensive biomechanical database for stair ascent and descent in adults aged over 60. It is the first study to explore the biomechanical changes due to increasing older age, and provides biomechanical information on adults aged over 80 which is novel in the literature. It is the first study to present full lower limb joint kinetics, kinematics and temporal data which enables a clearer picture of how older adults adapt to the demands of the activity.

The key findings are:

1. There were no significant changes between the 60 and 70 year old groups in all aspects of stair ascent and descent. This suggests that the physical changes of ageing do not affect performance until a critical point where strength, balance or flexibility limits are close to the functional demands of the activity.
2. The oldest group generally adopted a slower and more cautious gait on the stairs with increased time spent in stance.
3. The posture of the oldest group was generally more flexed than the younger groups. This was observed as flexed hips, with an anteriorly tilted pelvis and a forward lean in the trunk. This posture may facilitate hip extensors by placing them in a better part of their range, enabling the hip extensors to provide a greater contribution to the support moment. It may also be part of the mechanism that enables the ground reaction vector to be closer to the ankle joint in the oldest

group. The reasons for these postural changes cannot be explained by loss of joint ROM so there appears to be a functional reason for these adaptations.

4. There was a redistribution of joint moments in both stair ascent and descent seen in the oldest age group. Ankle plantarflexor moments were reduced and were compensated for by an increase in knee and hip extensor moments and hip abductor moments. The oldest age group appear to be bringing the ground reaction vector closer to the ankle joint to cause this effect. This compensation may be as a result of the ankle joint moment approaching the maximum achievable torque in older adults, when there is more reserve in the knee and hip muscles.
5. Use of a handrail improved mediolateral stability in all older adults, but more so in the oldest age group. The use of the handrail also resulted in more subjects being able to attempt the task, again more so in the oldest group. Handrail use reduced hip, knee and ankle moments, which may bring these into the safe limits of operation for the oldest adults.

6.2 Implications for health professionals working with older adults

The findings of this study are useful for health professionals working with older adults who may or may not be experiencing difficulties with stairs. The ROM demands of normal stair performance are presented and may be used as targets for older adults with ROM losses that prevent normal reciprocal gait.

The adaptations that the oldest participants made to their gait resulted in reduced demands at the ankle joint. Older adults have been found to operate close to their maximum limits at the ankle (Reeves et al., 2009) and it may be that age related strength reductions result in the need to alter the biomechanical strategy for stairs. It would seem appropriate that professionals working with older adults should assess muscle strength, especially at the ankle. A recent Cochrane review (Lui and Latham, 2009) concludes that progressive resistance strength training can improve muscle strength and function in older adults. Ferri et al. (2003), specifically investigated plantarflexor strength and power changes in response to resistance training in 16 older adults (aged 65-81 years). They found that a 16-week programme of training three times a week at 80% of the 1 repetition maximum resulted in an increase in peak isometric plantarflexor torque by 12.4%, and peak power by 33%. Improving strength and power would provide greater reserves for older adults operating near their functional limits.

The postural changes observed in the oldest group may be as a result of hip flexion contractures and it would be appropriate for health professionals to assess these. It is important to bear in mind that this may be a compensatory mechanism so it is important not to correct hip position without considering hip, knee and ankle strength.

The other role which health professionals can have in to help older adults to access stair rails in their home environments. Use of a handrail enhanced stability and reduced demands on the lower limbs both in stair ascent and descent. If an older adult is experiencing difficulty with stairs then attempting stairs with a handrail may improve independence.

6.3 Implications for designers

The EQUAL project, of which this work was a small part, had an aim to influence designers to create environments that would be inclusive to all. This work on stair ascent and descent gives designer information on some of the difficulties older adults experience negotiating the environment. It reinforces the benefits handrails have for older adults and demonstrates how increasing step height will increase the demands of stair negotiation.

In the UK new buildings must comply with building regulations (Building Regulations, 2000). These regulations have been considered in relation to the Disability Discrimination Act of 1995 (DDA, 1995) which ensures that the needs of adults with mobility impairments are met. Guidance is given on staircase design, advising the necessity for handrails and the need for the rise of steps to be between 150-170mm. These standards apply to new public buildings and should address some of the difficulties older adults experience in their environment.

6.4 Recommendations for further study

This study has highlighted the biomechanical changes experienced with increasing age in a large group of healthy older adults. Several hypotheses have been proposed to explain why these changes occur but there are many areas within this work where further development would be beneficial.

Due to the constraints of the equipment available to this project a staircase with a single force platform was used. Although this provided biomechanical information for the full gait cycle it was not possible to determine if changes may have been due to alterations in the distribution in push-off forces from the trailing leg and pull-up forces in the lead leg. Knowing this may assist in the training of stair gait with older adults experience difficulties as it may be possible to advise someone to push or pull harder. It is therefore recommended for further studies that a minimum of two force platforms be used on adjacent steps.

It is always difficult to assess biomechanics in a normal environment due to the nature of the laboratory equipment. In all reported trials, custom built staircases have been constructed in biomechanics laboratories. These staircases are limited in size due to the testing environment and there is a possibility that this will impact the style of gait adopted in a laboratory setting. As technology improves, testing on larger staircases which better represent the home environment may become possible and this would be a useful progression in research. This would highlight whether people maintain the same movement pattern as they progress up a staircase and may be able to demonstrate any problems resulting from fatigue.

In addition to larger staircases and multiple force platforms, it would also be beneficial to determine how the upper limbs are used on the handrails. There was a reduction in the lower limb moments when using a handrail which suggests that there is some load being taken through the upper limbs. At the current time there is no literature suggesting how much older adults rely on the upper limbs to reduce the loads on the lower limb, although it appears that with increasing age the dependency on hand rails becomes greater. In order to investigate this, it would be necessary to instrument the hand rails with force transducers. Comprehensive biomechanical modelling of the upper limb would give an impression of how these upper limb loads are dealt with by older adults.

In order to gain more clarity regarding the physical changes that influence the biomechanics, it would be beneficial to investigate isokinetic muscle strength of all lower limb joints and compare these findings to 3-D motion analysis of the same subjects. To date there is not a study that investigates the 3-D biomechanical demands of stair climbing and descent related to strength at the hip, knee and ankle. A study of this type would help in the understanding of the reasons for the redistribution of joint moments in the lower limb, whereas at the current time it can only be hypothesised that there is greater reserve at the hip than the ankle. In a study of this type it would also seem important to investigate joint range of moment to determine if the presence of hip flexion contractures may be resulting in the more flexed hip during stance. It would be important to consider the 2-joint muscles that cross the hip such as rectus femoris in addition to the iliopsaos muscle group, and to measure the presence of shortening of the hip flexors both statically and dynamically.

As age increases there is evidence that there is a decrease in the peak plantarflexor moment and power that can be generated. As discussed earlier in section 6.2 there is good evidence that ankle strength and power can be improved with resistive exercises. What is unknown is whether improving strength and power in the ankle plantarflexors will translate to improvements in function. It would seem beneficial to perform biomechanical analysis of stair climbing in subjects before and after a strength training programme to determine if some of the biomechanical changes observed in this study can be altered by improving ankle strength. The type of training should also be studied to determine if greater improvements are made by introducing exercises that target the rate of muscle contraction in addition to purely the amount of muscle strength.

This study, as with most studies to date, was constrained by ethics and safety to investigate only healthy older adults. This provides a data set that is useful and will assist with further research in the calculation of power values, but cannot be considered to be representative of all older adults. Future work should consider how more frail older adults could participate safely in biomechanical studies in order to gain valuable information on the strategies adopted by this group. It is likely that more frail subjects would adopt different strategies to stair climbing than a reciprocal step over step gait. The alterations that are typically made to cope with increasing frailty would be of benefit to study as these adaptations may themselves give indications of the muscle groups that would most benefit from rehabilitation.

To enable more frail participants it will be necessary to ensure that precautions are made to minimize the hazards that may result from falls. Current 3-D camera technology requires that the area is free from obstructions which make it difficult to position an assistant in the field of data capture. Improvements in technology again may make this possible or it may be that in future studies, safety feature such as overhead harnesses may become suitable options.

In conclusion this study provides biomechanical data for a large number of healthy older adults. Questions still remain as to the reasons that older adults struggle with stair ascent and descent and further research in this area would be beneficial.

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Appendix 1.1

Final report of the EQUAL project

EPSRC

Engineering and Physical Sciences Research Council

Engineering and Physical Sciences Research

Polaris House, North Star Avenue, Swindon, Wiltshire, United Kingdom SN2 1ET
Telephone +44 (0) 1793 444000. Internet http://www.epsrc.ac.uk

FINAL REPORT FORM

Grant Reference	GR/R26856/01	Programme	EQUAL
Organisation	University of Strathclyde	Scheme	
		Call	

CURRENT INVESTIGATOR DETAILS

Details	Principal Investigator	Co-investigator 1	Co-investigator 2
Title	Professor	Dr	Professor
Forename(s)	Alexander C	Bernard A	Alastair
Surname	Nicol	Conway	Macdonald
Organisation	University of Strathclyde	University ofStrathclyde	Glasgow School of Art
Division or Department	Bioengineering Unit	Bioengineering Unit	Product Design Engineering

PROJECT DETAILS

Title of Research Project	**Integration of biomechanical and psychological parameters of functional performance of older adults into a new computer aided design package for inclusive design**

Funds Awarded (£)	
Staff (£)	183049
Travel and Subsistence (£)	14491
Consumables	24412
Exceptional Items (£)	21233
Equipment	19975
Cost of Access to Services (£)	
Indirect Costs (£)	84203
Total Grant Value (£)	347363

Start Date:	1/10/2001
End Date:	30/6/2005

OBJECTIVES AND RESEARCH SUMMARY

Original objectives as presented in the grant proposal
Provide a spectrum of isometric joint strength versus joint angle, range of motion and force production endurance for older adults in the age groups of 60's, 70's and 80+ years. Provide three dimensional intersegmental loadings for the joints of the lower limbs and upper limbs during a range of daily living tasks undertaken by older adults. Provide detailed information on the perceived ability of older adults to perform functional tasks and to specify which aspects of the environment or devices result in positive or negative contributions. Generate a database of functional capacity of the older adult including data on functional impairment and functional ability. Produce a unique software package which will combine computer aided design principles with the biomechanical and psychological parameters which can be used to test the ability of older adults to use new products effectively.

These objectives remained valid throughout the research ☐ or
These objectives were changed ☒ (If so, please explain how/why in your report)

New objectives
Muscle strength and endurance measurements were modified following medical advice and after consideration of the risk of hypertension during the production of maximal effort by older adults. In order to comply with ethics and safety guidelines, the endurance measurements were omitted and the measures of maximal strength were restricted to the knee joint, the hip joint and the hand. The biomechanical measurements had initially involved all the major joints of the body for a wide spectrum of activities. All the instrumentation was available for such measures but the cumulative duration of these tests was deemed to be excessive for older adults. In order to limit the involvement of older adults to 2 visits to the Bioengineering Unit (4 test sessions, each of 2 hours), it was necessary to alter the measurements to 5 activiies of whole body movements and 4 acitivies of hand use. Details of these activities are listed in section 2.2 of the final report.

Project summary at time of proposal
This project will investigate three important factors which affect the functional performance of older adults. Firstly, the combinations and interdependency of range of motion, strength and fatigue will be measured at specific joints for 3 age groups (60's, 70's, 80+ yrs.). Secondly, the subjects' perceived ability to perform daily tasks will be assessed using standard psychological tests. Thirdly, three dimensional biomechanical measurements will be made as the subjects perform a range of functional tasks involving lower limbs and/or upper limbs (walking, stair negotiation, chair/bath/car, bottle opening, lifting kettles etc.). The results from these three measurement programmes will produce a database of functional capacity for the three age groups of older adult and will be used to develop a biomechanically based design package which will enable designers of new products to test if the product can be used effectively by older adults in terms of limb motion, strength and endurance.

Summary of Outcomes: In simple terms describe your work in such a way that it could be publicised to a general audience.
This research project has developed a new software package which can be used by designers to ensure that new products and devices will be functionally acceptable for older adults. Measurements were taken of the biomechanical requirements of older adults during numerous activities of daily living. In addition, measures were made of the maximal muscular strength of older adults together with several psychological parameters. A new database has been generated which provides designers with a wealth of information on the control and regulation of actions and the psychological factors which influence this. The research has produced detailed psychological profiles for how older adults deal with novelty and learn a new task and how they think and feel about their movement capabilities. Data from these measurements has been integrated into the new software package in the form of animations of a human model. These animations combine different sets of data to produce a "life like" motion which has coloured indicators for the functional demand needed at the joints of the body. This dynamic representation allows designers to test the functionality of new products and devices. Based on the results of the animation, the designer can modify their design until an optimum design has been achieved in terms of the use of the new device within the functional capability for older adults.
Web address with further details if you wish:

BENEFICIARIES

Intended beneficiaries at time of proposal
The combination of biomechanical and psychological parameters being integrated into the new design package software will be of great benefit to design professionals who adopt inclusive design principles. For new designs of products, which can sell to a wider age spectrum, the manufacturing base in the UK will have a competitive advantage in terms of being able to capitalise on the future 'grey' spending power of the increasing number of older adults. Such an advantage will also be effective in markets throughout Europe and the rest of the developed world. The users of products based on such inclusive design will benefit from ease of use, lower costs (compared to specialist aids) and will therefore be able to be in greater control of their daily activities. This independence will dramatically reduce the burden on the care professionals which in turn will produce a cost reduction for the care of the projected numbers of older adults in the UK.

The major benefits from the EPSRC supported research involve i) those who have joined the project and ii) the research results.

i) STAFF EMPLOYED ON THE RESEARCH PROJECT

Pages 248-249 redacted for Data Protection reasons

[redacted]

Pages 248-249 redacted for Data Protection reasons

[redacted]

ii) RESEARCH RESULTS

(a) Publications

(1) Please state how many research results arose principally as a result of the research funded through this grant and indicate the type (journal, conference, book, patent, software, other)

	JOURNAL	CONFERENCE PAPERS	BOOK	PATENT	SOFTWARE	OTHER
Total Number of Publications	1 accepted + 3 submitted	12 + 1 submitted + 1 invited (USA)			1 - as part of PhD thesis (Loudon, 2006)	2 PhD theses
Number of Refereed Publications (if different from above)						
Number of Publications That Contain An Industrial Co-Author						
Number of Publications That Contain An International Co-Author						

(2) Please list up to five significant publications that arose principally as a result of the research funded through this grant and indicate the type (journal, conference, book, patent, software, other); whether there was an international or industrial co-author and whether the publication was refereed.

AUTHOR(S)	TITLE	REFERENCE				TYPE	Refereed?	Industrial co-author?	International co-author?
		Journal/Conf.	year	vol.	page				
Potter,L M & Grealy M A	Aging and inhibitory errors on a motor shift of set task	Experimental Brain Research	2006	in press		Journal	☒	☐	☐
Loudon, D & Macdonald A S	Software tool for designers	Proceedings of "Include 2005" ISBN 1-905000-10-3	2005	CD	n/a	Conference	☒	☐	☐
Rowe PJ, Hood V, Loudon D,Samuel D, Nicol AC, Macdonald AS, Conway B	Calcuating and presenting biomechanical functional demand in older adults during activiies of daily living.	Proceedings of BioMech2005, Spain	2005	485	20	Conference	☒	☐	☐
Hood V, Nicol A C	Biomechanics of stair descent in older adults. (http://www.isb2005.org/proceedings/ISB05/prof235.html)	Proceedings of ISB	2005	website	n/a	Conference	☒	☐	☐

AUTHOR(S)	TITLE	REFERENCE				TYPE	Refereed?	Industrial co-author?	International co-author?
		Journal/Conf.	year	vol.	page				
Samuel D, Hood V, Rowe PJ	An investigation into hip muscle strength and biomechanical moments produced during chair sit down in older adults	Age and Aging	2006	in press		Journal abstracts	☒	☐	☐

(b) Other Results from the Research

Please indicate the timescale for potential exploitation (achieved, expected within 3 years, expected within 10 years, none foreseeable in 10 years) and include patent nos., dates, company names and addresses etc. as appropriate, in the 'details' and 'reference' columns.

	DETAILS	REFERENCE	TIMESCALE
Licences or Patents			
Spin Off Company			
Direct Consultancy			
Industrial Training Courses			
Other *(please specify)*			

(c) Follow-on Support

Please indicate the level of further research support that has arisen principally as a result of the work supported under this grant. Please do not include any contributions already listed under the section on project partners on the current project.

FUNDING SOURCE	DETAILS	SUPPORT (£)
EPSRC		
Other UK Research Council		
Other UK Government		
UK Industrial		
UK Other	2 PhD studentships funded by the University of Strathclyde. The projects involve further research into hand activities of older adults and into chair use by older adults	£110,000
European Commission		
Other Industrial		
Other Non-Industrial		

PROJECT PARTNERS (as identified in the original proposal)

Details	Project Partner 1 : Proposed Details		Project Partner 1 : Actual Details	
Title/Forename(s) of Contact	Ms Maureen		Ms Maureen	
Surname of Contact	O'Neil		O'Neil	
Name of partner organisation	Age Concern (Scotland)		Age Concern (Scotland)	
Division/Department	Director		Director	
Address Line 1	113 Rose Street		113 Rose Street	
Address Line 2				
Address Line 3				
Town/City	Edinburgh		Edinburgh	
Administrative Area/County				
Postal Code	EH2 3DT		E2 3DT	
Country				
DIRECT CONTRIBUTION TO RESEARCH PROJECT	**Description**	**Value (£)**	**Description**	**Value (£)**
a. cash				
b. equipment/materials/facilities				
c. secondment of staff	Vounteer subjects	8000	Volunteer subjects	4800
d. other				
Sub-Total		8000		4800
INDIRECT CONTRIBUTION TO RESEARCH PROJECT	**Description**	**Value (£)**	**Description**	**Value (£)**
a. use of facilities/equipment				
b. staff time	Advice and guidance	3000	Advice, guidance and contacts in other older adult organisations	4000
c. other				
Sub-Total		3000		4000
Total Contribution		**11000**		8800
Please feel free to comment on the nature of the partnership, particularly where the original partnership arrangements were modified or extended.				

.0

Details	Project Partner 2 : Proposed Details		Project Partner 2 : Actual Details	
Title/Forename(s) of Contact	Mrs Jean		Mrs Jean	
Surname of Contact	Miller		Miller	
Name of partner organisation	The Abbeyfield Society		The Abbeyfield Society	
Division/Department				
Address Line 1	27 Rodger Drive		27 Rodger Drive	
Address Line 2				
Address Line 3				
Town/City	Glasgow		Glasgow	
Administrative area/County				
Postal Code	G73 3QY		G73 3QY	
Country				
DIRECT CONTRIBUTION TO RESEARCH PROJECT	**Description**	**Value (£)**	**Description**	**Value (£)**
a. cash				
b. equipment/materials/facilities				
c. secondment of staff	Volunteer subjects	8000	Volunteer subjects	6000
d. other				
Sub-Total		8000		6000
INDIRECT CONTRIBUTION TO RESEARCH PROJECT	**Description**	**Value (£)**	**Description**	**Value (£)**
a. use of facilities				
b. staff time	Advice and guidance	3000	Advice, guidance and information/contacts about other older adult groups	4000
c. other				
Sub-Total		3000		4000
Total Contribution		**11000**		10000
Please feel free to comment on the nature of the partnership, particularly where the original partnership arrangements were modified or extended.				

Use of Services and Facilities

Service or Facility	Details of service or facility used (where appropriate)	Units Awarded	Units Used

GRANT ADDITIONAL/SPECIFIC CONDITIONS AND INFORMATION

Additional and Specific Conditions/Information

Signatures:

Principal Investigator	Head of Department	Research Grants Office
A. C. Reid	A. C. Reid	
Date: 4/11/05	Date: 4/11/05	Date:

COMPLIANCE WITH THE DATA PROTECTION ACT 1998

In accordance with the Data Protection Act 1998, the personal data provided on this form will be processed by EPSRC, and may be held on computerised databases and/or manual files.

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The information held will be retained for no longer than necessary.

If you have any queries regarding this statement please contact the Data Protection Officer for EPSRC, telephone 01793/444100.

ADDITIONAL CO-INVESTIGATORS

Name	Organisation	Division or Department
Dr Philip Rowe	Queen Margaret University College	Department of Physiotherapy
Dr Madeleine Grealy	University of Strathclyde	Department of Psychology

ADDITIONAL PROJECT PARTNERS

Name	Partner Organisation	Address	Value (£)
Mrs Lesley Hart	University of Strathclyde	Senior Studies Institute, George Street, Glasgow	10000
Mr Robert Murray	Glasgow City Council	Community Care, India Street, Glasgow	5000

FINAL REPORT SELF-ASSESSMENT

Grant Reference: GR/R26856/01	Project Title: Integration of biomechanical and psychological parameters of functional performance of older adults into a new computer aided design package for inclusive design.
This sheet is an integral part of the Final Report Form and should be returned with it.	Investigator: Professor Alexander C Nicol Institution/Organisation: University of Strathclyde

Please note that your self assessment form will be forwarded to the external assessors. If, in the context of this particular project, you consider one or more of the assessment criteria "not applicable" please tick the relevant box. Please indicate, in the box provided, your perception of the importance of each assessment criteria in relation to the objectives of the project (H=High, M=Medium, L=Low).

Criterion \ Assessment	Not Applicable	Unsatisfactory	Tending to ←	National standing	Tending to →	Internationally leading	Relevance to the Project H/M/L
Research Quality	☐	☐	☐	☐	☐	☒	H
Research Planning and Practice	☐	☐	☐	☐	☐	☒	H
Potential Scientific Impact	☐	☐	☐	☐	☐	☒	H

Criterion \ Assessment	Not Applicable	Unsatisfactory	Tending to ←	Good	Tending to →	Outstanding	Relevance to the Project H/M/L
Quality of Training & Experience Provided	☐	☐	☐	☐	☐	☒	H
Communication of Research Outputs	☐	☐	☐	☐	☒	☐	H
Potential Benefits to Society	☐	☐	☐	☐	☒	☐	H
Cost-effectiveness	☐	☐	☐	☐	☒	☐	H

Final Report: GR/R26856/01

Integration of biomechanical and psychological parameters of functional performance of older adults into a new computer aided design package for inclusive design.

1: Recruitment of volunteer participants

The recruitment of older adults was performed initially through the project collaborators. This was followed by a mail shot and several presentations to older adult organisations throughout central and west Scotland. In total, over 1200 recruitment leaflets were distributed, from which 125 people volunteered. In order to comply with the requirements of ethics and safety committees, participants were required to undertake medical screening to determine their suitability to participate in all aspects of the project, both cognitive and physical. This screening resulted in the exclusion of 41 potential volunteers which resulted in 84 volunteers who were able to participate in the study. The distribution of age and sex was as follows: 15 males and 15 females aged 60-69 years, 15 males and 15 females aged 70-79 years and 13 males and 11 females aged over 80 years. These subjects were from a wide range of social, economic and educational backgrounds. Subjects attended the Biomechanics Laboratory at the University of Strathclyde for four 2-hour sessions (2 sessions per visit). The first visit involved the measurement of muscle strength followed by lunch then data collection for hand biomechanics. The second visit involved data collection for the whole body biomechanics followed by lunch then the psychological assessment.

2: Methods

2.1 Strength measurements:

To estimate the maximal isometric strength of large muscle groups, a special strain gauged torque transducer was designed, constructed and validated for use. The torque transducer proved capable of measuring muscle moments with a precision of 0.1 Nm through the operating range of +/- 300 Nm.
The torque transducer was attached to a custom built test couch using a mounting system which allowed the torque transducer to be aligned with the centre of rotation of the joint and which also allowed isometric strength to be measured at a range of joint angles. Initially, it was the intention to measure maximum strength at all the major joints of the body. However, the pilot study of younger adults indicated that such a scheme would be unduly taxing for older adults and could place participants at risk of developing significant hypertension during the course of prolonged periods of maximal strength production. In addition, a Health Promotion England report published by the DTI in 2001 also indicated that the main causes of fatal accidents for those aged over 65years are from falls on stairs and steps (62%), transits between two levels such as rising from a chair (15%), and mobility on the same level (13%). Accordingly, for this aspect of the project, the decision was taken to focus on the measurement of hip and knee strength for the lower limb and to measure hand grip strength for the upper limb.
For each participant, considerable care was taken to ensure that measurements of strength at the hip and knee were consistent with regard to whole body posture and that all force measurements were isolated to the muscle groups under investigation by using adjustable postural restraints. Maximal muscle moments at the knee joint were measured in flexion and extension at knee angles of 90, 60 and 20 degrees of flexion. Maximal flexion/extension moments at the hip joint were measured at 45, 30 and 0 degrees of hip flexion. Hip abduction/adduction moments were measured with the hip in the neutral position. Hand grip strength was measured using a commercially available strain gauged dynamometer. In addition, functional ability was recorded using a self-reported questionnaire for daily activities and an indication of quality of life was obtained using the SF36 general health questionnaire.

2.2 Biomechanics of whole body movements and hand activities:

Five activities of daily living involving the whole body and four activities involving the more dextrous movements of the hand were chosen to be investigated. The activities were selected after a review of literature relating to the activity profiles of older adults and to the frequency of accidents in the home. The selected activities were considered to be the fundamentals that could be integrated into the proposed design tool to enable designers to work with a range of postural and dynamic activities.
The whole body movements were:

1. Straight line level walking (normal gait)
2. Walking combined with opening and closing a door
3. Ascending and descending a staircase with and without the use of a handrail
4. Rising and sitting on a chair with and without the use of arm rests
5. Lifting a can to a high and low shelf

The tasks focussing on hand use were:

1. Operating a television remote control
2. Removing and replacing a jar lid
3. Drinking from a glass
4. Turing a key in a lock

For the whole body movements, three-dimensional motion analysis was performed using an 8 camera Vicon Motion Analysis System (120 Hz), combined with ground reaction forces measured by Kistler force platforms. A full body marker placement protocol was developed to enable identification of bony landmarks whilst minimising artefacts caused by soft tissue movement. This followed best practice recommendations from the International Society of Biomechanics. A doorframe, staircase and chair were commissioned for use in the measurement sessions and they were designed to facilitate normal activity and to enable the 8 cameras to view the skin markers on the subjects. Subjects performed three practice sessions of each activity and data were captured for three subsequent repetitions of each activity.

For the hand activities, the Vicon system was used with the cameras in a small volume setup. An experimental marker system was developed in order to indicate the positions of the major joints of the fingers and hand and to enable reconstruction of the bony anatomy. The remote control activity involved each subject performing 10 simple button presses on a standard television remote control (i.e. Turn ON, Turn OFF, Change channel to 1 etc.) with a rest period between these actions. This was followed by tasks which required the striking of a more complex set of key numbers such as would be required for using teletext or entering a phone number on a hand held phone. Retro-reflective markers were attached to the remote control so that it was possible to determine the accuracy of button presses, together with the preferred orientation of the hand. Three repetitions were performed of the jar and drinking tasks with no constraints on the method of performing this activity. To examine key turning, a purpose built strain gauged transducer was used to simulate the turn of a key in a lock. Resistance to motion could be modified by adjustment of a rotary viscous damper. Three repetitions of a key turn were performed at two fixed resistances. The forces from the transducer were combined with the marker position data to determine the moments applied to the joints of the index finger.

Vicon data and the data from the force transducers were combined and processed using Vicon *Bodybuilder* software. Custom written software models were developed to enable the reconstruction of each subject's anatomy and the calculation of 3-dimensional joint angles and moments for the appropriate joints. Further software was developed to enable export of the body segment positions and joint moments into the software environment used for development of the design tool.

2.3: Assessment of psychological parameters:

2.3.1 The Perceived Motor Efficacy Scale.

As a precursor to the main project, a large pilot study was undertaken to develop a scale for measuring the beliefs older people have about their motor abilities. Questionnaire items were developed through interviews with older adult volunteers and academics, and these were administered to 300 healthy older adults aged 60-96 years. Factor analysis of their scores resulted in 10 subscales which demonstrated strong internal reliability. This was replicated using a second sample of 167 older adults aged 60-92 years. The analysis of the questionnaire data indicated that were no significant differences between the perceived motor-efficacy scores of older adults in their 60's and 70's, except for perceived confidence in motor ability in the face of aging where people in their 60's scored significantly higher than those in their 70's ($p < .05$). Participants in their 60's showed significantly higher perceived ability than those in their 80's for all subscales ($p < .05$), except for perceived ability to co-ordinate precise movements, physical endurance, manual ability, and motor ability relative to same age peers for which those in their 80's showed significantly higher scores than those in their 60's ($p < .05$). Finally, participants in their 70's scored significantly higher than those in their 80's for their perceived ability to co-ordinate precise movements and motor ability in demanding and novel motor contexts (all $p < .05$). The final questionnaire was then administered to the 84 participants who took part in main research project.

2.3.2 Assessing inhibitory abilities.

a: The first task assessed the ability to inhibit a previously established action in favour of a novel one. In everyday life there are many circumstances when this type of inhibition is essential, for example, when driving a new car in which the switches are on the opposite side to previous vehicle. To assess this ability, participants were given a task which involved reaching and grasping a cup. During the experiment, participants were repeatedly presented with the cup in an easy grasp orientation and then the cup was unexpectedly switched to a novel and more complex grasp orientation.

b: The second assessment of inhibition was the ability to resist executing an obvious but inappropriate action. Again this ability is required in many everyday tasks in which an individual's planned or current actions may be rendered inappropriate because of changes in how objects are to be used. To assess this, a wiping task was developed in which a sponge was moved to the left and to the right over a horizontal surface. Each participant was asked to copy all of the experimenter's movements except when the experimenter lifted their sponge up/down. In these circumstances, the participant was to attempt to continue with the ongoing action of wiping to the left and right instead. The number of inhibition failures was recorded, as well as the type and number of other motor errors that occurred when participants succeeded to inhibit the lifting action but did not manage to continue undisturbed wiping back and forth on the left/right plane of movement.

c: The third experiment was concerned with the ability to inhibit irrelevant visual information. The investigation assessed the inhibition of visual distractors in a task which involved searching for particular jars of coloured pasta amongst a display of similar or dissimilar jars. The experiment was designed to address four questions:(i) Do older adults have greater problems in finding objects when the surrounding objects share one of more feature to the target item? (ii) Do older adults find it harder to ignore shape or colour when searching(iii) Are older adults able to stop themselves from searching a display when they have found what they are looking for?(iv) Do older adults show a greater tendency to return to a previous target location when starting a new search?

3: Results

3.1 Muscle strength and quality of life:

In summary, the knee flexion/extension moments measured through a range of joint angles showed significant differences with joint angle, gender and age. The knee flexion moments generated throughout the range of joint angles varied by +/- 10% of the mean value while knee extension moments varied by +/- 35%. Females produced 59 – 68% less moment than males and the subjects in their 80's had only 76-84% of the strength of the subjects who were in their 60's and 70's. Similar trends were noted for hip flexion and extension and for hip abduction and adduction. Left and right hand grip strength showed significant gender based differences but did not show an age-based difference. These results demonstrate that the strength available to an individual during a functional task varies significantly depending on joint angle, gender and age. These variations in strength are large. For example, the isometric strength capacity of an elderly female subject can be half to one third of the strength of a younger male subject.

There was little associated loss of range of motion of the joints in the population studied and hence it would seem that flexibility of the joints is maintained with increasing age when no overt pathology intervenes. However, loss of strength was found to correlate with lower functional ability and reduced quality of life as measured by the SF36 questionnaire. In particular a reduction in muscle strength was associated with lower "physical functioning", more "bodily pain", less "vitality", less "social function", and less "role emotional" ($p<0.05$ for all aspects). These findings are significant and indicate that reduced strength is an important factor in the reduction of quality of life for older adults.

3.2 Biomechanical parameters:

The data collection sessions involved 900 hours of laboratory time and produced a large quantity of information. In addition to these results being integrated into the design tool, they were also analysed in order to determine age related changes in performance of activities of daily living. For example, the activity of stair descent was shown to be an activity that required a high level of muscular strength. Subjects performed three stair descents on an instrumented staircase with and without the use of a handrail. All subjects in their 60's could do this activity, however 10% of the 70 year olds and 17% of the 80 year olds were unable to descend the stairs without using the handrail. The strategy of stair descent changed in the 80 year olds with subjects being slower and producing a larger knee extensor moment in the early stance phase compared to younger subjects who tended to use their ankle extensors to a greater extent. The peak moments generated throughout the whole step cycle did not differ significantly between the ages. The moments produced by the ankle extensors were between 1.02 and 1.13Nm/kg of body mass which suggests that there is a minimum strength requirement for this activity. Since muscle strength was found to decline across the decades for these subjects, this result indicates that some of the oldest adults were working very close to their maximal functional ability.

In order to explore the contribution which muscle strength may have in determining functional ability, the muscle strength data were combined with the biomechanical moment and angle data for the 5 functional tasks to determine the "functional demand" placed on the muscles by a particular functional task.

Functional demand for a muscle group was defined as the muscle moment required at a particular joint angle, divided by the maximum isometric muscle strength available at that joint angle (expressed as a percentage).

For the knee flexors, the functional demand was lowest during stand-to-sit (8%) and sit-to-stand (10%), it increased for stair ascent (42%) and was highest in stair descent (73%) and gait (75%). For the knee extensors, the functional demand was higher for all 5 mobility tasks. Three quarters of the available isometric extensor muscle strength was utilised when using a chair (69% for stand-to-sit and 73% for sit-to-stand), while gait and stair negotiation required moments in excess of the maximum isometric muscle strength available (gait 101%, stair ascent 103%, stair descent 120%). These values which exceed 100% can be explained by the fact that dynamic contractions have been shown to be able to generate moments in excess of the maximum isometric moment.

For the hip, the levels of demand were still high but somewhat lower than for the knee. For the hip flexors the functional demand was lowest during sit-to-stand (22%) and stand-to-sit (24%) , it increased for stair ascent and stair descent (each 43%) and was highest during gait (101%). For the hip extensors, the functional demand was higher for all mobility tasks except gait (sit-to-stand 88%, stand-to-sit 74%, stair ascent 89%, stair descent 51% and gait 68%).

From these data it can be concluded that mobility functional tasks represent a demanding challenge to the musculoskeletal system of the older adult. When these findings are incorporated with those reported above on quality of life it is evident that muscle strength is an important contributor to functional ability and to quality of life. While the work presented in this report is a first attempt to quantify the functional demand of everyday tasks as experienced by older adults, the data presents a case for interventional studies aimed at improving muscle strength in older adults. Such a strategy would ensure that older adults could benefit from safer, independent lives which would increase their quality of life.

3.3 Psychological parameters:

In the experiment which assessed the ability to switch from a well learnt task to a new task, it was found that although 10% of older adults performed similarly to young adults, the majority failed to plan a new movement. Nearly 30% of the older adults failed to learn on the second attempt, and of these 80% in their 70's-80's failed to learn at all. They were aware of these difficulties even though they were unable to stop themselves from making mistakes. Thus designers must be aware that objects or environments which require the ability to shift from an established to a novel 'way of doing things' will present problems for people over the age of 60 and that during these tasks the person's attention is likely to be focused on correcting for unintentional errors.

The second experiment studied the ability to resist executing an obvious but inappropriate action. The data indicated that from the age of 60, the frequency and variety of motor errors became more extreme with age. Apart from making gross mistakes, there were significantly more pauses or stoppages in movements than for younger adults. Lacking in this ability can have serious consequences, for example, a common cause of car accidents with older drivers is that, when suddenly required to stop, they instinctive press down with their foot but fail to move it from the accelerator to the brake pedal.

The third experiment looked at how people search for objects. It was found that older adults can automatically extract distinctive features (just like younger adults), so the design of new objects should incorporate distinctive features as a way of directing the user's attention to the relevant part of the object (such as a particular switch). However, the search time is much slower when the complexity is increased. For example, it is more difficult to search for multiple symbols, colours and shapes, as is often the case for visual displays on objects like mobile telephones and computer screens.

4: The Design Tool

In order to develop the specification for a new product, and ultimately to embody features and qualities into a product solution, designers must synthesise many different types and qualities of information. For the process of inclusive design, it is vital to consider information on the capability of the user. However, to date there has been very limited information on the biomechanics involved in product interaction, particularly on the effects of ageing on biomechanical capability.

In addition to the lack of information, currently available information is too often presented in a format that is at best onerous to use, at worst incomprehensible, and which fails to acknowledge the needs of the designer or the processes of design. As the interactions of people with products and environments are dynamic, the typically static representation of information in statistical tabulation formats proves of questionable value to a designer. There is a real risk of a designer misapplying the information, which requires specialist knowledge, interpretation and calculation. These current information sources also fail to recognise the environment and tools which designers work with, especially the predominant use of computer aided design (CAD) software in the design process.

4.1 Development of Inclusive CAD Software:

The biomechanics data and functional demand data would be of great interest to designers if such information could be accessible and were made usable within established design processes. However, the observation of numerical data or graphs of joint moments, joint angles and functional demand data was found to be time consuming, to require skill in interpretation and a level of biomechanical comprehension and training beyond most of the community which we wish to reach, involve and educate – particularly designers.

For these reasons, the 3D movement of the subject was reanimated for the activities listed above. This was achieved by the creation of a software application, implemented in Visual C++ and OpenGL, which displayed real-time generated 3D animated visualisations of participants performing the activities. The animated human model consists of simple cylindrical or block representations of the body segments which were rendered from the Vicon data and displayed in a light blue colour. The hands were displayed as 3D hand models shaded as flesh tone. The real-time animation shows the subject performing the functional task in a realistic way. In order to visualise the functional demand of joints, each segment was connected to the next segment by a node representing the joint as a variable-colour sphere. The level of functional demand was represented on a continuous colour gradient from green through yellow to red. A green colour is displayed when functional demand is below 40%. When the functional demand for that joint becomes higher but still within acceptable limits (between 40% and 80%), the sphere turns from green to yellow to orange. Finally, if the joint were to require a level of functional demand deemed to be unacceptable (above 80%), the sphere turns from orange to red. This 'traffic light' system is considered to be a clear and immediate way of allowing the designer to understand the functional demands placed on an individual while performing daily activities. At the same time, the colour coding will allow designers to understand the design implications of the products and environments developed in response to need.

The software was integrated with the engineering CAD package SolidWorks, a popular tool with product designers, implemented as a plug-in in C++, using the Solidworks Application Interface. This enables a virtual model of a product to be imported and attached to the hand of the human model, and the effects of the product weight on the arm moments to be calculated. The arm of the human model can be manipulated by moving the hand with the mouse pointer, and the arm position resolved using inverse kinematics. Hence it is possible to evaluate the effects of different upper limb configurations and different designs of objects held in the hand on the moments generated during the functional task. The effects of varying the parameters in the CAD model are calculated immediately and can be quickly visualised. For the example of a jug kettle, parameters such as the shape of a kettle, material properties, handle position/orientation, and water content can be varied and immediate feedback is given of the change in demands the design of a particular kettle places on the older adult.

5: Research Impact

Successful performance of everyday tasks requires both cognitive and motor abilities. The new database from the results of this project provides designers with a wealth of information on the control and regulation of actions and the psychological factors which influence this. The research has produced detailed psychological profiles for how older adults; deal with novelty and learn a new task, make or avoid action errors in demanding situations, search complex scenes for a desired object, and how they think and feel about their movement capabilities. The specific parameter of "functional demand" has been shown to be an effective way to reduce the need for the design process to rely on specialist knowledge. This is achieved by combining moment data, angle data, and muscle strength, and simply expressing it as *how hard the muscles are working as a percentage of their maximum capability*. This is something which the lay person can understand, and provides a medium for the communication of important biomechanical results to those beyond the boundary of the biomechanics knowledge domain.

The choice of animation combined with the visual "traffic light" representation of functional demand is an effective method of communicating the functional demand of tasks of everyday living, as it is in a form which can be immediately appreciated and understood across all interested parties, with potential applications in both design and healthcare.

The integration of the tool with CAD software enables the designer to obtain feedback quickly, encouraging several iterations of product design/analysis. The first generation model of the biomechanical interaction of a virtual product model with a virtual human model has shown potential as a means of assessing the demands which a product places on users (eg. weight, centre of mass, handle position). Initial feedback from a sample of professional designers who have used the software has been very encouraging.

This EQUAL research project has produced a first generation design tool. The next stage has been commissioned at the Glasgow School of Art and will involve an evaluation by designers, health and social care providers, older adults and their carers, and human factors experts. The results of this evaluation will generate a series of guidelines and strategies for designers. These will allow designers to determine the design characteristics of potential solutions for products, interfaces and environments that are able to safely accommodate the functional limitations of older users. Already, from discussions with biomechanics researchers, several guidelines and 'rules of thumb' have been identified which can be given to designers when considering the biomechanics of older users. As the animations of the individuals performing the tasks are from motion capture, the motions have a realistic 'life-like' quality. There are differences, often subtle, in the way that different individuals perform the same activity and which have a corresponding effect on the functional demand at joints. The design tool is able to give the designer a "feel" for the experience of the user. The potential of the software package to be used as an empathetic design tool will be evaluated as part of the above commission. Such a package could be used in the early stages of design practice where freehand sketches are often used in preference to CAD modelling.

6: Dissemination

6.1: User groups and designers:

A special seminar was held in June 2005 at the University of Strathclyde to present the results of the research and to obtain comments about the database and the design tool. The seminar was attended by all of the investigators, the collaborators and 40 of the volunteer older adults who took part in the research. There was active discussion about the results in terms of the aging process and independent living. The delegates were very impressed with the series of animations produced by the design tool and commented that the "traffic light" system of functional demand was easy to understand and gave a dynamic image of the muscular effort produced.

Following this seminar, the database and the design tool have been presented to designers who are active within the field of inclusive design. The initial response has been extremely encouraging and the ability to perform iterations of design detail has been given particular praise. It is planned to extend this dissemination to a wider group of designers via conferences/seminars in 2006. Funding has also been secured to enable David Loudon to finalise a website which will contain the content of the database and examples of the application of the design tool.

6.2: Conference presentations:

The output of this multidisciplinary research is of interest to numerous professional bodies and learned societies. Twelve papers reporting the progress of the research have already been presented to the following national and international conferences:

CWUAAT 2004, 2nd Cambridge Workshop for Universal Access and Assistive Technologies.
Physiotherapy Research Society, Manchester, 2004.
British Geriatrics Society, Birmingham, 2004.
BioMech2005, International Association of Science and Technology for Development conference on Biomechanics, Benidorm, Spain.
ISB2005, The Congress of the International Society of Biomechanics, Cleveland, USA (4 papers).
ESMAAC2005, European Society of Movement Analysis of Adults and Children, Barcelona, Spain.
Society for Rehabilitation Research, Southampton, 2005.
Include 2005 The international conference on inclusive design at the Helen Hamlyn Research Centre, Royal College of Art, London.
SPARC 2005 The launch of the new initiative for research into aging, University of Strathclyde, Glasgow.
CWUAAT2006, 3rd Cambridge Workshop for Universal Access and Assistive Technologies (submitted).
35th Annual meeting of the American Aging Association, Boston, USA (June 2006 – invited paper)
Abstracts from these conference presentations have been accepted for publication in the following peer-reviewed journals: Clinical Rehabilitation, Physiotherapy, Gait and Posture, Age and Aging.

6.3: Journal publications:

The results of the research are worthy of publication in a wide range of journals. To date, four papers have been submitted to the following peer-reviewed journals:

Experimental Brain Research (accepted September 2005)
Journal of Experimental Psychology
British Journal of Psychology (2 papers)

<u>Appendix 3.1</u>
Information for participants

You can decline to participate in this study giving a reason and your decision to decline will be accepted without question.

Purpose of the study:

There are two specific aims to this project. Firstly, it is proposed to provide a database of biomechanical information, which will describe the factors responsible for the limitation of functional performance in the senior citizen. Secondly, the information contained in the database will be coupled with the philosophy of inclusive design to produce a novel design package, which will animate the muscolo-skeletal system combined with the object, device or environment under design. In this way it will be possible to test whether the older adult will be able to perform the activity or task using the new object or device.

There is no financial reward associated with your participation in this experiment. The Bioengineering Unit will provide free transport and lunch for all subjects.

Who should volunteer?
This study requires the participation of senior citizens with no history of neurological conditions.

You should not volunteer if you are:

- Ill for any reason
- Suffer and/or suffered previously from a cardiovascular, respiratory or neurological condition
- Known to have previously received treatment for any neurological condition.
- Known to have a diagnosed skin condition
- Known to have diabetes
- Known to have any infectious disease
- Known to have an allergy to sticking plasters or tapes
- Known to have a history of thrombosis or have been diagnosed with blood clotting disorder
- On medicine that makes you drowsy or influences your balance

The experiment:

The experiments require that you complete a series of movement tasks of the upper and of the lower limbs. All experiments will involve the recording of the movements you have been instructed to perform. This will be achieved by the use of special markers that will be placed on your arms and/or legs and this also involves the use of special camera equipment that detects the motion of your limbs. The movements you may be asked to perform will be explained, and shown to you prior to the experiment. Generally you will be asked to perform the following tasks with 5 minutes breaks between each section.

Section 1. Displacement tasks a, Standing up from a standard arm chair (height of the seat: 43 cm). b, Making two steps ahead.
c, Going up and coming down 4 stairs with standard height (17.5 cm) and with standard run (30 cm).

Section 2. Skill Tasks
a, Opening a door.
b, Opening a window. c, Reaching for and lifting a mock or simulated kettle.

Section 3. Fine motor skills a, Grasping a remote control and entering a four-digit code b, Picking up a key, putting it in the keyhole and turning the key.

Procedure	**Risk**
Attachment/Removal ofsensors or markers to/from the skin.	1. Potential allergic reaction to the materials used to make and attach the sensors and markers. 2. Transitory discomfort as sensors or/and markers are removed from the skin.
Movement execution	1. Studies on walking, standing up from a chair and climbing 4 stairs may result in a situation where a fall could occur. The probability of falling increases with age but the experimental design will minimize this risk by allowing you to stop at any time of the experiment for a break.

Risks summarized:

The risk levels associated with participation in this study are considered to be low. However, some of the above procedures may result in short term and transitory discomfort and in studies on standing up, walking, climbing on stairs there is a risk of accidental tripping.

As a volunteer you are free to demand that an experiment is stopped and that you can withdraw from the study at any time.

IF YOU AGREE TO PARTICIPATE IN THIS STUDY YOU SHOULD COMPLETE THE ATTACHED CONSENT FORM

Declaration of Consent

Project Title: Integration of biomechanical and psychological parameters of functional performance of older adults into a new computer aided design package for inclusive design.

<u>To be completed by the subject</u>

	YES	NO
Have you read the information for participants?	☐	☐
Have you had opportunity to discuss the study?	☐	☐
Have you received satisfactory answers to your questions?	☐	☐

Who have you spoken to?…………………………………

Are you aware you are free to withdraw at any time?	☐	☐
Do you agree to participate in this study?	☐	☐

Name (please print):………………………………………………………………………

Signature:……………………………………………………………………………………..

Witness signature:………………………………… Date:…………………………….

Addresses and Telephone numbers (optional)

Subject: Wittness:

~~Appendix~~ ~~3.2~~

Bodybuilder Code

```
{This programme is the Bodybuilder code developed by Victoria Hood for determining joint}
{angles and moments collected using Vicon Workstation and the protocol and marker set}
{developed by Victoria Hood for the EQUAL project}


{*Start of macro section*}
{*=====================*}

macro SUBSTITUTE4(p1,p2,p3,p4)
{*Replaces any point missing from set of four fixed in a
segment*} s234 = [p3,p2-p3,p3-p4] p1V =
Average(p1/s234)*s234 s341 = [p4,p3-p4,p4-p1] p2V =
Average(p2/s341)*s341 s412 = [p1,p4-p1,p1-p2] p3V =
Average(p3/s412)*s412 s123 = [p2,p1-p2,p2-p3]
p4V = Average(p4/s123)*s123

p1 = p1 ? p1V p2
= p2 ? p2V p3 =
p3 ? p3V p4 =
p4 ? p4V
endmacro


macro POINTER(Anatomy,Segment)
{*Calculates the position of the end of the pointer for calibration in the technical frame it belongs
to*}
{*1st determine the "point" in the Global system and outputs it as point#Calib. Then converts the
point into*}
{*the appropriate technical reference frame and stores it as parameter $%#point#Calib*}

unitPointer=((POI1-POI2)/DIST(POI1,POI2))
Anatomy#Calib=POI1+123*unitPointer
OUTPUT(Anatomy#Calib)
PARAM(Anatomy#Calib)
%#Anatomy#Calib=Anatomy#Calib/Segment
PARAM(%#Anatomy#Calib)
endmacro


macro SEGVIS(Segment)
{*outputs a visual representaion of the segment to be viewed in the Workspace*}
{*0(Segment) is the origin of the segment*}
ORIGIN#Segment=0(Segment)
XAXIS#Segment=0(Segment)+(1(Segment)*100)
YAXIS#Segment=0(Segment)+(2(Segment)*100)
ZAXIS#Segment=0(Segment)+(3(Segment)*100)
OUTPUT(ORIGIN#Segment,XAXIS#Segment,YAXIS#Segment,ZAXIS#Segment) endmacro

macro AXES(Segment)
{*Outputs the 3 orthogonal unit vectors for the segment in order that the rotation matrices can be
defined*}
{*This is for the animation package*}
X#Segment=1(Segment)
```

Y#Segment=2(Segment)
Z#Segment=3(Segment)
OUTPUT(X#Segment,Y#Segment,Z#Segment)
endmacro

macro ColeJCS(seg1,seg2,joint)
{* Procedure to calculate the rotations about defined embedded axes using the joint co-ordinate system.

References: Cole,G.K. et al (1993). Application of the Joint Co-ordinate System to Three-dimensional Joint Attitude and Movement Representation : A Standardization Proposal. Journal of Biomechanical Engineering. November 1993 : Vol 115 : pp 344-349

aEone,aEtwo,aEthree =unit vector describing the attitude of the 1st,2nd and 3rd axis of the joint co-ordinate system between the reference segment (seg1) and the target segment (seg2), relative to an inertial reference system.

If the axes of a body segment co-ordinate system are identified as an axis of Flexion, a Longitudinal axis and a Third axis, then Fone, Lone, Tone are unit vectors that describe the attitude of the Flexion, Longitudinal and Third axes respectively, in an inertial reference system.

Input: 'seg1', 'seg2' describing the axes of the co-ordinate systems embedded in each segment.
Fone, Lone, Tone describe the flexion, longitudinal and third co-ordinate axes of the proximal segment.
Ftwo, Ltwo, Ttwo describe the flexion, longitudinal and third co-ordinate axes of the distal segment.
'joint' is the name given to the joint at which the specified segments interact.

Output: Angles of rotation about axes aEone,aEtwo,aEthree, flexion, abduction and rotation respectively. Counterclockwise rotations are chosen as positive*}

Fone=3(seg1)
Lone=2(seg1)
Tone=1(seg1)
Ftwo=3(seg2)
Ltwo=2(seg2)
Ttwo=1(seg2)

{*Defines e1 and e3*}
aEone=Fone
aEthree=Ltwo

{*Calculate the Vector or Cross Product between the Vectors*}
Va={2(aEthree)*3(aEone)-3(aEthree)*2(aEone),3(aEthree)*1(aEone)-1(aEthree)*3(aEone),1(aEthree)*2(aEone)-2(aEthree)*1(aEone)}
Vb=DIST({2(aEone)*3(aEthree)-3(aEone)*2(aEthree),3(aEone)*1(aEthree)1(aEone)*3(aEthree),1(aEone)*2(aEthree)-2(aEone)*1(aEthree)},{0,0,0})
Vc={2(Va)*3(aEthree)-3(Va)*2(aEthree),3(Va)*1(aEthree)-1(Va)*3(aEthree),1(Va)*2(aEthree)-

```
2(Va)*1(aEthree)}

{*Calculate the Scalar or Dot Product between the Vectors*}
DPone=(1(Va)*1(Ttwo))+(2(Va)*2(Ttwo))+(3(Va)*3(Ttwo))
DPtwo=(1(Vc)*1(Ftwo))+(2(Vc)*2(Ftwo))+(3(Vc)*3(Ftwo))

{*Calculates A (AA) and then e2*}
IF DPone < 0 AND DPtwo > 0 THEN AA=-1 ELSE AA=1 ENDIF
aEtwo=(Va/Vb)*AA

{*Calculate the value of r.*}
Rone={2(Fone)*3(aEtwo)-3(Fone)*2(aEtwo),3(Fone)*1(aEtwo)-
1(Fone)*3(aEtwo),1(Fone)*2(aEtwo)-2(Fone)*1(aEtwo)}
Rtwo=DIST(Rone,{0,0,0})
r=Rone/Rtwo

{*Calculate the Scalar or Dot Product between the Vectors.*}
aEtwoTonedp=(1(aEtwo)*1(Tone))+(2(aEtwo)*2(Tone))+(3(aEtwo)*3(Tone))
aEtwoLonedp=(1(aEtwo)*1(Lone))+(2(aEtwo)*2(Lone))+(3(aEtwo)*3(Lone))
rLtwodp=(1(r)*1(Ltwo))+(2(r)*2(Ltwo))+(3(r)*3(Ltwo))
FoneLtwodp=(1(Fone)*1(Ltwo))+(2(Fone)*2(Ltwo))+(3(Fone)*3(Ltwo))
aEtwoTtwodp=(1(aEtwo)*1(Ttwo))+(2(aEtwo)*2(Ttwo))+(3(aEtwo)*3(Ttwo))
aEtwoFtwodp=(1(aEtwo)*1(Ftwo))+(2(aEtwo)*2(Ftwo))+(3(aEtwo)*3(Ftwo))

IF aEtwoLonedp >= 0 THEN aEtwoLonesign=1 ENDIF
IF aEtwoLonedp < 0 THEN aEtwoLonesign=-1 ENDIF
IF FoneLtwodp >= 0 THEN FoneLtwosign=1 ENDIF
IF FoneLtwodp < 0 THEN FoneLtwosign=-1 ENDIF
IF aEtwoFtwodp >= 0 THEN aEtwoFtwosign=1 ENDIF
IF aEtwoFtwodp < 0 THEN aEtwoFtwosign=-1 ENDIF

joint#Flex=(acos(aEtwoTonedp))*(aEtwoLonesign)
joint#Abd=(acos(rLtwodp))*(FoneLtwosign)
joint#Rot=(acos(aEtwoTtwodp))*(aEtwoFtwosign)
joint#JCSAngles=<joint#Flex,joint#Abd,joint#Rot>

{*For later calculations of moments*} {*x
axis will be the floating axis*}
joint#JCS=[0(Seg1),aEtwo,aEone,xyz]
XAXISjcs#joint=aEtwo

ENDMACRO

macro FORCEVECTOR(FP)
{*This defines the quantities of force(F), moment(M) and Centre(C) from the reaction (FP)*}
{*P_#FP is the centre of pressure and is set at the forceplate centre if load is below 10N*}

If ExistAtAll( FP )
	F_#FP = FP(1)
	M_#FP = FP(2)
C_#FP = FP(3)
	if ( ABS ( F_#FP ) > 10 )
```

```
                    P_#FP = C_#FP + { -M_#FP(2)/F_#FP(3), M_#FP(1)/F_#FP(3), -C_#FP(3)
}           else                P_#FP = C_#FP           endif
            F_#FP = F_#FP + P_#FP
            OUTPUT ( P_#FP, F_#FP )
EndIf endmacro

{*Macro for Dot Product*}

MACRO DotProduct (One,Two,DotProd)
            DotProd = (1(One)*1(Two)+2(One)*2(Two)+3(One)*3(Two))
ENDMACRO

{* Macro to do a cross product *}

MACRO CrossProduct ( First, Second, Result )
            Result = { First(2)*Second(3)-First(3)*Second(2),
            First(3)*Second(1)-First(1)*Second(3),
            First(1)*Second(2)-First(2)*Second(1)}
ENDMACRO

MACRO MATRIX(Seg1,Seg2,joint)
{*Determines the 3x3 rotation matrix for rotating from one segment to the next*}

X#Seg1=1(Seg1)
Y#Seg1=2(Seg1)
Z#Seg1=3(Seg1)
X#Seg2=1(Seg2)
Y#Seg2=2(Seg2)
Z#Seg2=3(Seg2)
RotX#Seg1#Seg2={1(X#Seg1)*1(X#Seg2)+2(X#Seg1)*2(X#Seg2)+3(X#Seg1)*3(X#Seg2),1(X#
Seg1)*1(Y#Seg2)+2(X#Seg1)*2(Y#Seg2)+3(X#Seg1)*3(Y#Seg2),1(X#Seg1)*1(Z#Seg2)+2(X#S
e g1)*2(Z#Seg2)+3(X#Seg1)*3(Z#Seg2)}
RotY#Seg1#Seg2={1(Y#Seg1)*1(X#Seg2)+2(Y#Seg1)*2(X#Seg2)+3(Y#Seg1)*3(X#Seg2),1(Y#
S
eg1)*1(Y#Seg2)+2(Y#Seg1)*2(Y#Seg2)+3(Y#Seg1)*3(Y#Seg2),1(Y#Seg1)*1(Z#Seg2)+2(Y#Se
g 1)*2(Z#Seg2)+3(Y#Seg1)*3(Z#Seg2)}
RotZ#Seg1#Seg2={1(Z#Seg1)*1(X#Seg2)+2(Z#Seg1)*2(X#Seg2)+3(Z#Seg1)*3(X#Seg2),1(Z#S
eg1)*1(Y#Seg2)+2(Z#Seg1)*2(Y#Seg2)+3(Z#Seg1)*3(Y#Seg2),1(Z#Seg1)*1(Z#Seg2)+2(Z#Seg
1)*2(Z#Seg2)+3(Z#Seg1)*3(Z#Seg2)}
OUTPUT(RotX#Seg1#Seg2,RotY#Seg1#Seg2,RotZ#Seg1#Seg2)
trans#joint=joint/Seg1
OUTPUT(trans#joint)

ENDMACRO

macro LINVELACC(Point,Segment)
{*When called, this macro calculates the linear velocity in m/s and the linear acceleration in
m/s^2 of a
point, using numerical differentiation.  For numerical differentiation, reference one of the
following:
Hildebrand, F.B. (1974).  Introduction to Numerical Analysis, 2nd Edition, pp.111
Kreyszig, Erwin (1983).  Advanced Engineering Mathematics, 5th Edition, pp.793
```

Yakowitz, Sydney and Szidarovsky, Ferenc (1989). An Introduction to Numerical Computations, 2nd Edition, pp.185*}

```
$SamplingRate= 120
$FrameTimeLength=1/$SamplingRate LVel#Point=((Point[-2]-(8*Point[-1])+(8*Point[1])-
Point[2])/(12*$FrameTimeLength))/1000
LAccel#Point=((LVel#Point[-2]-(8*LVel#Point[-1])+(8*LVel#Point[1])-
LVel#Point[2])/(12*$FrameTimeLength))
output(LVel#Point,LAccel#Point)

ENDMACRO

{*END OF MACRO SECTION*}

{*Anthropometric Data (Dempsters)*}
{*==================*}

AnthropometricData
DefaultFemur 0.1 0.567 0.323 0
DefaultShank 0.0465 0.567 0.302 0
DefaultFoot 0.0195 0.5 0.475 0
DefaultHumerus 0.0280 0.564 0.322 0
DefaultForearm 0.0160 0.570 0.303 0
DefaultHand 0.0060 0.494 0.297 0
EndAnthropometricData

{*Optional points are points which may not be present in every trial*}
{*=================================================================*}

OptionalPoints(COM,THO1,THO2,THO3,THO4,C7,T8,XYPH,JUG,RUA1,RUA2,RUA3,RUA4,LU
A1,LUA2,LUA3,LUA4)
OptionalPoints(RFA1,RFA2,RFA3,RFA4,RMCP3d,RMCP3p,RMCP5d,LFA1,LFA2,LFA3,LFA4,L
MCP3d,LMCP3p,LMCP5d)
OptionalPoints(RASIS,LASIS,RPSIS,LPSIS,RPEL,LPEL,RPPE,LPPE,RTH1,RTH2,RTH3,RTH4,
RMEF,RLEF)
OptionalPoints(LTH1,LTH2,LTH3,LTH4,LMEF,LLEF,RSH1,RSH2,RSH3,RSH4,RHEE,RMFO,RL
FO,RMM,RLM)
OptionalPoints(LSH1,LSH2,LSH3,LSH4,LHEE,LMFO,LLFO,LMM,LLM,LMM1,LLM1,RACR,LAC
R
,RUS1,LUS1,RRS1,LRS1,POI1,POI2)
OptionalPoints(CalRMEH,CalRLEH,CalLMEH,CalLLEH,CalRUS,CalRRS,CalLUS,CalLRS,Can1,
Can2,Can3,DoorTR,DoorTL,DoorBR,DoorBL)

{*Substitutes missing markers based on clusters of 4 markers*}
{*=========================================================*}
SUBSTITUTE4(C7,T8,XYPH,JUG)
SUBSTITUTE4(RASIS,LASIS,RPSIS,LPSIS)
SUBSTITUTE4(RTH1,RTH2,RTH3,RTH4)
SUBSTITUTE4(LTH1,LTH2,LTH3,LTH4)
SUBSTITUTE4(RSH1,RSH2,RSH3,RSH4)
SUBSTITUTE4(LSH1,LSH2,LSH3,LSH4)
SUBSTITUTE4(RUA1,RUA2,RUA3,RUA4)
SUBSTITUTE4(LUA1,LUA2,LUA3,LUA4)
```

```
SUBSTITUTE4(RFA1,RFA2,RFA3,RFA4)
SUBSTITUTE4(LFA1,LFA2,LFA3,LFA4)

{*Defines technical axis systems for the segments from the clusters*}
{*=======================================================================*}
RightThigh=[RTH2,RTH2-RTH3,RTH3-RTH4,yxz]
LeftThigh=[LTH2,LTH2-LTH3,LTH3-LTH4,yxz]
RightShank=[RSH1,RSH1-RSH4,RSH2-RSH4,yxz] LeftShank=[LSH1,LSH1-LSH2,LSH2-
LSH3,yxz]
RightUpperArm=[RUA2,RUA2-RUA3,RUA3-RUA4,yxz]
LeftUpperArm=[LUA1,LUA1-LUA3,LUA2-LUA3,yxz]
RightForearm=[RFA1,RFA1-RFA4,RFA2-RFA4,yxz]
LeftForearm=[LFA1,LFA1-LFA2,LFA3-LFA2,yxz]
Trunk1=[JUG,C7-T8,JUG-C7,yxz]
Trunk2=[T8,JUG-XYPH,T8-XYPH,yxz]
Trunk3=[T8,C7-T8,XYPH-T8,yxz]
Trunk4=[JUG,C7-XYPH,JUG-C7,yxz]
Pelvis1=[RASIS,LASIS-RASIS,RPSIS-RASIS,yxz]
Pelvis2=[RASIS,RPSIS-RASIS,RPSIS-LPSIS,yxz]
Pelvis3=[LASIS,LPSIS-LASIS,LPSIS-RPSIS,yxz]

{*Anatomical calibration from static/pointer trials*}
{*==================================================*}
If $Static==1

%RMEF=RMEF/RightThigh
%RLEF=RLEF/RightThigh
%LMEF=LMEF/LeftThigh
%LLEF=LLEF/LeftThigh
RKJC=(RMEF+RLEF)/2
LKJC=(LMEF+LLEF)/2
%RKJC=RKJC/RightThigh
%LKJC=LKJC/LeftThigh
%RMM=RMM/RightShank
%RLM=RLM/RightShank
%LMM=LMM/LeftShank
%LLM=LLM/LeftShank
%XYPH=XYPH/Trunk1
%C7=C7/Trunk2
%JUG=JUG/Trunk3
%LPSIS1=LPSIS/Pelvis1
%LASIS1=LASIS/Pelvis2
%RASIS1=RASIS/Pelvis3
PARAM(%RMEF,%RLEF,%LMEF,%LLEF,%RMM,%RLM,%LMM,%LLM,%XYPH,%LPSIS1,%C
7
,%JUG,%LASIS1,%RASIS1)


{*From Wang (1996)*}
{*SJC is 37mm inferior, 14mm lateral and 8mm anterior to ACjt*}

Point1=JUG+{0,0,100}
Trunk=[JUG,Point1-JUG,JUG-C7,yzx]
```

```
SEGVIS(Trunk)

%RightShoulderCentreOffset={-8,-44,14} {* Accounts for half marker width of 7mm*}
%LeftShoulderCentreOffset={-8,-44,-14}
%RACR=RACR/Trunk
%LACR=LACR/Trunk
%RSJC=%RACR+%RightShoulderCentreOffset
%LSJC=%LACR+%LeftShoulderCentreOffset
RSJC=%RSJC*Trunk
LSJC=%LSJC*Trunk
%RSJC=RSJC/RightUpperArm
%LSJC=LSJC/LeftUpperArm
%RSJC1=RSJC/Trunk1
%LSJC1=LSJC/Trunk1

OUTPUT(RSJC,LSJC)
PARAM(%RSJC,%LSJC,%RSJC1,%LSJC1)

{*For locating HJC in femoral technical system for subjects who continually occlude ASIS*}
%LeftHipOffsetFactor={-0.19,-0.3,0.14}
%RightHipOffsetFactor={-0.19,-0.3,-0.14}
{*%HipOffset2={-7.25,0,0}*}
%HipOffset2={-7.25,0,0}

midASIS=(LASIS+RASIS)/2
midPSIS=(LPSIS+RPSIS)/2
Pelvis=[midASIS,RASIS-LASIS,midPSIS-midASIS,zyx]

IntASISdist=DIST(LASIS,RASIS)
%RASIS=RASIS/Pelvis
%LASIS=LASIS/Pelvis
%RHJC=%RASIS+(IntASISdist*%RightHipOffsetFactor)+%HipOffset2 {*Last bit corrects for
marker width*}
%LHJC=%LASIS+(IntASISdist*%LeftHipOffsetFactor)+%HipOffset2 {*Gives position of HJC in
pelvic frame*}
RHJC=%RHJC*Pelvis
LHJC=%LHJC*Pelvis
%RHJCthigh=RHJC/RightThigh
%LHJCthigh=LHJC/LeftThigh
PARAM(%RHJCthigh,%LHJCthigh)

{*For pointers*}
{* Will give parameter Anatomy#calib  *}

If EXIST(CalRMEH)
POINTER(RMEH,RightUpperArm)
EndIf

If EXIST(CalRLEH)
POINTER(RLEH,RightUpperArm)
EndIf

If EXIST(CalLMEH)
```

```
POINTER(LMEH,LeftUpperArm)
EndIf

If EXIST(CalLLEH)
POINTER(LLEH,LeftUpperArm)
EndIf

If EXIST(CalRUS)
POINTER(RUS,RightForearm)
EndIf

If EXIST(CalRRS)
POINTER(RRS,RightForearm)
EndIf

If EXIST(CalLUS)
POINTER(LUS,LeftForearm)
EndIf

If EXIST(CalLRS)
POINTER(LRS,LeftForearm)
EndIF


EndIf




{*Dynamic trials*}
{*==============*}
If $Static==0

{*Anatomical frame definition*}

{*Pelvis Segment*}
{*==============*}
{*This segment uses 4 markers RASIS,RPSIS,LASIS,LPSIS left on for all trials and ISB standard
for pelvis*}
{*Pelvis Offset Factors*}
{*From Bell et al. (1990) , HipOffset2 corrects for marker width*}


%LeftHipOffsetFactor={-0.19,-0.3,0.14}
%RightHipOffsetFactor={-0.19,-0.3,-0.14}
%HipOffset2={-7.25,0,0}

midASIS=(LASIS+RASIS)/2
midPSIS=(LPSIS+RPSIS)/2
Pelvis=[midASIS,RASIS-LASIS,midPSIS-midASIS,zyx]
SEGVIS(Pelvis)
```

```
IntASISdist=DIST(LASIS,RASIS)
%RASIS=RASIS/Pelvis
%LASIS=LASIS/Pelvis
%RHJC=%RASIS+(IntASISdist*%RightHipOffsetFactor)+%HipOffset2 {*Last bit corrects for
marker width*}
%LHJC=%LASIS+(IntASISdist*%LeftHipOffsetFactor)+%HipOffset2 {*Gives position of HJC in
pelvic frame*}
RHJC=%RHJC*Pelvis
LHJC=%LHJC*Pelvis


OUTPUT(midASIS,midPSIS,RHJC,LHJC)
PARAM(LHJC,RHJC)


{*HipSegments*}
{*==========*}
RHip=[RHJC,RASIS-LASIS,midPSIS-midASIS,zyx]
LHip=[LHJC,RASIS-LASIS,midPSIS-midASIS,zyx]


{*Right Thigh Segment*}
{*==================*}
{*Proposed by ISB hip comittee July17th2000*}
{*Defined from HJC calculated from the Pelvis anatomical landmarks and MEF and LEF from
calibration*}

RMEF=%RMEF*RightThigh
RLEF=%RLEF*RightThigh

midRFEs=(RLEF+RMEF)/2
RFemur=[midRFEs,RHJC-midRFEs,RMEF-RLEF,yxz]
RFemurA=[RHJC,RHJC-midRFEs,RMEF-RLEF,yxz]
SEGVIS(RFemur)
Axes(RFemur)
RKJC=midRFEs
OUTPUT(RKJC)
PARAM(RKJC)

{*Left Thigh Segment*}
{*=================*}

LMEF=%LMEF*LeftThigh
LLEF=%LLEF*LeftThigh

midLFEs=(LLEF+LMEF)/2
LFemur=[midLFEs,LHJC-midLFEs,LLEF-LMEF,yxz]
LFemurA=[LHJC,LHJC-midLFEs,LMEF-LLEF,yxz]
SEGVIS(LFemur)
AXES(LFemur)
LKJC=midLFEs
OUTPUT(LKJC)
```

```
PARAM(LKJC)

{*Right Shank System*}
{*==================*}
{*ISB standard for the tib/fibula coordinate system Biomechanics 35(2002) 543-548 is concerned
with the ankle*}
{*Have used KJC in the shank axis system from calibration and AJC (midMM) for the definition of
the y-axis*}

RMM=%RMM*RightShank
RLM=%RLM*RightShank
RAJC=(RMM+RLM)/2
RShank=[RAJC,RKJC-RAJC,RMM-RLM,yxz]
RShankA=[RKJC,RKJC-RAJC,RMM-RLM,yxz]
SEGVIS(RShank)
AXES(RShank)
OUTPUT(RAJC)
PARAM(RAJC)

{*Left Shank System*}
{*=================*}

LMM=%LMM*LeftShank
LLM=%LLM*LeftShank
LAJC=(LMM+LLM)/2
LShank=[LAJC,LKJC-LAJC,LLM-LMM,yxz]
LShankA=[LKJC,LKJC-LAJC,LLM-LMM,yxz]
AXES(LShank)
SEGVIS(LShank)
OUTPUT(LAJC)
PARAM(LAJC)

{*Right Ankle System*}
{*==================*}
{*Sets tib/fib axis system as per ISB recommendations for the ankle (see above)*}
{*The flexion axis for the ankle will be the Zaxis of this arrangement*}

RAnkle=[RAJC,RLM-RMM,RKJC-RAJC,zxy]
SEGVIS(RAnkle)

{*Left Ankle System*}
{*=================*}

LAnkle=[LAJC,LMM-LLM,LKJC-LAJC,zxy]
SEGVIS(LAnkle)

{*Right Foot System*}
{*=================*}
{*Consider this to represent the shoe rather than the foot. The markers are put on so they lie in
a*}
{*plane perpendicular to the sole of the shoe*}
```

```
If $FootLength==0
FootLength=0.152*$Height
ELSE
FootLength=$FootLength
EndIf

RmidFOOT=(RMFO+RLFO)/2
RToe=((RmidFOOT-RHEE)/ABS(RmidFOOT-RHEE)*FootLength)+RHEE
RFoot=[RToe,RHEE-RmidFOOT,RMFO-RLFO,yxz]
SEGVIS(RFoot)
AXES(RFoot)
OUTPUT(RmidFOOT,RToe)

{*Left Foot System*}

LmidFOOT=(LMFO+LLFO)/2
LToe=((LmidFOOT-LHEE)/ABS(LmidFOOT-LHEE)*FootLength)+LHEE
LFoot=[LToe,LHEE-LmidFOOT,LLFO-LMFO,yxz]
SEGVIS(LFoot)
AXES(LFoot)
OUTPUT(LmidFOOT,LToe)

{*Trunk system*}
{*============*}
{*ISB standard for the trunk from the recommendations of the ISG*}
{*Have changed the X and Z axis from ISG to be consistent with the LL system*}
{*Uses C7, T8, IJ (JUG) and PX (XYPH)*}

If EXIST(XYPH) {*This replaces XYPH with it's virtual point if it is excluded*}
ELSE
XYPH=%XYPH*Trunk1
ENDIf
If EXIST(C7) {*This replaces C7 with it's virtual point if it is excluded*}
ELSE
C7=%C7*Trunk2
ENDIf
If EXIST(JUG)
ELSE
JUG=%JUG*Trunk3
OUTPUT(JUG)
ENDIf

midC7andIJ=(C7+JUG)/2 midT8andPX=(T8+XYPH)/2
Trunk=[JUG,midC7andIJ-midT8andPX,JUG-C7,yzx]
TrunkA=[midT8andPX,midC7andIJ-midT8andPX,JUG-C7,yzx]
SEGVIS(Trunk)
AXES(Trunk)


{*Right humeral system*}
{*===================*}

RSJC1=%RSJC1*Trunk1
```

```
RSJC=%RSJC*RightUpperArm
RMEH=%RMEHcalib*RightUpperArm               {*position of point in global system*}
RLEH=%RLEHcalib*RightUpperArm
OUTPUT(RMEH,RLEH)
REJC=(RMEH+RLEH)/2
OUTPUT(RSJC,REJC,RSJC1)
RHumerus=[REJC,RSJC-REJC,RMEH-RLEH,yxz]
RHumerusA=[RSJC,RSJC-REJC,RMEH-RLEH,yxz]
SEGVIS(RHumerus)
AXES(RHumerus)

{*Left humeral system*}
{*===================*}

LSJC1=%LSJC1*Trunk1
LSJC=%LSJC*LeftUpperArm
LMEH=%LMEHcalib*LeftUpperArm
LLEH=%LLEHcalib*LeftUpperArm
LEJC=(LMEH+LLEH)/2
OUTPUT(LSJC,LEJC,LSJC1)
LHumerus=[LEJC,LSJC-LEJC,LLEH-LMEH,yxz]
LHumerusA=[LSJC,LSJC-LEJC,LLEH-LMEH,yxz]
SEGVIS(LHumerus)
AXES(LHumerus)

{*Shoulder systems*}
RShoulder=[RSJC,midC7andIJ-midT8andPX,JUG-C7,yzx]
LShoulder=[LSJC,midC7andIJ-midT8andPX,JUG-C7,yzx]

{*Right forearm system*}
{*====================*}


RUS=%RUScalib*RightForearm
RRS=%RRScalib*RightForearm

RWJC=(RUS+RRS)/2
PARAM(RWJC)
OUTPUT(RWJC)
RForearm=[RWJC,REJC-RWJC,RUS-RRS,yxz]
RForearmA=[REJC,REJC-RWJC,RUS-RRS,yxz]
SEGVIS(RForearm)
AXES(RForearm)

{*Left forearm system*}
{*====================*}

LUS=%LUScalib*LeftForearm
LRS=%LRScalib*LeftForearm

LWJC=(LUS+LRS)/2
PARAM(LWJC)
```

```
OUTPUT(LWJC)
LForearm=[LWJC,LEJC-LWJC,LRS-LUS,yxz]
LForearmA=[LEJC,LEJC-LWJC,LRS-LUS,yxz]
SEGVIS(LForearm)
AXES(LForearm)

{*Right hand system*}
{*=================*}

RmidMCP3=(RMCP3p+RMCP3d)/2
RHand=[RMCP3d,RMCP3p-RMCP3d,RMCP5d-RMCP3d,yxz]
SEGVIS(RHand)
AXES(RHand)
OUTPUT(RmidMCP3)

{*Left hand system*}
{*================*}

LmidMCP3=(LMCP3p+LMCP3d)/2
LHand=[LMCP3d,LMCP3p-LMCP3d,LMCP3d-LMCP5d,yxz]
SEGVIS(LHand)
AXES(LHand)
OUTPUT(LmidMCP3)

{*OUTPUT FOR ANIMATION PACKAGE FOR GLASGOW SCHOOL OF ART*}
{*============================*}
AXES(Pelvis)
MATRIX(Pelvis,RFemurA,RHJC)
MATRIX(RFemurA,RShankA,RKJC)
MATRIX(RShankA,RFoot,RAJC)
MATRIX(Pelvis,LFemurA,LHJC)
MATRIX(LFemurA,LShankA,LKJC)
MATRIX(LShankA,LFoot,LAJC)
MATRIX(Pelvis,TrunkA,midT8andPX)
MATRIX(TrunkA,RHumerusA,RSJC)
MATRIX(RHumerusA,RForearmA,REJC)
MATRIX(RForearmA,RHand,RWJC)
MATRIX(TrunkA,LHumerusA,LSJC)
MATRIX(LHumerusA,LForearmA,LEJC)
MATRIX(LForearmA,LHand,LWJC)


{*KINEMATIC CALCULATIONS*}
{*=====================*}

{*Euler angles for output into computer programme*}
GlobalPelvis=<Pelvis,1>
GlobalTrunk=<Trunk,1>
OUTPUT(GlobalPelvis,GlobalTrunk)

{*Angles calculated using the floating axis method*}
ColeJCS(RHip,RFemur,RightHip)
SEGVIS(RightHipJCS)
```

```
ColeJCS(RFemur,RShank,RightKnee)
ColeJCS(RAnkle,RFoot,RightAnkle)
ColeJCS(Pelvis,LFemur,LeftHip)
ColeJCS(LFemur,LShank,LeftKnee)
ColeJCS(LAnkle,LFoot,LeftAnkle)
ColeJCS(RShoulder,RHumerus,RightShoulder)
ColeJCS(LShoulder,LHumerus,LeftShoulder)
ColeJCS(RHumerus,RForearm,RightElbow)
ColeJCS(LHumerus,LForearm,LeftElbow)
ColeJCS(RForearm,RHand,RightWrist)
ColeJCS(LForearm,LHand,LeftWrist)
ColeJCS(Pelvis,Trunk,Trunk)

{*corrects so that flexion (dorsiflexion), abduction and external rotation are positive*}
{*Order of angles is flexion, abd, ER*}

RightHipJCSAngles=<1(RightHipJCSAngles),-2(RightHipJCSAngles),-3(RightHipJCSAngles)>
LeftHipJCSAngles=<1(LeftHipJCSAngles),2(LeftHipJCSAngles),3(LeftHipJCSAngles)>
RightKneeJCSAngles=<-1(RightKneeJCSAngles),-2(RightKneeJCSAngles),-
3(RightKneeJCSAngles)>
LeftKneeJCSAngles=<-1(LeftKneeJCSAngles),2(LeftKneeJCSAngles),3(LeftKneeJCSAngles)>
RightAnkleJCSAngles=<(1(RightAnkleJCSAngles)-90),-2(RightAnkleJCSAngles),-
3(RightAnkleJCSAngles)>
LeftAnkleJCSAngles=<(1(LeftAnkleJCSAngles)90),2(LeftAnkleJCSAngles),3(LeftAnkleJCSAngle
s)>
RightShoulderJCSAngles=<1(RightShoulderJCSAngles),-2(RightShoulderJCSAngles),-
3(RightShoulderJCSAngles)>
LeftShoulderJCSAngles=<1(LeftShoulderJCSAngles),2(LeftShoulderJCSAngles),3(LeftShoulder
JCSAngles)>
RightElbowJCSAngles=<1(RightElbowJCSAngles),-2(RightElbowJCSAngles),-
3(RightElbowJCSAngles)>
LeftElbowJCSAngles=<1(LeftElbowJCSAngles),2(LeftElbowJCSAngles),3(LeftElbowJCSAngles)
>
RightWristJCSAngles=<1(RightWristJCSAngles),-2(RightWristJCSAngles),-
3(RightWristJCSAngles)>
LeftWristJCSAngles=<1(LeftWristJCSAngles),2(LeftWristJCSAngles),3(LeftWristJCSAngles)>

Output(RightHipJCSAngles,LeftHipJCSAngles,LeftKneeJCSAngles,RightKneeJCSAngles,LeftAn
kleJCSAngles,RightAnkleJCSAngles)
Output(RightShoulderJCSAngles,LeftShoulderJCSAngles,RightElbowJCSAngles,LeftElbowJCS
A ngles,RightWristJCSAngles,LeftWristJCSAngles,TrunkJCSAngles)

EndIF
{*Ends dynamic trials*}


{*KINETIC CALCULATIONS*}
{*===================*}

BODYMASS=$BODYMASS

{*Build the kinetic hierarchy*}
{*Considers the pelvis to be the root segment for the lower limb and the trunk for the UL*}
```

```
{* Segment=[Child,Parent,Connection Point,Anthropometric Data]   *}

RFemur=[RFemur,Pelvis,RHJC,DefaultFemur]
LFemur=[LFemur,Pelvis,LHJC,DefaultFemur]
RShank=[RShank,RFemur,RKJC,DefaultShank]
LShank=[LShank,LFemur,LKJC,DefaultShank]
RFoot=[RFoot,RShank,RAJC,DefaultFoot]
LFoot=[LFoot,LShank,LAJC,DefaultFoot]
RHumerus=[RHumerus,Trunk,RSJC,DefaultHumerus]
LHumerus=[LHumerus,Trunk,LSJC,DefaultHumerus]
RForearm=[RForearm,RHumerus,REJC,DefaultForearm]
LForearm=[LForearm,LHumerus,LEJC,DefaultForearm]
RHand=[RHand,RForearm,RWJC,DefaultHand]
LHand=[LHand,LForearm,LWJC,DefaultHand]

{*Force Vectors*}
{*=============*}

OptionalReactions(ForcePlate1,ForcePlate2,ForcePlate3)
ForceVector(ForcePlate1)
ForceVector(ForcePlate2)
ForceVector(ForcePlate3)



{* The correction makes so +ve moments tend to abduct, externally rotate and flex*}
{* These moments are external moments*}

RHipMoment=2(REACTION(RFemur))
RHipMoment=RHipMoment/(1000*BODYMASS)
RightHipMoment={1(RHipMoment),2(RHipMoment),-3(RHipMoment)}
RKneeMoment=2(REACTION(RShank))
RKneeMoment=RKneeMoment/(1000*BODYMASS)
RightKneeMoment={1*RKneeMoment(1),1*RKneeMoment(2),1*RKneeMoment(3)}
RAnkleMoment=2(REACTION(RFoot))
RAnkleMoment=RAnkleMoment/(1000*BODYMASS)
RightAnkleMoment={1(RAnkleMoment),2(RAnkleMoment),-3(RAnkleMoment)}
LHipMoment=2(REACTION(LFemur))
LHipMoment=LHipMoment/(1000*BODYMASS)
LeftHipMoment={-1(LHipMoment),-2(LHipMoment),-3(LHipMoment)}
LKneeMoment=2(REACTION(LShank))
LKneeMoment=LKneeMoment/(1000*BODYMASS)
LeftKneeMoment={-1(LKneeMoment),-2(LKneeMoment),3(LKneeMoment)}
LAnkleMoment=2(REACTION(LFoot))
LAnkleMoment=LAnkleMoment/(1000*BODYMASS)
LeftAnkleMoment={-1(LAnkleMoment),-2(LAnkleMoment),-3(LAnkleMoment)}

{*Currently not normalised to body mass*}

RShoulderMoment=2(REACTION(RHumerus))
RShoulderMoment=RShoulderMoment/(1000)
RElbowMoment=2(REACTION(RForearm))
RElbowMoment=RElbowMoment/(1000)
```

```
RWristMoment=2(REACTION(RHand))
RWristMoment=RWristMoment/(1000)
LShoulderMoment=2(REACTION(LHumerus))
LShoulderMoment=LShoulderMoment/(1000)
LElbowMoment=2(REACTION(LForearm))
LElbowMoment=LElbowMoment/(1000)
LWristMoment=2(REACTION(LHand))
LWristMoment=LWristMoment/(1000)

OUTPUT(RightHipMoment,RightKneeMoment,RightAnkleMoment,LeftHipMoment,LeftKneeMom
ent,LeftAnkleMoment)
OUTPUT(RShoulderMoment,RElbowMoment,RWristMoment,LShoulderMoment,LElbowMoment
,LWristMoment)
{*Ends dynamic trials*}

{*Whole Body Centre of Mass Calculation*}
{*============================*}
MidTRUNK = (RSJC1+LSJC1)/2
MIDHIP = (LHJC+RHJC)/2
HNT_CM = MIDHIP+(0.66*(MidTRUNK-MIDHIP))
LFEMUR_CM = LHJC+(0.433*(LKJC-LHJC))
RFEMUR_CM = RHJC+(0.433*(RKJC-RHJC))
LTIBIA_CM = LKJC+(0.433*(LAJC-LKJC))
RTIBIA_CM = RKJC+(0.433*(RAJC-RKJC))
LSHOE_CM = 0.5*(LAJC+LToe)
RSHOE_CM = 0.5*(RAJC+RToe)
RARM_CM = RSJC+(0.436*(REJC-RSJC))
LARM_CM = LSJC+(0.436*(LEJC-LSJC))
RFOREARM_CM = REJC+(0.682*(RUS-REJC)) {*includes hand*}
LFOREARM_CM = LEJC+(0.682*(LUS-LEJC))
FEMURmass = 0.100*$BODYMASS
TIBIAmass = 0.0465*$BODYMASS
SHOEmass = 0.0195*$BODYMASS
HNTmass = 0.578*$BODYMASS
ARMmass = 0.028*$BODYMASS
FOREARMmass = 0.022*$BODYMASS
{*COM =
(((HNTmass*(HNT_CM))+(FEMURmass*(LFEMUR_CM))+(FEMURmass*(RFEMUR_CM))+(TIB
IAmass*(LTIBIA_CM))+(TIBIAmass*(RTIBIA_CM))+(SHOEmass*(LSHOE_CM))+(SHOEmass*(
RSHOE_CM))+(FOREARMmass*(RFOREARM_CM))+(FOREARMmass*(LFOREARM_CM))+(A
RMmass*(RARM_CM))+(ARMmass*(LARM_CM)))/(HNTmass+2*(FEMURmass)+2*(TIBIAmass
)+2*(SHOEmass)+2*(ARMmass)+2*(FOREARMmass)))*}
COM =
(((HNTmass*(HNT_CM))+(FEMURmass*(LFEMUR_CM))+(FEMURmass*(RFEMUR_CM))+(TIB
IAmass*(LTIBIA_CM))+(TIBIAmass*(RTIBIA_CM))+(SHOEmass*(LSHOE_CM))+(SHOEmass*(
RSHOE_CM))+(ARMmass*(RARM_CM))+(ARMmass*(LARM_CM)))/(HNTmass+2*(FEMURmas
s)+2*(TIBIAmass)+2*(SHOEmass)+2*(ARMmass)))
{*COM =
(((HNTmass*(HNT_CM))+(FEMURmass*(LFEMUR_CM))+(FEMURmass*(RFEMUR_CM))+(TIB
IAmass*(LTIBIA_CM))+(TIBIAmass*(RTIBIA_CM))+(SHOEmass*(RSHOE_CM))+(SHOEmass*(
LSHOE_CM))+(ARMmass*(LARM_CM)))/(HNTmass+2*(FEMURmass)+2*(TIBIAmass)+2*(SHO
Emass)+1*(ARMmass)))*}

OUTPUT(COM)
```

LINVELACC(COM,Pelvis)

Appendix 3.3

Details of Kinetic Functions in BodyBuilder `reactionS`

`= REFER(reactionR,pointP)`

This function „moves" the reference point for the given reaction reactionR, to create a new reactionS. The Force component of the reaction remains unchanged. The moment is changed by adding the additional moment caused by the movement of the reference point. As the reference point moves away from the line of application of the force, the moment increases. This is found from the cross product of the point movement and the force.

NewMoment = OldMoment + (OldPoint - NewPoint) x Force;

`Reaction = ForcePlate1`

to get force plate data gives the closest single sample from the analogue data (which is sampled at a higher rate than the marker data) for the given force plate. There is no attempt to take a mean of a period of data.

`ReactionR = REACTION(segmentS)`

The reaction calculation needs to add all the reactions acting on the given segment, making the assumption that only one reaction (acting at the proximal end) is unknown. All of the components are added to give the compensating reaction that needs to be applied to the segment to keep it in dynamic equalibrium.
The forces due to acceleration (including gravity) and moments of inertia are calculated from the position of the centre of mass of the segment (as given in the anthropometric table) from the current frame, and frames +- 0.25 seconds from the current frame (represented by „Next" and „Previous"). This gives a moving average filter (of width 0.5 seconds).
See Winter (2nd edition p47-48), and 3D analysis of human movement, Ed.Allard/Stokes/Blanchi 1995 ISBN 0-87322-623-2 for fuller explanations of the equations.

```
LinearAccel = Next.CoM + Previous.CoM - 2 * Current.CoM
                    / m_FramePeriod * m_FramePeriod*SampleWidth*SampleWidth
LinearAccel += Gravity;
Reaction.Force = Segment.Mass * LinearAccel / 1000; //  kilos and millimetres
```

The angular velocities are calculated from the dot products („&‟) of the axes of the segment +- 0.25 seconds from the current time.

```
AngularVeloc.X = Next.YAxis & Previous.ZAxis;
AngularVeloc.Y = Next.ZAxis & Previous.XAxis;
AngularVeloc.Z = Next.XAxis & Previous.YAxis;
AngularVeloc /= 2*FramePeriod*SampleWidth;
```

and accelerations are calculated similarly, also using the Current position :-

```
AngAccel.X = (Next.YAxis & Current. ZAxis) -  (Current.YAxis & Previous.ZAxis);
AngAccel.Y = (Next.ZAxis & Current.XAxis) -  (Current.ZAxis & Previous.XAxis);
AngAccel.Z = (NextXAxis & Current.YAxis) -  (Current.XAxis & Previous.YAxis);
AngularAccel /= m_FramePeriod * m_FramePeriod*SampleWidth*SampleWidth;
```

These are then used to calculate the gyroscopic component, then the moment of inertia acting round the centre of mass.

```
GyrComp.X = Inertia.X * Mass * AngularVeloc.X;
GyrComp.Y = Inertia.Y * Mass * AngularVeloc.Y;
GyrComp.Z = Inertia.Z * Mass * AngularVeloc.Z;
Moment = AngularVeloc x GyrComponent;

Moment.X += Inertia.X * AngularAccel.X;
Moment.Y += Inertia.Y * AngularAccel.Y;
Moment.Z += Inertia.Z * AngularAccel.Z;
```

The other reactions acting on the segment are then added, refering them all to the CoM:-

```
for ( R=0; R<NumReactions; R++ )
{
  Reaction.Force += Reaction[R]. Force;
  Reaction.Moment += Reaction[R]. Moment;
  Reaction.Moment += (Reaction[R]. Point - Reaction.Point) x Reaction[R].Force; }
```

And then if a force plate‟s connected, add its values, noting that the data from the force plates is expressed as the reaction being applied to the segment, rather than to the plate, again refering to the CoM :-

```
for ( FP=0; FPIter <NumForcePlates; FP++ )
{
  if ( ForcePlates[FP]->IsConnected() )
  {
    Reaction.Force -= ForcePlates[FP].Force;
    Reaction.Moment -= ForcePlates[FP].Moment;
    Reaction.Moment -= (ForcePlates[FP].Point - Reaction.sm_Point) x
ForcePlates[FP].Force;
    FP = NumForcePlates; // skip the rest of the plates
  }
}
```

Convert the result to refer to the attachment point.

```
Reaction.Moment += (Reaction.Point - Segment.Attachment) x Reaction.Force;
Reaction.Point = Segment.Attachment;
```

`powerI = POWER( Seg1, Seg2 )`

The same time separations are used to calculate the angular velocity for this as was used for the REACTION function above. „Previous‟ „Current‟ and „Next‟ positions separated by +- 0.25 seconds of the two segments are found.

Initially the Seg2 axes (represented in a 3x3 matrix) are converted to be relative to Seg1 coordinate space for each sample time.

```
Seg2Previous.Axes = Seg1Previous. Axes * Seg2Previous. Axes;
Seg2Current.Axes = Seg1Current. Axes * Seg2Current. Axes;
Seg2Next.Axes = Seg1Next. Axes * Seg2Next. Axes;
```

find the relative angular velocity between the two segments

```
AngularVeloc.X = Seg2Next.Axes.Y & Seg2Previous.Axes.Z;
AngularVeloc.Y = Seg2Next.Axes.Z & Seg2Previous.Axes.X;
AngularVeloc.Z = Seg2Next.Axes.X & Seg2Previous.Axes.Y;
AngularVeloc /= 2*m_FramePeriod*SampleWidth;
```

Convert the moment of the reaction between the segments (found using the REACTION function) to Segment 1 coordinate space too, and do the dot product with the angular velocity

LocalMoment = Transpose(Seg1Current. Axes) * Reaction. Moment
Power = LocalForce & AngularVeloc;

Appendix 4.1

Matlab programming code for processing of stair data

```
 clear;

%Create variable list  of filenames in matlab

% Find out number of rows in file
 r=0;
x=0;

% Open Data File
fid1 = fopen('C:\EQUAL\Data\Stairs\upstairsin.txt','rt'); fid2
= fopen('C:\EQUAL\Data\Stairs\upstairsout.txt','rt');

% Loop through data file until we get a -1 indicating EOF
while(x~=(-1))        x=fgetl(fid1);        r=r+1; end r
= r-1;
disp(['Number of rows = ' num2str(r)])
 frewind(fid1);
for i = 1:r
     name = fscanf(fid1,'%s',1); % Filter out string at beginning of
line          if(i==1)
       names = name; % Add 1st text string

else
       names = str2mat(names,name); % Add next string
      end end  r=0; x=0;  while(x~=(-
1))        x=fgetl(fid2);
r=r+1; end r = r-1;  disp(['Number of
rows = ' num2str(r)])
 frewind(fid2);
for i = 1:r
     nameout = fscanf(fid2,'%s',1); % Filter out string at beginning of
line
     if(i==1)
       namesout = nameout; % Add 1st text string

else
       namesout = str2mat(namesout,nameout); % Add next string
      end end
fclose(fid1);
fclose(fid2);
```

```
%Loads filenames

y=length(names(:,1));
for f=1:y
%Load data
%in_name=input('Enter data file : ','s');
%in_name=input('Input filename to be processed (in '') and include .txt
:  ');
%out_nameL=input('Input output filename(in '') and include .xls :   ');
 %fid1=fopen(out_nameL,'w');

in_name=names(f,:); %fprintf('%s\n',names(f,:));
uiimport(in_name); %will bring up the import wizard. Will need to select
create matrix from column function

    Subject=input('Input Subject Number: ');
    Trial=input('Input Trial Number:  ');
    RLeg=[RightAnkleJCSAngles_X RightKneeJCSAngles_X
RightKneeJCSAngles_Y RightKneeJCSAngles_Z RightHipJCSAngles_X
RightHipJCSAngles_Y RightHipJCSAngles_Z GlobalTrunk_Y
RightAnkleMoment_Z RightKneeMoment_Z RightHipMoment_Z RightHipMoment_X
COM_X COM_Y COM_Z LVelCOM_X LVelCOM_Y LVelCOM_Z];

%Select one step
    start=input('Input trajectory start frame number: ');
a=input('Input Right Heel Strike 1:  ');     a=(a-start+1);
    b=input('Input Right Heel Strike 2:  ');
b=(b-start)+1;     timeRgait=(b-a)+1;

%Selects the relevant gait cycle data Rleg
    t=(a:b);
    RGaitData=RLeg(t,:);

%Normalises the data to 100 data points for both L and R stance phases
%Interpolate gait cycle data to 100 points, using cubic spline
interpolation     timeR=(1:timeRgait);     Lpoints=timeRgait/100;
yy=0:Lpoints:timeRgait;
    PcL=yy./Lpoints;
%Sets baseline=0
    Interpol_RGaitData=0;

%Interpolates data over 100 points
    Interpol_RGaitData=interp1(timeR,RGaitData,yy,'spline');

%Finds maximum values and at what percentage of the gait cycle these
occur     for p=1:18
        maxR_data(:,p)=max(Interpol_RGaitData(:,p));
 %Find at what point maximum values occur
for j=1:101
            if Interpol_RGaitData(j,p)==maxR_data(:,p);
Percent_Rcycle_max(:,p)=PcL(j);             end;
end;     end
```

```
 %Find mimimum values and at what point of the gait cycle these occur
        for
p=1:18
        minR_data(:,p)=min(Interpol_RGaitData(:,p));
%Find at what point minimum values occur         for
j=1:101
            if Interpol_RGaitData(j,p)==minR_data(:,p);
Percent_Rcycle_min(:,p)=PcL(j);             end;
end;     end

%Outputs maximum and minumum moments
    fid3=fopen('C:\equal\data\stairs\UptestmaxminR.xls','a');
fprintf(fid3,'%4.0f\t %4.0f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t
%8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t
%8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t ',Subject,Trial,maxR_data);
fprintf(fid3,'%8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t
%8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t
%8.2f\t %8.2f\t %8.2f\t',minR_data);
    fprintf(fid3,'%8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t
%8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t
%8.2f\t %8.2f\t %8.2f\t',Percent_Rcycle_max);
    fprintf(fid3,'%8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t
%8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t
%8.2f\t %8.2f\t %8.2f\t\n',Percent_Rcycle_min);
%Write processed data to OUT files
 fileout=namesout(f,:);
 fid=fopen(fileout,'w');

%Outputs normalised angles and moments
   fprintf(fid,'Right Ankle Flexion\t Right Knee Flexion\t Right Knee
Abduction\t Right Knee ER\t Right Hip Flexion\t Right Hip Abduction\t
Right Hip ER\t GlobalTrunk_Y\t Right Ankle Mz\t Right Knee Mz\t Right
Hip Mz\t Right Hip Mx\t COM_X\t COM_Y\t COM_Z\t VelCOM_X\t VelCOM_Y\t
VelCOM_Z\t\n');
for i=1:101
fprintf(fid,'%8.2f\t
%8.2f\t %8.2f\t
%8.2f\t %8.2f\t
%8.2f\t %8.2f\t
%8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t %8.2f\t
%8.2f\t %8.2f\t %8.2f\t\n',Interpol_RGaitData(i,:));
end;        figure(1);      h1=subplot(3,2,1);
plot(Interpol_RGaitData(:,1));     title('Right
Ankle Flexion');     h2=subplot(3,2,2);
plot(Interpol_RGaitData(:,2))     title('Right Knee
Flexion');     h3=subplot(3,2,3);
plot(Interpol_RGaitData(:,5));     title('Right Hip
Flexion');     h4=subplot(3,2,4);
plot(Interpol_RGaitData(:,9));     title('Right
Ankle Mz');     h5=subplot(3,2,5);
plot(Interpol_RGaitData(:,10));     title('Right
Knee Mz');     h6=subplot(3,2,6);
```

```
plot(Interpol_RGaitData(:,11));     title('Right Hip
Mz');
 end;
fclose('all');
[u,v]=size(Interpol_RGaitData); v=v-1;

%To bring together multiple trials
%Create variable list  of filenames in matlab
% Find out number of rows in file
 r=0;
x=0;

% Open Data File
fid4 = fopen('C:\EQUAL\Data\Stairs\upstairsout.txt','rt');

% Loop through data file until we get a -1 indicating EOF
while(x~=(-1))        x=fgetl(fid4);        r=r+1; end r
= r-1;  disp(['Number of rows = ' num2str(r)])
 frewind(fid4);
for i = 1:r
     name = fscanf(fid4,'%s',1); % Filter out string at beginning of
line          if(i==1)
       names = name; % Add 1st text string

else
names =
str2mat(nam
es,name); %
Add next
string

end end
fclose(fid4);

    out_namec=input('Input output filename for trial (in '') and
include .xls :   ');     fid5=fopen(out_namec,'w');

%Loads filenames

y=length(names(:,1)); number=y;

%sets size of file to be imported range
= [1 0 u v];
 for n=1:number
in_name=names(n,:);
    data=dlmread(in_name,'\t',range);
    [c,d]=size(data);
a=(n*d)-(d-1);
b=(n*d);         if(n==1)
data1=data;         else
data1(:,a:b)=data;
end
            if(n~=number)
```

```
            fprintf(fid5,'Right Ankle Flexion\t Right Knee Flexion\t
Right Knee Abduction\t Right Knee ER\t Right Hip Flexion\t Right Hip
Abduction\t Right Hip ER\t Trunk Inclination\t Right Ankle Mz\t Right
Knee Mz\t Right Hip Mz\t Right Hip Mx\t COM_X\t COM_Y\t COM_Z\t
VelCOM_X\t VelCOM_Y\t VelCOM_Z\t') ;
else
            fprintf(fid5,'Right Ankle Flexion\t Right Knee Flexion\t
Right Knee Abduction\t Right Knee ER\t Right Hip Flexion\t Right Hip
Abduction\t Right Hip ER\t Trunk Inclination\t Right Ankle Mz\t Right
Knee Mz\t Right Hip Mz\t Right Hip Mx\t COM_X\t COM_Y\t COM_Z\t
VelCOM_X\t VelCOM_Y\t VelCOM_Z\t\n') ;
end end  for i=1:101    for k=1:(d*n)
if(k~=d*n)
       fprintf(fid5,'%8.2f\t',data1(i,k));
        else
       fprintf(fid5,'%8.2f\t\n',data1(i,k));
end    end end fclose('all');
```