# The effects of ankle foot orthoses on the gait pattern of stroke patients with equines deformity of the foot

Students' Name Sharon Kinsella MSc , BSc

Supervisors' Name Dr Kieran Moran

Doctorate of Philosophy
School of Health and Human Performance
Dublin City University

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I hereby certify that this material, which I now submit for assessment on the programme of study leading to the award of Doctorate of Philosophy is entirely my own work and has not been taken from the work of others save and to the extent that such work has been cited and acknowledged within the text of my work.

Signed: Sharon Knell (Candidate) ID No.: 51158191

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## Abbreviations

| | |
|---|---|
| A1 | Ankle flexion at heel strike |
| A2 | Maximum plantarflexion flexion of the ankle during loading response |
| A3 | Maximum dorsiflexion of the ankle in stance phase |
| A4 | Flexion at toe-off at the ankle |
| A5 | Maximum ankle dorsiflexion in swing phase |
| A6 | Total sagittal plane excursion of the ankle |
| A7 | Maximum ankle plantarflexion in swing |
| AAFO | Anterior leaf ankle foot orthosis |
| Aff | Affected leg |
| AFO | Ankle foot orthosis |
| AM1 | Maximum ankle plantarflexor moment |
| AM2 | Maximum ankle dorsiflexor moment |
| AP1 | Power absorption by the ankle plantarflexors as the leg rotates over the foot |
| AP2 | The ankle push off power burst |
| AR1 | Maximum external rotator moment of the foot |
| AR2 | Maximum internal rotator moment of the foot |
| CM | Custom made |
| COM | Center of mass |
| CVA | Cerebrovascular accident |
| EMG | Electromyographic |
| H A T | Head, arms and trunk |
| H1 | Hip flexion at heel strike |
| H2 | Maximum flexion of the hip at loading response |
| H3 | Maximum extension of the hip in stance phase |

| | |
|---|---|
| H4 | Flexion at toe-off at the hip |
| H5 | Maximum flexion in swing phase of the hip |
| H6 | Total sagittal plane excursion of the hip |
| H8 | Maximum adduction of the hip in stance |
| H9 | Maximum abduction of the hip in swing |
| HM1 | Maximum hip flexor moment |
| HM2 | Maximum hip extensor moment |
| HP1 | Hip extensor power generation |
| HP2 | Hip flexor power absorption |
| HP3 | Hip flexor power generation |
| HR2 | Maximum rotation of the pelvis in the coronal plane |
| HR3 | Maximum rotation of the pelvis in the coronal plane |
| K1 | Knee flexion at heel strike |
| K2 | Maximum flexion at loading response at the knee |
| K3 | Maximum extension in stance phase at the knee |
| K4 | Flexion at toe-off at the knee |
| K5 | Maximum flexion of the knee in swing phase |
| K6 | Total sagittal plane excursion of the knee |
| KM1 | 1st knee maximum extensor moment |
| KM2 | Maximum knee flexor moment |
| KM3 | 2nd maximum knee extensor moment |
| KP1 | Power absorption by the eccentrically contracting quadriceps during weight acceptance |
| KP2 | A small amplitude power phase resulting from the knee extensors shortening to reduce knee flexion during midstance |
| KP3 | Energy lost in decelerating the backward rotating leg |

| | |
|---|---|
| KP4 | Energy absorbed by the hamstrings late in swing as they decelerate the forward swinging leg and foot |
| M W S | Maximum walking speed |
| NAFO | Not wearing an ankle foot orthosis |
| NAMgroup | Normal age matched subject group who were age matched with that of the stroke group |
| Ngroup | Normal healthy subjects |
| PAFO | Posterior leaf ankle foot orthosis |
| R1 | Rotational position of the foot at heel strike |
| R2 | Maximum external rotation of the foot during stance |
| R3 | Maximum internal rotation of the foot during stance |
| R4 | Rotational position of the foot at foot off |
| R5 | Maximum external rotation of the foot during swing |
| Sgroup | Stroke patients with equinus deformity of the foot |
| Unaff | Unaffected leg |
| $VO_2$ | Oxygen uptake |

# The effects of ankle foot orthoses on the gait pattern of stroke patients with equinus deformity of the foot

**Background** Following a stroke many patients develop a deformity of the foot referred to as equinus deformity Equinus deformity is commonly treated by prescribing an ankle foot orthosis (AFO) Few studies have assessed the effects of wearing AFOs or the effects of different types of leaf forms of AFOs on hemiplegic gait

**Aim** To establish the effects of an equinus deformity of the foot on gait and to examine the effect of wearing AFOs on hemiplegic gait

**Methods** 30 normal subjects (Ngroup), 10 normal subject's age matched with the stroke group (NAMgroup) and 23 stroke patients with equinus deformity of the foot (Sgroup) were recruited to participate in this study The kinematics and kinetics of each subject's gait were analysed using a Vicon Motion Analysis system and two AMTI forceplates [250Hz] Gait was assessed under the conditions of walking 1) without an ankle foot orthosis, 2) with an anterior leaf AFO, and 3) with a posterior leaf AFO Statistical analyses were carried out using repeated measures ANOVA, correlations and a cluster analysis The p value was set at 0 05

**Results** The gait pattern of the Sgroup were described and found to be non-homogeneous and consisted of three distinct patterns Only one subgroup (SG1) had significantly improved gait with the wearing of an AFO SG1 had the more normalised gait pattern prior to wear of an AFO in comparison to the other two subgroups, having a greater walking speed, stride and step length Furthermore, this study identified that different leaf forms of AFOs have different effects on joint angular kinematics Wearing an AFO did not significantly decrease oxygen uptake, energy cost or COM displacement of the Sgroup

**Conclusion** The results of this study will aid in improved prescription of AFOs to stroke patients with equinus deformity of the foot The results of this study would also suggest that wearing AFOs has no significant effect on energy expenditure

# Chapter 1: Introduction

Equinus deformity of the foot is a motor deficiency which is caused by total or partial central paralysis of the muscles innervated by the common peroneal nerve and occurs often unilaterally following a stroke (Voigt and Sinkjaer, 2000) The mechanism behind the formation of the deformity is complex but is thought to be caused by a combination of 1) an incorrect strength ratio of the dorsiflexor muscles to the plantarflexor muscles due to the presence of muscle stiffness in the plantarflexor muscles, 2) an inability to generate sufficient force m the dorsiflexor muscles to cause dorsiflexion, 3) an inability of the plantarflexor muscles to eccentrically lengthen to allow dorsiflexion to occur, 4) co-activation of the dorsiflexor and plantarflexor muscles of the foot at inappropriate phases of the gait cycle, or 5) an unknown cause resulting in an equinus deformity (Burridge et al , 1997) In this thesis stroke patients will be operationally defined as having equinus deformity of the foot if 1) they have an inability to actively dorsiflex their ankle, 2) if they are wearing or have a history of wearing an ankle foot orthosis (AF0) or 3) if they have been referred from a physiotherapist who believe that they would benefit from wearing an AFO Approximately 298 to 596 stroke patients in the Republic of Ireland develop equinus deformity of the foot each year (Burridge et al , 1997, Leane et al , 1998, Verdie et al , 2004) Patients who develop this deformity require a more specialised, intensive and prolonged rehabilitation over patients without this deformity (Verdie et al , 2004) Stroke patients with equinus deformity of the foot are believed to have an altered gait which results in an increased energy expenditure (Lehmann, 1979, Acimovic-Janezic et al , 1984)

The literature to date on the gait pattern of equinus deformity of the foot in stroke patients is limited and has mainly described its effects on temporal distance parameters (Lehmann et al , 1987, Wong et al , 1992, Granat et al , 1996, Gok et al , 2003, Iwata et al , 2003) Only three studies so far have examined aspects of the sagittal joint kinematics and kinetics of such a patient group, each only focussing on the affected limb (Lehmann et al , 1987, Wong et al , 1992, Gok et al , 2003) One of these studies describes the gait deformity present in these patients when compared to normal age matched subjects walking at a comparable speed (Lehmann et al , 1987) They reported that many of the alterations observed in the gait pattern were due to the effects of their decreased walking speed, rather than due to the deformity itself This thesis will take the effect of walking speed into consideration when describing the gait pattern of these patients By establishing the differences between the stroke patients with equinus deformity of the foot and normal age matched subjects walking at a comparable speed, the true negative effect of the deformity can be established Identification of the gait pattern of these patients is necessary to aid the development of appropriate treatment methods such as AFOs

A method commonly prescribed to patients with equinus deformity of the foot is the use of AFOs There are many different types of AFOs but they can normally be categorised into posterior leaf and anterior leaf (Wong et al , 1992, Chen et al , 1999, Danielsson and Sunnerhagen, 2004) It is believed that as a result of wearing an AFO, a more normal gait will be approximated and thus the energy cost is reduced (Lehmann, 1979) Although AFOs are frequently prescribed and numerous authors have reported on their possible benefits, many of the leading names in stroke rehabilitation research (Davies, 1985, Bobath, 1990, Carr and Shepherd, 1998) feel that insufficient evidence has been provided to support their prescription (Lehmann, 1979, Lehmann et al , 1983, Perry and Montgomery, 1987, Condie and Meadows, 1993, DeVries's et al , 1991, Teasell et al , 2001) In general positive benefits on temporal distance parameters have been reported following the wear of an AFO (Mojica et al , 1988, Heese et al , 1996, Churchill et al , 2003) However, only a few studies have examined the effects of wearing of an AFO on sagittal joint angular kinematics (Burdett et al , 1988, Gok et al , 2003) and angular joint kinetics (Lehmann et al , 1987, Gok et al , 2003) Burdett et al (1988) and Gok et al (2003) reported that wearing an AFO significantly improved the angle of the ankle at heel strike and during midswing Lehmann et al (1987) and Gok et al (2003) examined the angular kinetics of the knee joint and reported contradictory findings To date, no study has examined the effects of wearing an AFO on coronal joint angular kinematics or the effects of different leaf forms of AFO on joint angular kinematics Also, only one study to date has assessed the impact of wearing an AFO on the unaffected limb (Hesse et al , 1999) and this research focussed solely on temporal distance parameters In this thesis, the effects of wearing an AFO on the joint kinematics and kinetics of both the affected and unaffected limb will be examined

A compounding factor in describing patients with equinus deformity of the foot and determining the effectiveness of AFO intervention is that unlike normal subjects the equinus deformity gait pattern may not be homogenous (Mulroy et al , 2003) No study to date has examined if this is the case If subgroups of gait patterns were identified in this patient group, it would aid in the development of more specific treatment (for each group) Previous research has not assessed the effects of different leaf forms of AFOs on the gait pattern of stroke patients with equinus deformity It is possible that the design characteristics of the different leaf forms of AFOs may be beneficial to the gait pattern of some stroke patients and have a detrimental effect on the gait pattern of other stroke patients

Individuals with a disturbed movement pattern may have higher energy expenditure than normal subjects while walking (Zampero et al , 1995) Higher energy expenditure results in a greater fatigue (Onley et al , 1986, Zampero et al , 1995) AFOs, by improving the gait pattern, are believed to lower energy expenditure (Franceschini et al , 2003) Three studies have indirectly examined the effects of wearing of an AFO on the energy expenditure by assessing oxygen uptake and the energy cost of gait (Corcoran et al , 1970, Franceschini et al , 2003, Danielsson and Sunnerhagen, 2004) One study reported that oxygen uptake decreased (Corcoran et al , 1970) and all three studies reported a decrease in energy cost with the wearing of an AFO (Corcoran et al , 1970, Franceschini et al , 2003, Danielsson and Sunnerhagen, 2004) Energy cost of gait is significantly influenced by walking speed (Zampero et al , 1995) and future studies need to control this variable when assessing the effects of AFO wear on energy expenditure

The aims of the present study were

(i) to describe the gait pattern of stroke patients with equinus deformity of the foot in comparison to normal subjects matched for age and speed of walking

(ii) to examine if the gait pattern of stroke patients with equinus deformity of the foot is homogeneous and if not, to examine how the gait pattern of the sub groups differed from normal subjects

(iii) to examine if the gait pattern of normal subjects is affected by speed of walking

(iv) to assess if wearing an AFO returns the gait of the patient group to a more normal pattern and to examine if wearing of an AFO has differing effects on the gait pattern of the subgroups

(v) to assess if different leaf forms of AFOs have different effects on the gait pattern of stroke patients and normal subjects

(vi) to examine if the gait pattern while wearing an AFO is altered by walking speed

(vii) to investigate the effects of walking with an AFO on the energy expenditure of gait and to examine if the measurement of the center of mass (COM) displacement is a viable method of assessing the energy expenditure of gait in stroke patients with equinus deformity of the foot

# Chapter 2: Human Gait

## 2 0 Key functions of gait

For a person to walk the locomotor system must be able to meet many demands Each leg must be able to support the weight of the body in turn without collapsing (Lamoreux, 1971) Static and dynamic balance must be maintained, especially during single-leg stance (Whittle, 2000) The swinging leg must be able to advance to a position where it can take on the supporting role without adversely affecting the normal movement of the other joints (Lamoreux, 1971) Sufficient power must also be available to produce and control the necessary limb movements to propel the body forward (Whittle, 2000) The overall displacement of the center of mass during the gait cycle should be minimal thereby requiring little energy input (Kerrigan et al , 2000, Croce et al , 2001) For most people gait is natural, efficient and is normally accomplished by employing a bipedal gait pattern that is repeatable, cyclical and symmetrical in nature (Lamoreux, 1971) In order to understand abnormal gait it is necessary firstly to understand normal gait since this provides the standard against which comparisons can be made

The examination of gait patterns can involve assessment of the temporal distance parameters, joint angular kinematics and kinetics, electromyography and the energy expenditure of gait (Lehmann et al , 1987, Mojica et al , 1988, Dieli et al , 1997, Tyson et al , 1998, Tyson and Thornton, 2001, Gok et al , 2003) The measurement of joint angular kinetics and the use of dynamic electromyography help to identify the specific functional actions that occur at critical points in the gait cycle Abnormalities at the critical points in the gait cycle can lead to a disruption of gait events and may result in incorrect timing for the events that follow This can also lead to changes in the distance traveled by each lower limb which can be measured by examining the angular joint kinematics and temporal distance variables Overall, theses changes lead to asymmetries in the gait pattern Measurement of temporal distance variables of symmetry and distances covered are extremely useful in characterising gait performance These measures aid practitioners in the rehabilitation of gait-related pathologies, assist the clinician in the selection of an orthosis or prosthetic components, and help in the identification of other variables that may enhance performance In the rehabilitation of stroke patients, views differ on what effect rehabilitation should have on the gait pattern Some therapists judge a successful rehabilitation based on the patient developing a more normal gait pattern (Bobath, 1978), whereas other therapists view a successful rehabilitation where the gait pattern allows effective day to day functioning, such as enabling the patient to get from A to B even if it is with a so called abnormal gait (Davies, 1985)

The aim of this review of literature is to examine the components of normal gait and the effect of age on gait pattern, as stroke patients are normally over the age of 60 The causes and effects of stroke on the mechanisms of gait are presented The causes of equinus deformity of the foot in stroke patients are then outlined and the effects of this deformity on gait patterns are examined The effects of ankle foot orthoses on temporal distance parameters, joint angular kinematics and kinetics and the energy cost of gait are examined

The literature review will highlight the lack of research into examining both the gait pattern of stroke patients with equinus deformity of the foot and the effects of wearing an ankle foot orthosis on the gait pattern of adult stroke patients with equinus deformity of the foot

## 2.1 Normal Gait

This section provides the reader with an overview of the rudiments of normal gait In the literature the terminology used to describe gait varies In the following chapter and throughout this thesis all gait events will be described using definitions within this section

A variety of quantitative methods have been used to examine normal gait, including joint and whole body kinematics and kinetics and dynamic electromyography (EMG) recording (Winter, 1991) In general each joint and whole body kinematics and kinetics has been examined in terms of peak values, values at key points or the total excursion during the gait cycle (Winter et al , 1987, Benedetti et al , 1998, Whittle, 2000) In the following sections only kinematic and kinetic data relevant to the present study will be discussed All values reported are for healthy adults

**2 1 1 Temporal distance parameters**

Walking speed is normally defined as the average speed attained after approximately three steps, expressed in m s$^{1}$ (Rose and Gamble, 1994) Normal self selected walking speed in healthy adults ranges between 1 2 (±0 2) and 1 4 (±0 2) m s$^{-1}$ (Kadaba et al , 1989, Oberg et al , 1993, Benedetti et al , 1998, Riley et al , 2001) Cadence is defined by the number of steps per minute and ranges between 111 6 (±8 3) and 124 8 (±10 2) steps min$^{-1}$(Kadaba et al , 1989, Oberg et al , 1993, Riley et al , 2001, Titianova, 2003, Cho et al , 2004) Step length is defined by the distance between the same points on each foot (usually the heel) during double limb support (Rose and Gamble, 1994) Normal self selected step length ranges between 0 6 (±0 1) and 0 8 (±0 1) m (Oberg et al , 1993, Titianova, 2003) Stride length is defined by the distance travelled between two successive foot strikes of the same foot (Rose and Gamble, 1994) Therefore each stride is composed of one right and one left step length Normal self-selected stride length in healthy adults is between 1 0 (±0 1) and 1 5 (±0 1) m (Kadaba et al , 1989, Benedetti et al , 1998, Riley et al , 2001, Titianova, 2003, Cho et al , 2004) Step width ıs the sıde-to-sıde dıstance between the centers of each heel durıng a strıde (Whıttle, 2000) and ıs normally reported to range from 0 1 (±0 2) to 0 2 (±0 2) m (Cho et al , 2004, Owıngs and Grabıner, 2004)

### 2.1 2 Gaıt cycle perıods

In normal healthy adults walkıng at theır customary walkıng speed the followıng events occur (Fıgure 2 1) [Whıttle, 2000],

- Stance phase 0- 62% of the gaıt cycle
- Swıng phase 62-100% of the gaıt cycle
- Inıtıal double lımb support occurs between 0-12% of the gaıt cycle
- Sıngle lımb support occurs between 12-50% of the gaıt cycle
- Second double lımb support occurs between 50-62% of the gaıt cycle
- Opposıte foot contact occurs at 50% of the gaıt cycle
- Opposıte foot off occurs at 12% of the gaıt cycle
- Inıtıal swıng occurs between 62-75% of the gaıt cycle
- Mıd swıng occurs between 75-85% of the gaıt cycle
- Termınal swıng 85-100% of the gaıt cycle

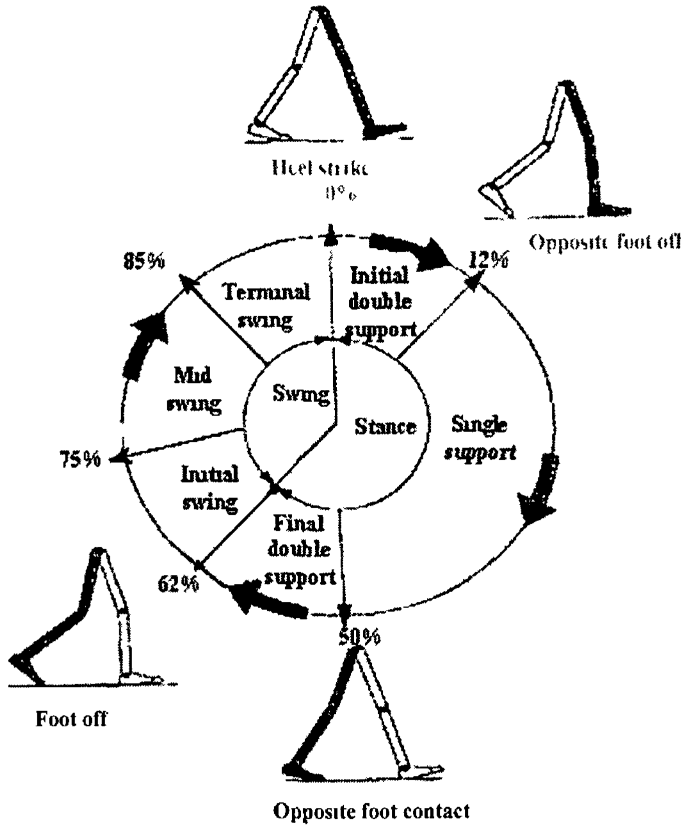



**Figure 2 1** Typical normal walk cycle illustrating the events of gait (adapted from Carollo and Matthews, 2002)

Limp Index is the total single and double support for the selected leg divided by the total single and double support for the unselected leg Normal value for this variable should be 1 Symmetry is calculated by dividing the single limb support time of one limb by the single limb support time of the other limb (Tyson et al , 1998) A normal value for this variable is 1 Step symmetry is calculated by dividing the right step length by the left step length (Tyson and Thornton, 2001) Normal value for this variable is 1 Swing symmetry is calculated by dividing the duration of the swing phase of one limb by the duration of the swing phase of the other limb (Hesse et al , 1999) A normal value for this variable is 1 Stance symmetry is calculated by dividing the duration of the stance phase of one lımb by the duratıon of the stance phase of the other lımb (Hesse et al , 1999) A normal value for thıs varıable ıs 1

**2 1 3 Joınt angular kınematıcs and kınetıcs durıng gaıt**

The flexıon and extensıon pattern of the hıp and knee and the ankles' dorsıflexıon and plantarflexıon pattern ın the sagıttal plane have been maınly examıned ın the lıterature ın three ways Fırstly, a number of authors have reported maxımum flexıon and extensıon or dorsıflexıon and plantarflexıon angles durıng the gaıt cycle (Wınter et al , 1991, Oberg et al , 1994) Secondly, the angular posıtıons of the hıp, knee and ankle durıng varıous poınts of the gaıt cycle (Benedettı et al , 1998) Fınally, the total excursıons of the hıp, knee and ankle have been reported (Wınter et al , 1991, Oberg et al , 1994, Benedettı et al , 1998)

**Hıp flexıon and extensıon**

At heel strıke the hıp ıs ın a flexed posıtıon (27 ±5°) and may slıghtly flexes further durıng the loadıng response [29 ±6°] (Benedettı et al , 1998) [Fıgure 2 2] The hıp then extends untıl opposıte foot contact occurs (50% of the gaıt cycle) at whıch poınt maxımum extensıon of the hıp ıs found, rangıng from -10 (±5) to -20 (±4)° [Wınter, 1984, Wınter et al , 1987, Benedettı et al , 1998, Kerrıgan et al , 2001, Yavuzer and Ergın, 2001] Hıp extensıon durıng the stance phase ıs ımportant as ıt allows the vertıcal trunk segment to move forward over the stance foot, enablıng the opposıte leg to advance forward, resultıng ın a step Followıng heel strıke the hıp extends due to a strong extensor moment (HM1) actıng concentrıcally (HP1) whıch has been reported to be -3 (±1) %BW H (Benedettı et al , 1998) [Fıgure 2 2] The presence of thıs strong extensor moment arrests the forward acceleratıon of the combıned head, arms and trunk segment (H A T ) Thıs extensor moment ıs a result of the sudden posterıor hıp reactıon

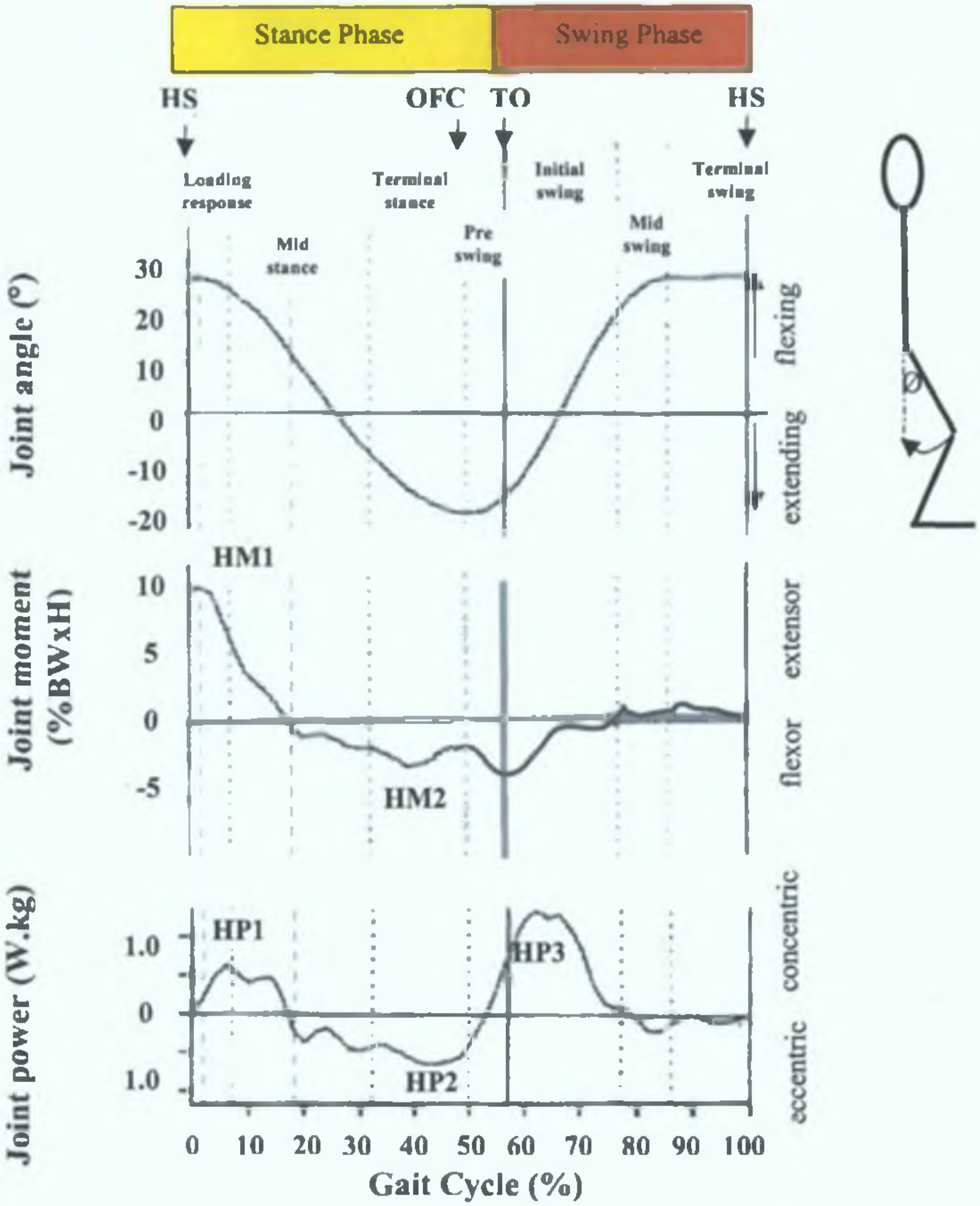



**Figure 2.2**: Hip flexion/extension angle, moment and power in the sagittal plane during normal gait while walking barefoot (adapted from Winter et al., 1984; Benedetti et al., 1998)

HS = Heel strike
OFC = Opposite foot contact
TO = Toe off

force, which begins at heel strike and lasts for the first half of stance In the region of 20 to 50% of the gait cycle the moment pattern reverses with the hip flexor moment (HM2) acting eccentrically (HP2) to control hip extension (Figure 2 2) This hip flexion moment (HM2) has been reported to be 9 (±2) %BWxH (Benedetti et al , 1998) Following opposite foot contact the hip starts to flex due to the flexor moment acting eccentrically and then concentrically just prior to toe off By toe-off the hip has flexed approximately to a range of -4 (±6) to 8 (±7)° and it continues to flex to a peak ranging from 19 (±5) to 30 (±5)° by mid-swing due to the hip flexors acting concentrically to cause hip flexion (Winter et al , 1987, Benedetti et al , 1998, Whittle, 2000) This is followed by slight hip extension just prior to foot strike due to a small hip extensor moment acting eccentrically to prevent further hip flexion prior to heel strike (Figure 2 2) The total excursion of the hip has been reported to be in the range of 32 to 48 (±7)° [Winter et al , 1991, Oberg et al , 1994, Benedetti et al , 1998]

**Knee flexion and extension**

The action of the knee in the sagittal plane can be described as two periods of flexion each starting in relative extension, progressing into flexion, and then returning to extension (Figure 2 3) At heel strike the knee is in an extended position Maximum knee extension occurs just prior to this point during terminal swing ranging from 0 (±5) to 5° [Benedetti et al , 1998, Richards et al , 2003] Following heel strike the knee then starts to flex to generate its first flexion period during the loading response and has been reported to range from 15 (±5) to 20 (±6)° [Oberg et al , 1994, Benedetti et al , 1998, Kerrigan et al , 2001] Immediately following heel strike knee flexion is produced as the body's weight is accepted, by a knee flexor moment acting concentrically to pull the body forward This flexor moment is followed by a knee extensor moment (KM1) acting eccentrically (KP1) at approximately 7% of the gait cycle, which has been reported to reach a peak of -4 (±1)%BW H (Benedetti et al , 1998) At approximately 15% of the gait cycle this knee extensor moment begins to act concentrically (KP2) to extend the knee during midstance By 25% of the gait cycle this moment pattern is reversed with a knee flexor moment (KM2) acting eccentrically which has been reported to be 2 (±2)%BW H (Benedetti et al , 1998) [Figure 2 3] After initial flexion the knee begins to extend during midstance until about 35% of the gait cycle where it reaches a reported extension ranging from 2 (±4)° to 11° (Benedetti et al , 1998, Kerrigan et al , 2001, Yavuzer and Ergin, 2001, Richards et al , 2003) At around 35% of the gait cycle the knee then begins to flex again in order to clear the foot off the ground in the early swing phase At the beginning of this second period of knee flexion, knee flexion is controlled by a knee flexor moment acting concentrically until approximately 45% of the gait cycle The flexion of the knee there after is controlled by an extensor moment (KM3) acting eccentrically (KP3) which is reported to be approximately -3 (±1)%BW H (Benedetti et al , 1998) At toe off the knee is reported to be in approximately 37 (±8)° of flexion (Benedetti et al , 1998) The knee is flexed rapidly after toe off to a maximum during initial swing which has been reported to be in the range of 59 (±4) to 68 (±6)° [Oberg et al , 1994, Benedetti et al , 1998, Kerrigan et al , 2001, Yavuzer and Ergin, 2001, Richards et al , 2003] Following maximum knee flexion the knee begins to extend rapidly to achieve nearly full extension prior to heel strike The extension of the knee during mid and terminal swing is controlled by a small knee flexor moment acting eccentrically (KP4), which decelerates the leg and foot prior to the next heel strike The total excursion of the knee has been reported to be 59 (±5) to 61 (±5)° (Judge et al , 1996, Benedetti et al , 1998)

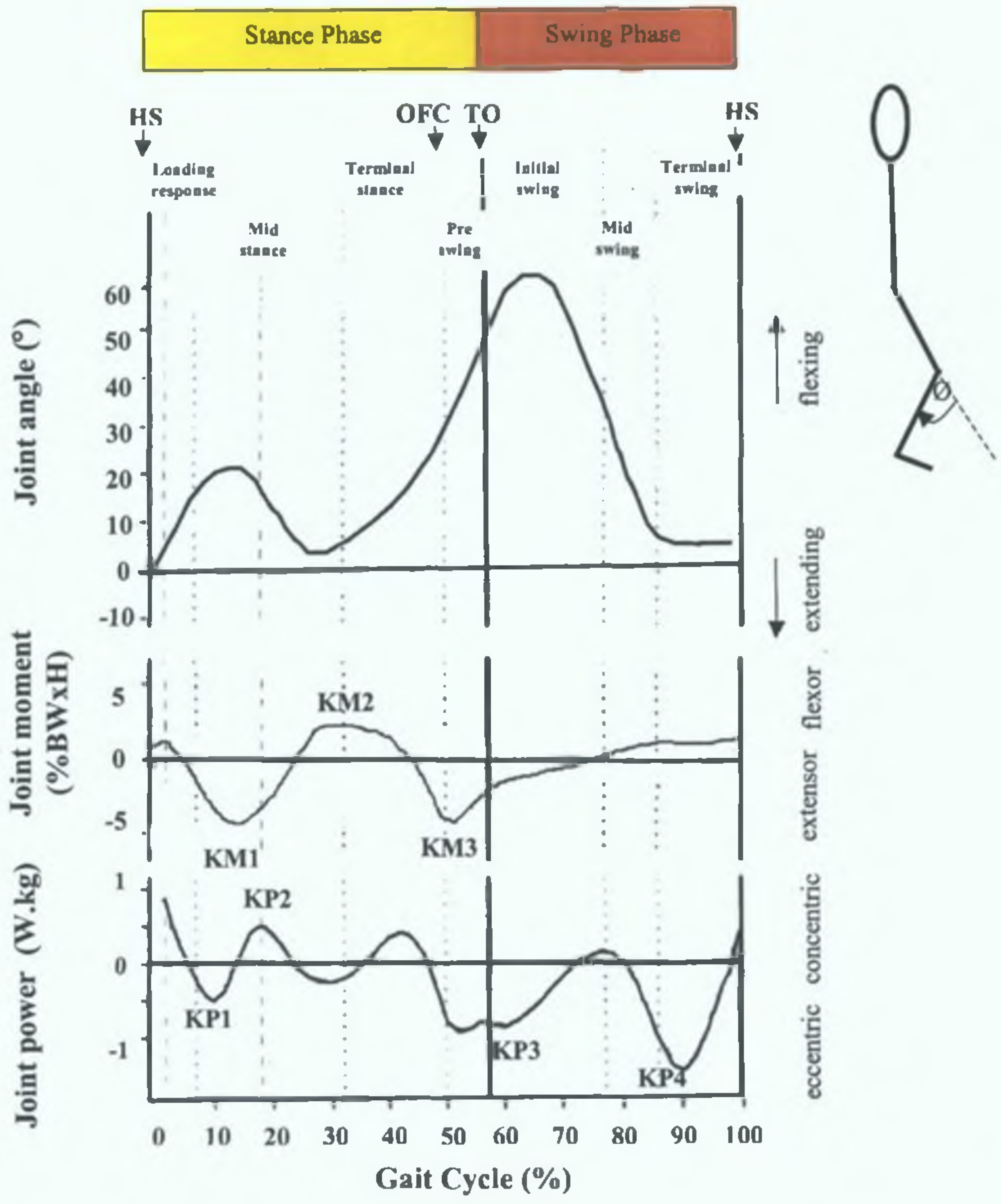



**Figure 2.3**: Knee flexion/extension angle, moment and power in the sagittal plane during normal gait while walking barefoot (adapted from Winter et al., 1984; Kadaba et al., 1989; Benedetti et al., 1998)

HS = Heel strike
OFC = Opposite foot contact
TO = Toe off

**Ankle plantar flexion and dorsiflexion**

The pattern of the ankle in the sagittal plane has normally been described in four sections (Figure 2 4) The first section occurs between heel strike and opposite foot off (0-12% of gait cycle) The angle of the ankle at heel strike has been reported to vary from 4 (±6)° to 5° of dorsiflexion (Benedetti et al 1998, Richards et al , 2003) The foot then moves into a plantarflexed position during the loading response, ranging from -1 to -13 (±5)° [Lehmann et al , 1986, Benedetti et al , 1998, Kerrigan et al , 2001, Cho et al , 2004] Ankle plantarflexion during the loading response is controlled by a small dorsiflexor moment (AM1) acting eccentrically which has been reported to be approximately -1 (±1) %BW H (Benedetti et al , 1998) The second section of the pattern of the ankle joint occurs during single limb stance (12-50% of the gait cycle) The ankle begins to dorsiflex due to the weight of the body passing over the foot Maximum ankle dorsiflexion occurs during midstance and has been reported to be in the range of 8 (±2) to 23 (±7)° [Winter et al , 1987, Benedetti et al , 1998, Kerrigan et al , 2001] Ankle dorsiflexion during this phase of the gait cycle is necessary to allow forward inclination of the leg, which is essential to allow hip extension and forward transport of the vertical trunk to occur From 5 to 42% of the gait cycle there is a plantarflexor moment acting eccentrically (AP1) to control the amount of dorsiflexion achieved during midstance By opposite foot contact the ankle is less dorsiflexed but has not yet returned to a neutral position The third phase begins with opposite foot contact and ends with toe-off Rapid plantar flexion occurs ranging from -7 (±4)° to -22 (±7)° [Winter et al , 1987, Judge et al , 1996, Benedetti et al , 1998, Kerrigan et al , 2001] This is produced by a strong plantarflexor moment (AM2), reported to range from 1 6 Nm/kg (Winter, 1991) to 8 2(±1 1)%BW H (Benedetti et al , 1998), acting concentrically (AP2) This rapid plantarflexion causes a forceful push off The fourth phase is rapid ankle dorsiflexion from toe off till the end of terminal swing which has

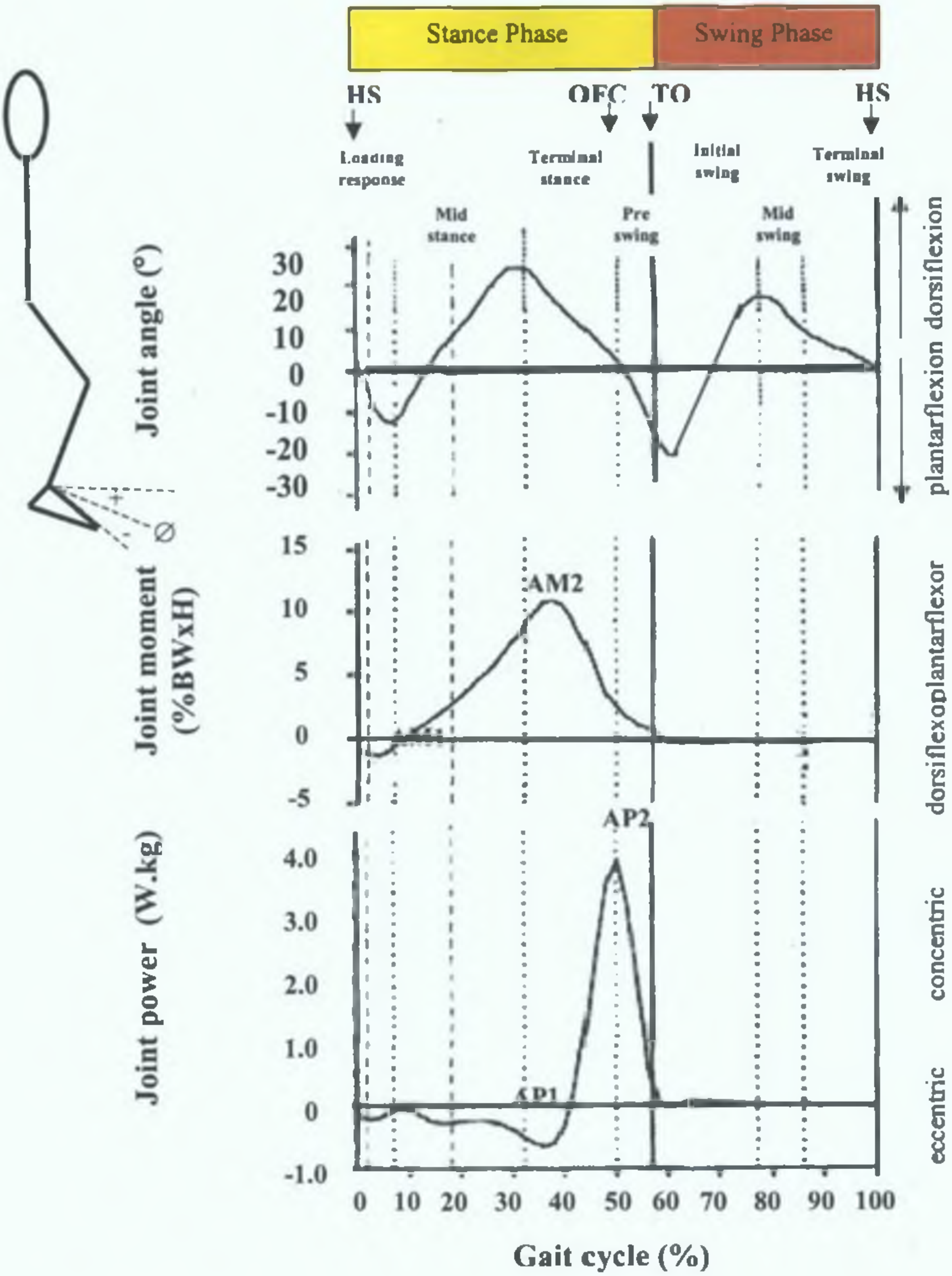



**Figure 2.4:** Ankle plantarflexion/dorsiflexion angle, moment and power in the sagittal plane during normal gait while walking barefoot (adapted from Winter et al., 1984; Benedetti et al., 1998)
HS = Heel strike
OFC = Opposite foot contact
TO = Toe off

been described to range from 2 (±3)° to 8° (Lehmann et al , 1987, Kerrigan et al , 2001, Yavuzer and Ergin, 2001) Dorsiflexion during swing is achieved by a small dorsiflexor moment acting concentrically The total excursion of the ankle has been reported to be 34 (±7)° [Benedetti et al , 1998]

**Coronal Plane**

**Pelvic Obliquity**

In normal subjects the first pelvic obliquity peak occurs during midstance and has been reported to be approximately -2 (±2)° [Benedetti et al , 1998] (Figure 2 5) There is a small secondary rise during the acceleration period of the single stance phase and then the pattern inversely duplicates itself The second pattern begins at opposite foot contact as the opposite leg repeats the same functional tasks and reaches a maximum peak during initial swing which has been reported to range from 0 7 (±0 4)° to 2 0 (±2 0°) [Benedetti et al , 1998, Yavuzer and Ergin, 2001] The pelvic obliquity rise during swing functionally shortens the limb and decreases the vertical translation of the trunk, thus decreasing the displacement of the centre of mass

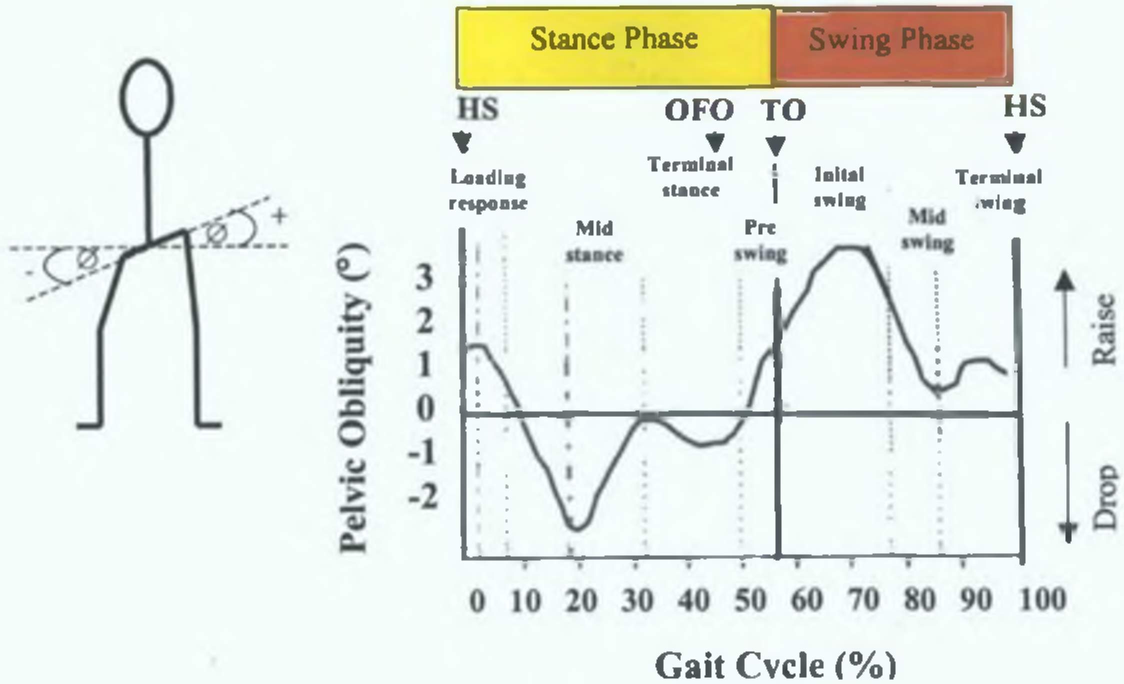



**Figure 2.5:** Pelvic obliquity in the coronal plane during normal gait while walking barefoot (adapted from Benedetti et al., 1998)

HS = Heel strike
OFC = Opposite foot contact
TO = Toe off

**Hip abduction and adduction**

In normal subjects the hip abduction and adduction curve is remarkably similar to the pelvic obliquity curve (Figure 2.6). Maximum adduction of the hip occurs at opposite toe off and has been reported to be -5 (±3)° [Benedetti et al., 1998]. Maximum abduction of the hip occurs at toe off and has been reported to range from 3 (±2)° to 6 (±3)° [Benedetti et al., 1998; Yavuzer and Ergin, 2001].

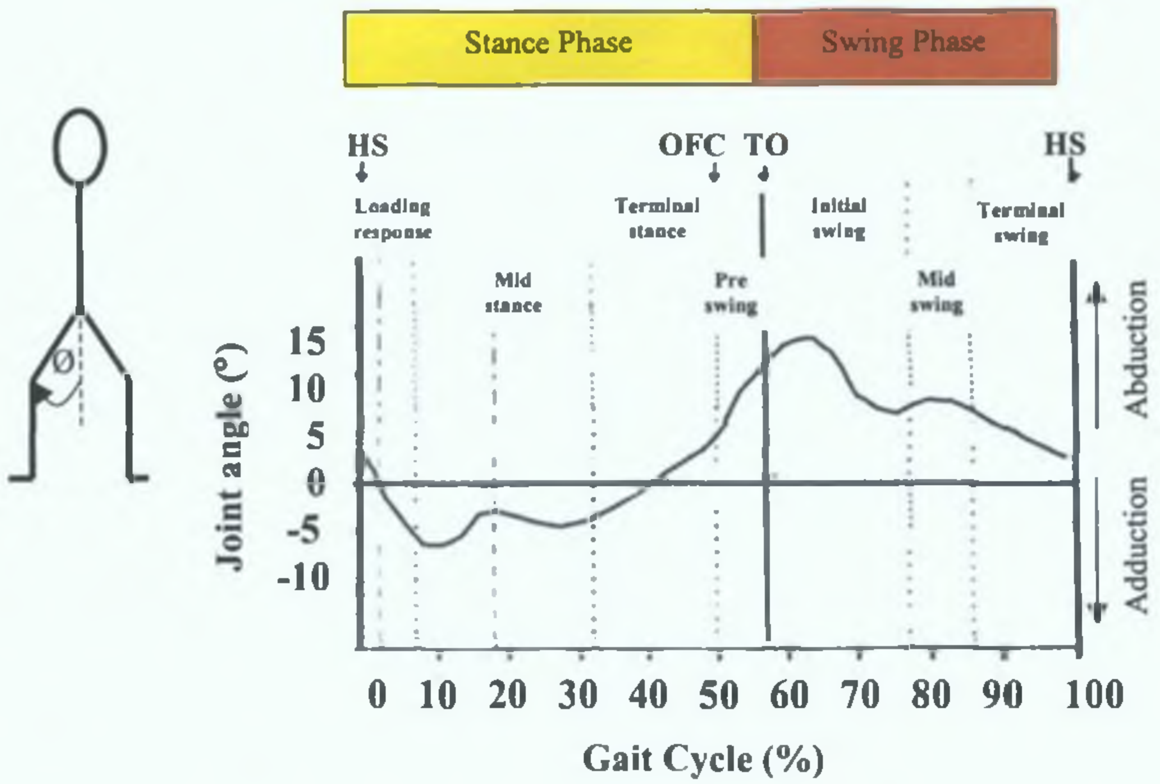



**Figure 2.6:** Hip adduction and abduction angle in the coronal plane during normal gait while walking barefoot (adapted from Benedetti et al., 1998)
HS = Heel strike
OFC = Opposite foot contact
TO = Toe off

### 2.2 Gait Pattern in the elderly

Gait pattern is affected by age. As a person gets older temporal distance parameters and joint kinematic and kinetic variables alter. Since the majority of stroke patients are over 60 years of age, consideration must be given to the likelihood that some of the deviations observed in hemiplegic gait may be due to age related changes. Therefore the gait pattern of a stroke patient should be compared to an age matched group.

### 2 2.1 Kinematic variables

**Temporal distance parameters**

In summary, the effect of age on temporal distance variables is a significant decrease in stride length, cadence and walking speed (Wade et al , 1987, Lehmann et al , 1987, Winter et al , 1990, Elble et al , 1991, Oberg et al , 1993, Turnbull et al , 1995, Von Schroeder et al , 1995, Goldie et al , 1996[b], Lord et al, 1996, Maki, 1997, Witte and Carlsson, 1997, Bohannon, 1997, Kerrigan et al , 1999, Riley et al , 2001, Sadeghi et al , 2002, Hahn and Chou, 2003, Kressig et al, 2003) Younger subjects employ a cadence in the range of 111 6 (±8 3) to 124 8 (±10 2) steps min$^{1}$ (Kadaba et al , 1989, Oberg et al , 1993, Titianova, 2003) whereas the elderly have a reported cadence in the range of 92 3 (±10 9) to 123 6 (±10 8) steps mm$^{1}$ (Table 2 1) The stride length of the younger subjects is in the range of 1 0 (±0 1) to 1 5 (±0 1) m (Kadaba et al , 1989, Benedetti et al , 1998, Titianova, 2003) where as the elderly have a reported stride length in the range of 0 8 (±0 2) to 1 4 (±0 1) m (Table 2 1 and Figure 2 7) Walking speed in the elderly is reported to range from 0 6 to 1 4 m s$^{1}$ (Table 2 1) which is slower than that of younger subjects, which has been reported to range from 1 2 (±0 2) to 1 4 (±0 2) m s$^{-1}$ (Kadaba et al , 1989, Oberg et al , 1993, Benedetti et al , 1998) The reduction in walking speed in the elderly subjects is probably responsible for the observed changes in cadence and stride length (Lehmann et al , 1987)

**Table 2 1** Examples of temporal distance parameters in subjects aged 50+

| Author | No of subjects | Age (yrs) | Method of measurement | Speed (m sec $^{1}$) | Cadence (steps min $^{1}$) | Step length (m) | Stride length (m) | Stance % gait cycle | Swing % gait cycle | Double support % gait cycle | Single Support % gait cycle |
|---|---|---|---|---|---|---|---|---|---|---|---|
| Wade et al (1987[b]) | 34 | 70-79 | Stopwatch | 0 6 | | | | | | | |
| Lehmann et al (1987) | 7 | 57-73 | Motion analysis | 1 0 | | 0 6 | | | | 34 0 | |
| Winter et al (1990) | 15 | 68 9(±4 0) | | 1 3 | 111 8(±8 7) | | 1 4(±0 1) | 65 5(±1 7) | 34 5 | 31 0 | |
| Elble et al (1991) | 20 | 74 7(±6 6) | | 0 9(±0 1) | 104 0(±10 0) | | 0 9(±0 1) | | | | |
| Oberg et al (1993)[Φ] | 15 M<br>15 M<br>14 M | 50-59<br>60-69<br>70-79 | | 1 3(±0 2)*<br>1 3(±0 1)*<br>1 2(±0 2)* | 117 6(±10 8)<br>117 0(±8 0)<br>114 6(±8 4) | 0 6(±0 6)<br>0 7(±0 4)<br>0 6(±0 5) | | | | | |
| | 15 F<br>15 F<br>15F | 50-59<br>60-69<br>70-79 | | 1 1(±0 1)*<br>1 2(±0 2)*<br>1 1(±0 1)* | 121 8(±7 8)<br>123 6(±10 8)<br>121 8(±8 4) | 0 5(±0 3)<br>0 6(±0 4)<br>0 5(±0 4) | | | | | |
| Turnbull et al (1995) | 20 | 61 5(±13 0) | Computerized grid walkway | 1 2(±0 4) | | | | | | | |
| Von Schroeder et al (1995) | 24 | 64 2 | Stride analyzer | 1 1(±0 2) | 95 0(±12 5) | | 1 3(±0 5) | 67 0(±6 6)<br>70 8(±8 6) | | 35 1(±6 2) | 29 2(±8 6)<br>33 0(±6 6) |
| Judge et al (1996)[Φ] | 26 | 79(±6) | Motion analysis | 1 0(±0 1)* | 116 0(±7 0)* | 0 7(±0 1)* | | | | | |
| Goldie et al (1996[b]) | 42 | 62 0 | Stopwatch | 1 2(±0 2) | | | | | | | |
| Lord et al (1996) | 80 | 71 1(±5 2) | | 1 1(±0 2) | 115 4(±11 2) | | 1 2(±0 1) | 64 2(±1 8) | | | |
| Maki (1997) | 26 | 82 0(±6 0) | | 0 7(±0 2) | | | 0 8(±0 2) | | | 19 8(±5 5) | |
| Witte and Carlsson (1997) | 11 | 53 0 | Stop watch | 1 7(±0 1) | | | | | | | |
| Bohannon (1997) | 36 | 65 0 | Stopwatch | 1 3(±0 2) (F)<br>1 4(±0 2) (M) | | | | | | | |
| Bohannon (1997) | 42 | 73 0 | Stopwatch | 1 3(±0 2) (F)<br>1 3(±0 2) (M) | | | | | | | |
| Kerrigan et al (1999) | 20 | 52 0(±17 6) | Motion analysis | 1 3(±0 2) | | | | | | | |
| Mills and Barrett (2001)[Φ] | 8 | 68 9(±0 4) | Motion analysis | 1 4(±0 0) | | | 1 7(±0 0) | | 37 7(±0 6) | 24 6(±1 2) | |
| Riley et al (2001)[Φ] | 14 | 72 9(±5 6) | Motion analysis | 1 2(±0 1)* | 114 0(±26 0) | | 1 2(±0 1)* | | | | |
| Sadeghi et al (2002)[Φ] | 20 | 72 0(±5 5) | Motion analysis | 1 0(±0 2)* | 92 3(±10 9)* | | 1 2(±0 1)* | 63 0(±3 0)* | | | |
| Hahn and Chou (2003) | 9 | 72 0(±6 4) | Motion analysis | 1 2(±0 1) | | | | | | | |
| Kressig et al (2003) | 50 | 79 6(±5 8) | Motion analysis | 1 0(±0 2) | 105 7(±12 7) | | 1 1(±0 2) | 66 0(±3 1) | 34 0(±3 1) | 32 1(±5 8) | |

F= Female M=Male

[Φ] Performed statistical tests to assess if the elderly group was significantly different to the younger group *Statistically different to a younger subject group

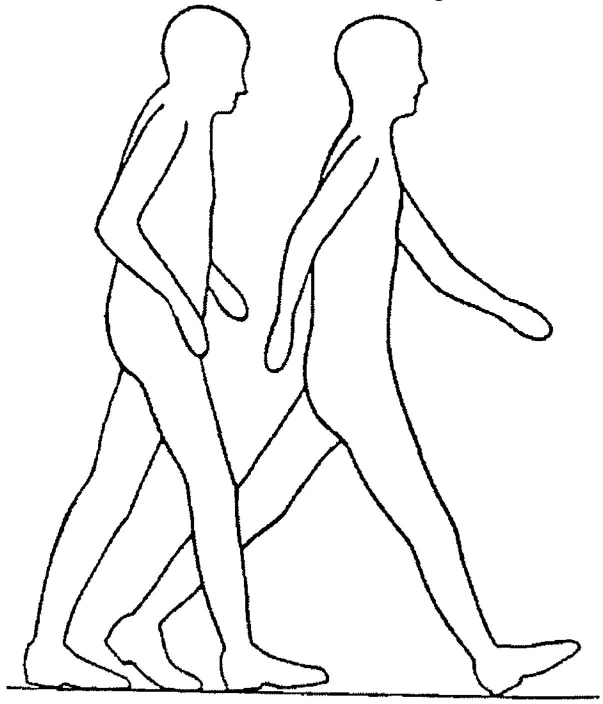



**Fıgure 2 7** Illustratıon of the foot placement and kınematıc gaıt dıfferences between the young and very old (Murray et al , 1969)

**Gaıt cycle perıods**

Unless otherwıse stated, normatıve data for younger subjects are taken from Carollo and Matthews, (2002) In summary, the reported effects of age on gaıt cycle perıods has been an ıncreased percentage of the gaıt cycle spent ın the stance phase and the double support phase and a decreased percentage of the gaıt cycle spent ın the swıng phase and the sıngle lımb support phase (Table 2 1) The majorıty of studıes reportıng the dıfferences ın gaıt cycle perıods between both groups have not statıstıcally tested theır findıngs and have not controlled for the dıfferences ın walkıng speeds between the groups The percentage of the gaıt cycle spent m the stance phase has been reported to be greater ın the elderly subjects rangıng from 64 (±2) to 71 (±9)% [Table 2 1] ın comparıson to younger subjects (62%) The double support phase has been reported to be greater ın the elderly subjects ranging from 20 (±6) to 35 (±6)% [Table 2 1] ın comparison to younger subjects (24%) Mills and Barrett (2001) however, reported no significant difference between young and older subjects for double support percentage of the gait cycle Single limb support phase percentage has also been reported to decrease with age ranging from 29 (±9) to 33 (±7)% [Table 2 1] ın comparison to younger subjects (38%) The percentage of the gait cycle spent in the swing phase has been reported to decrease in the elderly subjects to 34% (Table 2 1) in comparison to 38% in younger subjects Mills and Barrett (2001) however, reported no significant difference between young and older subjects for double support percentage of the gait cycle These gait cycle period changes are believed to be due to the general decrease in muscle strength due to loss of motor neurons, muscle fibrils and aerobic capacity (Trueblood and Rubenstein, 1991, Bendall et al , 1989, Bassey et al , 1988) and a general decrease in dynamic and static balance (Hatzitaki et al , 2004, Jonsson et al , 2005)

**Joint angular kinematics**

Unless otherwise stated normative data for younger subjects are taken from Kerrigan et al , (1998) Age has been reported to have a significant effect on sagittal, coronal and transverse plane joint angular displacements Judge et al (1996) reported that aging resulted in a significant decrease in the range of movement of the pelvis in the coronal and transverse planes Judge et al (1996) and Kerrigan et al (1998) also reported significantly increased range of motion during an anterior pelvic tilt in the elderly subjects in comparison to the younger groups Studies examining the effect of age on total sagittal plane hip excursion have reported both an increase in joint range of movement (Winter, 1991) and also no change (Oberg et al , 1994) Riley et al (2001) and Kerrigan et al (1998) reported that the maximum extension angle of the hip was significantly lower (approximately 6°) in the elderly subjects compared to the younger subjects In contrast the study of Judge et al (1996) reported no significant difference for this variable between the two groups Judge et al (1996) reported that the hip abduction angle at toe off was significantly decreased in the elderly in comparison to the younger subjects by 3 2° The knee sagittal plane angle has also been reported to alter with age The total sagittal plane knee range of motion has been reported by Judge et al (1996) to be significantly less in his elderly subjects by 2 9° in comparison to their younger subjects The knee extension angle at midstance has been reported to increases by about 0 5° per decade while during the swing phase the knee flexion angle has been reported to decrease by 0 5 to 0 8° per decade (Oberg et al , 1993 and Judge et al , 1996) Winter (1991) reported that the elderly maintain a slight knee flexion at the end of swing by 5° in comparison to younger subjects who normally displayed almost a full knee extension This knee flexion at terminal swing in the elderly may facilitate a decrease in the demand for quadriceps activity during loading response (Prince et al , 1997) Prince et al (1997) believe that the knee flexion at terminal swing may account for the observed decreased step length noted in elderly subjects (Oberg et al , 1993, Winter et al , 1990) Judge et al (1996) and Kerrigan et al (1998) did not reported any significant differences between elderly and younger subjects in knee joint angles during the gait cycle Judge et al (1996) and Kerrigan et al (1998) reported a significantly reduced maximum ankle plantarflexion angle by 2 9° and 5 9° during stance in the elderly in comparison to their younger subjects Ankle joint sagittal range of motion has been reported to be decreased m the elderly (25°) in comparison to younger subjects (34 (±7)°) [Winter, 1991, Oberg et al , 1994] Trueblood and Rubenstein, (1991) and Bendall et al , (1989) believed the decrease in ankle joint sagittal range of motion in elderly subjects was due to a weakness of the ankle plantarflexor and dorsiflexor muscles

**Joint angular kinetics**

Unless otherwise stated normative data for younger subjects are taken from Winter et al (1991) Kerrigan et al (1998) reported that the hip flexor moment at initial contact was significantly lower in elderly subjects in comparison to younger subjects by 0 08Nm kg $^{1}$ Mills and Barrett (2001) reported that during the swing phase of the gait cycle elderly subjects produced a significantly greater hip extensor moment by 0 05 Nm Kg $^{1}$ Kerrigan et al (1998) reported that the knee extensor moment at initial contact and the knee flexor moment during midstance were significantly lower in elderly subjects by 0 02 and 0 14Nm kg $^{1}$ Sadeghi et al (2002) reported that the peak knee flexor moment during swing was significantly lower by 0 12 Nm kg $^{1}$ The maximum ankle plantarflexor moment (AM2) has been reported to be lower in the elderly, ranging from 0 5 to 1 4 Nm kg$^{-1}$ [Winter, 1991 and Lamontagne et al , 2002] compared to younger subjects (1 6 Nm kg $^{1}$) [Winter, 1991] Hip power acting concentrically (HP3) has been reported to be significantly greater in the elderly group by 0 15 W kg $^{1}$, if an adjustment is made to the calculations to compare them across a similar walking speed (Judge et al , 1996) Winter et al (1990) examined the knee power curve reported in both normal and elderly subjects They reported that knee power acting eccentrically (KP3) during the transition between stance and swing was higher in the elderly and that the elderly were found to absorb almost half the energy generated during the push-off phase while the young absorbed only 16% However, during both mid and late swing they reported that peak knee power acting eccentrically (KP4) tended to decrease with respect to age Kerrigan et al (1998) reported that the knee power acting concentrically during the loading response and eccentrically during midstance was significantly lower in elderly subjects by 0 16 and 0 17 W kg$^{1}$ In comparison, Judge et al (1996) reported no significant differences in knee power during the gait cycle between a young and elderly group The ankle power acting concentrically has been reported to significantly decrease in the elderly during toe off (AP2), ranging from a 0 44 to 0 6W Kg$^{1}$ (Judge et al , 1996, Riley et al , 2001) This finding has also been supported by the work of Winter et al (1991) In contrast Kerrigan et al (1998) reported that the ankle power acting concentrically during pre swing was significantly higher in the elderly subjects by 0 04 W kg$^{1}$ In all the above studies which compared the joint kinetics of the young and elderly, subjects were not matched for walking speed Future studies need to examine if the differences in joint kinetics would exist if walking speed was matched across age groups

**2 3 Stroke**

The term 'stroke' is synonymous with cerebrovascular accident (CVA), and is a purely clinical definition, which, according to the World Health Organisation, can be defined as clinical signs of a disturbance in cerebral function which lasts longer than 24 hours and is normally vascular in origin (WHO, 1978) Strokes are by far the most common cause of neurological disability in the adult population (Wade, 1992[a]) Stroke is the third largest cause of death in the developed countries and accounts for much disability in the elderly (Bonita, 1992) The onset of stroke is usually sudden with maximum functional deficit at the outset, thus resulting in extreme shock to the subjects Only 66% of people who suffer a stroke and survive regain the ability to walk within one year (Wade et al , 1987), however, the gait will be slow and many will never walk out doors (Wade et al , 1992[b], Wade et al , 1987)

It ıs estımated that 2,393 people dıed ın the Republıc of Ireland ın 2002 from a stroke (Central Statıstıcs Office, 2004) representıng 0 1% of the adult populatıon Unfortunately there are no figures avaılable for Ireland for the number of people who suffer a stroke each year However, based on UK data, whıch reports a survıval rate of 56% (Royal College of Physıcıans, 1989), approxımately 2,978 people ın the Republıc of Ireland wıll survıve a stroke

**2 3 1 Types of Stroke**

There are three major types of stroke whıch can effect dıfferent parts of the braın and result ın characterıstıc deficıts These are ıschemıc, haemorrhagıc and subarachnoıd haemorrhage strokes An ıschemıc stroke ıs the most common (Downıe, 1992) It occurs due to the obstructıon of one of the major cerebral arterıes or theır smaller perforatıng branches The hemıplegıa ıs ınıtıally flaccıd but wıthın a few days thıs gıves way to the typıcal spastıc type The most common artery to be obstructed ıs the mıddle cerebral artery If occluded a dense contralateral hemıplegıa occurs affectıng the arm, face and leg Optıc radıatıon ıs often affected leadıng to a contralateral homonymous hemıanopıa and there may be a cortıcal type of sensory loss (Izzo and Aravabhumı, 1989) Aphasıa can be severe ın left hemısphere lesıons and there may be neglect of the contralateral sıde (Downıe, 1992) In rıght hemısphere lesıons parıetal damage can lead to vıso-spatıal dısturbances (Izzo and Aravabhumı, 1989)

About 5 to 10% of strokes are caused by haemorrhage ınto the deeper parts of the braın If a person survıves the ınıtıal ınsult, profound hemıplegıa and hemısensory sıgns may be elıcıted A homonymous vısual field defect may also be apparent The ınıtıal prognosıs ıs grave but those who do recover often do surprısıngly well as the haematoma ıs reabsorbed (Downıe, 1992)

Between 5 to 10% of strokes are due to subarachnoid haemorrhage with bleeding into the subarachnoid space usually arising from a berry aneurysm situated at or near the circle of Willis (Downie, 1992) The person complains of sudden intense headache often associated with vomiting and neck stiffness Consciousness may be lost and about 10% will die in the first hour or two (Downie, 1992) Of those that remain, 40% will die within the first two weeks and the survivors have a substantially increased risk from re-bleeding for the next six weeks (Downie, 1992) A hemiplegia may be evident at the outset if the blood erupts into the deeper parts of the brain Other focal neurological signs may evolve over the first two weeks because there is a tendency for blood vessels, tracking through the bloody subarachnoid space, to go into spasm leading to secondary ischemic brain damage (Downie, 1992)

**2 3 2 Effects of stroke on gait**

The hemiparesis that results from stroke may have a profound effect upon a person's ability to walk Even those who achieve functional ambulation display gait patterns that are very different from those of able-bodied persons (Lehmann, 1987) The walking pattern of each stroke patient is variable and depends upon the site and severity of the lesion and the manner of compensation employed (Perry, 1969) Impaired walking ability greatly contributes to general functional disability after stroke The term hemiplegic gait has been frequently used by many clinicians to describe a pattern of limb movement and posture during ambulation characterised by slow, laborious, partially uncoordinated limb movement

Many stroke patients believe the most important goal of rehabilitation is to restore walking ability as this determines whether they will go home or to a nursing home and whether they will return to the pre-stroke level of productivity (Shiavi et al , 1987[a], Shiavi et al , 1987[b], Wall, 1986, Bohannon et al , 1995) Walking is a significant part of functional recovery following stroke and depends on several factors, including size and location of the infarct (Perry, 1969) and pre-stroke health The following section will discuss the mechanisms which underpin the altered gait patterns evident following a stroke

### 2 3 3 Mechanisms underlying abnormal gait pattern in patients with stroke

Of the 56% of patients who survive stroke, 75 to 84% of these will be left with a physical disability which will affect their gait (Sheikh et al , 1983, Wade et al , 1985) These gait deviations can lead to pain, joint damage, a risk of falling and increased energy expenditure (Morris et al , 1992, Olney et al , 1986) Gait disorders resulting from stroke are believed to be caused by many factors (Musa, 1986, Dietz et al , 1981, Conrad et al , 1983, Pierrot-Deseilligny et al , 1983), including muscle weakness or poor muscle activation, sensory loss, spasticity (hyperactive stretch reflexes), loss of normal balance mechanisms and ankle joint problems These factors will be discussed in more detail below

#### 2 3 3 1 Muscle weakness and poor muscle activation

Weakness or decreased muscle activity is a prominent feature in stroke patients Research has demonstrated that weakness or decreased agonist activity exists in both the upper and lower extremities of stroke patients (Gardiner, 1996) In addition, weakness has been demonstrated not only in the affected, but also in the unaffected limb of the stroke patient (Gardiner, 1996) This may be caused by generalised inactivity following the stroke

Weakness of lower limb muscles may also be considered as a possible cause of the poor weight-acceptance pattern found in hemiplegic gait Although investigators have found EMGs recorded during hemiplegic gait to be decreased, they still often consider them to be within normal limits (Hirshberg and Nathanson, 1952, Bogardh and Richards, 1981, Knutsson and Richards, 1979) The investigations of Peat et al (1976) and Richards and Knutsson (1974) showed that the subgroup of muscles most frequently termed 'weak' were the hip adductors, hamstrings, gastrocnemius, and pretibial muscles Because the majority of patients with a weakness usually walk with an aid of some sort, their affected leg does not bear full weight at floor-contact However, Peat et al (1976) found that the previously inactive unloaded muscles were activated if the patients bore more weight on their affected leg Any increased load on a muscle stretches the muscle spindle receptors, increasing the Ia discharge and facilitates a muscle contraction (Ezyaguire and Fidone, 1975) The patients in the study by Peat et al (1976) exhibited a decreased stretch reflex capacity when walking The authors suggest that patients may need increased weight bearing to activate their muscles Nevertheless, the EMG pattern of these muscles was incorrect with the normal peaks of activity during locomotion being absent and periods of maximum activity occurring during midstance (Peat et al , 1976) Clearly the proper phasing of muscle activation is essential (Knutsson, 1983) The co-activation of agonist and antagonist muscles has often been considered a cause of improper foot ground contact in spastic hemiplegia (Brunnstrom, 1965, Dietz and Berger, 1983, Perry, 1975) Knutsson and Richards (1979) observed that in their Type III stroke patients co-activation existed between the gastrocnemius and tibialis anterior muscle thus preventing correct ankle dorsiflexion at heel strike

The mechanism behind muscle weakness and poor muscle activation is believed to be due to a loss in the number and rate of firing of motor units, a change in the recruitment order and a change ın the stıffness of muscles In normal muscle, force productıon ıs augmented by ıncreasıng the number of actıve motor unıts and the fırıng rates of already actıve motor unıts, or by a combınatıon of both (Bourbonnaıs and Vanden Noven, 1989, Gardıner, 1996) In hemıparesıs there ıs a reductıon m the number of motor unıts actıvated, a decrease ın fırıng rates and a change ın the recruıtment order of the motor unıts (Bourbonnaıs and Vanden Noven, 1989, Gardıner, 1996) Mc Comas et al (1973) estımated that between the second and sıxth month followıng a stroke only half the normal number of motor unıts were functıonıng Dıetz et al (1986) attrıbuted thıs to degeneratıon of the cortıcospınal tract after a stroke wıth resultıng transynaptıc changes ın motor neurons Clınıcally, as a result of these physıologıcal changes stroke patıents may dısplay an ınabılıty to generate normal levels of force and subsequent have dıfficulty ın performıng fast movements Stıffness of muscles may be another ımportant factor contrıbutıng to the movement defıcıts of stroke patıents wıth spastıcıty Studıes have reported that the ıncreased resıstance that was found durıng a muscle contractıon ın stroke patıents was due to a non-reflex medıated stıffness (Dıetz et al , 1991, Ibrahım et al , 1993, Lee et al , 1987, Sınkjaer and Magnussen, 1994) These authors attrıbuted thıs non-reflex medıated stıffness to changes ın mechanıcal elastıc characterıstıcs of muscle and connectıve tıssues followıng damage to the central nervous system These changes may be secondary to dısuse resultıng from the hemıparesıs

### 2 3 3 2 Sensory loss

Clınıcal experıence has shown that loss of sensatıon ın patıents wıth stroke can result ın extreme dıfficulty wıth re-learnıng movements (Ashburn, 1997) Buskırk and Webster, as long ago as 1955, reported that persıstent sensory loss was related to poor prognosıs and an extended length of hospıtal stay Lesıons of the cerebral cortex cause defıcıts ın sensory perceptıon (Izzo and Aravabhumı, 1989) Sensory perceptıon defıcıts may include loss of two-point discrimination, discrimination of size, shape, texture or point localisation, or loss of stereognosis (Izzo and Aravabhumi, 1989) Simple sensory deficits include impairment of touch, pinprick and proprioception In the lower extremity, loss of proprioception at the hip or knee may make walking unsafe or impossible Stroke patients may also experience a distorted body image and a perceptual neglect of one side of the body There may be disturbance of awareness of certain body parts and their relationship to each other and their position in space (Halligan and Marshall, 1994) Overall a loss of sensation may impair the subject's ability to move and balance normally (Halligan and Marshall, 1994)

**2 3 3 3 Spasticity (Hyperactive stretch reflexes)**

Spasticity is a motor disorder characterised by a velocity-dependent increase in tonic stretch reflexes with exaggerated tendon jerks It is a result of hyper excitability of the stretch reflex and can be one symptom of an upper motor neuron lesion (Gardiner, 1996) The presence of spasticity or increased muscle tone is assessed by the examiner by feeling for excessive resistance as a limb is moved passively through its range of motion (Katz and Rymer, 1989) It may also be diagnosed by an increase in EMG activity during the passive stretch of a muscle (Ibrahim et al , 1993, Powers et al , 1989) Changes of muscle tone may not be entirely of neural origin In an activity contracting muscle it has been reported that the increase in muscle tone may be due to an increase in stiffness of the muscle (Lee et al , 1987, Sinkjaer and Magnussen, 1994) It has been proposed that this stiffness was due to changes in the mechanical properties of muscle following a stroke (Dietz et al , 1991, Ibrahim et al , 1993, Lee et al , 1987)

#### 2.3.3.4 Balance

Berg et al. (1989) stated that balance has three basic dimensions: maintenance of position, postural adjustment to voluntary movements and reaction to external disturbances. Following a stroke, impairment in balance may occur (Collen, 1995; Leonard, 1990). It has been noted that the maintenance of position, as measured by postural sway during quiet standing, is increased following stroke (Shumway-Cook et al., 1988). Some authors have also attempted to measure the postural adjustments of the stroke patients to voluntary movements by examining their ability to transfer weight during gait (Leonard, 1990; Goldie et al., 1996[a]). These studies have reported that there was unequal weight transference between the affected and unaffected legs in both the lateral and forward directions (Leonard, 1990; Goldie et al., 1996[a]). Dettmann et al. (1987) reported that in the absence of good balance, the affected limb of the stroke patient apparently compensates by taking smaller steps and walking more slowly.

#### 2.3.3.5 Ankle joint problems

Any ankle joint problems may contribute to the phenomenon of poor weight acceptance. For instance contractures, pain and any previous ankle injury may well lead to decreased ankle dorsiflexion. However, in a review of the literature, no reference to ankle joint problems was found. Clinical experience would suggest that it is a frequent cause of poor weight acceptance.

## 2.4 Equinus deformity of the foot

Postural abnormalities and deformities of the lower extremity can develop frequently following a stroke (Whittle, 2000). The most commonly reported problems include hip adductor spasticity and contracture, reduced knee flexion, knee flexion contracture, ankle equınus, equınovarus or planovalgus and toe flexıon deformıtıes (Izzo and Aravabhumı, 1989, Craık, 1991, Whıttle, 2000)

Equınus deformıty of the foot ıs a motor defıcıency caused by total or partıal central paralysıs of the muscles ınnervated by the common peroneal nerve and occurs often unılaterally followıng a stroke (Voıgt and Sınkjaer, 2000) Approxımately 60 papers have been wrıtten on the effects of wearıng an AFO on equınus deformıty of the foot ın adult stroke Wıthın these papers there appears to be no agreed upon crıterıa on whıch to dıagnosıs the presence of equınus deformıty of the foot Few papers dıscuss how they dıagnosed the presence of equınus deformıty of the foot Two papers stated that they assessed ıf actıve dorsıflexıon was present ın the foot (Burdett et al , 1988, Mueller et al , 1992), another two papers assessed ıf actıve ankle control was present (Gok et al , 2003, Lehmann et al , 1987) and lastly sıx papers made an attempt to assess the level of spastıcıty of the ankle dorsıflexors and plantarflexors (Corcoran et al , 1970, Tyson et al , 1998, Burdett et al , 1988, Hesse et al , 1996, Hesse et al , 1999, Gok et al , 2003) None of these papers however, reported the level of dorsıflexıon strength, ankle control or spastıcıty whıch lead to the patıents been classıfied as havıng equınus deformıty of the foot Another paper (Ohsawa et al , 1992) stated that they examıned the gaıt pattern of the stroke patıent for equınus durıng swıng Thıs method of assessment however, may be ınapproprıate at present, as research has yet to prove conclusıvely that these patıents place theır foot ın a plantarflexed posıtıon throughout the swıng phase of the gaıt cycle (Gok et al , 2003, Burdett et al , 1988, Lehmann et al , 1987) The mam crıterıa set by the majorıty of papers on equınus deformıty of the foot for patıent ınclusıon was a hıstory of wearıng of an AFO (Smıth et al , 1982, Mojıca et al , 1988, Yamamoto et al , 1993[b], Hesse et al , 1996, Yamamoto et al , 1997, Butler et al , 1997, Hachısuka, 1998, Chen et al , 1999, de Wıt et al , 2004, Danıelsson and Sunnerhagen,

2004) or a referral from a physiotherapist as a patient who would benefit from wearing an AFO to facilitate their rehabilitation progress (Wong et al , 1992, Churchill, 2003)

### 2 4 1 Incidence of equinus deformity of the foot

The incidence of equinus deformity of the foot in adult stroke patients has been reported to vary between 10 and 20% (Burridge et al , 1997, Leane et al , 1998, Verdie et al , 2004) Based on the number of people in Ireland suffering a stroke per annum this would mean that between 298 to 596 stroke patients develop equinus deformity of the foot per annum

### 2 4 2 Mechanism behind equinus deformity

The overall result of an equinus deformity of the foot is an inability to dorsiflex the foot To obtain normal dorsiflexion of the foot the dorsiflexor muscles must be of sufficient strength to lift the foot during swing and the plantarflexor muscles must be able to eccentrically lengthen to allow and control dorsiflexion to occur during midstance and swing In patients with equinus deformity due to stroke it is suggested that the dorsiflexors become unable to dorsiflex due to weakness and the plantarflexor muscles become unable to lengthen eccentrically due to many factors Theses factors include increased plantarflexor muscle spasticity, the presence of a contracture in the muscles of the affected limb or increased muscle stiffness in the plantarflexor muscles (Kerrigan and Sheffler, 1995, Perry, 1978, Dietz et al , 1981) While weakness of the dorsiflexor muscles has been postulated to be a major contributor to the formation of equinus deformity of the foot, no study to date has directly examined this variable (Perry, 1978) Lamontagne et al (2002) reported that in four acute stroke patients adequate dorsiflexion of the foot during swing was not achieved because the dorsiflexor muscles were not strong enough to overcome the muscle stiffness of the plantarflexors muscles

This was evident by decreased amplitude in the EMG pattern of the dorsiflexor muscles (Lamontagne et al , 2002)

Burridge (1997) examined the effect of spasticity on the formation of equinus deformity of the foot with the use of surface EMG She reported that these patients demonstrated four patterns of activation of their dorsiflexor and plantarflexor muscles Pattern one consisted of the dorsiflexors and plantarflexor muscles contracting together at the same time when dorsiflexion should be occurring In this EMG pattern, dorsiflexion could not be achieved as the moments generated by both muscles at the same time might result in no movement of the ankle EMG pattern two consisted of patients with the ability to selectively activate only the dorsiflexor muscles and not the plantarflexor muscles Dietz et al (1981) reported a similar pattern in stroke patients using EMG during the swing phase of gait In these patients the dorsiflexor muscles could contract but the plantarflexor muscles would not eccentrically lengthen to allow dorsiflexion to occur EMG pattern three consisted of patients with an inability to modulate the force of the contraction of the dorsiflexor and plantarflexion muscles In these patients sufficient force could not be generated to dorsiflex the foot concentrically when necessary during the gait cycle The last EMG pattern was an equmus deformity of unknown cause

In stroke patients, studies have suggested that as a result of chronic spasticity a permanent shortening of the muscle tissue occurs, resulting in increased muscle stiffness (Thilmann et al , 1991, Wolf and Minkiwitz, 1989) Muscle stiffness is defined as the mechanical response to a tensile load on the non contracting muscle (Harlaar et al , 2000) Two studies have reported the presence of muscle stiffness in the plantarflexor muscles in stroke patients at the ankle joint (Thilmann, 1991, Lamontagne et al , 2002) Thilmann et al (1991) reported that this stiffness resulted in 50% more resistance to dorsiflexion than a normal unaffected ankle Therefore for dorsiflexion to occur in patients with increased muscles stiffness, the dorsiflexor muscles need to be approximately twice the strength of normal

In summary the literature would suggest that equinus deformity of the foot is caused by,

1 Incorrect strength ratio of the dorsiflexor muscles to the plantarflexor muscles due to the presence of muscle stiffness in the plantarflexor muscles
2 An inability to generate sufficient force in the dorsiflexor muscles to cause dorsiflexion
3 An inability of the plantarflexor muscles to eccentrically lengthen to allow dorsiflexion to occur
4 Co-activation of the dorsiflexor and plantarflexor muscles of the foot at inappropriate phases of the gait cycle
5 Equinus deformity of unknown cause

**2 4 3 The gait of stroke patients with equinus deformity**

It is difficult to discuss the isolated effects of an equinus deformity of the foot on the gait pattern of stroke patients as this deformity does not tend to occur in isolation It is normally only one of many deformities which may be present in the stroke patients In addition in clinical trials describing the gait pattern of stroke patients it is not commonly stated whether the deformity was present of not The following section examines the reported gait pattern of adult stroke patients where equinus deformity was specifically reported The gait patterns of stroke patients were included in this review if they were classified as having equmus deformity of the foot or if the patients were assumed to have this condition as they were being treated for it as indicated by use of either an ankle foot orthosis or a drop foot stimulator The reader should in particular draw their attention to the large standard deviations reported in the literature for this patient group This would suggest that stroke patients with equinus deformity of the foot may have many different neuromuscular deficits as a result of their stroke and may be employing various different compensating strategies

**Temporal Distance Parameters**

The preferred walking speed of stroke patients with equinus deformity walking varies greatly from 0 1 to 0 9 (±0 5)m s$^{1}$(Table 2 2) Only two studies examined stroke patients with equinus deformity of the foot walking at a maximum speed and found it to be in the range of 0 5 (±0 2) to 0 8 (±0 3)m s$^{1}$ (Waters et al , 1975, Iwata, 2003) In stroke patients with equinus deformity of the foot cadence varies greatly from 60 (±5) to 92step min$^{1}$ for the affected side and 79 (±7)step min$^{1}$ for the unaffected side, indicating an asymmetry of gait (Table 2 2) Only one study examined the cadence of stroke patients with equmus deformity of the foot walking at their maximum speed and found that it could increase by 20% (Table 2 2) It must be remembered however, that Lehmann et al (1987) reported that the reduced cadence of the stroke group was statistically similar to normal subjects walking at a matched speed Therefore the reported cadence of this group may be caused directly by their walking speed and not the deformity The step length of stroke patients with equmus deformity of the foot varied between 0 1 and 0 3m for the unaffected leg and 0 2 to 0 4 m for the affected sides (Table 2 2) In general the step length of the affected side was found to be greater than that of the unaffected side, with the exception of the study by Wong (1992), demonstrating that their gait is asymmetrical Lehmann et al (1987) believed that the difference in step length between the affected and unaffected leg was due to a reduction in hip extension on the affected leg Accompanying this was a shorter duration of stance

**Table 2 2** Gait cycle parameters in stroke patients with equinus deformity walking at preferred and maximum walking speed (MWS)*

| Author | Age (years) | No of patients | Assessment method | Walking speed (m sec⁻¹) | Cadence (steps min⁻¹) | Step length (m) | Stride length (m) | Stride time (sec) | Cycle duration (sec) |
|---|---|---|---|---|---|---|---|---|---|
| Corcoran et al (1970) | 45 1 | 13 | Timed | 0 76 | | | | | |
| Waters et al (1975) | Not stated | 13 | Not stated | 0 6(±0 3)<br>0 8(±0 3) MWS | Aff 79 0(±24 0)<br>Aff 95 0(±19 0) MWS | | Aff 0 8(±0 2)<br>Aff 1 0(±0 2) MWS | | |
| Hale and Wall (1987) | 51-73 | 7 | Footswitches | 0 1 - 0 6 | | | | Aff 1 6 – 3 0 | |
| Lehmann et al (1987) | 59-75 | 7 | Motion Analysis | 0 5 | | Unaff 0 3<br>Aff 0 4 | | | |
| Cozean et al (1988) | 52 0 | 10 | 3D Motion Analysis | | | | Aff 0 5(±0 1) | | Aff 2 7(±1 0) |
| Burdett et al (1988) | Not stated | 19 | Not stated | 0 4(±0 2) | | Unaff 0 3(±0 1)<br>Aff 0 3(±0 1) | Aff 0 6(±0 2) | Aff 2 7(±0 9) | |
| Mojica et al (1988) | 46-66 | 8 | Timed | 0 6 | Aff 91 8(±25 4) | | Aff 0 6(±0 4) | | |
| Wong et al (1992) | Not stated | 6 | Footswitches | | | Unaff 0 3<br>Aff 0 3 | | | |
| Granat et al (1996) | 56 0 | 16 | Foot-switch | 0 9(±0 6) | | | | | |
| Hesse et al (1996) | 55 2 | 19 | Stop watch | 0 3(±0 2) | Aff 65 0(±21 0) | | Aff 0 6(±0 2) | | |
| Dieli et al (1997) | | 3 | Stride Analyser | 0 5 | Aff 77 0 | | Aff 0 7 | | |
| Tyson (1998) | 42 5 | 4 | Stop watch | 0 2 | | Unaff 0 1<br>Aff 0 2 | Aff 0 3 | | |
| Taylor et al (1999[b]) | 55 4(±18 2) | 111 | Stop watch | 0 6(±0 3) | | | | | |
| Hesse et al (1999) | 58 2 | 21 | Timed | 0 3(±0 2) | Aff 62 0(±17 0) | | Aff 0 6(±0 2) | | |
| Voigt and Sinkjaer (2000) | 59 1(±8 7) | 8 | Motion Analysis | | Aff 83 0(±7 5)<br>Unaff 79 2(±7 1) | | | Aff 1 4(±0 1)<br>Unaff 1 5(±0 1) | |
| Gok et al (2003) | 54(39-65) | 12 | 3D Motion analysis | 0 3(±0 1) | Aff 62 3(±20 6) | Aff 0 3(±0 1) | | | |
| Iwata (2003) | 62 2(±7 8) | Not stated | Timed | 0 5(±0 2) MWS | Aff 82 6(±23 2) | | Aff 0 7(±0 2) | | |
| Churchill et al (2003) | 25-60 | 5 | 2D Motion Analysis | 0 3(±0 02) | Aff 60(±4 8) | | Aff 0 6(±0 1) | | Aff 2 1(±1 8) |
| De Wit et al (2004) | 61 2 | 20 | IR Beam | 0 5 | | | | | |
| Wang et al (2005) | 59 9(±13)<br>62 3(±11 8) | 42 SDG<br>61 LDG | Stopwatch | 0 6(±0 3)<br>0 6(±0 3) | Aff 75 1(±27 5)<br>Aff 75 8(±31 2) | | | | |

Aff= Affected leg SDG = Short duration group
Unaff=Unaffected leg LDG = Long duration group

* Values are for preferred walking speed unless indicated as maximum walking speed (MWS)

on the affected side and a short duration of swing on the unaffected side Lehmann et al (1987) suggests that these gait deviations are due to instability of the affected leg In stroke patients with equinus deformity of the foot stride length varied between 0 3 to 0 8m (Table 2 2) In stroke patients with equinus deformity of the foot stride time varied between 1 4 to 2 7s on the affected side and 1 5s on the unaffected side (Table 2 2) A longer stride time was present on the unaffected side in comparison to the affected side indicating an asymmetry of gait Only two studies assessed the gait cycle duration in the stroke patients with equinus deformity of the foot and it was reported to vary between 2 1 (±1 8) to 2 7 (±1)s (Table 2 2) In stroke patients with equinus deformity of the foot the temporal distance parameters of cadence, step length and stride time indicate that these patients lack symmetry in their gait pattern

**Gait cycle periods**

Stance phase duration of the gait cycle has been reported to vary between 56 to 69% on the affected leg and 75 to 80% on the unaffected leg, indicating an asymmetry of their gait pattern (Table 2 3) Therefore stroke patients with equinus deformity of the foot spend a significantly longer time placing weight through their unaffected leg Lehmann et al (1987) believes that the instability of the affected leg prompts the patient to shift their weight as early as possible to the unaffected leg Single limb support duration of the gait cycle varies between 14 and 39% on the affected side and between 43 and 44% on the unaffected side, indicating an asymmetry of their gait pattern (Table 2 3) These results supports the hypothesis that stroke patients with equinus deformity of the foot prefer to spend less time placing weight through their affected leg The percentage of the gait cycle spent in double support in stroke patients with equinus deformity of the foot varied from 17 to 49% in the affected leg and was reported to be 20% in the unaffected leg, indicating an asymmetry of their gait pattern (Table 2 3) However, it

**Table 2 3:** Percentage of tıme spent ın varıous aspects of the gaıt cycle of stroke patınets wıth equınus deformıty walkıng at a preferred pace

| Author | Age (years) | No of patıents | Assessment method | Stance % of gaıt cycle | Sıngle Support % of gaıt cycle | Double Support % of gaıt cycle | Heel Strıke % of gaıt cycle | Swıng % of gaıt cycle |
|---|---|---|---|---|---|---|---|---|
| Hale and Wall (1987) | 51-73 | 7 | Footswıtches | | Aff 14 4-39 0 | | | Aff 28 0-49 6 |
| Lehmann et al (1987) | 59-75 | 7 | Motıon Analysıs | Aff 69 0<br>Unaff 80 0 | | Aff 49 0 | Aff 3 0 | Aff 31 0<br>Unaff 20 0 |
| Wong et al (1992) | Not stated | 6 | Footswıtches | Aff 56 2<br>Unaff 75 6 | Aff 26 3<br>Unaff 44 4 | Aff 17 2<br>Unaff 20 1 | | |
| Granat et al (1996) | 56 0 | 16 | Foot-swıtch | | | | Aff 7 0(±7 9) | |
| Dıelı et al (1997) | | 3 | Strıde Analyser | Aff 57 3<br>Unaff 74 7 | Aff 25 3<br>Unaff 42 7 | | | Aff 42 7<br>Unaff 25 3 |
| Hesse et al (1999) | 58 2 | 21 | Tımed | Aff 69 1(±5 9)<br>Unaff 79 2(±4 0) | | Aff Inıtıal 21 6<br>Aff Termınal 27 0 | | Aff 30 8(±5 9)<br>Unaff 20 6(±3 9) |

Aff= Affected leg

Unaff = Unaffected leg

must be remembered that the double support percentage of gait is significantly influenced by walking speed (Lehmann, 1987) with double support percentage increasing as the walking speed decreases The definition of double support percentage is not clearly stated in any of the reviewed studies Therefore it is not possible to determine how a differentiation was made between double support percentage on the affected and unaffected leg The affected leg in stroke patients with equinus deformity of the foot was reported to spend from 3 to 7% of the gait cycle m heel strike (Table 2 3)

The percentage of the gait cycle spent in swing in stroke patients with equinus deformity of the foot was reported to range from 28 to 50% for the affected leg and from 20 to 25% in the unaffected leg (Table 2 3) Therefore stroke patients with equinus deformity of the foot have greater difficulty placing their unaffected leg in swing for a longer length of time as this limb is needed to support their body weight as much as possible during the gait cycle The research findings clearly show there is a lack of symmetry in their gait pattern The literature would also suggest that these patients had evidence of limb clearance problems with prolonged single support times on the unaffected side, a reduced percentage of gait spent in stance and an incorrect stance/swing ratio on the affected side (Sutherland, 1984)

**Joint kinematics**

There has been a lack of research examining the joint kinematics of stroke patients with equinus deformity of the foot Due to a lack of research in this area this section will also discuss the joint kinematics of chronic stroke patients who are a minimum of six months post cerebral vascular accident as up to 20% of these patients may also have this deformity present (Verdie et al , 2004) The addition of the literature on the angular kinematics and kinetics during the gait cycle of chronic stroke patients may give the reader a clearer picture of the possible gait deviations, which may be present in the stroke patients with equinus deformity of the foot

**Hip flexion and extension joint angular kinematics**

Lehmann et al (1987) appears to be the only study which examined the sagittal plane movement of the hip joint in patients with equinus deformity of the foot while patients wore shoes They stated that in these patients the hip was kept more flexed than normal patients during the gait cycle However, they reported that this finding may be due to the speed that the stroke patients were walking Lehmann et al (1987) also reported that the stroke patients exhibited an average of 14° less hip extension during the late midstance and pushoff phases which was significantly less than normal subjects walking at a matched speed This decrease in hip extension would be consistent with that reported for chronic stroke patients Lehmann et al (1987) also noted a significant 8 cm reduction in contralateral step length when their stroke patients were compared to an age and speed matched group They believed that stroke patients compensate for a lack of hip extension by excessively rotating their trunk forward on the swing side in late stance phase, which slightly decreases the contralateral step length In contrast Wong et al (1992) reported an increase in contralateral step length of 6 cm

Chronic stroke patients demonstrate differences in sagittal plane hip joint angular kinematics in comparison to normal healthy subjects Maximum hip flexion in swing for the affected leg has been reported to range from 19 (±6) to 21 (±9)° [Olney et al , 1991 and 1994, Detrembleur et al , 2005], which is below the normal levels (29 8±4 8°) as described by Benedetti et al (1998) Maximum hip extension has been reported to range between -5 (±10) to +10 (±7)° [Olney et al , 1991 and 1994, Detrembleur et al , 2005], which is above the values (-10 0±5 1°) described by Benedetti et al (1998) Kim and

Eng (2004) reported on the total excursion of the hip of the affected leg They reported a greater decrease in hip excursion in the affected leg in comparison to the non-affected leg (35 ±19° affected leg, 41 ±8° unaffected leg)

**Knee flexion and extension angular joint kinematics**

Lehmann et al (1987) reported that patients with equinus deformity of the foot, knee extension angle during midstance (4°) was greater but not significantly different than normal healthy age matched subjects (7°) Many authors have reported a similar alteration in knee angle during midstance in chronic stroke patients (Knutsson and Richards, 1979, Morris et al , 1991, Olney et al , 1988, Pinzur et al , 1986 and 1987, Richards and Knutsson, 1974, Takebe and Basmajian, 1976, Van Griethuysen et al , 1982) Lehmann et al (1987) also reported on the amount of knee flexion during midstance in patients with equinus deformity The amount of knee flexion was higher (17°) but not significantly different in the stroke patients compared to age match subjects walking at a matched speed (14°) Many authors have reported a similar alteration in knee angle during midstance in chronic stroke patients (Bogardh and Richards, 1981, Carlsoo et al , 1974, Knutsson and Richards, 1979, Olney et al , 1986, 1988 and 1989, Takebe and Basmajian, 1976, Trueblood et al , 1989) Increased knee flexion during the stance phase of gait may result in a decrease in peak ipsilateral hip extension, which was also noted by Lehmann (1987) These alterations in hip and knee angles may result in a decreased contralateral step length because decreased hip extension and excessive knee flexion decrease the extent to which the hips are transported forward over the stance foot

Lehmann (1987) reported that the peak knee flexion angle in swing in the stroke patients was significantly less (by 17°) than that of age matched subjects walking at a matched speed A decrease knee flexion during swing is commonly observed in people with chronic stroke There are reports in the literature of decreased knee flexion during pre-swing (Knutsson and Richards, 1979, Olney et al , 1988 and 1989, Takebe and Basmajian, 1976), at toe-off (Brandell, 1977, Burdett et al , 1988, Finley and Karpovich, 1964, Knutsson and Richards, 1979, Olney et al , 1986 and 1988, Takebe and Basmajian, 1976) and during the swing phase (Knutsson and Richards, 1979, Lehmann et al , 1987, Olney et al , 1986, 1988 and 1989, Takebe and Basmajian, 1976, Van Griethuysen et al , 1982) Values for peak knee flexion during swing in chronic stroke patients vary from 34 (±19) to 65° [Detrembleur et al , 2005, Olney et al , 1991, Richards et al , 2003], which is below normal values (65 7±5 2°) as described by Benedetti et al (1998)

A lack of knee extension prior to heel strike has commonly been reported for chronic stroke patients in the literature (Finely and Karpovich, 1964, Knutsson and Richards, 1979, Lehmann et al , 1987, Olney et al , 1988 and 1989, Van Griethuysen et al , 1982) Detrembleur et al (2005) reported that in a 'slow walking group' of patients with chronic stroke had a knee flexion of 14 (±12)° at heel strike which is above the values (0 4±4 9°) described by Benedetti et al (1998) for normal subjects

**Ankle dorsiflexion and plantarflexion angular kinematics**

Only three studies have examined the joint angular kinematics of the ankle in stroke patients with equinus deformity of the foot Gok et al (2003) reported that stroke patients with equinus deformity of the foot at heel strike, placed their ankle in -16 (±11)° of plantarflexion which is well below normal values (4±6°) as described by Benedetti et al (1998) Goks' study does not state whether their stroke patients gait was assessed while they wore shoes Burdett et al (1988) also reported a similar angle of -13

(±9)° of plantarflexion at heelstrike for their stroke group wearing shoes Lack of dorsiflexion at heel strike has been reported by many authors for patients with chronic hemiplegia (Berger et al , 1984, Burdett et al , 1988, Colaso and Joshi, 1971, Finley and Karpovich, 1964, Giuliani, 1990, Knutsson and Richards, 1979, Olney et al , 1986, 1988 and 1989, Wolf and Minkiwitz, 1989, Lamontagne et al , 2002) Lehmann et al (1987) and Burdett et al (1988) reported the amount of ankle dorsiflexion during midstance in patients with equinus deformity of the foot Burdett et al (1988) reported that the angle of the ankle during midstance was -2 (±11)° of plantarflexion and Lehmann et al (1987) reported that the angle of the ankle during midstance was 12° of dorsiflexion, which was significantly less than their age and speed matched group (15°) Lehmann et al (1987) and Burdett et al (1988) also reported the position of the ankle at toe off Burdett et al (1988) reported the angle to be -5 (±11)° and Lehmann et al (1987) reported the ankle angle to be 11° which was significantly different to their age and speed matched group (16°) Gok et al (2003), Burdett et al (1988) and Lehmann et al (1987) reported on the dorsiflexion angle during swing of stroke patients with equinus deformity of the foot Gok et al (2003) reported the maximum dorsiflexion angle during swing to be -12 (±13), Burdett et al (1988) reported the maximum dorsiflexion angle to be -6 (±6)° and Lehmann et al (1987) reported the maximum dorsiflexion ankle angle to be 3° which was significantly different from the age and speed matched group (9°) As can be seen from the results there is quite a large amount of patient variability in ankle dorsiflexion during swing, however all these maximum dorsiflexion angles during swing are significantly lower than normal values of 9° (Lehmann et al , 1987) These findings are similar to those reported for chronic stroke patients (Bogardh and Richards, 1981, Knutsson and Richards, 1979, Olney et al , 1988 and 1989) Kim and Eng (2004) reported on the total excursion of the ankle of chronic stroke patients They reported a greater decrease in ankle excursion in the affected leg (18 ±5°) in comparison to the unaffected leg (25 ±8°)

## Coronal Plane

### Pelvic obliquity joint kinematics angle

To date no study has examined pelvic obliquity in stroke patients with equinus deformity of the foot Clinical observations suggest that decreased pelvic obliquity is a relatively common kinematic deviation amongst people with stroke Tyson et al (1999) examined pelvic obliquity in chronic stroke patients These patients demonstrated large amplitudes of pelvic displacement and a reduced displacement of the pelvis towards the affected side compared to normal values

### Hip adduction and abduction joint kinematic angles

Lehmann et al (1987) appears to be the sole study which reported hip adduction joint angles in stroke patients with equinus deformity of the foot They reported a significant decrease in the peak hip adduction angle of the affected leg (6°) in comparison to normal healthy age and speed matched subjects (9°) Lehmann et al (1987) suggest that this significant decrease in peak hip adduction angle was probably associated with a decreased pelvic obliquity Decreased pelvic obliquity may also be associated with decreased ipsilateral hip extension during the stance phase (Lehmann et al , 1987) and a decrease in contralateral step length (Lehmann et al , 1987, Tyson 1998, Burdett et al , 1988) or a shortened duration of contralateral swing phase (Lehmann et al , 1987, Dieli et al , 1997) No study to date has assessed pelvic obliquity and its relationship to the kinematics of the hip or temporal distance parameters

**Kinetics**

There has been limited research examining the kinetics of patients with equinus deformity Kinetics however has been studied in chronic stroke patients It is possible that a percentage of this group, which are classified as chronic stroke patients will have equinus deformity of the foot Due to lack of research in kinetics in patients with equinus deformity, this section will also discuss the kinetics reported for patients with chronic stroke in an attempt to give the reader a clearer picture of the possible kinetic gait changes, which may be present in these patients

**Hip sagittal moments**

No study to date has assessed hip sagittal moment in stroke patients with equinus deformity of the foot Only one study to date has assessed this variable in chronic stroke patients (Kim and Eng, 2004) Kim and Eng (2004) reported that the hip extensor moment was less on the affected leg (0 3 ±0 2 Nm kg$^{1}$) compared to the unaffected side (0 6 ±0 3 Nm kg$^{1}$) and they reported a similar finding for the hip flexor moment which was less on the affected leg (0 3±0 2 Nm kg$^{-1}$) in comparison to the unaffected leg (0 4 ±0 2 Nm kg$^{1}$) The differences between the limbs were not tested for significance

**Knee sagittal moments**

Two studies (Lehmann et al , 1987 and Gok et al , 2003) examined the moments in the knee joint during the stance phase of gait in stroke patients with equinus deformity of the foot Lehmann et al (1987) reported no significant differences between the mean total knee extensor and flexor moment during midstance of stroke patients in comparison to normal subjects walking at comparable speeds, however in 6 out of the 7 patients the mean total knee extensor moment value was larger Gok et al (2003) reported that the knee flexor moment was 0 4 (±0 3) N m, which was similar to the reported results of Lehmann et al (1987) and Kim and Eng (2004) Lehmann et al (1987) also reported that the mean knee extensor and flexor moment due to a vertical force during midstance was significantly less in stroke patients with equinus deformity of the foot compared to normal age and speed matched subjects (Lehmann et al , 1987)

**Ankle sagittal moments**

No study has assessed the ankle moments in stroke patients with equinus deformity of the foot In chronic stroke patients the ankle plantarflexor moment on the affected leg has been reported to range from 0 6 (±0 2) Nm kg $^{-1}$ (Kim and Eng, 2004) to 0 6 ±0 4 Nm kg $^{-1}$ (Lamontagne et al , 2002) and for the unaffected leg from 0 5 ±0 2 Nm kg $^{-1}$ (Lamontagne et al , 2002) to 1 1 (±0 3) Nm kg $^{-1}$ (Kim and Eng, 2004) These values are lower than those reported for normal age matched subjects (1 6 Nm kg$^{-1}$) by Winter (1991) A study by Kim and Eng (2004) reported that the ankle joint sagittal kinetic patterns differed between the affected and unaffected legs in that there was an absence of a dorsiflexor moment at initial contact on the affected side The absence of the normal dorsiflexor moment acting eccentrically resulted in a reduction in the plantarflexion angle achieved after the loading response in 12 out of the 20 patients

**Hip power**

The hip power profiles during gait in a single patient with equinus deformity of the foot were noted by Voigt and Sinkjaer (2000) The HP1 burst was almost absent, the HP2 burst was extended over the first three quarters of the stance phase and HP3 burst covered the remainder of the stance phase The HP2 burst was 0 2 (±0 2)W kg $^{-1}$ and the HP3 burst ranged from 0 4 ±0 3 to 0 5 ±0 6 W kg $^{-1}$ on the affected side, and on the unaffected side the HP2 burst was 0 3 ±0 2 W kg$^{-1}$ and the HP3 burst ranged from 0 5 ±0 3 to 0 6 ±0 3 W kg $^{-1}$ These values are different in comparison to those reported by

Eng and Winter (1995) for normal subjects They reported a HP2 of 0 031 (±0 02)J kg$^{l}$ and a HP3 of 0 06 (0 02)J kg$^{1}$ Olney et al (1991) reported in their study of patients with chronic stroke that there was an increased HP3 burst at toe off They believe this might be due to diminished strength of the ankle plantarflexors

**Knee power**

The knee power profiles during gait in a single patient with equinus deformity were reported by Voigt and Sinkjaer (2000) The KP1 burst at the beginning of the stance phase was missing and the KP2 burst was very small (0 2 (±0 1) W kg$^{1}$) The KP3 bursts were clear (0 3 ±0 4 W kg$^{1}$) and only the contralateral leg displayed a KP4 burst (0 7 ±0 5 W kg$^{1}$) These values are significantly reduced in comparison to those reported by Eng and Winter (1995) for normal subjects who reported a KP2 of 0 31 (±0 3)J kg$^{1}$, a KP3 of -0 13 (±0 06)J kg$^{1}$ and a KP4 of -0 14 (±0 3)J kg$^{1}$ In the unaffected limb the KP2 burst had a slightly higher range (0 2 ±0 1 to 0 3 ±0 4 W kg$^{-1}$) which is lower than that of normal healthy subjects (Eng and Winter, 1995)

**Ankle Power**

The ankle power profiles during gait in a single patient with equinus deformity were reported by Voigt and Sinkjaer (2000) At both the affected and unaffected ankle joints the patient had clear AP1 and AP2 bursts In chronic stroke patients, Kim and Eng (2004) reported that the AP1 burst was 0 2 (±0 2)W kg$^{-1}$ on the affected leg and 0 4 (±0 3)W kg$^{1}$ on the unaffected leg and the AP2 burst was 0 4 (±0 4)W kg$^{1}$ on the affected leg and 1 6 (±1 2)W kg$^{1}$ on the unaffected leg These values are different than that for normal healthy subjects, who have an AP1 of -0 096 (±0 04)J kg$^{-1}$ and a AP2 of 0 39 (±0 082)J kg$^{1}$ (Eng and Winter, 2004) They also reported that power variables were strongly related to speed with the strongest being for sagittal hip power acting concentrically on the affected side (r=0 9) and the sagittal ankle power on the unaffected side (r=0 7)

In summary, the review of literature clearly indicates that there is a lack of research examining the effect of equinus deformity of the foot in stroke patients on joint kinematic and kinetics during gait Listed below are the general noted gait deviations in stroke patients with equinus deformity of the foot,

**Temporal distance variables**

- Decreased walking speed
- Increased cadence
- Decreased step and stride length
- Increased stride time
- Increased cycle duration
- Increased stance percentage of the gait cycle
- Decreased swing percentage of the gait cycle
- Decreased single limb support duration
- Decreased double support time

**Joint Kinematics**

- Lack of hip extension during late midstance and pushoff
- Decreased hip adduction
- Increased knee extension during midstance
- Increased knee flexion during midstance
- Decreased knee flexion during swing
- Decreased plantarflexion of the ankle during toe-off
- Increased plantarflexion of the ankle during midstance, at heel strike and during swing

**Joint kinetics**

- Increased mean knee extension vertical force
- Almost absent HP1, extended HP2 and HP3
- Absent KP1 and small KP2

## 2 5 Ankle-foot Orthoses

There are many treatments for equinus deformity of the foot These include physiotherapy, which consists of stretching and strengthening techniques and biofeedback (Basmajian et al, 1975), surgery (Waters et al, 1982), electrical stimulation (Granat et al, 1996, Taylor et al, 1999) and the use of ankle foot orthoses (Hesse et al, 1999, Chen et al, 1999) The following sections discuss ankle foot orthoses and their effects on the gait of stroke patients with equinus deformity of the foot

**2 5 1 Types of AFOs**

There are several types of AFOs They can be made up of either metal or plastic material, but the use of metal ankle foot orthoses for stroke patients is rare at present (Ofir and Sell, 1980) Plastic AFOs can be divided into two types based on the degree of activation temperature, high versus low temperature Most high temperature plastic AFOs are applied posterior to the leg and fabricated by a certified orthotist by means of lamination or vacuum-forming techniques over a modified plaster mold of the leg and foot rather than directly from the body part (Sarno and Lehneis, 1971) Posterior leaf plastic AFOs appear to be the most popular form (Ofir and Sell, 1980, Lehmann et al, 1983, Wong 1992, Dieli et al 1997), although, recent literature has started to examine the effectiveness of anterior leaf plastic AFOs (Wong et al, 1992, Chen et al, 1999, Danielsson and Sunnerhagen, 2004) To date no study has compared directly the differences between the effects of the anterior and posterior leaf AFOs on the gait of stroke patients

**2 5 2 Prescription of AFOs**

Ankle foot orthoses (AFOs) are amongst the most commonly prescribed categories of lower limb orthoses in clinical practice (Condie and Meadows, 1993) Approximately 22% of stroke patients are discharged from hospital with AFOs (Teasell et al, 2001, DeVries's et al, 1991) AFOs have been reported to be prescribed to stroke patients who demonstrated limited motor recovery following their stroke, paresis of the involved leg, poor balance or/and a decreased ambulatory function (Teasell et al , 2001)

There are three basic considerations in prescribing an AFO Firstly the AFO may not be necessary for the patient to walk However, the patient may walk safer with the AFO due to the fact that the AFO will provide mediolateral stability during stance thus preventing an ankle sprain or fracture (Lehmann, 1979) Secondly, since many patients with stroke have weakness around the ankle joint, the AFO should provide adequate foot clearance during the swing phase of gait so that the patient does not drag their foot and stumble or fall (Lehmann, 1979) Since many of the patients who have weakness around the ankle also have weakness around the knee the effect of the AFO on knee stability is also important (Lehmann, 1979) Thirdly, wearing of an AFO should aim to approximate the normal gait pattern as nearly as possible and thus reduce the energy expenditure (Lehmann, 1979) Prior to prescription of an AFO various factors should be considered Factors include the posture and the balance of the patient, muscle spasticity, muscle strength, the range of movement of the affected limb and the gait pattern present (Nagaya, 1997) If the plantarflexor muscles are weak then the AFO needs to facilitate plantarflexion but to limit the amount of dorsiflexion occurring at the ankle, thus facilitating toe off and preventing excessive dorsiflexion of the foot during the stance phase of the gait cycle If the dorsiflexor muscles are weak the AFO needs to be inflexible into plantarflexion but flexible into dorsiflexion Thus the AFO will improve the mode of initial contact allowing a heel strike to occur, prevent excessive inversion of the ankle and maintain the ankle in a dorsiflexed position during the swing phase of the gait cycle (Perry and Montgomery, 1987, Lehmann et al , 1983) Orthosis resistance to plantarflexion during the swing phase of gait should not be greater than necessary because it creates a flexor moment at the knee during heelstrike, which has to be controlled by voluntary action of the knee extensors (Miyazaki et al , 1997)

Plantarflexion resistance requirements will vary from minimal in a flaccid paralysis to maximal in a case of severe spasticity of the gastrocnemius and soleus Therefore it is desirable to have an orthoses designed specifically for each patient so that its biomechanical functioning matches the patients' requirements (Condie and Meadows, 1993)

Even though AFOs are frequently prescribed and many authors have reported on their possible benefits, many of the leading names in stroke rehabilitation research feel that sufficient evidence has not been provided to support their prescription in patients with stroke (Lehmann, 1979, Lehmann et al , 1983, Davies, 1985, Perry and Montgomery, 1987, Bobath, 1990, Condie and Meadows, 1993, DeVries's et al , 1991, Carr and Shepherd, 1998, Teasell et al , 2001) Carr and Shepherd (1998) believe that AFOs should be avoided in stroke patients, as they hold the foot in a fixed position throughout the gait cycle, thus preventing the restoration of a normal gait pattern Davies (1985) also supports this view, suggesting that these devices should not be used unless the ankle is unstable, no dorsiflexion present and that all attempts of training have failed Bobath (1990) suggests that many people do not need or benefit from AFOs, that they make the hip, knee and ankle less stable, and increase ankle clonus Hesse et al (1999) also reported that therapists are often hesitant to prescribe an AFO as they fear that the use of an orthosis might result in untimely over activity of the plantar flexor and non-use of the tibialis anterior muscle resulting in its atrophy

**2 5 3 Design characteristics of AFOs**

The design of AFOs must be adapted to the needs of each particular patient (Miyazaki et al , 1997) However, research describing how different designs of AFO function and how best to match these designs to the gait pattern present in the stroke patients is clearly lacking

Factors involved in the design of AFOs include its flexibility into dorsiflexion, plantarflexion, inversion and eversion, the initial setting of the dorsiflexor plantarflexor angle and the length of the sole (Miyazaki et al , 1993, Yamamoto et al , 1993[a]) The length of the sole is determined by the degree of toe clawing The flexibility of the AFO is probably one of the most important characteristics of the design of an AFO The flexibility of the AFO must match the degree of spasticity and muscle weakness of each patient The flexibility of the AFOs in dorsiflexion and plantarflexion are empirically decided A study by Yamamoto et al (1999) on the flexibility characteristics of various types of posterior leaf AFOs has demonstrated that the dorsiflexion and plantarflexor flexibility varies markedly between different forms of posterior leaf AFOs and that the inversion and eversion flexibility is similar across various types of posterior leaf AFOs Dorsiflexion and plantarflexion flexibility can be altered by selecting different positioning of the plastic of the AFO around the malleolli, by altering the width of the ankle joint area, by altering the thickness of the plastic and the height of the lateral wall (Yamamoto et al , 1993[b], Nagaya, 1997)

Miyazaki et al (1993) developed a measuring system to assess the effects of the flexibility of an AFO on the ankle joint and to calculate the moment generated by the AFO due to its deformation Yamamoto et al (1993[b]) then examined the effects of the flexibility of a posterior leaf AFO with the limb in situ on the moments generated at the ankle joint They measured the dorsiflexor moment generated by a spring AFO during the gait of 15 patients with stroke, who had no spasticity present in their plantarflexors Their results showed that the dorsiflexor moment required by the AFO to prevent equinus positioning of the foot during the swing phase was relatively small, however, the dorsiflexor moment required to prevent equinus positioning of the foot at heel strike was larger, although this difference was not statistically tested It was found that the required dorsiflexion moment needed for correct positioning of the foot for each patient differed (Yamamoto et al , 1993[b]) These results would suggest that the amount of dorsiflexor moment generated by an orthosis at the time of initial heel strike is the most important factor in selecting the function of an AFO suited to the individual patient Miyazaki et al (1997) studied the effect of the flexibility and the initial angle of the AFO on the moments generated at the ankle during gait in 20 stroke patients They reported that in stroke patients the dorsiflexor moment (AM1) following heel strike was negligible and that the AFO played an important role m replicating the moment normally provided by the eccentric contraction of the dorsiflexor muscles following heel strike In mid to late stance, the plantarflexor moment generated by AFO is very small compared with the plantarflexor moment (AM2) acting concentrically normally generated by the plantarflexors muscles They therefore suggest that the AFO plays only a small role in assisting the plantarflexor muscles However, in 11 out of 20 patients the flexibility and the initial angle of the AFO significantly effected the ankle plantarflexor moment generated Miyazaki et al (1997) suggested that the AFOs play only a limited role in assisting plantar flexors muscles during mid to late stance and a greater role in assisting the dorsiflexor muscles following heel strike Yamamoto et al (1997) suggested that the plantarflexor moment generated by the AFO in many cases may be unnecessary and have a negative effect by reducing the amount of dorsiflexion available during midstance thus decreasing the ability of the center of gravity to move forward

In summary the literature suggests that AFOs are frequently prescribed especially for stroke patients who have mediolateral instability at the ankle, decreased ankle joint motion, decreased strength of muscles around the ankle and spasticity present in the muscles around the ankle joint The views on the perceived benefits of AFO are conflicting and would suggest that further research is necessary to determine the effects of AFO on many aspects of gait There are many designs of AFOs available and each design potentially has a different biomechanical effect on gait The research would suggest that the biomechanical effects of an AFO should be matched to the individual requirements of each stroke patient

## 2 6 Effects of AFOs on hemiplegic gait A review

This section examines the effects of wearing an AFO on adult stroke patients with equinus deformity of the foot, using a number of biomechanical outcomes as the criteria measures Studies were reviewed provided

1 They included adult stroke patients

2 The effects of wearing an AFO were compared with that of not wearing an AFO during level ground walking or on a treadmill

3 That the biomechanical outcome criteria measures used within the studies included some of the following walking speed, kinetics (joint moments and foot loading patterns), kinematics (temporal distance data and joint angles), EMG patterns of lower limb muscles and measures of spasticity, energy cost of gait, balance and functional mobility

4 The trial designs consisted of randomised and non-randomised within subject comparisons or case studies

5 The language of the reported trials was in English

6 The results of the studies were statistically analysed Unfortunately the studies on the effects of wearing of an AFO on hemiplegic gait did not provide sufficient information to allow a meta-analysis to be performed

Only twenty-two papers fulfilled the criteria for this review These papers are listed in Table 2 4 chronologically The papers were then reviewed under the following headings kinematics (temporal distance data and joint angles), kinetics (joint moments

**Table 2.4**: Summary of papers reviewed which examined the effect of AFOs on hemiplegic gait

| Author | No. of patients | Age (yrs) | Time post stroke (mths) | Study design | Inclusion Criteria | Type of AFO | Biomechanical criteria of AFO | Length of time of wear of AFO before testing | Outcome measure |
|---|---|---|---|---|---|---|---|---|---|
| Corcoran et al. (1970) | 15 | 45.1 | 40.3 | Non randomised within subject comparisons* | Yes | Plastic & Metal AFO (CM) | Plastic & Metal AFO plantarflexion stop at 80° | Used test braces 1 week prior to testing | Energy expenditure<br>Walking speed |
| Lehmann et al. (1987) | 7 | 59-75 | 36-156 | Non randomised within subject comparisons* | Yes | Plastic AFO | Plantarflexor stop at 95° or Dorsiflexor stop at 85° | NG | Kinematics<br>Walking speed |
| Burdett et al. (1988) | 19 | 61.9 | 3.7 | Non randomised within subject comparisons* | Yes | Plastic & Metal AFO (CM) | Metal AFO had a plantarflexor stop at 90° dorsiflexor stop between 80-85°<br>Plastic AFO had a plantarflexor stop between 85-90° | 11 patients previously wore an AFO, average length of time NG | Kinematics<br>Walking speed |
| Mojica et al. (1988) | 8 | 46-66 | 5.0 | Non randomised within subject comparisons* | No | Plastic AFO (CM) | Not described | Mean 7.5 weeks of wear | Kinematics<br>Balance<br>Walking speed |
| Diamond & Ottenbacher (1990) | 1 | 32.0 | 9.0 | Case study | Yes | Tone inhibiting dynamic AFO (CM)<br>Prefabricated AFO | Not described | Did previously wear an AFO for 3 months 6 months before testing | Kinematics<br>Walking speed |
| Ohsawa et al. (1992) | 3 | NG | NG | Non randomised within subject comparisons* | Yes | Yunoko orthosis<br>Hemispiral orthosis<br>Shoehorn orthosis<br>FAFO II in 5 degrees of DF (CM) | Not described | NG | Kinematics<br>Walking speed<br>EMG |
| Mueller et al. (1992) | 1 | 55.0 | 24.0 | Case study | Yes | Dynamic ankle-foot orthosis | Not described | NG | Kinetics |
| Wong et al. (1992) | 6 | NG | 12.0 | Non randomised within subject comparisons* | No | Anterior and Posterior leaf AFO (CM) | Anterior leaf plantarflexor stop at 80-85°<br>Posterior leaf not stated | 3-7 days | Kinematics |
| Beckermann et al. (1996[a]) | 60 | 58.0 | 34.0 | Randomised Clinic trial | Yes | Polypropylene AFO (CM) | 85° plantarflexion stop | NG | Walking speed<br>Function |
| Beckermann et al. (1996[b]) | 60 | 58.0 | 34.0 | Randomised Clinic trial | Yes | Polypropylene AFO (CM) | 85° plantarflexion stop | NG | Walking speed<br>Spasticity<br>Function |

**Table 2.4** Summary of papers reviewed which examined the effect of AFOs on hemiplegic gait

| Author | No of patients | Age (yrs) | Time post stroke (mths) | Study design | Inclusion Criteria | Type of AFO | Biomechanical criteria of AFO | Length of time of wear of AFO before testing | Outcome measure |
|---|---|---|---|---|---|---|---|---|---|
| Hesse et al (1996) | 19 | 55 2 | 5 1 | Non randomised within subject comparisons* | Yes | Valens calliper | Plantarflexor stop at $90^0$ Dorsiflexor stop at $80^0$ Dorsiflexion was spring assisted | Less than 1 week of wear | Kinematic Walking speeds |
| Dicli et al (1997) | 4 | 25 62 | 8month –3 years | Case study | Yes | Hinged AFO (CM) | Metal ankle joint and an adjustable plantarflexion stop | Varied from non to 2 weeks | Walking speed Kinematics |
| Tyson et al (1998) | 4 | 43 5 | 16 5 | Non randomised within subject comparisons* | No | Hinged AFO (CM) | Adjustable plantar flexor stop Medio-lateral stability was present | 1 patient fitted day before testing and another patient had 2 weeks of wear | Walking speed Kinematics |
| Hesse et al (1999) | 21 | 58 2 | 4 9 | Non randomised within subject comparisons* | Yes | Valens calliper | Plantarflexor stop at $90^0$ Dorsiflexor stop at $80^0$ Dorsiflexion was spring assisted | Less than 1 week of wear of caliper | Kinematics Kinetics EMG Walking speed |
| Chen et al (1999) | 24 | 58 9 | 13 0 | Non randomised within subject comparisons* | Yes | Anterior AFO | Not described | NG | Balance |
| Tyson & Thornton (2001) | 25 | 49 9 | NG | Non randomised within subject comparisons* | Yes | Hinged AFO (CM) | Adjustable plantar flexor stop Medio-lateral stability was present | One month of wear | Walking speed Kinematics Function |
| Franceschini et al (2003) | 9 | 66 5 | 2-244 | Non randomised within subject comparisons* | Yes | Not stated | Not described | NG | Walking speed Kinematics Energy Expenditure |
| Gok et al (2003) | 12 | 54 | 1 5 | Non randomised within subject comparisons* | Yes | Seattle type AFO Metallic AFO | Dorsiflexion angle 90 degrees | Had an opportunity for walking practice | Kinematics Kinetics |
| Churchill et al (2003) | 5 | 25 60 | Not stated | Non randomised within subject comparisons* | No | Moulded plastic AFO (CM) | Not stated | Not stated | Walking speed Kinematics |

**Table 2 4 Summary of papers reviewed continued**

| Author | No of patients | Age (yrs) | Time post stroke (mths) | Study design | Inclusion Criteria | Type of AFO | Biomechanical criteria of AFO | Length of time of wear of AFO before testing | Outcome measure |
|---|---|---|---|---|---|---|---|---|---|
| De Wit et al (2004) | 20 | 61 2 | 25 6 | Non randomised within subject comparisons* | Yes | Plastic nonarticulated AFO | AFO with small posterior steel<br>AFO big posterior steel<br>AFO with 2 crossed posterior steel and open heel | 6 months | Walking speed<br>Functional tests |
| Danielsson and Sunnerhagen (2004) | 10 | 52 | 16 | Randomised within subject comparisons | Yes | Anterior Leaf AFO | Not described | 4-21 months | Energy Expenditure |
| Wang et al (2005) | 42 SDG<br>61 LDG | 59 9<br>62 3 | 3<br>34 | Randomised within subject comparisons* | Yes | Plastic AFO | Set in neutral position | Not stated | Walking speed<br>Balance |

NG- not given
CM – custom made
AFO – Ankle Foot Orthosis
SDG – Short Duration Group
LDG - Long Duration Group
* Each patient walked with and without and AFO

and foot loading patterns), EMG patterns of lower limb muscles, energy cost of gait, balance and functional mobility

### 2 6 1 Kinematics

#### 2 6 1 1 Temporal distance parameters

Sixteen studies have statistically examined the effects of AFOs on walking speed in stroke patients and they are summarised in Table 2 5 (Corcoran et al , 1970, Lehmann et al , 1987, Burdett et al , 1988, Mojica et al , 1988, Beckermann et al , 1996[a&b], Diamond and Ottenbacher, 1990, Hesse et al , 1996, Dieli et al , 1997, Tyson et al , 1998, Tyson and Thornton, 2001, Franceschini et al , 2003, Gok et al , 2003, Churchill et al , 2003, De Wit et al , 2004, Wang et al , 2005) Twelve of these studies reported that wearing an AFO significantly increased walking speed, ranging from 10 to 58% (Corcoran et al , 1970, Lehmann et al , 1987, Mojica et al , 1988, Diamond and Ottenbacher, 1990, Hesse et al , 1996, Dieli et al , 1997, Tyson et al , 1998, Tyson and Thornton, 2001, Franceschini et al , 2003, Gok et al , 2003, De Wit et al , 2004, Wang et al , 2005) In contrast, four studies reported that an AFO had no significant effect on walking speed ranging from a 1% decrease to a 13% increase (Burdett et al , 1988, Beckermann et al , 1996[a&b], Churchill et al , 2003) The largest increase in walking speed from wearing an AFO over walking with shoes was 58% (Diamond and Ottenbacher, 1990) [Table 2 5] Taking all sixteen studies into account, the average increase in walking speed across these studies when compared to walking with shoes was 21 (±14)% Methodological factors may explain, at least in part, why the findings from Burdett et al (1988), Beckermann et al (1996[a&b]) and Churchill et al (2003) differ from the vast majority of studies All the studies cited used different forms of AFOs whose biomechanical characteristics may not be beneficial for improving walking speed and different methods of assessing walking speed (Table 2 5) The randomised clinically controlled

**Table 2 5** Effect of AFO wear on walking speed

| Author | No of patients | Age (yrs) | Time post stroke (mths) | Type of AFO | Percentage change in walking speed | Significance level | Method of assessment | Instruction to patient |
|---|---|---|---|---|---|---|---|---|
| Corcoran et al (1970) | 15 | 45 1 | 40 3 | Plastic AFO | 13% increase | P<0 005 | Stopwatch | Walk at comfortable walking speed |
| | | | | Metal AFO | 12% increase | P<0 005 | | |
| Lehmann et al (1987) | 7 | 59 75 | 36-156 | Plastic AFO positioned in 5° PF | 1% decrease | Not significant | Motion analysis | Walk at comfortable walking speed |
| | | | | Plastic AFO positioned in 5°of DF | 8% increase | P<0 05 | Motion analysis | |
| Burdett et al (1988) | 11 | 64 6 | 4 2 | Plastic AFO | 15% increase | Not significant | Video tape | NG |
| Mojica et al (1988) | 8 | 46-66 | 5 0 | Metal AFO | 27% increase | P<0 01 | NG | Walk as fast as possible |
| Beckermann et al (1996[a]) | 60 | 58 0 | 34 0 | Polypropylene AFO (*CM*) | Not stated | Not significant | Infra red beams | Walk at their normal speed<br>Walk as fast as judged safe by themselves |
| Beckermann et al (1996[b]) | 60 | 58 0 | 34 0 | Polypropylene AFO (*CM*) | Not stated | Not significant | Infra red beams | NG |
| Diamond & Ottenbacher (1990) | 1 | 32 0 | 9 0 | Tone inhibiting dynamic AFO | 17% increase | p<0 05 | Stopwatch | NG |
| | | | | Prefabricated AFO | 58 % increase | p<0 05 | | |
| Hesse et al (1996) | 19 | 55 2 | 5 1 | Valens calliper | 66% increase over barefoot<br>28% over shoe wear | P<0 006 | Stopwatch | Walk at maximum speed |
| Dieli et al (1997) | 4 | 25-62 | 8-36 | Hinged AFO (*CM*) | Not stated | P<0 001 | Stopwatch | NG |
| Tyson et al (1998) | 4 | 43 5 | 16 5 | Hinged AFO | 39% increase | P<0 000 | Stopwatch | NG |
| Tyson & Thornton (2001) | 25 | 49 9 | NG | Hinged AFO | 39% increase | P<0 000 | Stopwatch | NG |

**Table 2 5** Effect of AFO wear on walking speed

| Author | No of patients | Age (yrs) | Time post stroke (mths) | Type of AFO | Percentage change in walking speed | Significance level | Method of assessment | Instruction to patient |
|---|---|---|---|---|---|---|---|---|
| Franceschini et al (2003) | 9 | 66 5 | 2-244 | Not stated | Not stated | p<0 01 | BTS telemetric system | NG |
| Gok et al (2003) | 12 | 54 0 | 1 5 | Seattle type AFO | 16% increase | P<0 05 | Motion analysis | Self-selected speed |
| | | | | Metallic AFO | 28% increase | Not significant | | |
| Churchill et al (2003) | 5 | 25 60 | Not stated | Moulded plastic AFO (CM) | 9% increase | Not significant | 2D motion analysis | NG |
| | | | | | 13% increase over barefoot | Not significant | | |
| De Wit et al (2004) | 20 | 61 2 | 25 6 | Plastic nonarticulated AFO | 10% increase | P<0 05 | Infrared beams | NG |
| Wang et al (2005) | 42 SDG | 59 9 | 3 | Plastic AFO | 19% increase | Not significant | Stopwatch | NG |
| | 61 LDG | 62 3 | 34 | | 16% increase | P<0 05 | | |

Average change 21 ±14% increase incomparison to shoe wear
Min percentage change 1% decrease incomparison to shoe wear
Max percentage change 58% increase incomparison to shoe wear

NG- not given
AFO – Ankle Foot Orthosis
CM – Custom Made
SDG – Short duration Group
LDG – Long duration Group

studies of Beckermann et al (1996[a&b]) measured walking speed over a much larger time frame where the initial and finial walking speed values were taken 12 weeks apart in contrast to the other studies which assessed the change in walking speed generally during one testing session Perhaps if a significant improvement occurs in walking speed with the wear of the AFO it is relatively short lived To date no study has examined if a 'carry-over effect' in walking speed occurs with wear of an AFO A 'carry-over effect' is where prolonged wear of the AFO results in an improvement in the speed of walking without its wear A significant 'carry-over effect' of 21% has been reported in the literature following prolonged wear a functional electrical stimulator as an ankle foot orthosis (Kinsella et al, 2000) It is therefore feasible that a carry-over effect might occur with continued wearing of an AFO The results of these studies in general indicate that the wearing of an AFO significantly increases walking speed Table 2 6 summarises the fourteen studies which have statistically examined the effects of wearing an AFO on other temporal distance parameters and there is a large variation in the reported results (Lehmann et al, 1987, Burdett et al, 1988, Mojica et al, 1988, Diamond and Ottenbacher, 1990, Mueller et al, 1992, Wong et al, 1992, Hesse et al, 1996, Dieli et al, 1997, Tyson et al, 1998, Hesse et al, 1999, Tyson and Thornton, 2001, Franceschini et al, 2003, Gok et al, 2003, Churchill et al, 2003) In the affected leg the wear of an AFO has been reported to significantly increase cadence on average by 0 2 (±17)% Cadence was reported to significantly increase in three out of eight studies (Mojica et al, 1988, Hesse et al, 1996, Tyson and Thornton, 2001) with the largest increase reported by Tyson and Thorton (2001) of 18% However, in direct contrast Diamond and Ottenbacher (1990) reported a significant decrease in the rate of cadence by 36% Step length on the affected leg with the wearing of an AFO was reported to significantly increase on average by 13 (±7)% in two out of five studies (Burdett et al, 1988, Diamond and Ottenbacher, 1990) with the largest increase being

**Table 2.6:** Percentage changes in kinematics with the application of an AFO

| Author | Type of AFO | Cadence (Step.min$^{-1}$) | Step symmetry | Step width (m) | Step length (m) | Stride length (m) | Swing Symmetry | Stance Symmetry | Symmetry | Stance Duration (sec) | Stance phase % of gait cycle | Swing Phase % of gait cycle | Single support % of gait cycle | Double support % of gait cycle | Midstance duration (s) | Toe Off duration (s) | Duration of heel strike (s) |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| Lehmann et al. (1987) | Plastic AFO positioned 5 degrees of DF | | | | | | | | | | | | | | 22%*↓ | 2.5%↑ | 270% *↑ |
| | Plastic AFO positioned 5 degrees of PF | | | | | | | | | | | | | | 26%*↓ | 42%*↑ | 114%*↑ |
| Burdett et al. (1988) | Plastic & Metal AFO | | | | 12.0% *↑ | 0.4% ↓ | | | | | | | | | | | |
| Mojica et al. (1988) | Plastic AFO | 12.0%*↑ | | | | 16.0% *↑ | | | | | | | | | | | |
| Diamond & Ottenbacher (1990) | Tone inhibiting dynamic AFO | 29.0%* ↓ | | | 24.0% *↑ | | | | | 5.0%*↑ | | | | | | | |
| | Prefabricated AFO | 36.0% *↓ | | | 14.0% *↑ | | | | | 5.0%*↑ | | | | | | | |
| Mueller et al. (1992) | Dynamic AFO | | | | | | | | | 17.0%*↓ | | | | | | | |
| Wong et al. (1992) | Anterior leaf AFO | | | 0.9%↓ | 15.0% ↑ | 0.9%↓ | | | | | 1.0%↑ | | 3.0%↓ | 14.0%↓ | | | |
| | Posterior leaf AFO | | | 8.0%↓ | 14.0% ↑ | 8.0%↓ | | | | | 7.0%↑ | | 25.0%↑ | 19.0%↑ | | | |
| Hesse et al. (1996) | Valens calliper from barefoot | 18.0%*↑ | | | | 38.0%*↑ | 5.0%↑ | 6.0%↓ | | | | | | | | | |
| | from shoes | 8.0% ↑ | | | | 19.0%*↑ | 3.0%↓ | 1.0%↓ | | | | | | | | | |
| Dieli et al. (1997) | Dynamic AFO | 2.0% ↑ | | | | | | | | | 2.0%↑ | 9.0↑% | | | | | |
| | Posterior leaf spring AFO | 0.4%↓ | | | | | | | | | 8.0%↑ | 2.0↑% | | | | | |

**Table 2 6** Percentage changes in kinematics with the application of an AFO continued

| Author | Type of AFO | Cadence (Step/min) | Step symmetry | Step width (m) | Step length (m) | Stride length (m) | Swing Symmetry | Stance Symmetry | Symmetry | Stance Duration (s) | Stance phase % of gait cycle | Swing Phase % of gait cycle | Single support % of gait cycle | Double support % of gait cycle | Midstance duration (s) | Toe Off duration (s) | Duration of heel strike (s) |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| Tyson et al (1998) | Hinged AFO (CM) | | | | 0 2% ↑ | 3 0%↑ | | | 37 0% ↓ | | | | 10 0% ↑ | | | | |
| Hesse et al (1999) | Valens caliper | 2 0%↑ | | 1 6%↑ | | 5 0%↑ | 19 0% *↑ | 2 0%↑ | | | 3 0%↓ | 7 0%*↑ | 2 0%↑ | | | | |
| Tyson & Thornton 2001 | Hinged AFO (CM) | 18 0% *↑ | 15%↑ | | 9 0%↑ | 12 0%*↑ | | | | | | | | | | | |
| Franceschini et al (2003) | Not stated | | | | | | | | | 19 0%*↓ | | | | | | | |
| Gok et al (2003) | Seattle-type AFO<br>Metallic AFO | 3 0%↑<br>5 0%↑ | | | | | | | | | | | | | | | |
| Churchill et al (2003) | Plastic AFO (CM) | 0% | | | | 8%*↑ | | | | 4%↑ | | | | | | | |
| **Average percentage change** | | **0 2%↑ ± 17 0** | **15 0%↑** | **2 4%↓ ± 5 0** | **12 6 ↑ ± 7 1** | **9 2% ↑ ± 13 0** | **7 0%↑ ± 11 1** | **1 6%↓ ± 4 0** | **37 0% ↓** | **4 4% ↓ ± 12 4** | **3 0%↑ ± 4 5** | **6 0%↑ ± 3 6** | **8 5%↑ ± 12 2** | **2 5%↑ ± 23 3** | **24 0↓ ±2 8%** | **22 3↑ ± 27 9%** | **192 0 ↑ ± 110%** |
| **Max change** | | **36 0% ↓** | | | **24 0% ↑** | **38 0%** | **19 0%** | **1 0%** | | **19%** | **8 0%** | **9 0%** | **25 0%** | **19 0%↑** | **26%↓** | **42%↑** | **270% ↑** |
| **Min change** | | **0%↓** | | | **0 2% ↑** | **0 4%** | **3 0%** | **6 0%** | | **4%** | **1 0%** | **2 0%** | **2 0%** | **14 0%↓** | **22%↓** | **2 5%↑** | **114%↑** |

* = Statistically significant ↑ = Increase ↓ = Decrease

24% Stride length was reported to be significantly increased with the wearing of an AFO on the affected limb in four out of eight studies on average by 9 (±13)% [Mojica et al , 1988, Hesse et al , 1996, Tyson and Thornton, 2001, Churchill et al , 2003] with the largest increase (38%) reported by Hesse et al (1996) The percentage of the gait cycle spent in swing was reported by Hesse et al (1996) to have increased significantly by 7%, however, it did not result in a significant change in walking speed Stance phase duration was also reported to significantly increase by 5% in a study by Diamond and Ottenbacher (1990) and to significantly decrease by 17% in a study by Mueller et al (1992) and by 19% in a study by Franceschini et al (2003) Again methodological differences are present between these studies which may account for the results, with the patients being different lengths of time post stroke [Franceschini et al (2003) 2 to 244 months, Mueller et al (1992) 24 months and Diamond and Ottenbacher (1990) 9 months] and the studies of Mueller et al (1992) and Diamond and Ottenbacher (1990) being single subject case studies Swing symmetry was also reported to significantly increase by 7 (±11)% in the affected limb with the wear of an AFO, with a maximum increase of 19% (Hesse et al , 1996) Lehmann et al (1987) reported a significant increase in the percentage of the gait cycle spent in heel strike, midstance and toe off with the wearing of an AFO positioned in both five degrees of plantarflexion and five degrees of dorsiflexion The following parameters were also reported to change but not significantly step symmetry increased, step width decreased, stance symmetry decreased, symmetry decreased, the percentage of the gait cycle spent in stance increased, the percentage of the gait cycle spent in single limb support increased and the percentage of the gait cycle spent in double limb support increased Clearly the gait of an individual is not only dependent upon the action of the affected limb but also the unaffected limb Unfortunately, the effects of wearing an AFO on the temporal distance parameters on the unaffected leg has only been assessed by only one study (Hesse et al , 1999). Hesse et al. (1999) reported that with the wearing of the AFO the length of time that the unaffected leg spent in swing significantly increased from 21 (±4)% to 27 (±4)%. This result would suggest that by wearing an AFO the stroke patients achieved better symmetry in their gait.

**2.6.1.2 Joint angular kinematics**

Only two studies to date have examined the effects of wearing an AFO on the sagittal joint angular kinematics (Burdett et al., 1988; Gok et al., 2003). Both Burdett et al. (1988) and Gok et al. (2003) reported that wearing an AFO significantly increased the angle of dorsiflexion at heel strike. Gok et al. (2003) reported that wearing of the AFO resulted in the ankle being placed in greater dorsiflexion from a position of -16 (±11)° without wearing the AFO to 6 (±6)° with the wearing of the plastic AFO, and -0 (±4)° with wearing of the metallic AFO. Burdett et al. (1988) reported that wearing of the AFO resulted in the ankle being placed in greater dorsiflexion from a position of -13 (±9)° without wearing the AFO to -5 (±5)° with the wearing of the AFO. They reported that wear of an AFO did not significantly alter the angle of the ankle during midstance, heel off and toe off. Gok et al. (2003) and Burdett et al. (1988) reported that wearing of the AFO resulted in the ankle been placed in significantly greater dorsiflexion during midswing. Gok et al. (2003) reported that wearing of the AFO altered the position of the ankle during midswing from -12 (±13)° to -1 (±6)° when a plastic AFO was worn and to 3 (±6)° when a metallic AFO was worn. Burdett et al. (1988) reported that wearing the AFO resulted in greater dorsiflexion during midswing changing from -6 (±6)° to 0 (±6)°. Both studies reported that wearing an AFO had no significant effect on sagittal plane hip and knee joint angular kinematics during the gait cycle. The results of Burdett et al. (1988) should perhaps be viewed with caution as the kinematic data was collected using a video camera and a goniometer was placed on the television screen to assess angle changes The authors do not report on the reliability of this method of analysis of joint angular kinematics The results of these studies demonstrate that wearing an AFO significantly affected the temporal distance parameters of gait There has been a lack of research examining the effects of wearing an AFO and different types of AFOs on joint angular kinematics Changes in the joint angular kinematics may be responsible for the changes noticed in the temporal parameters, however further studies are need to examine this Only one study to date has examined the effects of wearing an AFO on the unaffected limb Future studies need to assess if wearing an AFO has any detrimental affects on the unaffected limb A carry over effect from wearing of an AFO may also be possible as demonstrated by the case study of Mueller et al (1992) however, larger clinical trials will be necessary to establish this effect

**2 6 2 The effects of wearing an AFO on the kinetics of gait**

Four papers have statistically examined the effects of wearing an AFO on the kinetics of stroke patients with equinus deformity of the foot (Lehmann et al , 1987, Mueller et al , 1992, Hesse et al , 1999, Gok et al , 2003) Lehmann et al (1987) reported a significant increase in the knee flexor moment (KM2) with the wear of an AFO positioned in 5° of dorsiflexion during the stance phase of gait, from 5 to 12N m The results of Gok et al (2003) would contradict this finding as they reported a significant reduction in the knee flexor moment (KM2) with the wearing of a metallic AFO, from 0 4 (±0 3)N m to 0 2 (±0 2)N m The differences in the reported effects of AFOs on the knee flexor moments by Gok et al (2003) and Lehmann et al (1987) might possibly be accounted for by the different types of AFO worn In the study by Lehmann et al (1987) the knee flexor moment was decreased by positioning the AFO into 5° of plantarflexion, while the AFO used by Gok et al was positioned in neutral Gok et al (2003) also reported that no significant differences in mean hip flexor and extensor moments, knee extensor moment, knee valgus moments, ankle plantarflexor moment, total ankle power and first vertical force peak with the wearing of an AFO

Mueller et al (1992) in their case study examined the effects of a tone-inhibiting AFO, and a dynamic AFO (DAFO) on foot loading patterns Mueller et al found a significant increase in the maximum weight borne through the foot and the maximum area of the foots' plantar surface in contact with the ground during the single stance period of the gait cycle when the DAFO was worn The authors believe that these results indicate that use of a DAFO may result in greater stability in the foot during the stance phase of the gait cycle Mueller et al (1992) study also examined if any carry-over effect occurred from wearing an AFO for every 'waking hour' after 14 days They reported a significant change in total foot area from 190 [Standard error measurement (SEM) $\pm 3cm^2$] on day 1 in comparison to 183 (SEM $\pm 2$)$cm^2$ on day 14 They also reported a significant increase in the total force under the foot during stance from 1241 (SEM $\pm 33$)N on day 1 in comparison to 1345 (SEM $\pm 17$)N on Day 14 Hesse et al (1999) reported similar findings, with significantly improved loading of force under the affected leg with the wearing of an AFO from 1 60 ($\pm 0$ 05)N $s^{-1}$ to 1 62 ($\pm 0$ 4)N $s^{1}$ when an AFO was worn They also found a significant improved 'gait line', which is the force point of action under both feet given as a percentage of the anatomic foot length, on the affected and unaffected legs

The studies on the effects of wearing an AFO on the kinetics of gait are few The results of the studies would suggest that the wearing of an AFO can significantly effect the moments of the knee joint and the ability of the affected foot to accept weight bearing force The results of Mueller at al (1992) would also suggest that there may be carry-over effects from the wearing of an AFO on the kinetics of gait Additional studies are necessary to examine the effects of wearing an AFO on the kinetics of other lower limb joints, such as the ankle and hip and the effects of wearing of an AFO on the joint kinetics of the unaffected leg There is also a need for future studies to try and identify the possible kinetic causes of changes observed in the kinematics of gait

**2 6.3 Effect of wearing an AFO on balance**

Chen et al (1999), Mojica et al (1988) and Wang et al (2005) examined the effects of wearing of an AFO on static and dynamic postural stability in stroke patients This was measured by examining postural sway, via measurement of the displacement of the center of pressure and centre of gravity Mojica et al (1988) and Wang et al (2005) reported that the wearing of a posterior leaf AFO significantly decreased postural sway However, Chen et al (1999) examined the effects of wearing of an anterior leaf AFO on postural sway and found no significant change The difference in the reported results between the three studies for postural sway may be due to the longer length of time that Chen et al (1999) patients were post stroke (13 months) in comparison to the 3 months (Wang et al , 2005) and 5 months (Mojica et al , 1988) in the other two studies and/or due to the different type of AFO worn

Chen et al (1999) and Wang et al (2005) reported that the wearing of an AFO significantly improved the ability of the stroke patients to transfer their weight from the right to the left side and improved their ability to place a greater amount of weight through the affected leg

An improvement in postural sway and weight transference with the wearing of an AFO may be of great functional benefit to stroke patients resulting in an improved confidence in their ability to walk and potentially decreasing the incidence of falling in this patient group Further studies are necessary in this area to examine the effects of AFO on postural control, especially in dynamic conditions

**2 6 4 Effect of wearing an AFO on EMG patterns of lower limb muscles**

Hesse et al (1999) and Ohsawa et al (1992) reported on the effects of wearing AFOs on stroke patients with marked plantar flexor spasticity Both studies reported the presence of a different EMG pattern in the lower limb muscles when wearing an AFO was compared to barefoot However, only Hesse et al (1999) statistically analysed their results When an AFO was worn Hesse et al (1999) found that the Vastus Lateralis muscle activity of the affected leg increased significantly by 35 (±24)% However, the activity of the tibialis anterior muscle decreased significantly by 48 (±24)% Hesse et al (1999) suggested that wearing an AFO may cause atrophy of the tibialis anterior from disuse and result in dependence on the AFO Ohsawa et al (1992) examined the effect of wearing of an AFO called the FAFOII on the EMG pattern of the lower limb in one patient They reported a decreased activity of the hamstring and gastrocnemius muscles, which was in agreement with the study of Hesse et al (1999), and an increase in quadricep muscle activity The results of the study of Hesse et al (1999) would support the hypothesis of many therapists that wearing an AFO will diminish the activity of the ankle dorsiflexors during swing However, this possible negative effect may be counter balanced by the possible positive effects of the AFO in decreasing the spasticity of the calf and hamstring muscles and promoting increased activation of the quadriceps muscle Nevertheless, further randomised clinical trials are necessary to examine if the EMG pattern changes in the lower limb muscles are present when an AFO is worn and if these EMG changes result in alteration of the levels of lower limb spasticity, muscle stiffness, ankle clonus or tendon reflexes

### 2.6.5 Effect of wearing an AFO on the energy cost of gait

Gait is exemplified by a smooth advancement of the body with the least mechanical and physiological energy expenditure (Waters and Mulroy, 1999). The energy expenditure of walking for persons with stroke is dependent on the extent of neurological dysfunction and the level of spasticity present, but it has been found to be higher than for normal subjects (Zamparo et al., 1995). For example walking at $0.1 m.s^{-1}$ requires 29% more energy for patients with stroke in comparison to normal subjects (Zamparo et al., 1995). This higher energy expenditure results in greater fatigue in comparison to normal subjects performing the same tasks (Onley et al. 1986; Zampero et al. 1995). Energy expenditure is normally assessed by measuring the amount of litres of $O_2$ consumed multiplied by the kilo calories used per litre of $O_2$ (Wilmore and Costill, 2004). Energy expenditure can be converted into an energy cost of an activity by determining the average oxygen uptake per unit of time and then calculating the kilocalories of energy used per minute during the activity (Wilmore and Costill, 2004). In patients with stroke, energy expenditure has been normally assessed indirectly by examining the average oxygen uptake during gait (Corcoran et al., 1970; Franceschini et al., 2003; Danielsson and Sunnerhagen, 2004). This oxygen uptake can be converted into an energy cost by dividing the oxygen uptake by walking speed (Danielsson and Sunnerhagen, 2004). This form of assessment of energy expenditure involves the patients wearing a face mask to collect the air while they walk, generally in a confined area, for a period of time (Franceschini et al., 2003). Some patients with stroke find this method of assessment distressing and uncomfortable.

AFOs, by altering the kinematics and kinetics of gait, are believed to lower the energy cost of walking (Franceschini et al. 2003). The energy cost of gait is calculated by dividing the oxygen uptake during the walking trial by the walking speed of the patients. Only three studies to date have examined the effects of wearing an AFO on the energy cost of gait (Corcoran et al., 1970; Franceschini et al., 2003; Danielsson and Sunnerhagen, 2004). In the study of Corcoran et al. (1970) the oxygen uptake of 15 stroke patients was measured as they walked wearing a metal AFO, a plastic AFO and while they walked without an AFO. The speed at which they walked was kept constant across each condition and the patients were assessed at three different walking speeds: slow, comfortable and maximum. Oxygen uptake significantly decreased when the stroke patients wore either AFO (6 and 8%) but only at the slow walking speed. The slow walking speed was the patients' customary walking speed without the wearing of the AFO. Two other studies assessed energy cost of gait in stroke patients with equinus deformity of the foot. Franceschini et al. (2003) assessed the energy cost of gait in nine and Danielsson and Sunnerhagen (2004) in 10 stroke patients. However, in these two studies walking speed was not controlled and patients where assessed while walking over ground (Franceschini et al., 2003) and on a treadmill (Danielsson and Sunnerhagen, 2004). In these studies oxygen uptake did not significantly decrease with the wear of an AFO. Both studies however, reported a significant increase in walking speed and a significant decrease in energy cost of walking with the wear of an AFO by 36% (Franceschini et al., 2003) and 12% (Danielsson and Sunnerhagen, 2004). Since walking speed was not controlled it is important to consider what impact this variable may have had on the results. A study by Zamparo et al. (1995) examined the effects of walking speed on the energy cost of gait in stroke patients. Zamparos' results demonstrated that as walking speed increased the energy cost of gait decreased. In the studies of Franceschini et al. (2003) and Danielsson and Sunnerhagen (2004) wear of an AFO resulted in an increase in walking speed by 0.1 and $0.07 m.s^{-1}$. In the study of Zamparo et al. (1995) a similar increase in walking speed of $0.1 m.s^{-1}$, from a walking speed of 0.3 to $0.4\ m.s^{-1}$ would result in an automatic reduction in the energy cost of gait by 19%. This would suggest that the lack of control of walking speed may have had a significant effect of the results of the previous studies.

The results of these few studies are contradictory. Further studies are necessary to examine the effects of wearing an AFO on oxygen uptake and energy cost removing the variable of walking speed. Since many patients with stroke find the assessment of oxygen uptake uncomfortable and sometimes distressing, it would be beneficial to develop a more user friendly method of assessing energy expenditure in this patient group.

An alternative method of assessing the energy expenditure of walking with the wearing of an AFO may be to assess the displacement of the COM as it has been reported that the vertical displacement of the COM correlated strongly to oxygen uptake in normal subjects (Kerrigan et al., 1995). In normal walking the COM should move in a smooth sinusoidal curve in both medio-lateral and vertical directions (Saunders et al., 1953). The vertical displacement of the COM has been related to potential energy changes that occur in the body during walking (Willams, 1985; Tesio et al., 1985; Simon et al., 1978; Saunders et al., 1953). In normal walking the displacement of the COM is minimal thereby requiring little energy input. Many authors have assessed the variables of gait which serve to minimize and smooth the vertical and lateral displacement of the COM (Saunder et al., 1953; Croce et al., 2001; Kerrigan et al., 2000). It has been reported that the displacement of the COM is higher in patients with gait disorders compared to subjects with normal gait (Johnson, 1977; Simon et al., 1978; Kerrigan et al., 1995). The variables which have been suggested to influence the displacement of the COM are heel rise, pelvic obliquity, single stance knee flexion, ipsilateral and contralateral knee flexion and pelvic rotation (Saunder et al., 1953; Croce et al., 2001; Kerrigan et al.,

2000) With the wearing of an AFO in stroke patients with equinus deformity of the foot it may be possible that these variables may be improved, thus reducing the vertical and lateral displacement of the COM

Location of the COM can be estimated by many different methods, including the sacral marker method, which measures the displacement of a marker placed on the sacrum, and the segmental analysis method, which calculates the COM of segments in the body to provide an estimate of the vertical displacement of the overall COM (Saini et al , 1998) Both methods have been shown to result in significantly similar values in normal subjects and to correlate strongly (r=0 82) to one another (Saini et al , 1998) With the sacral marker method the COM location is fixed to a particular anatomical site This may introduce errors because during walking, the COM moves with respect to the subject's anatomy due to movement of the limbs (Simon et al , 1978) Saini et al (1998) reported that the segmental analysis method demonstrated a trend towards producing a lower value than the sacral marker displacement method, but were within the error of measurement

Kerrigan et al (1995) reported that the vertical displacement of the sacrum significantly correlated with oxygen uptake (r=0 9) and was sensitive enough to identify unilateral immobilization of the knee Kerrigan et al (1996) used sacral displacement to assess the effects of wearing an AFO on patients with various neurological disabilities They reported that wear of an AFO had no significant effect on the displacement of the sacrum (p=0 07) Examination of their data revels that all the patients with stroke in their trial obtained a decrease in sacral displacement with the wear of an AFO To date no study has assessed the displacement of the COM by sacral and segmental analyses methods in stroke patients with equinus deformity of the foot One would hypothesis that the displacement of the COM would be higher in this group compared to normal able-bodied subjects. If the displacement of the COM correlates with the oxygen uptake and can differentiate between the wearing of an AFO in stroke patients, it would prove to be a more user-friendly method of assessing energy expenditure in this subject group, as it is less invasive to the patients.

### 2.6.6 The effect of wearing an AFO on functional mobility

Stroke patients with equinus deformity of the foot tend to score poorly in general activities of daily living (Teasell et al. 2001). Five studies to date have assessed whether wearing of an AFO will improve the ability of stroke patients with equinus deformity of the foot to perform functional mobility tasks better (Hesse et al., 1996; Beckermann et al., 1996[a&b]; Tyson and Thornton, 2001; de Wit et al., 2004). Tyson and Thornton (2001) assessed the effects of wearing an AFO on functional mobility using a Functional Ambulation Categories Test (Holden et al., 1986), which assessed the amount of support and supervision needed during walking. They reported significantly improved walking ability with the wearing of an AFO, from a score of two when an AFO was not worn, which indicated that continuous support was need while walking, to a score of four, which meant that the stroke patients could walk independently. Hesse et al. (1996) used the trunk and leg section of the Rivermead Motor Assessment to assess mobility with and without the wearing of an AFO in 19 stroke patients. Patients were asked to stand on the affected leg and step with the unaffected leg on and off a block and to tap the ground lightly five times with the unaffected foot while standing on the affected leg. While barefoot four patients could perform one motor task and three patients both motor tasks. Wearing the AFO six patients could perform one task and four patients both tasks. Beckermann et al. (1996[a]) used the Sickness Impact Profile in particular the "ambulation" and "physical dimension" categories, to assess the effects of wearing an AFO on these parameters They again reported no significant alteration in the Sickness Impact Profile with the wearing of an AFO In another study Beckermann et al (1996[b]) used the Fugl-Meyer Assessment Scale to assess the change in motor function of the affected leg and the patients sitting and standing balance with the wearing of an AFO They reported no significant change in the Fugl – Meyer Assessment Scale with the wearing of the AFO De Wit et al (2004) assessed the effects of wearing an AFO on functional mobility using the timed up and go test and a stairs test (Richards et al , 1999) With the wearing of an AFO significantly less time was needed to perform the timed up and go and the stairs test

The reported findings on the effects of wearing an AFO on functional mobility have been conflicting Two studies have reported significant improvement in function and two reported no significant change with wearing of an AFO However, all of these studies have used different functional assessment tests and all have examined various different aspects of functional capacity

A meta-analysis was unable to be performed on the studies of the effects of wearing of AFOs on gait This was due to the fact that the majority of papers involved patients who were matched pairs and the information needed to perform a meta analysis on such subgroups was not available Therefore no pooling of the results from the papers published to date was feasible and consequently it was not possible to present an effect size on the variables studied It would also have been of interest to establish if there was a relationship between the types of AFO worn and the presented effects However, again not enough information was supplied within the papers to develop a subgroup analyses Future trial should report with greater detail the results obtained so that in the future a meta-analysis can be performed Future research may wish to examine what effects different types and leaf forms of AFOs may have on the gait of stroke patients with equinus deformity of the foot. This would aid in the development of better prescription protocols for AFOs. The effects of wearing an AFO on the unaffected limb are generally unknown. Future research may wish to establish that the effect is not detrimental. The majority of the research designs of the studies reviewed were not randomised clinical trials but were non-randomised within subject comparisons and case studies. Future studies would benefit from improved methodological quality and the use of measurement methods which have been reported by the authors to have established validity and reliability.

One of the main problems with research on AFOs to date is that stroke patients with equinus deformity of the foot have been treated as a homogeneous group. This may not be the case as the literature would suggest that stroke patients can be sub classified into different groups based on their temporal distance and kinematic parameters (Knutsson and Richards, 1979; Mulroy et al. 2003). This division of stroke patients based on gait characteristic into different groups would guide treatment planning for these patients. Future research should examine if subgroups of gait patterns are present within the stroke population with equinus deformity of the foot. If this form of analysis were possible it may aid in the development of a more appropriate treatment methods for these patients. Research has yet to establish what gait pattern in a stroke patient would warrant a prescription of an AFO and what form of AFO should be prescribed for particular types of patients.

**Conclusion**

Only twenty two papers out of a possible sixty papers reviewed on the effects of AFO on the gait of stroke patients fulfilled the inclusion criteria. The papers were reviewed under the headings of kinematics, kinetics, EMG patterns of lower limb muscles, energy cost of gait, balance and functional mobility Table 2 7 summaries the noted effects of wearing an AFO on stroke patients with equinus deformity of the foot Previous literature has identified that wearing of an AFO significantly increases walking speed and has a significant effect on many other temporal distance parameters of gait (Corcoran et al , 1970, Lehmann et al , 1987, Mojica et al , 1988, Diamond and Ottenbacher, 1990, Hesse et al , 1996, Dieli et al , 1997, Tyson et al , 1998, Tyson and Thornton, 2001, Franceschini et al , 2003, Gok et al , 2003, De Wit et al , 2004, Wang et al , 2005) The research on the effects of wearing an AFO on the kinetics of gait are limited but would suggest that wearing an AFO can significantly alter the knee flexor moment and the ability of the foot to accept force (Lehmann et al , 1987, Mueller et al , 1992, Hesse et al , 1999, Gok et al , 2003) When the posterior leaf AFO has been worn it has been demonstrated to significantly improve postural sway but in contrast the

**Table 2 7.** Summary of noted effects of wearing an AFO on stroke patients with equinus deformity of the foot on gait

## Temporal distance variables

- Increased walking speed
- Improved cadence, step and stride length, swing symmetry
- Improved percentage of gait cycle spent in swing, stance, heel strike, midstance and push off duration

## Joint Kinematics

- Increases angle of dorsiflexion at heel strike
- Increases midstance angle towards dorsiflexion
- Increases swing angle towards dorsiflexion
- No effect on hip or knee angles

## Joint Kinetics

- Affects knee moments causing both an increase and decrease in knee flexor moments
- Improves weight bearing through the affected limb

anterior leaf AFO has not (Chen et al., 1999; Mojica et al., 1988; Wang et al., 2005). Wearing an anterior leaf AFO has been reported to improve the ability of a stroke patient to transfer weight from the affected to the unaffected leg (Chen et al., 1999).

The EMG patterns of the lower limb muscles have been found to significantly alter when an AFO was worn (Hesse et al., 1999). Wearing an AFO resulted in a significant increase in vastus lateralis activity and a decrease in the activity of the hamstrings, gastrocnemius and tibialis anterior muscles. The studies of the effect of AFO on energy expenditure are inconclusive but would suggest that AFOs may have some beneficial effects on reducing the energy cost of gait (Corcoran et al., 1970; Franceschini et al., 2003; Danielsson and Sunnerhagen, 2004). The reported findings on the effects of wearing an AFO on functional mobility have been conflicting (Hesse et al., 1996; Beckermann et al., 1996[a&b]; Tyson and Thornton, 2001; de Wit et al., 2004). Two studies have reported significant improvement in function and two no significant change with wear.

In conclusion the research to date would suggest that the effects of wearing an AFO seem to be positive in nature. However, clearly there is a need for further clinical trials. There has been a lack of research examining the effects of wearing an AFO and different types and leaf forms of AFOs on joint angular kinematics and kinetics. Research has yet to establish that wearing of an AFO does not have a detrimental affect on the unaffected limb of the stroke patient. The effects of AFO on the energy cost of gait have been contradictory. Future research needs to establish the effects of wearing AFOs on the energy cost of gait removing the variable of walking speed. The criterion for prescription of AFOs needs to improve. Research needs to establish whether stroke patients with equinus deformity of the foot are a homogenous group, and if not attempt to establish a prescription criteria for AFOs for each subgroup

**2 7 Aims and hypotheses of study**

The aims of the study were

1 To describe the gait pattern of stroke patients with equinus deformity of the foot in comparison to normal subjects walking at matched speed

2 To determine if the gait pattern of stroke patients with equinus deformity of the foot is homogeneous

3 To examine if the gait pattern of the subgroups identified within the stroke patients with equinus deformity of the foot differs from normal subjects

4 To examine if the gait pattern of normal subjects is affected by walking speed

5 To assess if wearing an AFO returns the gait of the stroke patients with equinus deformity of the foot to a more normal pattern

6 To assess if wearing of an AFO has differing effects on the gait pattern of the subgroups of stroke patients with equinus deformity of the foot

7 To assess if different leaf forms of AFOs have different effects on the gait pattern of stroke patients with equinus deformity of the foot

8 To assess if the gait pattern while wearing an AFO is altered by walking speed in normal subjects

9 To investigate the effects of walking with an AFO on the energy expenditure of gait in stroke patients with equinus deformity of the foot

10 To investigate if the measurement of the COM assessed using whole body kinematics and sacral displacement, is a viable method of determining the energy expenditure of gait in stroke patients with equinus deformity of the foot

## Research Hypotheses

1 Stroke patients with equinus deformity of the foot will have a significantly different gait pattern than normal subjects at a matched speed

2 The gait patterns of stroke patients with equinus deformity of the foot are not homogenous

3 The gait pattern of each subgroup identified within the stroke patient group differs from normal subjects

4 The gait pattern of normal subjects is affected by walking speed

5 Wearing of an AFO has a significant effect on the gait pattern of stroke patients with equinus deformity of the foot

6 Wearing of an AFO has differing effects on the gait pattern of the subgroups of stroke patients with equinus deformity of the foot

7 Different leaf forms of AFOs have different effects on the gait pattern of stroke patients with equinus deformity of the foot

8 The gait pattern while wearing an AFO in normal subjects will be altered when walking speed is decreased

9 Wearing of an AFO will decrease the energy expenditure of gait in stroke patients with equinus deformity of the foot

10 Measurement of the COM assessed using whole body kinematics and sacral displacement, is a viable method to determine the energy expenditure of gait in stroke patients with equinus deformity of the foot

# Chapter 3: Methods

## 3 0 Introduction

This section describes the methods that were used to examine the aims and test the hypotheses of this study Three groups of subjects were assessed Group one consisted of thirty normal subjects (Ngroup) Group two consisted of ten normal subjects age matched to the stroke group (NAMgroup) Finally, group three consisted of twenty-three stroke patients (Sgroup) with equinus deformity of the foot (Figure 3 1) All groups were compared to one another in the following three chapters to assess if there were differences in the gait pattern between the three groups All subjects were firstly familiarised with the procedures and testing equipment used in these experiments prior to the commencement of the experiments For all groups the first experiment performed was the gait assessment All subjects were asked to perform the gait assessment under the conditions of 1) walking without an ankle foot orthosis (NAFO), 2) walking with a posterior leaf AFO (PAFO) and 3) walking with an anterior leaf AFO (AAFO) The order of the conditions for the gait assessment was randomised and all subjects worn trainers The NAMgroup were assessed walking at two different walking speeds during the gait assessment firstly at their normal walking speed (NAMgroup$_{ns}$) and secondly at the matched speed of the stroke group (NAMgroup$_{ms}$)

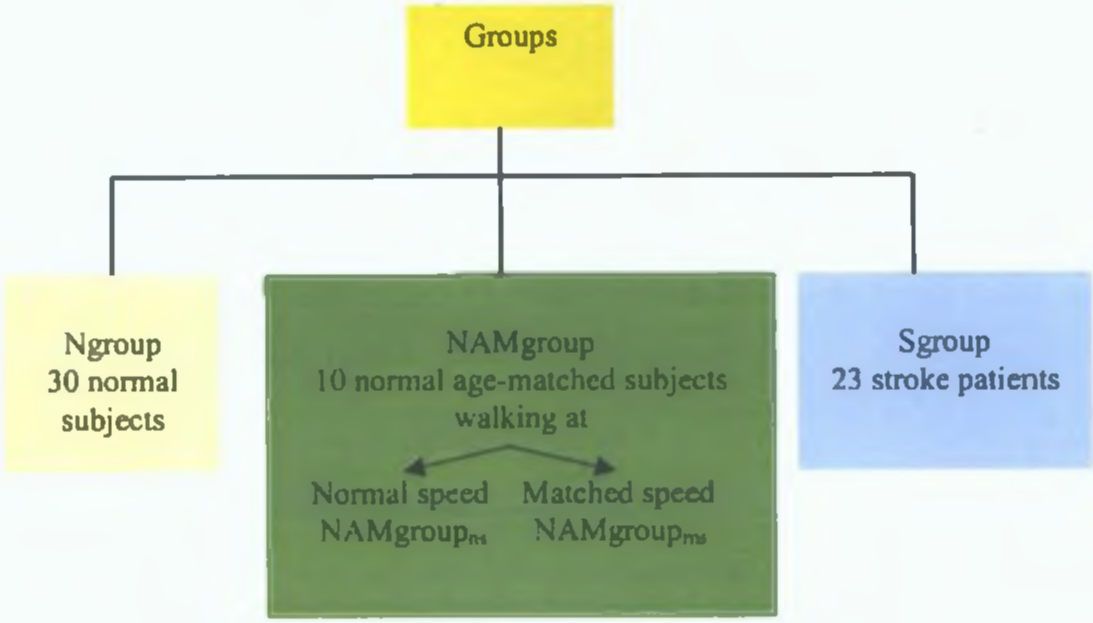



**Figure 3.1:** Subject experimental groups

Following the gait assessment, oxygen uptake during walking on a treadmill was assessed in the Ngroup and in six patients from the Sgroup (Figure 3.2). Oxygen uptake was measured while subjects were asked to walk under the conditions of NAFO, PAFO and AAFO (Figure 3.2). The order of the conditions was randomised.

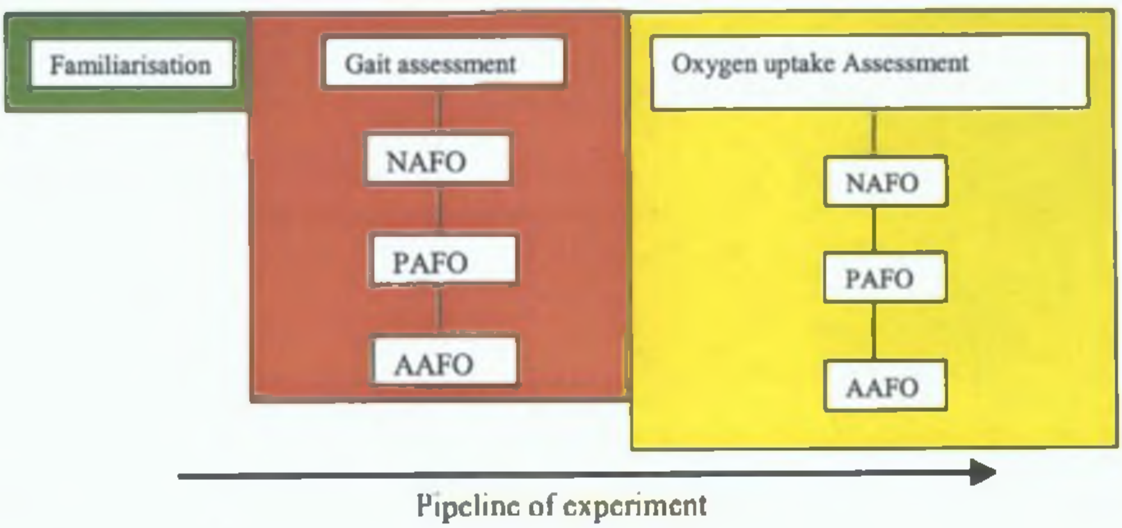



**Figure 3.2:** Experimental setup for trials

## 3 1 Delimitations

Six stroke patients who took part in this study had never worn an AFO before It may be possible that patients who were used to wearing an AFO reacted differently than those not used to wearing an AFO during testing, however, this variable was not assessed in this study

## 3 2 Subjects

Three groups of subjects where used in this trial, groups one and two consisted of normal subjects (Ngroup and NAMgroup) and group three consisted of stroke patients (Sgroup)

### 3 2 1 Stroke patients – Sgroup

Twenty-three patients (13 females and 10 males), diagnosed with hemiplegia (12 right and 11 left body side involvement) enrolled in this study (Table 3 1) All patients fulfilled the inclusion criteria (Table 3 2) The DCU ethics committee approved the experimental protocol and the patients gave their informed consent before participating in the experiments (Appendix A) The mean age of the group was 60 8 (±10 0) years, the mean height was 168 2 (±9 8) mm and the mean mass was 70 7 (±12 2) kg and mean time post stroke was 37 8 (±33 7) months (Table 3 3) All patients were community ambulators who used walking sticks and 17 currently wore an AFO on the involved side

### 3 2 2 Normal subjects – Ngroup and NAMgroup

Two separate groups of normal healthy subjects were enrolled in this study, the details of which can be found in Table 3 3 All subjects fulfilled the inclusion criteria (Table 3 4) The DCU ethics committee approved the experimental protocol and the subjects gave their informed consent before participating in the experiments (Appendix B)

**Table 3 1** Details of each stroke patient

| Patient No | Age of patient (yrs) | Gender | Time since stroke (months) | Side of stroke | Wear of orthosis |
|---|---|---|---|---|---|
| 1 | 65 | Female | 24 | Right | Yes PAFO |
| 2 | 44 | Female | 48 | Left | Yes PAFO CM |
| 3 | 64 | Male | 14 | Left | No |
| 4 | 64 | Female | 11 | Left | Yes PAFO CM |
| 5 | 59 | Male | 23 | Left | Yes Aircast |
| 6 | 44 | Female | 58 | Right | Yes PAFO CM |
| 7 | 49 | Female | 33 | Right | Yes Aircast |
| 8 | 57 | Female | 30 | Left | Yes PAFO CM |
| 9 | 53 | Male | 19 | Right | Yes Aircast |
| 10 | 65 | Male | 100 | Right | Yes PAFO CM |
| 11 | 59 | Male | 156 | Left | No |
| 12 | 52 | Male | 18 | Left | Yes PAFO |
| 13 | 43 | Female | 36 | Right | Yes PAFO CM |
| 14 | 70 | Female | 42 | Left | No |
| 15 | 68 | Male | 18 | Left | No |
| 16 | 80 | Female | 10 | Right | Yes PAFO |
| 17 | 72 | Female | 22 | Right | Yes AAFO |
| 18 | 73 | Male | 18 | Right | Yes PAFO |
| 19 | 73 | Female | 54 | Right | Yes Aircast |
| 20 | 55 | Male | 18 | Left | Yes PAFO CM |
| 21 | 65 | Female | 42 | Left | No |
| 22 | 62 | Female | 10 | Right | No |
| 23 | 63 | Male | 66 | Right | Yes PAFO CM |

PAFO = Posterior Leaf Orthosis
AAFO = Anterior Leaf Orthosis
CM = Custom Made

**Table 3 2** Inclusion criteria for stroke patients

**Inclusion Criteria**

1 Aged 18 years or older

2 Their stroke occurred more than 6 months prior to the study

3 Referred from a chartered physiotherapist with a diagnosis of equinus deformity of the foot

4 Were presently wearing an AFO or had a history of wearing an AFO for the treatment of equinus deformity of the foot or referred from a physiotherapist who believed that they would benefit from wearing an AFO

5 On static assessment they had an oxford scale rating of zero or one for the dorsiflexor muscles of the ankle joint

6 Passively could obtain 0° of ankle dorsiflexion

7 Currently in good health

8 Able to walk 10 metres independently

9 Have not had any lower limb joints replaced

10 Do not have severe arthritis in their ankles, knees or hips

11 Do not get excessively short of breath when they walk short distances

12 Do not have lower limb injuries at the time of testing which might effect their ability to walk

13 Do not suffer from Diabetes Mellitus

14 Do not suffer from Renal or Kidney dysfunction

15 Do not suffer from high blood pressure

16 Do not suffer from - metabolic disorders, acute myocardiac infarction, uncontrollable cardiac arrhythmias, active endocarditis, symptomatic severe aortic stenosis or acute pulmonary disease

**Table 3 3** Details of normal subject groups and stroke patient group

| | **Ngroup** | **NAMgroup** | **Sgroup** |
|---|---|---|---|
| **Number of subjects** | 30 Subjects | 10 Subjects | 23 Subjects |
| **Age of subjects (years)** | 41 4 (±16 9) | 51 0 (±6 0)* | 60 8 (±10 0) |
| **Mean height (cm)** | 170 0 (±6 2) | 172 4 (±10 8) | 168 2 (±9 8) |
| **Mean mass (kg)** | 69 9 (±11 7) | 74 7 (±10 1) | 70 7 (±12 2) |

* subjects were not exactly age matched with Sgroup

**Table 3 4** Inclusion criteria for normal subjects

**Inclusion Criteria**

1 Aged 18 years or older
2 Currently in good health
3 Able to walk 10 metres independently
4 Have not had any lower limb joints replaced
5 Do not have severe arthritis in their ankles, knees or hips
6 Do not get excessively short of breath when they walk short distances
7 Do not have lower limb injuries at the time of testing which might effect their ability to walk
8 Do not suffer from Diabetes Mellitus
9 Do not suffer from Renal or Kidney dysfunction
10 Do not suffer from high blood pressure
11 Do not suffer from - metabolic disorders, acute myocardiac infarction, uncontrollable cardiac arrhythmias, active endocarditis, symptomatic severe aortic stenosis or acute pulmonary disease

## 3 3 Measurement protocol

Each subject attended the laboratory once Prior to the commencement of testing subjects were familiarized with the experimental protocol and equipment All subjects wore shorts and the females also wore bras Anthropometric measures of height, mass, leg length, widths of the ankle, knees, wrist, elbow and thickness of the hand were taken from each individual according to the requirements for static and dynamic modelling

All subjects were instructed to walk through the laboratory at their self-selected walking speed, without paying attention to the force plates concealed in the floor The subjects started about four to five steps in front of the force plates and the starting point was adjusted so the affected foot contacted the force plate

Each subject had their gait analysed wearing both types of orthoses Three successful trials were collected for each of the following conditions 1) NAFO, 2) AAFO and 3) PAFO The order of each was randomised In the NAMgroup the above conditions were also measured when these subjects walked at the matched speed (0 3m s$^{1}$) of the stroke patients ($NAMgroup_{ms}$). In all groups the subjects determined the inter-trial intervals themselves to avoid fatigue.

### 3.4 Orthoses

Two types of AFOs were used in this study. The posterior leaf ankle foot orthosis (Figure 3.3) and the anterior leaf ankle foot orthosis (Figure 3.4).

**Figure 3.3:** Posterior Leaf Ankle Foot Orthosis

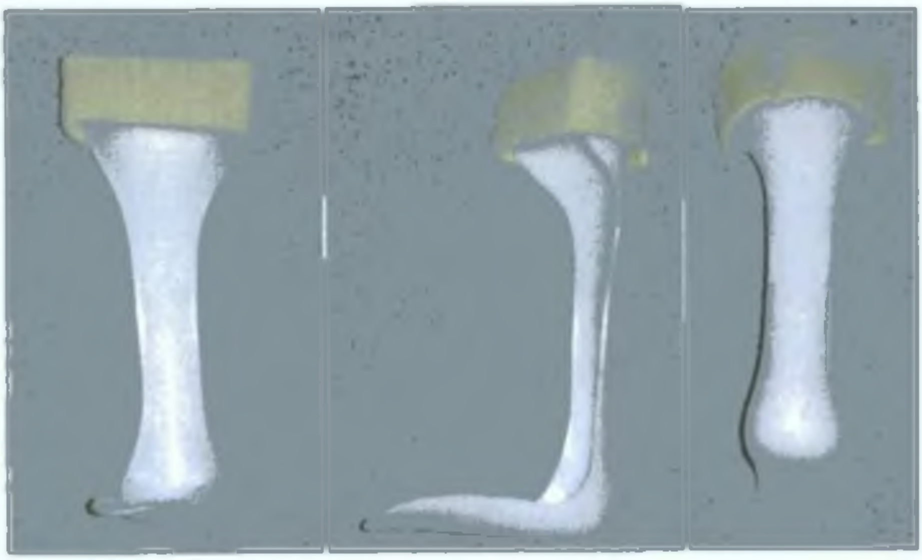
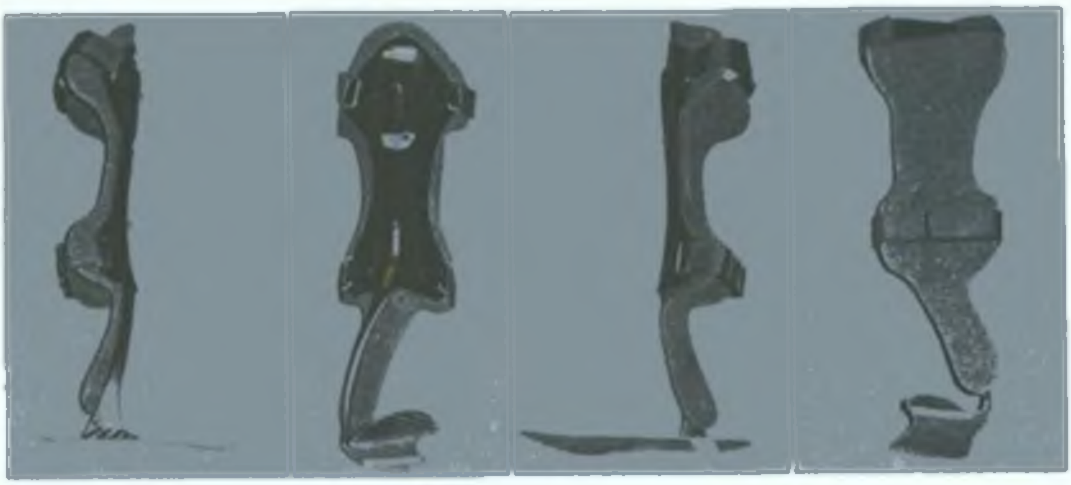


**Figure 3.4:** Anterior Leaf Ankle Foot Orthosis



Neither of these orthoses have any published peer review research articles detailing their clinical effects or biomechanical characteristics. The posterior leaf ankle foot orthosis (Roylan Inc.) extends to just below the knee and its flat footplate extends to the metatarsal joints. The manufacturers claim that the orthosis should block ankle plantarflexion but allow free dorsiflexion. The anterior leaf ankle foot orthosis was the Aesthetes Modular AFO (Tycon Inc.). The manufacturers claim that the orthosis allows a more controlled movement around the ankle area and exhibits less stress concentration on the skin due to the materials used in the device. It consists of a carbon fibre and glass fibre composite footplate and anterior strut. Attached to the anterior strut is an adjustable, trimmable, thermoplastic anterior shell in natural polypropylene. The shell is pre-fitted with trimmable padding and velcro straps. The footplate has a pre-fitted insole attached, which is trimmable. The manufacturers claim that the design and the materials used in this orthosis leads to improved product durability and wearer performance.

### 3.5 Kinematic and kinetic measurements

Thirty-four spherical passive reflective markers were affixed to the skin of subjects with double-sided tape at specific anatomical landmarks bilaterally consistent with Vicon's Plug–In-Gait model (Oxford Metrics, Oxford, England) [Figure 3.5 and Table 3.5]. A slightly modified version of the marker configuration described by Davis et al. (1991) was used. The markers formed a segmental rigid body model. Figure 3.6 details the joint angles formed. Each subject's movements were collected with an eight camera Vicon 370 system with two AMTI forecplate (AMTI OR6-5, Watertown, USA) embedded in the floor. Before each measurement session the system was calibrated both statically and dynamically. The residual values for calibration were set at 1.2mm.

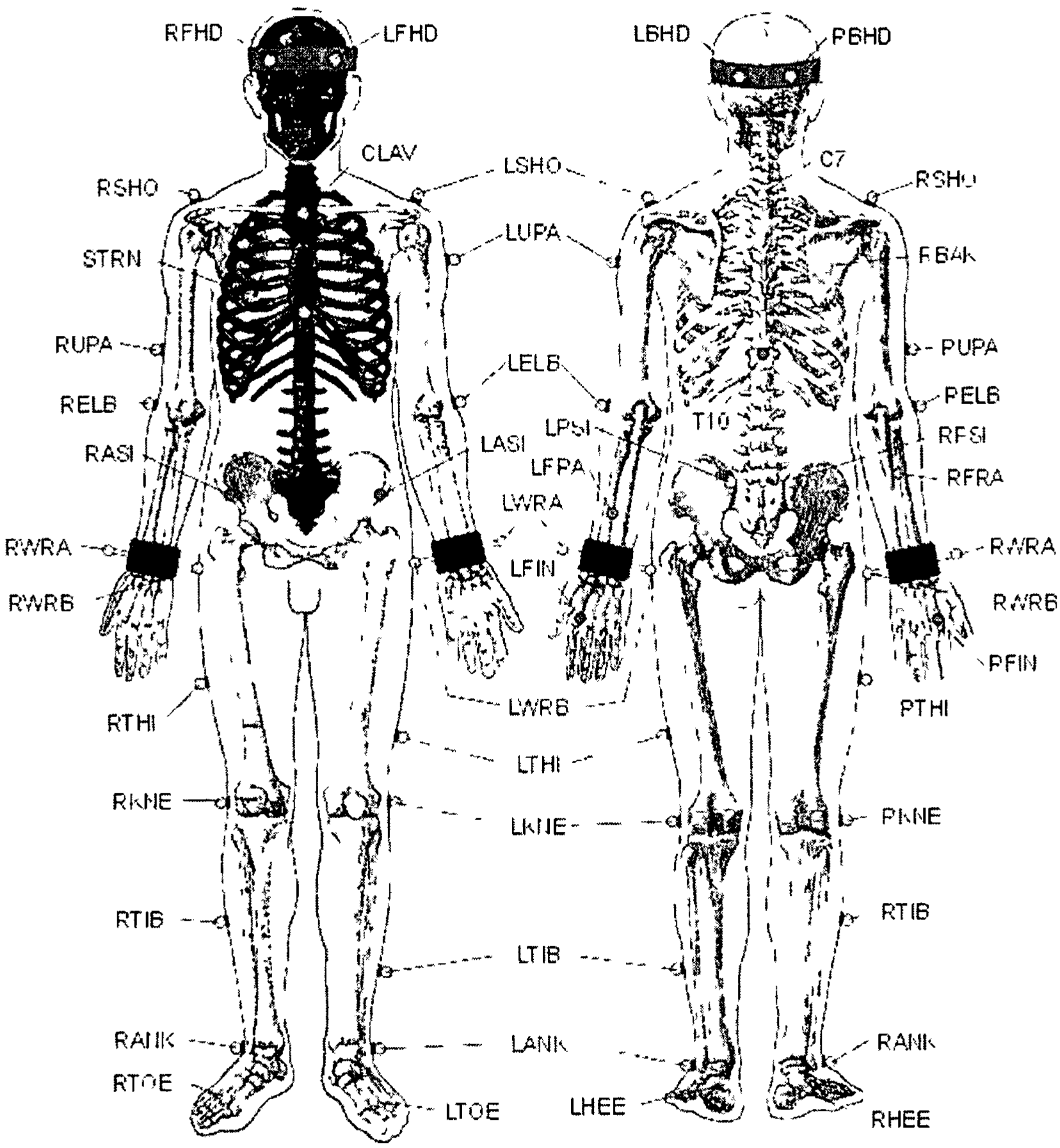



**Figure 3 5** Placement of markers on the body (Plug in Gait Marker Placement, Vicon Manual 2002)

| | |
|---|---|
| LFHD | Left front head |
| RFHD | Right front head |
| LBHD | Left back head |
| RBHD | Right back head |
| C7 | Seventh cervical vertebrae |
| T10 | Tenth thoracic vertebrae |
| CLAV | Clavicle |
| STRN | Sternum |
| LSHO | Left shoulder |
| LELB | Left elbow |
| LWRA | Left wrist bar thumb side |
| LWRB | Left wrist bar pinkie side |
| LFIN | Left finger |
| RSHO | Right shoulder |
| RELB | Right elbow |
| RWRA | Right wrist bar thumb side |
| RWRB | Right wrist bar pinkie side |
| RFIN | Right finger |
| RASI | Right anterior superior iliac spine |
| LASI | Left anterior superior iliac spine |
| RPSI | Right posterior superior iliac spine |
| LPSI | Left posterior superior iliac spine |
| LTHI | Left thigh |
| LKNEE | Left knee |
| LTIB | Left tibia |
| LANK | Left ankle |
| LHEE | Left heel |
| LTOE | Left toe |
| RTHI | Right thigh |
| RKNEE | Right knee |
| RTIB | Right Tibia |
| RANK | Right ankle |
| RHEE | Right heel |
| RTOE | Right toe |

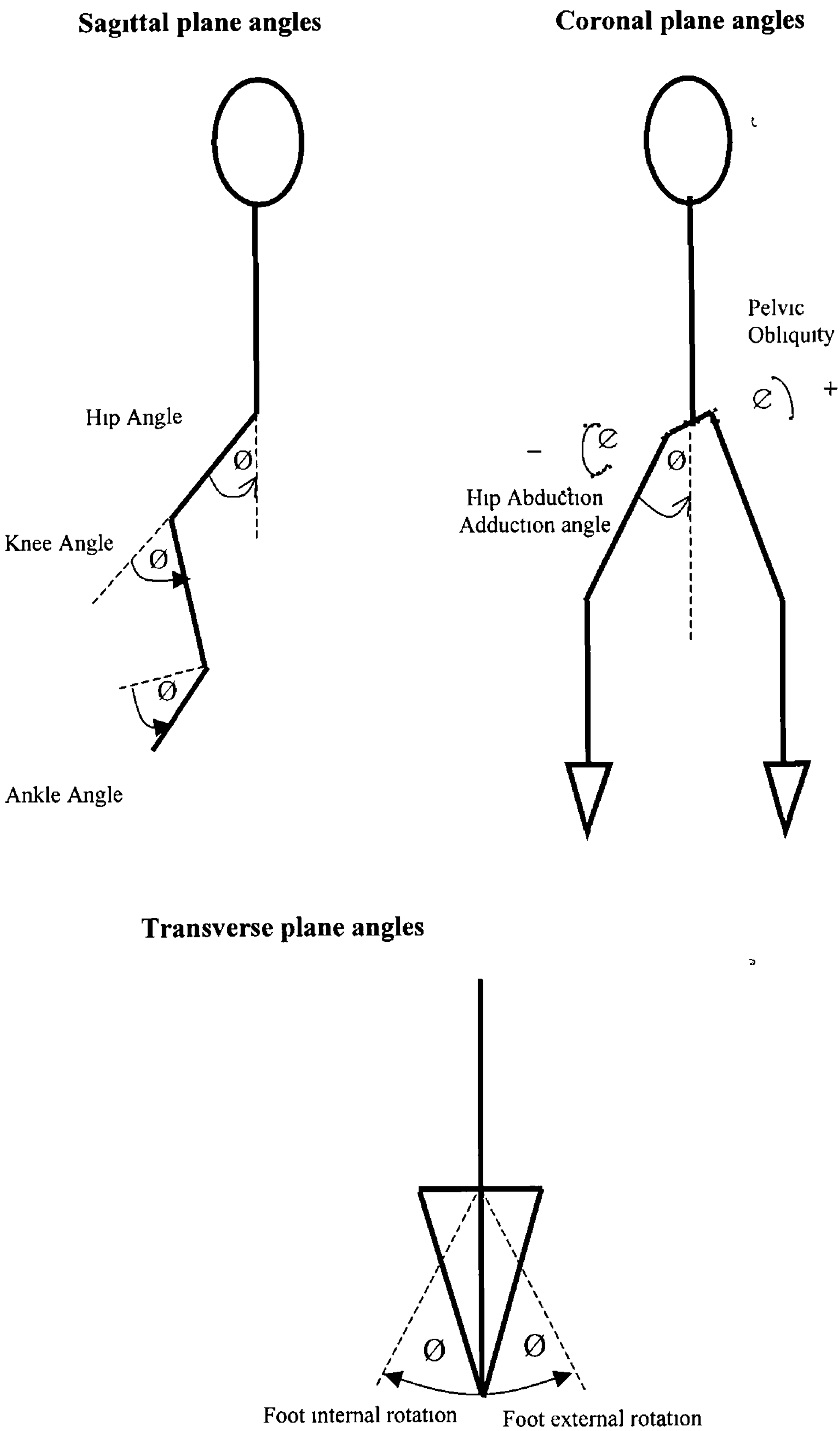



**Figure 3 6** Joint angles formed from the segmental rigid body model

**Table 3 5** Placement of markers on the body

| | | |
|---|---|---|
| Upper body | Head | Over the right and left temple, on the back of the head, left and right in line with the front markers, roughly on a horizontal plane |
| | Torso | Over the spinous process of the 7$^{th}$ cervical vertebrae, over the spinous process of the 10$^{th}$ thoracic vertebrae, over the jugular notch, over the xiphoid process |
| | Arm marker | On the acromio-clavicular joint, lateral epicondyle, wrist laterally and medially and dorsum of the hand just below the head of the second metacarpal |
| Lower body | Pelvis | On the anterior superior iliac spine, posterior superior iliac spine and sacral marker midway between the posterior superior iliac spines |
| | Leg markers | On the lateral epicondyle of the knee, on the lower lateral 1/3 surface of the thigh, lateral malleolus, on the lower lateral 1/3 of the leg |
| | Foot markers | Over the second metatarsal head and on the calcaneous |

Infrared video data was captured at 250 Hz and synchronized with the forceplate data (250 Hz) Forceplate data were taken under the affected foot in the stroke patients and the right foot in the normal subjects Testing continued until a minimum of three acceptable trials were computed for forceplate data Trials were accepted provided that a force signal was obtained for a full gait cycle and the subjects reported no gait difficulties during the trial

Raw marker trajectories were filtered using Vicon's Woltring quintic spline algorithm with a MSE value of 15 The markers were labelled, foot contact events were determined and 3D kinematics and kinetics were calculated for each trial using Plug-In-Gait Joint rotations were calculated as *YXZ* Cardan angles of the distal versus the proximal segment using a floating axis system, respectively, representing flexion (+)/extension (−), abduction (−)/adduction (+) and internal (+)/external (−) rotation The reported moments were the net joint moments (sum of all the internal moments in a particular plane in a particular joint) that balance inertia and the moments caused by external forces (e g gravity, ground reaction forces) The net joint moments were normalized for body weight Each subject's center of mass was calculated by segmental analysis within Vicon's Workstation software using body segment masses and moments of inertia from Dempster (1995)

**3 6 Oxygen Uptake**

Oxygen uptake was measured in the Ngroup and six patients (3 males and 3 females) from the Sgroup Patients in the latter group fulfilled the additional criteria of

- The ability to maintain a walking speed of 1km hr $^{1}$
- No aversion to wearing a nose clip or gas mask
- An ability to walk for 6 minutes without a break
- The capacity to safely walk on a treadmill

The mean age of the Sgroup who took part in this section of the study was 53 8 (±12 3) years and their mean time post stroke was 44 0 (±28 9) months All patients were community ambulators who used walking sticks and an AFO on the involved side Oxygen uptake ($VO_2$) was measured using the Sensormedics Vmax229 Metabolic system (Sensormedics Corp , USA) and calibrated for both gas and volume prior to the commencement of each test Baseline $VO_2$ was collected during 10 minutes of sitting in a relaxed state Subjects were then placed on the treadmill and walked at a speed of 3 5km hr $^{1}$ (Ngroup) or 1 0 km hr $^{1}$(Sgroup) These speeds were chosen as they are the customary walking speeds for these groups and optimum oxygen uptake during walking occurs at a customary walking speed (Zamparo et al , 1995) $VO_2$ was then measured while the subject walked on a treadmill for six minutes for each trial for each orthosis condition NAFO, PAFO and AAFO The order of testing was randomised The net ıncrease ın relatıve $VO_2$ was determıned for the last two mınutes of each condıtıon Kınematıc data were also collected durıng the last two mınutes of each trıal

### 3 7 Data processıng

The temporal dıstance data for each trıal and condıtıon were recorded for the rıght and left foot of the Sgroup and the rıght foot for the Ngroup and NAMgroup Sagıttal and coronal plane kınematıc and kınetıc parameters were recorded usıng the protocol of Benedettı et al (1998) These are detaıled ın figures 3 7-3 13

**Kınematıc and kınetıc parameters**

All temporal dıstance parameters, sagıttal, coronal and transverse angular kınematıcs and kınetıc data for each of the three trıals per condıtıon were recorded and values were averaged

**Oxygen uptake**

For oxygen uptake the relatıve ıncrease ın oxygen uptake was calculated for each trıal by subtractıng the restıng oxygen uptake from the absolute value measured (Baker et al , 2001) Thıs was converted ınto energy cost by dıvıdıng the oxygen uptake value by walkıng speed (Danıelsson and Sunnerhagen, 2004)

**Centre of Mass**

The sacral and segmental dısplacement of the center of mass (COM) over one gaıt cycle was calculated for each condıtıon

**Pelvis rotation**
HR2 Maximum negative rotation in the coronal plane
HR3 Maximum positive rotation in the coronal plane

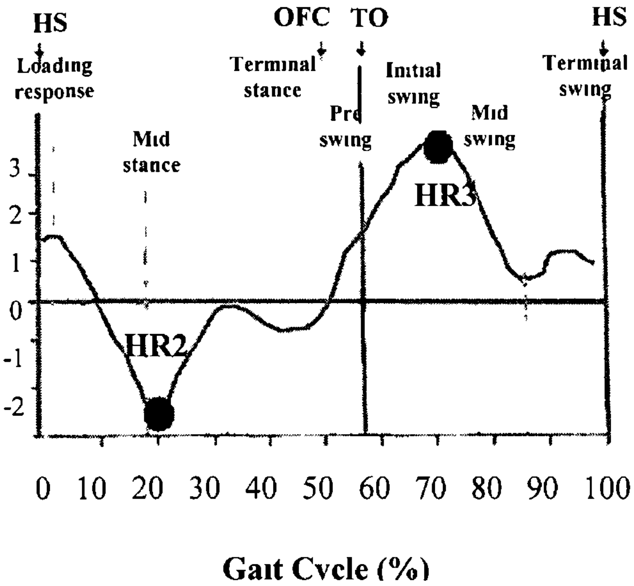



**Figure 3 7** Pelvis rotation angle (adapted from Benedetti et al 1998)

**Hip angle parameters**
H1 Flexion at heel strike
H2 Maximum flexion at loading response
H3 Maximum extension in stance phase
H4 Flexion at toe-off
H5 Maximum flexion in swing phase
H6 Total sagittal plane excursion

**Hip joint moments**
HM1 Maximum flexor moment
HM2 Maximum extensor moment

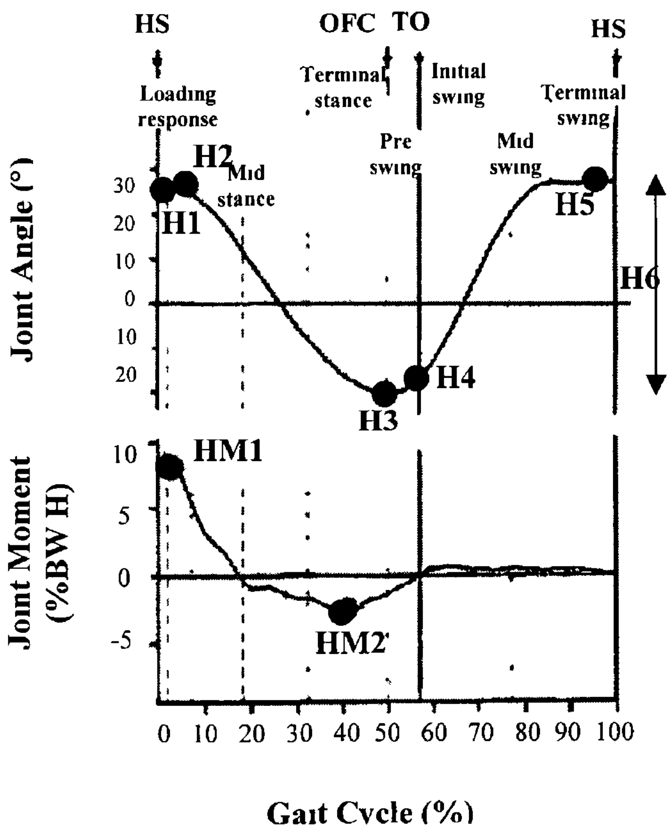



**Figure 3 8** Hip flexion/extension angles and hip moments (adapted from Benedetti et al 1998)

**Hip coronal plane angles**
H7 Total coronal plane excursion of the hip
H8 Maximum adduction in stance
H9 Maximum abduction in swing

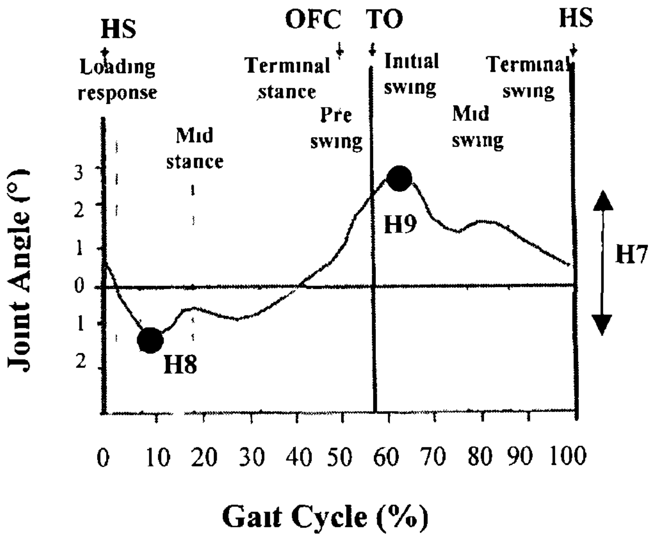



**Figure 3 9** Hip abduction/adduction angles (adapted from Benedetti et al 1998)

**Knee angle parameters**
K1 Flexion at heel strike
K2 Maximum flexion at loading response
K3 Maximum extension in stance phase
K4 Flexion at toe-off
K5 Maximum flexion in swing phase
K6 Total sagittal plane excursion

**Knee joint moments**
KM1 1st maximum extensor moment
KM2 Maximum flexor moment
KM3 2nd maximum extensor moment

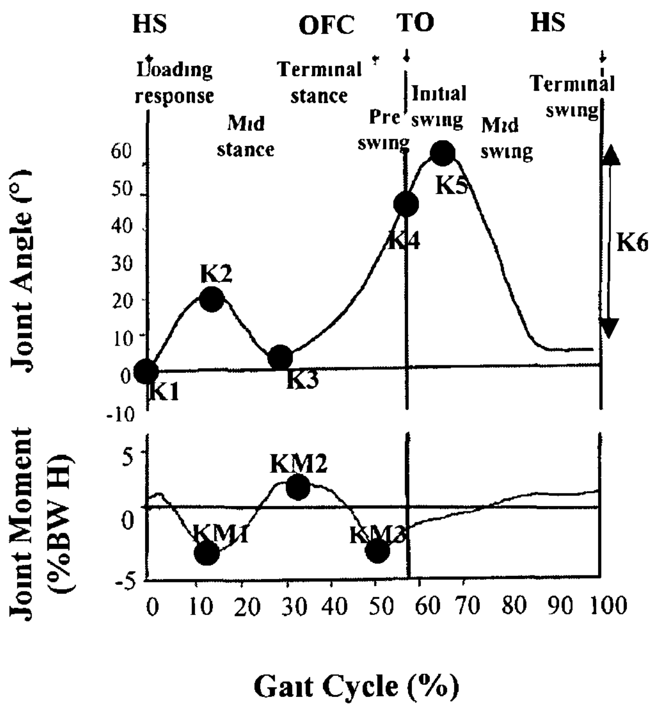



**Figure 3 10** Knee flexion/extension angles and moments (adapted from Whittle et al , 2000, Benedetti et al 1998 and Kadaba et al , 1989)

**Ankle angles parameters**

| | |
|---|---|
| A1 | Flexion at heel strike |
| A2 | Maximum plantarflexion flexion during loading response |
| A3 | Maximum dorsiflexion in stance phase |
| A4 | Flexion at toe-off |
| A5 | Maximum dorsiflexion in swing phase |
| A6 | Total sagittal plane excursion |
| A7 | Maximum plantarflexion in swing |

**Ankle joint moments**

| | |
|---|---|
| AM1 | Maximum plantarflexor moment |
| AM2 | Maximum dorsiflexor moment |

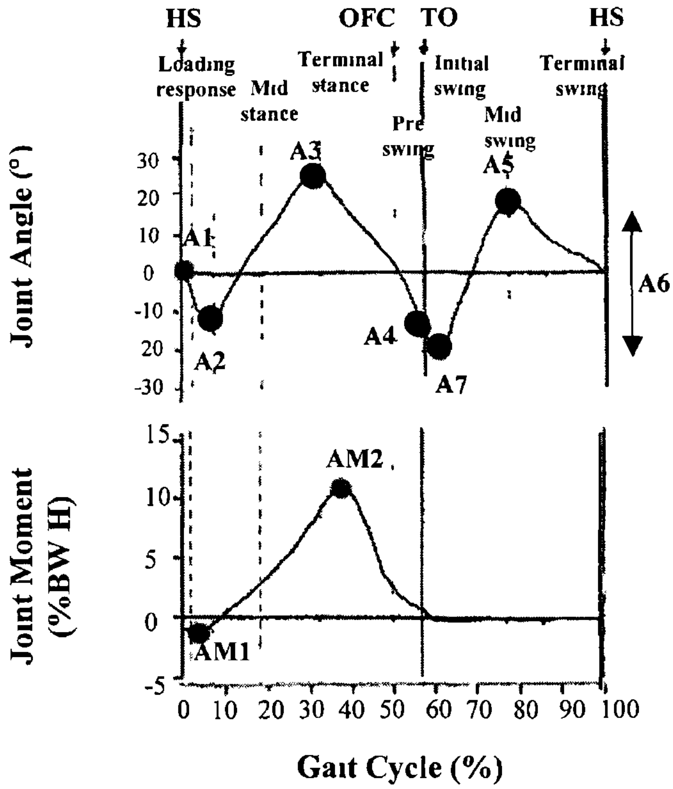
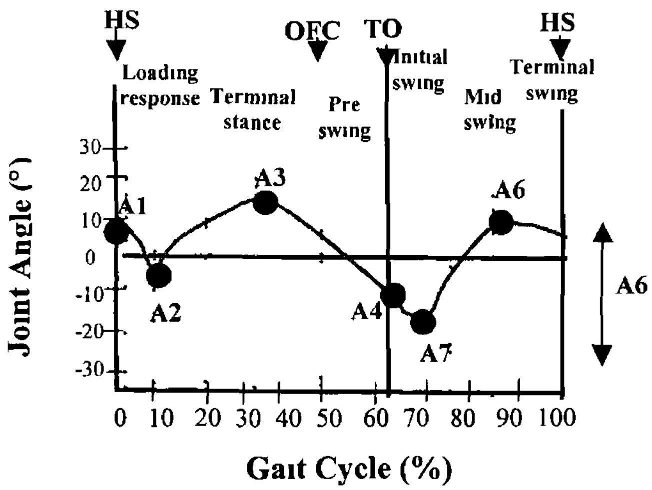



**Figure 3 11** Ankle plantarflexion/dorsiflexion angle and moments (adapted from Benedetti et al 1998)




**Ankle joint angular displacement curve in the sagittal plane**

This graph is included in this section since the ankle joint angular displacement curve of the NAMgroup$_{ns}$ in Chapter 4 is different to graph 3 11 This is believed to be due to the influence of footwear on the angular displacement curve in this joint

**Figure 3 12·** Ankle plantarflexion/dorsiflexion angle from the results of NAMgroup$_{ns}$ in Chapter 3

**Foot rotation angles**

- R1 Rotational position of the foot at heel strike
- R2 Maximum external rotation of the foot during stance
- R3 Maximum internal rotation of the foot during stance
- R4 Rotational position of the foot at foot off
- R5 Maximum external rotation of the foot during swing

**Foot rotation moment**

- AR1 Maximum external rotation moment of the foot
- AR2 Maximum internal rotation moment of the foot

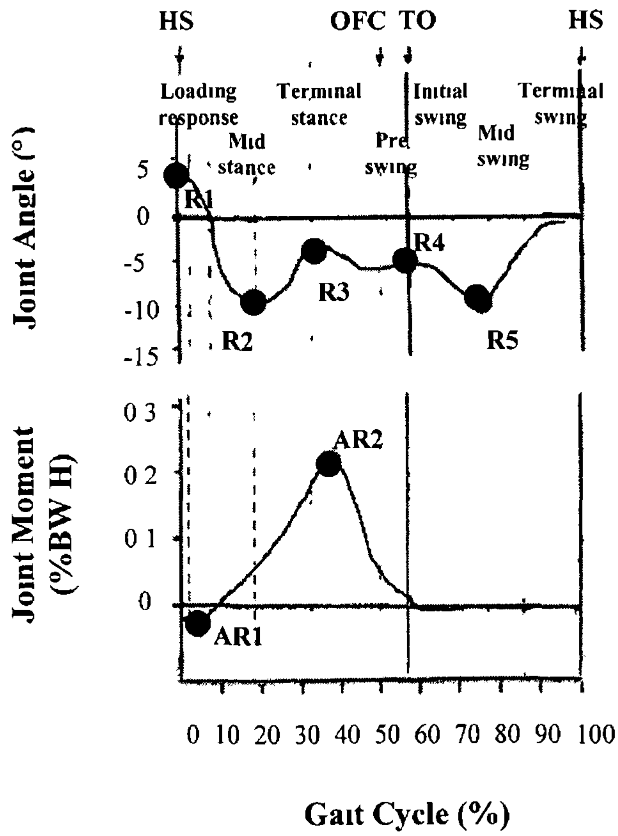



**Figure 3 13** Graph of foot rotation angles and moments

## 3 8 Statıstıcal Analyses

Statıstıcs package for socıal scıences versıon 11 (SPSS 11 0) was used for all statıstıcal analyses Means and standard devıatıons where calculated for all data and for each condıtıon All data were assessed for normalıty usıng the Kolmogorov-Smırnov normalıty test If data were not normally dıstrıbuted non-parametrıc tests were used The followıng sectıons descrıbe the statıstıcs used to address the aıms and objectıves of thıs study

**To examıne ıf the gaıt pattern of the NAMgroup was affected by walkıng speed and ıf thıs varıable affects the dıfferences noted between the Sgroup and the NAMgroup**

A repeated measures ANOVA was used to assess the effects of walkıng speed on the NAMgroup (two condıtıons normal and matched speed) and Sgroup Post hoc analysıs, to determıne the specıfic locatıon of dıfferences was completed usıng paır wıse adjusted comparısons usıng eıther a paıred or ındependent student t test were approprıate If the data were not normally dıstrıbuted, a Frıedman test was used ın place of the repeated measures ANOVA and post hoc analysıs were carrıed out usıng eıther the Wılcoxon or the Mann Whıtney U test where approprıate

**To examıne ıf the gaıt patterns of Sgroup was homogeneous**

A cluster analyses was used to assess the homogeneousness of the Sgroup It was hypothesısed that the Sgroup was not homogenous and ıf not, could be dıvıded ınto dıfferent subgroups based on gaıt characterıstıcs The abılıty to dıvıde the Sgroup ınto subgroups wıll aıd ın the understandıng of thıs deformıty and allow better dıagnoses based ınterventıon and custom fıttıng of AFO Cluster analysıs has been prevıously carrıed out successfully by Mulroy et al (2003) ın acute stroke patıents to ıdentıfy different subgroups based on gait characteristics Mulroy's study was able to identify four subgroups within the acute stroke patients

The cluster analysis was carried out on the gait data of the Sgroup when they were assessed during the NAFO condition Firstly, the mahanalobis distance criterion was used to eliminate outliers (two) from the sample Therefore the final sample included 21 patients The variables entered into the cluster variate were initially standardized and subsequently submitted to hierarchical cluster analysis The clustering methods applied were the Ward's linkage method and squared Euclidean distance measures (Hair et al , 1998) The decision on the number of clusters to be included in the solution was guided by both empirical as well as theoretical considerations With respect to the latter, the previous study that employed cluster analysis to examine gait patterns in stroke patients identified four distinct profiles (Mulroy et al , 2003) In terms of empirical consideration, Hair and colleagues (1998) suggest examining the agglomeration schedule for fairly substantial increases in the average within-cluster distance as a guideline for identifying the final cluster solution In the present study, there was a noticeable increase in the schedule coefficients when moving from a three to a four-cluster solution, suggesting a substantial decrease in homogeneity of each cluster within the three-cluster solution Based on these results, participants were grouped according to their assignment to one of the three identified clusters The final population distribution across the clusters yielded 9 patients in the first cluster, 7 patients in the second cluster and 5 patients in the third cluster Post-Hoc analyses employing a MANOVA were used to reveal significant between-cluster differences

**To assess if the gait pattern of the subgroups within Sgroup were different from the NAMgroup$_{ms}$**

If the data were normally distributed an independent student t-test was used to assess the differences in the gait pattern between the NAMgroup$_{ms}$ and each subgroup within the Sgroup If the data were not normally distributed a Mann Whitney U test was used

**To assess if the wearing of an AFO had an effect on the gait pattern of the Sgroup, Ngroup and the NAMgroup$_{ms}$ and to assess if different leaf forms of an AFO have differing effects on the gait pattern of the groups**

If the data were normally distributed, a repeated measure ANOVA ($p<0.05$) was used to calculate the effects of wearing an AFO and the effects of wearing different leaf forms of AFO's on the Ngroup, NAMgroup$_{ms}$ and Sgroup Post hoc analysis, to determine the specific location of differences, was completed using pair wise adjusted comparisons If the data were not normally distributed a Friedman test was used in place of the repeated measures ANOVA and post hoc analyses were carried out using the Wilcoxon test

**To assess if wearing an AFO and different leaf forms of AFO's caused different changes in the gait pattern within the subgroups of the Sgroup**

A repeated measures ANOVA ($p<0.05$), with each subgroup as a between subject factor, were used to assess if the wearing of an AFO affected the gait pattern of each subgroup differently Post hoc analyses, to determine the specific location of differences, were completed using pair wise adjusted comparisons

**To investigate the effects of walking with an AFO on the energy expenditure of gait in Sgroup and Ngroup.**

A repeated measures ANOVA ($p<0.05$) were used to assess the effects of wearing an AFO on oxygen uptake, energy cost of gait and the displacement of the COM. If the data were not normally distributed a Friedman test was used in place of the repeated measures ANOVA. Post hoc analyses to determine the specific location of differences were completed using pair wise adjusted comparisons. A Pearsons' correlation or a Spearman rank order correlation ($p<0.05$), if the data were not normally distributed, were used to asses the relationship between the different methods of assessment of the displacement of the COM, segmental analysis and sacral displacement methods and to assess the relationship between the displacement of the COM and oxygen uptake.

**Power Calculation for the variable of walking speed**

MINTAB statistical software version 13.1 (MINITAB 13.1) was used to calculate the sample size need for a statistical power of 0.8, for the variable of walking speed. An increase in walking speed in the order of 0.2 $m.s^{-1}$ was suggested by Perry et al. (1995) to be of clinical significance in stroke patients and was used as the effect size in the analysis. The estimate standard deviation for the variable of walking speed was set at $0.1 m.s^{-1}$ based on the standard deviation of walking speed of the study of Tyson and Thorton (2001). The results of the statistical test suggested that a sample size of six stroke patients with equinus deformity of the foot should provide enough power to detect a statistically significant difference in the variable walking speed.

Retrospectively, the statistical power for walking speed for stroke patients with equinus deformity of the foot walking with and without an AFO was calculated. The sample size was 23 patients and the increase of walking speed with the wearing of an AFO was

$0.07 m.s^{-1}$ and $0.09 m.s^{-1}$. The standard deviation of walking speed of the stroke patients was $0.25 m.s^{-1}$. The statistical power was low at 0.15 and 0.22.

# Chapter 4: The gait of stroke patients with equinus deformity of the foot

### 4.0 Introduction

This chapter focuses on comparing the differences in gait between the Sgroup, the $NAMgroup_{ms}$ and the $NAMgroup_{ns}$. Comparisons were made for the NAMgroup walking at both their normal ($1.3\pm0.2 m.s^{-1}$) and age matched speed ($0.4\pm0.1 m.s^{-1}$), where the matched speed was comparable with that of the average of the Sgroup ($0.3\pm0.3 m.s^{-1}$). This latter comparison was undertaken because previous research has demonstrated that certain kinematic and kinetic gait parameters in patients with stroke appeared to be due to the speed that patients walked at and not due solely to their condition (Lehmann et al., 1987; Chen et al., 2005). Examination of the gait differences between stroke patients and normal age matched subjects while walking at the same speeds may provide insights concerning the stroke impairment and related compensatory strategies that are in addition to the observation of slow walking speed. Absolute differences in variables between normal and slow walking speeds for normal subjects are presented in less detail. This section also describes the results of the cluster analysis, which examined if the Sgroup could be divided into subgroups based on their gait kinematic data. Each subgroup was then individually examined in an attempt to establish how their gait pattern differed from normal subjects walking at a matched speed. This comparison was undertaken in an attempt to establish the gait deformities present within each subgroup, thus aiding in the development of specific treatment protocols.

The following were the aims of the experimentation:

— To describe the gait pattern of stroke patients with equinus deformity of the foot in comparison to normal subjects walking at matched speed.

— To determine if the gait pattern of stroke patients with equinus deformity of the foot is homogeneous.

— To examine if the gait pattern of the subgroups identified within the stroke patients with equinus deformity of the foot differs from normal subjects.

— To examine if the gait pattern of normal subjects is affected by walking speed.

The following hypotheses were made prior to experimentation

— Stroke patients with equinus deformity of the foot will have a significantly different gait pattern than normal subjects at a matched speed.

— The gait patterns of stroke patients with equinus deformity of the foot are not homogenous.

— The gait pattern of each subgroup identified within the stroke patient group differs from normal subjects

— The gait pattern of normal subjects is affected by walking speed.

The methods and statistical analysis of this experimental work were previously presented in Chapter 2.

**Results**

In light of the aims of the study each result section will in the first instance report on the comparison between the Sgroup and the $NAMgroup_{ms}$ for the affected and unaffected legs. Then secondly the comparison of the affected and unaffected leg of the Sgroup will be reported, followed by the report on the comparison of the Sgroup affected and unaffected legs to the $NAMgroup_{ns}$. Lastly the effects of walking speed on the gait of the NAMgroup will be reported, but in less detail. Only statistically significant differences between the groups will be reported in the text in the following sections. The non statistically significant differences will be presented also in Tables 4.1-4.21.

## 4.1 Temporal distance parameters

The means and standard deviations of the temporal distance parameters for the Sgroup's affected and unaffected limb and the $NAMgroup_{ms}$ and $NAMgroup_{ns}$ are presented in Tables 4.1 and 4.2. When the affected limb of the Sgroup was compared to that of the $NAMgroup_{ms}$ double support time (DS) was significantly lower by 1.0s, limp index (LI) was significantly higher by 0.2, single limb support time (SS) was significantly decreased by 0.8s, step length (SL) was significantly decreased by 0.1m, step width (SW) was significantly increased by 0.1 m and stride length (SRL) was significantly decreased by 0.4m in the Sgroup. When the unaffected limb of the Sgroup was compared to that of the $NAMgroup_{ms}$ double support time was significantly lower by 1.1s, foot off percentage (FO) was significantly higher by 11.5%, limp index was significantly higher by 0.6, single limb support time was significantly decreased by 0.6s, step length was significantly decreased by 0.2m, step time (ST) was significantly decreased by 0.5s, step width was significantly increased by 0.1m and stride length was significantly decreased by 0.4m. When the unaffected limb of the Sgroup was compared to that of the affected limb significant differences were found for the percentage at which foot off occurred which was significantly higher on the unaffected limb by 11.5 (±6.2)%, limp index was significantly greater on the unaffected limp by 0.3 (±0.2), single support time was significantly increased by 0.3 (±0.2)s on the unaffected limb, step length was significantly decreased on the unaffected limb by 0.1 (±0.1)m and step time (ST) was significantly decreased on the unaffected limb compared to the affected limb by 0.4 (±0.4)s [Tables 4.1 and 4.2]. The affected limb of the Sgroup had a significantly lower cadence (C) by 47.5step.min$^{-1}$, a significantly longer stance percentage by 6.3%, a significantly greater limp index by 0.1, a significantly decreased single limb support time by 0.2s, a significantly decreased step length by 0.4m, a significantly increased step time by 0.8s, a significantly decreased stride length by 1.0m, a significantly increased stride time (SRT) by 1.0s and a significantly decreased walking speed (WS) by 0.9m.s$^{-1}$ when compared to that of the NAMgroup$_{ns}$. The unaffected limb of the Sgroup had a significantly lower cadence by 48.0step.min$^{-1}$, a significantly longer stance percentage by 17.8%, a significantly greater limp index by 0.5, a significantly decreased step length by 0.5 m, a significantly increased step time by 0.4s, a significantly decreased stride length by 0.9m, a significantly increased stride time by 1.0s and a significantly decreased walking speed by 1.0s when compared to that of the NAMgroup$_{ns}$. When the effect of walking speed on the NAMgroup was analysed speed had a significant effect on many variables. As walking speed decreased it resulted in a significant decrease in cadence by 50.7 (± 9.7)step.min$^{-1}$, a significant increase in double support duration by 1.4 (±0.3)s, a significant increase in foot off percentage by 6.4 (±3.1)%, a significant increase in single support time by 0.6 (±0.1)s, a significant decrease in step length by 0.3 (±0.4)m, a significant increase in step time by 0.2 (±0.3)s, a significant decrease in step width by 0.03 (±0.02)m, a significant decrease in stride length by 0.5 (±0.2)m, a significant increase in stride time by 1.0 (±0.1)s and a significant decrease in walking speed by 0.9 (±0.2)m.s$^{-1}$(Table 4.1).

### 4.2 Kinematic variables

**Pelvic Obliquity**

The means and standard deviations of the joint angles for the pelvis for the Sgroup, NAMgroup$_{ns}$ and the NAMgroup$_{ms}$ are presented in Tables 4.3 and 4.4. The pelvis of the stroke patients rose on the affected limb during the swing phase of the gait cycle (HR3) significantly more than that of the NAMgroup$_{ms}$ by 3.0° [Table 4.3]. During the stance phase the drop of the pelvis (HR2) in the Sgroup was significantly more than that of the

**Table 4.1**: Means (± standard deviation) of temporal distance parameters of NAMgroup walking at both (a) a normal and (b) a slow speed and (c) the Sgroups affected limb.

| | (a) | (b) | (c) | Statistical Analysis | | |
|---|---|---|---|---|---|---|
| | $NAM_{ns}$ group | $NAM_{ms}$ group | Sgroup | axb (p) | axc (p) | bxc (p) |
| **C** ($step.min^{-1}$) | 109.3 (±10.1) | 58.5 (±2.4) | 61.8 (±18.7) | <0.05 | <0.05 | NS |
| **DS** (sec) | 0.9 (±0.3) | 2.2 (±0.3) | 1.2 (±0.6) | <0.05 | NS | <0.05 |
| **FO** (%) | 63.5 (±2.8) | 69.8 (±2.9) | 69.8 (±8.0) | <0.05 | <0.05 | NS |
| **LI** | 0.7 (±0.2) | 0.6 (±0.1) | 0.8 (±0.1) | NS | <0.05 | <0.05 |
| **SS** (sec) | 0.6 (±0.1) | 1.2 (±0.1) | 0.4 (±0.1) | <0.05 | <0.05 | <0.05 |
| **SL** (m) | 0.70 (±0.14) | 0.43 (±0.1) | 0.34 (±0.16) | <0.05 | <0.05 | <0.05 |
| **ST** (sec) | 0.5 (±0.3) | 0.7 (±0.2) | 1.3 (±0.5) | <0.05 | <0.05 | NS |
| **SW** (m) | 0.18 (±0.27) | 0.19 (±0.03) | 0.25 (±0.06) | <0.05 | NS | <0.05 |
| **SRL** (m) | 1.47 (±0.14) | 0.96 (±0.18) | 0.54 (±0.29) | <0.05 | <0.05 | <0.05 |
| **SRT** (sec) | 1.1 (±0.1) | 2.1 (±0.1) | 2.1 (±0.7) | <0.05 | <0.05 | NS |
| **WS** ($m.s^{-1}$) | 1.33 (±0.17) | 0.4 (±0.05) | 0.31 (±0.25) | <0.05 | <0.05 | NS |

NS = Not statistically significantly different
<0.05= statistically significantly different

C= Cadence
DS = Double support time
FO = Foot off percentage
LI = Limp Index
SS = single limb support time
SL = Step length
ST = Step time
SW = Step width
SRL = Stride length
SRT = Stride time
WS = Walking speed

**Table 4.2**: Means (± standard deviation) of temporal distance parameters of NAMgroup walking at both (a) a normal and (b) a slow speed and (c) the Sgroups affected limb and (d) the Sgroups unaffected limb.

| | (a) | (b) | (c) | (d) | Statistical Analysis | | |
|---|---|---|---|---|---|---|---|
| | $NAM_{ns}$ group | $NAM_{ms}$ group | Sgroup Affected limb | Sgroup Unaffected limb | axd (p) | bxd (p) | cxd (p) |
| C ($step.min^{-1}$) | 109.3 (±10.1) | 58.5 (±2.4) | 61.8 (±18.7) | 61.0 (±18.2) | <0.05 | NS | NS |
| DS (sec) | 0.9 (±0.3) | 2.2 (±0.3) | 1.2 (±0.6) | 1.1 (±0.5) | NS | <0.05 | NS |
| FO(%) | 63.5 (±2.8) | 69.8 (±2.9) | 69.8 (±8.0) | 81.3 (±6.5) | <0.05 | <0.05 | <0.05 |
| LI | 0.7 (±0.2) | 0.6 (±0.1) | 0.8 (±0.1) | 1.2 (±0.1) | <0.05 | <0.05 | <0.05 |
| SS (sec) | 0.6 (±0.1) | 1.2 (±0.1) | 0.4 (±0.1) | 0.6 (±0.2) | NS | <0.05 | <0.05 |
| SL(m) | 0.70 (±0.14) | 0.43 (±0.01) | 0.34 (±0.16) | 0.24 (±0.16) | <0.05 | <0.05 | <0.05 |
| ST (sec) | 0.5 (±0.3) | 0.7 (±0.2) | 1.3 (±0.5) | 0.9 (±0.3) | <0.05 | <0.05 | <0.05 |
| SW (m) | 0.18 (±0.27) | 0.19 (±0.03) | 0.25 (±0.06) | 0.24 (±0.05) | NS | <0.05 | NS |
| SRL (m) | 1.47 (±0.14) | 0.96 (±0.18) | 0.54 (±0.29) | 0.54 (±0.29) | <0.05 | <0.05 | NS |
| SRT (sec) | 1.1 (±0.1) | 2.1 (±0.1) | 2.1 (±0.7) | 2.1 (±0.6) | <0.05 | NS | NS |
| WS ($m.s^{-1}$) | 1.33 (±0.17) | 0.4 (±0.05) | 0.31 (±0.25) | 0.32 (±0.26) | <0.05 | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 115 for a description of the abbreviations

**Table 4.3:** Means (± standard deviation) of hip and pelvic angles (°) of the NAMgroup walking at both (a) a normal and (b) a slow speed and (c) the Sgroups affected limb.

| | (a) | (b) | (c) | Statistical Analysis | | |
|---|---|---|---|---|---|---|
| | $NAM_{ns}$ Group | $NAM_{ms}$ group | Sgroup | axb (p) | axc (p) | bxc (p) |
| H1 | 24.9(±9.9) | 16.1(±8.2) | 14.1(±5.5) | <0.05 | NS | NS |
| H2 | 23.5(±11.4) | 28.3(±5.3) | 22.8(±10.3) | NS | NS | NS |
| H3 | -23.4(±8.9) | -15.6(±8.7) | -4.1(±5.3) | <0.05 | <0.05 | <0.05 |
| H4 | -13.2(±10.4) | -5.0(±9.6) | 4.9(±6.1) | <0.05 | <0.05 | <0.05 |
| H5 | 25.3(±8.8) | 18.4(±8.3) | 18.4(±16.9) | NS | NS | NS |
| H6 | 48.7(±4.1) | 33.0(±5.4) | 23.4(±10.1) | NS | <0.05 | <0.05 |
| H7 | 11.3(±4.9) | 7.9(±3.2) | 8.2(±4.4) | NS | <0.05 | NS |
| H8 | -4.4(±3.1) | -2.6(±2.8) | -7.8(±6.7) | <0.05 | <0.05 | <0.05 |
| H9 | 6.9(±4.3) | 5.5(±3.0) | 0.3(±6.1) | NS | <0.05 | <0.05 |
| HR2 | -2.3(±2.3) | -1.1(±2.1) | -7.2(±5.5) | NS | <0.05 | <0.05 |
| HR3 | 4.8(±3.2) | 1.7(±2.4) | 4.7(±5.5) | <0.05 | <0.05 | <0.05 |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 4.4:** Means (± standard deviation) of hip and pelvic angles (°) of the NAMgroup walking at both (a) a normal and (b) a slow speed (c) the Sgroups affected limb (d) Sgroups unaffected limb.

| | (a) | (b) | (c) | (d) | Statistical Analysis | | |
|---|---|---|---|---|---|---|---|
| | $NAM_{ns}$ Group | $NAM_{ms}$ group | Sgroup Affected limb | Sgroup Unaffected limb | axd (p) | bxd (p) | cxd (p) |
| **H1** | 24.9(±9.9) | 16.1(±8.2) | 14.1(±5.5) | 26.3(±11.4) | NS | <0.05 | <0.05 |
| **H2** | 23.5(±11.4) | 28.3(±5.3) | 22.8(±10.3) | 37.3(±8.1) | NS | NS | NS |
| **H3** | -23.4(±8.9) | -15.6(±8.7) | -4.1(±5.3) | -7.9(±11.8) | <0.05 | NS | <0.05 |
| **H4** | -13.2(±10.4) | -5.0(±9.6) | 4.9(±6.1) | 9.6(±8.9) | <0.05 | <0.05 | NS |
| **H5** | 25.3(±8.8) | 18.4(±8.3) | 18.4(±16.9) | 28.8(±13.1) | NS | <0.05 | <0.05 |
| **H6** | 48.7(±4.1) | 33.0(±5.4) | 23.4(±10.1) | 31.2(±15.4) | <0.05 | NS | <0.05 |
| **H7** | 11.3(±4.9) | 7.9(±3.2) | 8.2(±4.4) | 3.7(±2.8) | NS | <0.05 | <0.05 |
| **H8** | -4.4(±3.1) | -2.6(±2.8) | -7.8(±6.7) | 7.1(±5.5) | NS | <0.05 | <0.05 |
| **H9** | 6.9(±4.3) | 5.5(±3.0) | 0.3(±6.1) | 3.4(±6.2) | NS | NS | <0.05 |
| **HR2** | -2.3(±2.3) | -1.1(±2.1) | -7.2(±5.5) | -7.9(±5.9) | NS | <0.05 | <0.05 |
| **HR3** | 4.8(±3.2) | 1.7(±2.4) | 4.7(±5.5) | -1.7(±5.1) | <0.05 | NS | <0.05 |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

$NAMgroup_{ms}$ by 6.1° [Table 4.3]. On the unaffected limb of the Sgroup during the stance phase the drop of the pelvis (HR2) was significantly decreased in comparison to that of the $NAMgroup_{ms}$ by 9.0° [Table 4.4]. On the unaffected limb in the Sgroup, during both the swing and stance phases of the gait cycle the movement of the pelvis was significantly different from that of the affected limb. During the stance phase of the gait cycle (HR2) the pelvis on the unaffected limb was more raised than the affected limb by 15.1 (±5.2)° and during the swing phase of the gait cycle it was dropped more (HR3) by 6.4 (±5.4)°. When the gait of the affected limb of the Sgroup was compared to that of the $NAMgroup_{ns}$ the drop of the pelvis (HR2) was significantly less by 4.9° in the Sgroup in comparison to that of the $NAMgroup_{ns}$ [Table 4.3]. When the gait of the unaffected limb of the Sgroup was compared to that of the $NAMgroup_{ns}$ the position of the pelvis during swing (HR3) rose significantly more by 9.5° in the Sgroup in comparison to that of the $NAMgroup_{ns}$ [Table 4.4]. When the effect of walking speed on the NAMgroup was analysed, speed had a significant effect on the rise of the pelvis during the swing phase of gait (HR3) causing it to increase by 3.1 (±1.6)° when walking speed decreased [Table 4.3].

**Hip angles**

The means and standard deviations of the joint angles for the pelvis, hip, knee and ankle for the Sgroup, $NAMgroup_{ns}$ and the $NAMgroup_{ms}$ are presented in Tables 4.3 and 4.4. When the gait of the affected limb of the Sgroup was compared to that of the $NAMgroup_{ms}$ the maximum extension angle of the hip in stance (H3) was significantly decreased by 11.5°, the flexion angle of the hip at toe-off (H4) was significantly increased by 9.9° and the total sagittal plane excursion angle of the hip (H6) was significantly decreased by 9.6° in the Sgroup in comparison to that of the $NAMgroup_{ms}$ [Table 4.3]. When the gait of the unaffected limb of the Sgroup was compared to that of the $NAMgroup_{ms}$ the angle of the hip at heel strike (H1) was significantly increased by 10.2°, the flexion angle of the hip at toe-off (H4) was significantly increased by 14.6° and the maximum flexion angle of the hip during swing (H5) was significantly increased by 10.4° in the Sgroup in comparison to that of the $NAMgroup_{ms}$ [Table 4.4]. When the unaffected limb of the Sgroup was compared to that of the affected limb the angle of the hip at heel strike (H1) was significantly increased by 12.2 (±10.2)°, the maximum extension angle of the hip during stance (H3) was significantly increased by 4.6 (±8.4)°, the maximum flexion angle of the hip during swing (H5) was significantly increased by 9.4 (±9.9)° and the total excursion angle of the hip (H6) was significantly greater by 7.8 (±12.6)° on the unaffected limb [Table 4.4]. When the affected leg of the Sgroup was compared to that of the $NAMgroup_{ns}$ the maximum extension angle of the hip in stance (H3) was significantly decreased by 19.4°, the angle of the hip at toe-off (H4) was significantly increased by 18.1° and the total sagittal plane excursion angle of the hip (H6) was significantly decreased by 25.3° in the Sgroup in comparison to the $NAMgroup_{ns}$ [Table 4.3]. When the unaffected leg of the Sgroup was compared to that of the $NAMgroup_{ns}$ the maximum extension angle of the hip in stance (H3) was significantly decreased by 15.5°, the angle of the hip at toe-off (H4) was significantly increased by 22.8° and the total sagittal plane excursion angle of the hip (H6) was significantly decreased by 16.6° in the Sgroup in comparison to the $NAMgroup_{ns}$ [Table 4.4]. When the effect of speed on the NAMgroup was analysed, speed had a significant effect on the hip flexion angle at heel strike (H1) which was found to significantly decrease by 8.7 (±3.8)°, the maximum extension angle of the hip in the stance phase (H3) was found to significantly increase by 7.8 (±2.5)° and the flexion angle of the hip at toe-off (H4) was found to significantly increase by 8.1 (±1.3)° when walking speed decreased [Table 4.3].

Significant differences were also found for the movements of the hip in the coronal plane when the gait of the affected limb of the Sgroup was compared to that of the $NAMgroup_{ms}$. The maximum adduction angle of the hip in the stance phase of the gait cycle (H8) was significantly increased by 5.2° and the maximum abduction angle of the hip in the swing phase of the gait cycle (H9) was significantly decreased by 5.2° in the affected limb of the Sgroup in comparison to that of the $NAMgroup_{ms}$ [Table 4.3]. When the gait of the unaffected limb of the Sgroup was compared to that of the $NAMgroup_{ms}$ the maximum adduction angle of the hip in the stance phase of the gait cycle (H8) was significantly decreased by 9.7° and the total coronal plane excursion angle of the hip was significantly decreased by 4.3° in the unaffected limb of the Sgroup in comparison to that of the $NAMgroup_{ms}$ [Table 4.4]. The movement of the hip in the coronal plane was significantly different between the affected and unaffected limb in the Sgroup with a significantly decreased total coronal plane excursion angle (H7) by 4.5 (±3.9)°, a significantly decreased adduction angle of the hip (H8) in stance by 13.6 (±9.5)° and a significantly increased abduction angle of the hip in swing (H9) by 3.1 (±6.4)° in the unaffected limb [Table 4.4]. When the affected limb of the Sgroup was compared to that of the $NAMgroup_{ns}$ the maximum adduction angle of the hip in the stance phase of the gait cycle (H8) significantly increased by 3.4°, the maximum abduction angle of the hip in the swing phase of the gait cycle (H9) significantly decreased by 6.6° and the total coronal plane excursion angle of the hip (H7) significantly decreased by 3.1° in the affected limb of the Sgroups in comparison to that of the $NAMgroup_{ns}$ [Table 4.3]. When the unaffected limb of the Sgroup was compared to that of the $NAMgroup_{ns}$ no coronal plane variables of the hip were significantly different. When the effect of walking speed on the NAMgroup was analysed, speed had a significant affect on the maximum adduction angle of the hip in the stance phase (H8) which was significantly decreased by 1.8 (±1.3)° when the walking speed was decreased [Table 4.3].

**Knee angles**

The means and standard deviations of the joint angles for the knee for the Sgroup, $NAMgroup_{ns}$ and the $NAMgroup_{ms}$ are presented in Tables 4.5 and 4.6. When the affected limb of the Sgroup was compared to that of the $NAMgroup_{ms}$, the angle of the knee at heel strike (K1) was significantly increased by 6.4°, the angle of the knee at toe-off (K4) was significantly decreased by 14.8°, the maximum flexion angle of the knee during swing (K5) was significantly decreased by 34° and the total excursion angle of the knee (K6) was significantly decreased by 36.7° in the Sgroup in comparison to that of the $NAMgroup_{ms}$ (Table 5.5).

**Table 4.5:** Means (± standard deviation) of knee angles (°) of the NAMgroup walking at both (a) a normal and (b) a slow speed and (c) the Sgroups affected limb.

| | (a) $NAM_{ns}$ group | (b) $NAM_{ms}$ group | (c) Sgroup | Statistical Analysis axb (p) | axc (p) | bxc (p) |
|---|---|---|---|---|---|---|
| **K1** | -0.2(±4.9) | 0.0(±5.2) | 6.2(±8.8) | NS | <0.05 | <0.05 |
| **K2** | 16.9(±5.1) | 4.1(±4.6) | 11.3(±9.1) | <0.05 | <0.05 | NS |
| **K3** | -4.2(±4.1) | -3.1(±5.6) | -2.7(±9.7) | NS | NS | NS |
| **K4** | 33.0(±4.1) | 31.3(±6.6) | 18.2(±13.5) | NS | <0.05 | <0.05 |
| **K5** | 57.0(±3.7) | 43.7(±6.7) | 23.0(±14.6) | <0.05 | <0.05 | <0.05 |
| **K6** | 61.2(±3.8) | 45.4(±9.5) | 24.5(±13.3) | <0.05 | <0.05 | <0.05 |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 4.6:** Means (± standard deviation) of knee angles (°) of the NAMgroup walking at both (a) a normal and (b) a slow speed (c) the Sgroups affected limb (d) the Sgroups unaffected limb.

| | (a) $NAM_{ns}$ group | (b) $NAM_{ms}$ group | (c) Sgroup Affected limb | (d) Sgroup Unaffected limb | Statistical Analysis axd (p) | bxd (p) | cxd (p) |
|---|---|---|---|---|---|---|---|
| **K1** | -0.2(±4.9) | 0.0(±5.2) | 6.2(±8.8) | 17.4(±10.6) | NS | <0.05 | NS |
| **K2** | 16.9(±5.1) | 4.1(±4.6) | 11.3(±9.1) | 19.7(±8.8) | NS | <0.05 | <0.05 |
| **K3** | -4.2(±4.1) | -3.1(±5.6) | -2.7(±9.7) | 6.7(±6.6) | NS | <0.05 | <0.05 |
| **K4** | 33.0(±4.1) | 31.3(±6.6) | 18.2(±13.5) | 41.3(±10.9) | <0.05 | <0.05 | <0.05 |
| **K5** | 57.0(±3.7) | 43.7(±6.7) | 23.0(±14.6) | 51.2(±13.1) | <0.05 | NS | <0.05 |
| **K6** | 61.2(±3.8) | 45.4(±9.5) | 24.5(±13.3) | 38.6(±16.7) | <0.05 | NS | <0.05 |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

When the unaffected limb of the Sgroup was compared to that of the $NAMgroup_{ms}$ the angle of the knee at heel strike (K1) was significantly increased by 17.4°, the angle of the knee during the loading response (K2) was significantly increased by 15.6°, the maximum extension angle of the knee during stance (K3) was significantly increased by 9.8° and the knee angle at toe-off (K4) was significantly increased by 10° in the Sgroup in comparison to that of the $NAMgroup_{ms}$ (Table 4.6). When the unaffected limb of the Sgroup was compared to that of the affected limb, the angle of the knee during the loading response (K2) was significantly increased by 8.4 (±8.7)°, the maximum extension of the knee during stance (K3) was significantly decreased by 9.7 (±7.3)°, the knee flexion angle at toe-off (K4) was significantly increased by 23.1 (±11.1)°, the maximum flexion angle of the knee during swing (K5) was significantly increased by 28.2 (±14.6)° and the total sagittal plane excursion angle of the knee (K6) was significantly increased by 14.1 (±17.2)° in the unaffected limb. When the affected limb of the Sgroup was compared to that of the $NAMgroup_{ns}$, the angle of the knee at heel strike (K1) was significantly increased by 6.4°, the angle of the knee during the loading response (K2) was significantly decreased by 7.1°, the angle of the knee at toe-off (K4) was significantly decreased by 14.8°, the maximum flexion angle of the knee during swing (K5) was significantly decreased by 34.0° and the total excursion angle of the knee (K6) was significantly decreased by 36.7° in the Sgroup in comparison to that of the $NAMgroup_{ns}$ (Table 4.5). When the unaffected limb of the Sgroup was compared to that of the $NAMgroup_{ns}$ the angle to the knee at toe-off (K4) was significantly increased by 8.3°, the maximum flexion angle of the knee during swing (K5) was significantly decreased by 5.8° and the total excursion angle of the knee (K6) was significantly decreased by 22.6° in the Sgroup in comparison to that of the $NAMgroup_{ns}$ (Table 4.6). When the effect of walking speed on NAMgroup was analysed, speed significantly affected the angle of the knee during the loading response (K2) which significantly decreased by 12.8 (±4.3)°, the maximum flexion angle of the knee during swing (K5) which significantly decreased by 13.3 (±5.1)° and the total excursion angle of the knee (K6) which significantly decreased by 15.8 (±9.0)°, as the walking speed decreased [Table 4.5].

**Ankle angles**

The means and standard deviations of the joint angles for the ankle for the Sgroup, NAMgroup$_{ns}$ and the NAMgroup$_{ms}$ are presented in Tables 4.7, 4.8, 4.9 and 4.10. When the gait of the affected limb of the Sgroup was compared to that of the NAMgroup$_{ms}$, the maximum dorsiflexion angle of the ankle in the stance phase (A3) was significantly less by 4.3° and the total sagittal plane excursion angle of the ankle (A6) significantly less by 6.6° in the Sgroup in comparison to that of the NAMgroup$_{ms}$ [Table 4.7]. When the gait of the unaffected limb of the Sgroup was compared to that of the NAMgroup$_{ms}$, the maximum plantarflexion angle of the ankle during the loading response (A2) significantly decreased by 7.8° and the maximum dorsiflexion angle of the ankle in the stance phase (A3) significantly increased by 5.6° in the Sgroup in comparison to that of the NAMgroup$_{ms}$ [Table 4.8].

**Table 4.7:** Means (± standard deviation) of ankle sagittal angles (°) of the NAMgroup walking at both (a) a normal and (b) a slow speed and (c) the Sgroups affected limb.

| | (a) | (b) | (c) | Statistical Analysis | | |
|---|---|---|---|---|---|---|
| | NAM$_{ns}$ Group | NAM$_{ms}$ group | Sgroup | axb (p) | axc (p) | bxc (p) |
| A1 | 8.9(±7.1) | 3.8(±7.8) | 1.5(±6.4) | <0.05 | <0.05 | NS |
| A2 | -1.0(±8.3) | -2.2(±6.9) | -2.0(±5.0) | NS | NS | NS |
| A3 | 14.9(±7.5) | 16.6(±6.9) | 12.3(±8.5) | <0.05 | NS | <0.05 |
| A4 | -10.6(±10.1) | 1.4(±8.4) | 3.7(±6.5) | <0.05 | <0.05 | NS |
| A5 | 8.9(±5.9) | 9.3(±7.2) | 4.8(±6.1) | NS | NS | NS |
| A6 | 25.7(±5.4) | 18.9(±3.0) | 12.3(±7.0) | <0.05 | <0.05 | <0.05 |
| A7 | -12.3(±12.1) | -0.6(±7.6) | -0.2(±5.7) | <0.05 | <0.05 | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 4.8:** Means (± standard deviation) of ankle sagittal angles (°) of the NAMgroup walking at both (a) a normal and (b) a slow speed (c) the Sgroups affected limb and (d) Sgroups unaffected limb.

| | (a) | (b) | (c) | (d) | Statistical Analysis | | |
|---|---|---|---|---|---|---|---|
| | $NAM_{ns}$ Group | $NAM_{ms}$ group | Sgroup Affected limb | Sgroup Unaffected limb | axd (p) | bxd (p) | cxd (p) |
| **A1** | 8.9(±7.1) | 3.8(±7.8) | 1.5(±6.4) | 9.5(±7.9) | <0.05 | NS | NS |
| **A2** | -1.0(±8.3) | -2.2(±6.9) | -2.0(±5.0) | 5.6(±7.4) | NS | <0.05 | <0.05 |
| **A3** | 14.9(±7.5) | 16.6(±6.9) | 12.3(±8.5) | 22.2(±6.1) | <0.05 | <0.05 | <0.05 |
| **A4** | -10.6(±10.1) | 1.4(±8.4) | 3.7(±6.5) | 5.4(±8.1) | NS | NS | <0.05 |
| **A5** | 8.9(±5.9) | 9.3(±7.2) | 4.8(±6.1) | 14.3(±6.6) | <0.05 | NS | <0.05 |
| **A6** | 25.7(±5.4) | 18.9(±3.0) | 12.3(±7.0) | 15.2(±6.5) | <0.05 | NS | <0.05 |
| **A7** | -12.3(±12.1) | -0.6(±7.6) | -0.2(±5.7) | 3.9(±7.9) | NS | NS | <0.05 |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

When the gait of the unaffected limb of the Sgroup was compared to that of the affected limb, the maximum plantarflexion angle of the ankle during the loading response (A2) was significantly less by 7.6 (±8.2)°, the maximum dorsiflexion angle during stance (A3) was significantly increased by 9.9 (±9.7)°, the angle of the ankle at toe-off (A4) was significantly decreased by 1.7 (±7.8)°, the maximum dorsiflexion angle of the ankle during swing (A5) was significantly increased by 9.5 (±6.3)°, the maximum plantarflexion angle of the ankle during swing (A7) was significantly decreased by 4.2 (±6.9)° and the total excursion angle of the ankle in the sagittal plane (A6) was significantly increased by 2.9 (±6.9)° in the unaffected limb. When the gait of the affected limb of the Sgroup was compared to that of the $NAMgroup_{ns}$, the angle of the ankle at heel strike (A1) was significantly decreased by 7.4°, the angle of the ankle at toe-off (A4) was significantly decreased by 14.3°, the total sagittal plane excursion angle of the ankle (A6) was significantly decreased and the maximum plantarflexion angle of the ankle during swing (A7) was significantly decreased by 12.1° in the Sgroup in comparison to that of the NAMgroup$_{ns}$ [Table 4.7]. When the gait of the unaffected limb of the Sgroup was compared to that of the NAMgroup$_{ns}$, the angle of the ankle at heel strike (A1) was significantly increased by 0.6°, the maximum dorsiflexion angle of the ankle during stance (A3) was significantly increased by 7.3°, the maximum dorsiflexion angle of the ankle during swing (A5) was significantly increased by 5.4° and the total sagittal plane excursion angle of the ankle (A6) was significantly decreased by 10.5° in the Sgroup in comparison to that of the NAMgroup$_{ns}$ [Table 4.8]. When the effect of walking speed on NAMgroup was analysed, the angle of the ankle at heel strike (A1) significantly decreased by 5.1 (±2.5)°, the maximum dorsiflexion angle of the ankle during stance (A3) significantly increased by 1.7 (±1.5)°, the angle of the ankle at toe-off (A4) significantly decreased by 12.0 (±8.0)°, the maximum plantarflexion angle of the ankle during swing (A7) significantly decreased by 11.7 (±7.8)° and the total excursion angle of the ankle in the sagittal plane (A6) significant decreased by 6.8 (±5.9)°, as walking speed decreased (Table 4.7).

**Table 4.9**: Means (± standard deviation) of foot rotation angles (°) in the transverse plane of the NAMgroup walking at both (a) a normal and (b) a slow speed and (c) the Sgroups affected limb.

| | (a) | (b) | (c) | Statistical Analysis | | |
|---|---|---|---|---|---|---|
| | NAM$_{ns}$ Group | NAM$_{ms}$ group | Sgroup | axb (p) | axc (p) | bxc (p) |
| **R1** | 2.4(±4.5) | -0.1(±3.8) | -7.9(±19.1) | <0.05 | <0.05 | NS |
| **R2** | -7.2(±11.2) | -6.6(±9.9) | -18.0(±18.2) | NS | NS | NS |
| **R3** | -4.1(±10.5) | -4.7(±5.6) | -5.4(±18.9) | NS | NS | NS |
| **R4** | -6.6(±5.9) | -9.9(±4.4) | -12.2(±18.5) | <0.05 | NS | NS |
| **R5** | -12.2(±5.7) | -12.3(±3.7) | -15.8(±18.6) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 4.10**: Means (± standard deviation) of foot rotation angles (°) in the transverse plane of the NAMgroup walking at both (a) a normal and (b) a slow speed (c) the Sgroups affected limb and (b) the Sgroups unaffected limb.

| | (a) | (b) | (c) | (d) | Statistical Analysis | | |
|---|---|---|---|---|---|---|---|
| | $NAM_{ns}$ group | $NAM_{ms}$ group | Sgroup Affected limb | Sgroup Unaffected limb | axd (p) | bxd (p) | cxd (p) |
| R1 | 2.4(±4.5) | -0.1(±3.8) | -7.9(±19.1) | -4.4(±17.3) | NS | NS | NS |
| R2 | -7.2(±11.2) | -6.6(±9.9) | -18.0(±18.2) | -15.9(±14.3) | NS | NS | NS |
| R3 | -4.1(±10.5) | -4.7(±5.6) | -5.4(±18.9) | -3.6(±16.2) | NS | NS | NS |
| R4 | -6.6(±5.9) | -9.9(±4.4) | -12.2(±18.5) | -11.6(±13.8) | NS | NS | NS |
| R5 | -12.2(±5.7) | -12.3(±3.7) | -15.8(±18.6) | -14.1(±14.1) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

No significant differences were found in foot rotation angle when the gaits of the affected or unaffected limbs of the Sgroup were compared to that of the $NAMgroup_{ms}$ (Table 4.9). No significant differences were found for the foot rotation angles during the gait cycle between the affected and unaffected limb (Table 4.10). When the gait of the affected limb of the Sgroup was compared to that of the $NAMgroup_{ns}$, the rotation angle of the foot at heel strike (R1) was significantly increased into external rotation by 10.3° in the Sgroup in comparison to that of the $NAMgroup_{ns}$ (Table 4.9). When the gait of the unaffected limb of the Sgroup was compared to that of the $NAMgroup_{ns}$, no significant differences were noted between the foot rotation angles (Table 4.10). When the effects of walking speed on foot rotation angle in the NAMgroup was assessed, the rotation angle of the foot at heel strike (R1) significantly increased into external rotation by 2.5 (±3.0)° and the rotation angle of the foot at toe-off (R4) significantly increased into external rotation by 3.3 (±3.0)° as walking speed decreased [Table 4.9].

### 4.3 Kinetic Variables

The means and standard deviations of joint moments for the hip, knee and ankle for the Sgroup, the $NAMgroup_{ns}$ and the $NAMgroup_{ms}$ are presented in Table 4.11.

**Table 4.11:** Means (± standard deviation) of hip, knee and ankle moments (Nm/kg) for the NAMgroup walking at both (a) a normal and (b) a slow speed and (c) the Sgroup's affected limb.

| | (a) | (b) | (c) | Statistical Analysis | | |
|---|---|---|---|---|---|---|
| | $NAM_{ns}$ Group | $NAM_{ms}$ group | Sgroup | Axb (p) | axc (p) | bxc (p) |
| **HM1** | 1.3(±0.7) | 0.3(±0.2) | 0.5(0.4) | <0.05 | <0.05 | <0.05 |
| **HM2** | -0.8(±0.4) | -0.3(±0.2) | -0.3(±0.3) | <0.05 | <0.05 | NS |
| **KM1** | -0.5(±0.3) | -0.1(±0.1) | -0.2(±0.2) | <0.05 | <0.05 | NS |
| **KM2** | 0.3(±0.2) | -0.0(±0.1) | -0.0(±0.2) | <0.05 | <0.05 | NS |
| **KM3** | -0.7(±0.2) | -0.4(±0.1) | -0.4(±0.2) | <0.05 | <0.05 | NS |
| **AM1** | -0.1(±0.1) | -0.0(±0.0) | -0.0(±0.0) | <0.05 | <0.05 | NS |
| **AM2** | 1.7(±0.2) | 1.1(±0.2) | 0.7(±0.4) | <0.05 | <0.05 | <0.05 |
| **AR1** | -0.0(±0.0) | -0.0(±0.0) | -0.0(±0.0) | <0.05 | <0.05 | NS |
| **AR2** | 0.3(±0.1) | 0.2(±0.1) | 0.1(±0.1) | <0.05 | <0.05 | <0.05 |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

When the gait of the affected limb of the Sgroup was compared to that of the $NAMgroup_{ms}$ the maximum hip flexor moment (HM1) was significantly greater by 0.2Nm/kg, the ankle plantarflexor moment (AM2) was significantly decreased by 0.4Nm/kg and the foot internal rotation moment (AR2) was significantly decreased by 0.1Nm/kg in the Sgroup in comparison to that of the $NAMgroup_{ms}$ (Table 4.11). When the gait of the affected limb of the Sgroup was compared to that of the $NAMgroup_{ns}$ the hip flexor moment (HM1) was significantly decreased by 0.8Nm/kg, the hip extensor moment (HM2) was significantly decreased by 0.5Nm/kg, the first knee extensor moment (KM1) was significantly decreased by 0.3Nm/kg, the knee flexor moment (KM2) was significantly decreased by 0.3Nm/kg, the second knee extensor moment

(KM3) was significantly decreased by 0.3Nm/kg, the ankle dorsiflexor moment (AM1) was significantly decreased by 0.1Nm/kg, the ankle plantarflexor moment was significant decreased by 1.0Nm/kg, the maximum external rotation moment of the foot (AR1) was significantly decreased by 0.02Nm/kg and the maximum internal rotation moment of the foot (AR2) was significantly decreased by 0.2Nm/kg in the Sgroup in comparison to that of the $NAMgroup_{ns}$ (Table 4.11). Walking speed had a significant effect on the generation of all moments of the lower limb causing them all to be reduced. When walking speed decreased the hip flexor moment (HM1) was significantly decreased by 1.0 (±0.7)Nm/kg, the hip extensor moment (HM2) was significantly decreased by 0.6 (±0.4)Nm/kg, the first knee extensor moment (KM1) was significantly decreased by 0.4 (±0.3)Nm/kg, the knee flexor moment (KM2) was significantly decreased by 0.3 (±0.1)Nm/kg, the second knee extensor moment (KM3) was significantly decreased by 0.3 (±0.2)Nm/kg, the ankle dorsiflexor moment (AM1) was significantly decreased by 0.1 (±0.1)Nm/kg, the ankle plantarfiexor moment was significant decreased by 0.6 (±0.4)Nm/kg, the maximum external rotation moment of the foot (AR1) significantly decreased by 0.03 (±0.02)Nm/kg and the maximum internal rotation moment of the foot (AR2) was significantly decreased by 0.1 (±0.1)Nm/kg in the Sgroup in comparison to that of the $NAMgroup_{ns}$ (Table 4.11)

**4.4 Results of Cluster Analysis**

The final cluster group solution included 21 stroke patients, as two stroke patients were removed from the final sample. These two patients were removed from the sample as the mahanalobis distance criterion identified them as been outliers and their continued inclusion in the sample would have lead to a poor cluster solution. The cluster analysis identified three subgroups within the Sgroup. The variables identified by the MANOVA as being able to distinguish between the groups were: walking speed, stride and step length, double support time, total excursion of the hip, maximum adduction of the hip in stance, maximum abduction of the hip in swing, maximum extension of the knee in stance, maximum dorsiflexion of the ankle at heel strike, maximum plantarflexion angle of the ankle during loading response, maximum dorsiflexion angle of the ankle in stance phase, plantarflexion angle of the ankle at toe-off, total sagittal excursion angle of the ankle and maximum plantarflexion angle of the ankle in swing. The means and standard deviations of the kinematic and kinetic variables for the stroke cluster groups are presented in Tables 4.12-4.16.

**Table 4.12:** Means (± standard deviation) of the temporal distance parameters of the Sgroup divided into their cluster subgroups.

| | SG1 | SG2 | SG3 | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | SG1 x SG2 | SG1 x SG3 | SG2x SG3 |
| **C**(step.min$^{-1}$) | 67.5 (±16.2) | 52.6 (±14.1) | 59.2 (±18.0) | NS | NS | NS |
| **DS**(sec) | 0.9 (±0.4) | 1.5 (±0.6) | 1.2 (±0.6) | NS | NS | NS |
| **FO**(%) | 65.0 (±5.6) | 77.1 (±6.9) | 69.5 (±7.6) | <0.05 | NS | NS |
| **LI** | 0.8 (±0.1) | 0.9 (±0.1) | 0.8 (±0.1) | NS | NS | NS |
| **SS** (sec) | 0.4 (±0.0) | 0.4 (±0.1) | 0.4 (±0.1) | NS | NS | NS |
| **SL** (m) | 0.45 (±0.12) | 0.19 (±0.10) | 0.30 (±0.12) | <0.05 | NS | NS |
| **ST** (sec) | 1.1 (±0.3) | 1.3 (±0.5) | 1.5 (±0.6) | NS | NS | NS |
| **SW** (m) | 0.24 (±0.03) | 0.24 (±0.09) | 0.28 (±0.10) | NS | NS | NS |
| **SRL** (m) | 0.73 (±0.17) | 0.27 (±0.19) | 0.45 (±0.13) | <0.05 | <0.05 | NS |
| **SRT** (sec) | 1.9 (±0.2) | 2.4 (±0.2) | 2.3 (±0.9) | NS | NS | NS |
| **WS** (m.s$^{-1}$) | 0.42 (±0.16) | 0.13 (±0.11) | 0.23 (±0.13) | <0.05 | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 115 for a description of the abbreviations

**Table 4.13:** Means (± standard deviation) of the hip and pelvic angles (°) of the Sgroup divided into their cluster subgroups.

| | SG1 | SG2 | SG3 | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | SG1 x SG2 | SG1 xSG3 | SG2xSG3 |
| **H1** | 11.8(±13.1) | 16.1(±17.0) | 11.0(±20.8) | NS | NS | NS |
| **H2** | 16.9(±10.4) | 22.2(±9.0) | 35.2(±7.1) | NS | NS | NS |
| **H3** | -10.4(±12.6) | 4.0(±12.8) | -5.8(±21.4) | NS | NS | NS |
| **H4** | 0.7(±16.5) | 11.7(±13.8) | 2.7(±20.9) | NS | NS | NS |
| **H5** | 17.5(±17.2) | 20.4(±17.9) | 13.2(±20.2) | NS | NS | NS |
| **H6** | 29.3(±6.1) | 16.4(±12.1) | 20.8(±3.4) | <0.05 | NS | NS |
| **H7** | 8.7(±5.5) | 8.3(±4.9) | 7.5(±1.2) | NS | NS | NS |
| **H8** | -6.8(±6.0) | -12.5(±4.6) | -2.4(±7.6) | NS | NS | <0.05 |
| **H9** | 1.9(±4.7) | 4.2(±4.2) | 5.1(±7.3) | NS | NS | <0.05 |
| **HR2** | -7.1(±3.9) | -8.7(±5.7) | -4.7(±8.6) | NS | NS | NS |
| **HR3** | 4.7(±4.7) | 5.1(±6.0) | 4.1(±7.8) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 4.14:** Means (± standard deviation) of the knee angles (°) of the Sgroup divided into their cluster subgroups

| | SG1 | SG2 | SG3 | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | SG1 x SG2 | SG1 xSG3 | SG2xSG3 |
| **K1** | 3.4(±7.8) | 9.6(±5.5) | 0.4(±4.7) | NS | NS | NS |
| **K2** | 4.8(±3.0) | 12.5(±3.2) | 4.2(±5.6) | NS | NS | NS |
| **K3** | -6.6(±5.7) | 2.3(±4.0) | -10.0(±8.0) | <0.05 | NS | <0.05 |
| **K4** | 21.2(±15.7) | 19.3(±12.1) | 6.7(±6.2) | NS | NS | NS |
| **K5** | 28.3(±17.7) | 20.5(±11.9) | 11.8(±3.9) | NS | NS | NS |
| **K6** | 31.1(±15.5) | 18.1(±11.1) | 23.2(±6.9) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 4.15:** Means (± standard deviation) of the ankle sagittal angles (°) of the Sgroup divided into their cluster subgroups.

| | SG1 | SG2 | SG3 | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | SG1 x SG2 | SG1 xSG3 | SG2xSG3 |
| **A1** | -0.2(±4.4) | 3.8(±2.8) | -3.9(±5.5) | NS | NS | <0.05 |
| **A2** | -3.5(±3.8) | 3.6(±2.6) | -6.2(±4.3) | <0.05 | NS | <0.05 |
| **A3** | 13.7(±3.3) | 11.4(±2.1) | 3.6(±5.3) | NS | <0.05 | <0.05 |
| **A4** | 3.2(±4.6) | 5.8(±3.0) | -2.0(±6.2) | NS | NS | <0.05 |
| **A5** | 4.3(±5.2) | 5.7(±2.9) | -0.1(±5.4) | NS | NS | NS |
| **A6** | 17.2(±3.5) | 8.3(±3.4) | 6.2(±8.4) | <0.05 | <0.05 | NS |
| **A7** | -1.8(±2.6) | 3.3(±2.8) | -5.5(±6.7) | <0.05 | NS | <0.05 |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 4.16:** Means (± standard deviation) of the foot rotation angles (°) in the transverse plane of the stroke group divided into their cluster subgroups.

| | SG1 | SG2 | SG3 | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | SG1xSG2 | SG1XSG3 | SG2XSG3 |
| **R1** | -5.6(±24.7) | -10.2(±16.4) | -2.2(±11.1) | NS | NS | NS |
| **R2** | -17.7(±21.5) | -17.3(±16.6) | -2.0(±9.4) | NS | NS | NS |
| **R3** | -1.8(±23.9) | -6.9(±16.1) | 3.6(±5.3) | NS | NS | NS |
| **R4** | -10.9(±24.9) | -11.1(±14.6) | -7.7(±8.2) | NS | NS | NS |
| **R5** | -15.6(±25.7) | -14.6(±14.7) | -11.4(±7.8) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 4.17:** Means (± standard deviation) of the hip, knee and ankle moments (Nm/kg) of the Sgroup divided into their cluster subgroups.

| | SG1 | SG2 | SG3 | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | SG1 x SG2 | SG1 xSG3 | SG2xSG3 |
| **HM1** | 0.6(±0.4) | 0.3(±0.3) | 0.5(±0.5) | NS | NS | NS |
| **HM2** | -0.5(±0.2) | -0.2(±0.1) | -0.2(±0.4) | NS | NS | NS |
| **KM1** | -0.3(±0.2) | -0.2(±0.3) | -0.3(±0.3) | NS | NS | NS |
| **KM2** | 0.0(±0.3) | -0.1(±0.3) | 0.0(±0.1) | NS | NS | NS |
| **KM3** | -0.5(±0.2) | -0.3(±0.3) | -0.5(±0.2) | NS | NS | NS |
| **AM1** | 0.0(±0.0) | 0.0(±0.1) | 0.0(±0.0) | NS | NS | NS |
| **AM2** | 0.8(±0.3) | 0.6(±0.3) | 0.5(±0.4) | NS | NS | NS |
| **AR1** | 0.0(±0.0) | 0.0(±0.0) | 0.0(±0.0) | NS | NS | NS |
| **AR2** | 0.1(±0.1) | 0.1(±0.0) | 0.1(±0.0) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Subgroup 1**

This subgroup (SG1) in comparison to SG2 had a significantly faster walking speed (WS) by 0.3 (±0.1)m.s$^{-1}$ (Table 4.12). This was as a result of SG1 step length (ST) being significantly increased in comparison to SG2 by 0.3 (±0.1)m and SG1 stride length (SRL) being significantly increased by 0.3 (±0.1) when compared to SG3 and 0.5 (±0.1)m when compared to SG2 (Table 4.12). SG1 also had a significantly decreased foot off percentage (FO) by 12.1 (±3.3)% when compared to SG2 (Table 4.12). At the hip joint SG1 had significantly increased sagittal plane excursion angle of the hip (H6) in comparison to SG2 by 21.8 (±4.4)° (Table 4.13). At the knee joint SG1 extended their knee during stance (K3) to a significantly greater extent than SG2 by 8.9 (±2.9)° (Table 4.14). At the ankle joint SG1 had a significantly increased ankle plantarflexion angle during the loading response (A2) by 7.1 (±1.8)° in comparison to SG3, a significantly increased maximum dorsiflexion angle of the ankle during stance (A3) by 10.1 (±2.0)° in comparison to SG3, a significantly increased total sagittal plane excursion angle of the ankle (A6) in comparison to SG2 and SG3 by 8.9 (±2.5)° and

11.0 (±2.8)° and SG1 had a significant increase plantarflexion angle during swing (A7) than SG2 by 5.1 (±2.0)° (Table 4.15).

When the gait of SG1 was compared to that of the $NAMgroup_{ms}$ significant differences were noted between the groups for some gait variables (Tables 4.18-4.23). Double support time (DS) was significantly decreased in the SG1 in comparison to that of the $NAMgroup_{ms}$ by 1.3s (Table 4.18). The percentage at which foot off (FO) occurred was significantly increased in SG1 in comparison to that of the $NAMgroup_{ms}$ by 4.8% (Table 4.18). The limp index (LI) was significantly increased in the SG1 in comparison to that of the $NAMgroup_{ms}$ by 0.2 (Table 4.18). Single limb support time (SS) was significantly decreased in the SG1 in comparison to that of the $NAMgroup_{ms}$ by 0.8s (Table 4.18). Step length (SL) was significantly greater in the SG1 in comparison to that of the $NAMgroup_{ms}$ by 0.02m (Table 4.18). Step width (SW) was significantly greater in SG1 in comparison to $NAMgroup_{ms}$ by 0.05m (Table 4.18). At the pelvis the SG1 had a significantly greater fall of the pelvis in the coronal plane during the stance phase of the gait cycle (HR2) by 6.0° and a significantly greater rise of the pelvis in the coronal plane during the swing phase (HR3) by 6.4° in comparison to that of the $NAMgroup_{ms}$ (Table 4.19). When the knee angles of the SG1 were compared to that of the $NAMgroup_{ms}$ the maximum knee flexion angle obtained during the swing phase (K5) was significantly decreased by 15.4° and the total sagittal plane excursion angle of the knee (K6) was significantly decreased by 14.3° in the SG1 in comparison to that of the $NAMgroup_{ms}$ (Table 4.20). When the moments of the lower limbs in SG1 were compared to that of the $NAMgroup_{ms}$, the maximum hip flexor moment (HM1) was significantly increased by 0.3 Nm/kg and the maximum hip extensor moment (HM2) was significantly increased by 0.2 Nm/kg, the first maximum knee extensor moment (KM1) was significantly increased by 0.2 Nm/kg and the maximum foot internal moment (AR2) was significantly decreased by 0.1 Nm/kg in the SG1 in comparison to that of the $NAMgroup_{ms}$ (Table 4.23).

When a comparison is made between the significant differences of the gait of SG1 and that of the $NAMgroup_{ms}$ and the Sgroup affected limbs gait and that of the $NAMgroup_{ms}$ many differences can be noted. SG1 had a greater foot off percentage compared to that of the Sgroup by 8.5% [Table 4.18].

**Table 4.18:** Means (± standard deviation) of temporal distance parameters of SG1, SG2, SG3 and the $NAMgroup_{ms}$

| | $NAM_{ms}$ | SG1 | SG2 | SG3 | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|---|
| | | | | | NAM X SG1 | NAM X SG2 | NAM X SG3 |
| C(step.min$^{-1}$) | 58.5 (±2.4) | 67.5 (±16.2) | 52.6 (±14.1) | 59.2 (±18.0) | NS | NS | NS |
| DS(sec) | 2.2 (±0.3) | 0.9 (±0.4) | 1.5 (±0.6) | 1.2 (±0.6) | <0.05 | <0.05 | <0.05 |
| FO(%) | 69.8 (±2.9) | 65.0 (±5.6) | 77.1 (±6.9) | 69.5 (±7.6) | <0.05 | <0.05 | NS |
| L1 | 0.6 (±0.1) | 0.8 (±0.1) | 0.9 (±0.1) | 0.8 (±0.1) | <0.05 | <0.05 | <0.05 |
| SS (sec) | 1.2 (±0.1) | 0.4 (±0.0) | 0.4 (±0.1) | 0.4 (±0.1) | <0.05 | <0.05 | <0.05 |
| SL (m) | 0.43 (±0.1) | 0.45 (±0.12) | 0.19 (±0.10) | 0.30 (±0.12) | <0.05 | <0.05 | <0.05 |
| ST (sec) | 0.7 (±0.2) | 1.1 (±0.3) | 1.3 (±0.5) | 1.5 (±0.6) | NS | NS | NS |
| SW (m) | 0.19 (±0.03) | 0.24 (±0.03) | 0.24 (±0.09) | 0.28 (±0.10) | <0.05 | NS | NS |
| SRL (m) | 0.96 (±0.18) | 0.73 (±0.17) | 0.27 (±0.19) | 0.45 (±0.13) | NS | <0.05 | <0.05 |
| SRT (sec) | 2.1 (±0.1) | 1.9 (±0.2) | 2.4 (±0.2) | 2.3 (±0.9) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
Grey highlighting signifies that the reaction of the subgroup to the condition is different to the Sgroup when compared to the $NAMgroup_{ms}$
See pages 115 for a description of the abbreviations

At the hip joint SG1 had greater extension of the hip during stance (H3) by 6.3°, greater extension of the hip at toe-off (H4) by 4.2° and greater total sagittal plane excursion of the hip (H6) by 5.9° in comparison to the affected limb of the Sgroup (Table 4.19). The SG1 also had decreased adduction of the hip in stance (H8) by 1.0° and increased abduction of the hip in swing (H9) by 1.6° [Table 4.19].

**Table 4.19:** Means (± standard deviation) of hip and pelvic angles (°) of SG1, SG2, SG3 and the $NAMgroup_{ms}$

| | $NAM_{ms}$ | SG1 | SG2 | SG3 | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|---|
| | | | | | NAM x SG1 | NAM x SG2 | NAM x SG3 |
| **H1** | 16.1(±8.2) | 11.8(±13.1) | 16.1(±17.0) | 11.0(±20.8) | NS | NS | NS |
| **H2** | -3.7(±39.9) | 16.9(±10.4) | 22.2(±9.0) | 35.2(±7.1) | NS | NS | NS |
| **H3** | -15.6(±8.7) | -10.4(±12.6) | 4.0(±12.8) | -5.8(±21.4) | NS | <0.05 | NS |
| **H4** | -5.0(±9.6) | 0.7(±16.5) | 11.7(±13.8) | 2.7(±20.9) | NS | <0.05 | NS |
| **H5** | 18.4(±8.3) | 17.5(±17.2) | 20.4(±17.9) | 13.2(±20.2) | NS | NS | NS |
| **H6** | 33.0(±5.4) | 29.3(±6.1) | 16.4(±12.1) | 20.8(±3.4) | NS | <0.05 | <0.05 |
| **H7** | 7.9(±3.2) | 8.7(±5.5) | 8.3(±4.9) | 7.5(±1.2) | NS | NS | NS |
| **H8** | -2.6(±2.8) | -6.8(±6.0) | -12.5(±4.6) | 2.4(±7.6) | NS | <0.05 | NS |
| **H9** | 5.5(±3.0) | 1.9(±4.7) | -4.2(±4.2) | 5.1(±7.3) | NS | <0.05 | NS |
| **HR2** | -1.1(±2.1) | -7.1(±3.9) | -8.7(±5.7) | -4.7(±8.6) | <0.05 | <0.05 | NS |
| **HR3** | 1.7(±2.4) | 4.7(±4.7) | 5.1(±6.0) | 4.1(±7.8) | <0.05 | <0.05 | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
Grey highlighting signifies that the reaction of the subgroup to the condition is different to the Sgroup when compared to the $NAMgroup_{ms}$
See pages 101-104 for a description of the abbreviations

The SGI had decreased flexion of the knee at heel strike (K1) by 2.8° and increased flexion of the knee at toe-off (K4) by 3.0° in comparison to the affected limb of the Sgroup (Table 4.20). At the ankle joint the SG1 had an increased dorsiflexion angle of the ankle during stance (A3) by 1.4° and an increased total sagittal plane excursion angle of the ankle (A6) by 4.9° in comparison to the affected limb of the Sgroup (Table 4.21). The SG1 had different moments in the lower limb compared to the Sgroup, with an increased hip extensor moment (HM2) by 0.2 Nm/kg, increased first knee extensor moment (KM1) by 0.1 Nm/kg and an increased ankle plantarflexor moment by 0.1 Nm/kg [Table 4.23].

**Table 4.20:** Means (± standard deviation) of knee angles (°) of SG1, SG2, SG3 and the $NAMgroup_{ms}$

| | $NAM_{ms}$ | SG1 | SG2 | SG3 | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|---|
| | | | | | NAM x SG1 | NAM x SG2 | NAM x SG3 |
| **K1** | 0.0(±5.2) | 3.4(±7.8) | 9.6(±5.5) | 0.4(±4.7) | NS | <0.05 | NS |
| **K2** | 4.1(±4.6) | 4.8(±3.0) | 12.5(±3.2) | 4.2(±5.6) | NS | <0.05 | NS |
| **K3** | -3.1(±6.6) | -6.6(±5.7) | 2.3(±4.0) | -10.0(±8.0) | NS | <0.05 | NS |
| **K4** | 31.3(±6.6) | 21.2(±15.7) | 19.3(±12.1) | 6.7(±6.2) | NS | <0.05 | <0.05 |
| **K5** | 43.7(±6.7) | 28.3(±17.7) | 20.5(±11.9) | 11.8(±3.9) | <0.05 | <0.05 | <0.05 |
| **K6** | 45.4(±9.5) | 31.1(±15.5) | 18.1(±11.1) | 23.2(±6.9) | <0.05 | <0.05 | <0.05 |

NS = Not statistically significantly different
<0.05= Statistically significantly different
Grey highlighting signifies that the reaction of the subgroup to the condition is different to the Sgroup when compared to the $NAMgroup_{ms}$
See pages 101-104 for a description of the abbreviations

**Table 4.21:** Means (± standard deviation) of ankle angles (°) of SG1, SG2, SG3 and the $NAMgroup_{ms}$

| | $NAM_{ms}$ | SG1 | SG2 | SG3 | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|---|
| | | | | | NAM x SG1 | NAM x SG2 | NAM x SG3 |
| **A1** | 3.8(±7.8) | -0.2(±4.4) | 3.8(±2.8) | -3.9(±5.5) | NS | NS | NS |
| **A2** | -2.2(±6.9) | -3.5(±3.8) | 3.6(±2.6) | -6.2(±4.3) | NS | NS | NS |
| **A3** | 16.6(±6.9) | 13.7(±3.3) | 11.4(±2.1) | 3.6(±5.3) | NS | NS | <0.05 |
| **A4** | 1.4(±8.4) | 3.2(±4.6) | 5.8(±3.0) | -2.0(±6.2) | NS | NS | NS |
| **A5** | 9.3(±7.2) | 4.3(±5.2) | 5.7(±2.9) | -0.1(±5.4) | NS | NS | <0.05 |
| **A6** | 18.9(±3.0) | 17.2(±3.5) | 8.3(±3.4) | 6.2(±8.4) | NS | <0.05 | <0.05 |
| **A7** | -0.6(±7.6) | -1.8(±2.6) | 3.3(±2.8) | -5.5(±6.7) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
Grey highlighting signifies that the reaction of the subgroup to the condition is different to the Sgroup when compared to the $NAMgroup_{ms}$
See pages 101-104 for a description of the abbreviations

**Table 4.22:** Means (± standard deviation) of foot rotational angles (°) in the coronal plane of SG1, SG2, SG3 and the NAMgroup$_{ms}$

| | NAM$_{ms}$ | SG1 | SG2 | SG3 | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|---|
| | | | | | SG1xSG2 | SG1XSG3 | SG2XSG3 |
| **R1** | -0.1(±3.8) | -5.6(±24.7) | -10.2(±16.4) | -2.2(±11.1) | NS | NS | NS |
| **R2** | -6.6(±9.9) | -17.7(±21.5) | -17.3(±16.6) | -2.0(±9.4) | NS | NS | NS |
| **R3** | -4.7(±5.6) | -1.8(±23.9) | -6.9(±16.1) | 3.6(±5.3) | NS | NS | NS |
| **R4** | -9.9(±4.4) | -10.9(±24.9) | -11.1(±14.6) | -7.7(±8.2) | NS | NS | NS |
| **R5** | -12.3(±3.7) | -15.6(±25.7) | -14.6(±14.7) | -11.4(±7.8) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
Grey highlighting signifies that the reaction of the subgroup to the condition is different to the Sgroup when compared to the NAMgroup$_{ms}$
See pages 101-104 for a description of the abbreviations

**Table 4.23:** Means (± standard deviation) of hip, knee and ankle moments (Nm/kg) of SG1, SG2, SG3 and the NAMgroup$_{ms}$

| | NAM$_{ms}$ | SG1 | SG2 | SG3 | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|---|
| | | | | | NAM x SG1 | NAM x SG2 | NAM x SG3 |
| **HM1** | 0.3(±0.2) | 0.6(±0.4) | 0.3(±0.3) | 0.5(±0.5) | <0.05 | NS | NS |
| **HM2** | -0.3(±0.2) | -0.5(±0.2) | -0.2(±0.1) | -0.2(±0.4) | <0.05 | NS | NS |
| **KM1** | -0.1(±0.1) | -0.3(±0.2) | -0.2(±0.3) | -0.3(±0.3) | <0.05 | NS | NS |
| **KM2** | 0.0(±0.1) | 0.0(±0.3) | -0.1(±0.3) | 0.0(±0.1) | NS | NS | NS |
| **KM3** | -0.4(±0.1) | -0.5(±0.2) | -0.3(±0.3) | -0.5(±0.2) | NS | NS | NS |
| **AM1** | 0.0(±0.0) | 0.0(±0.0) | 0.0(±0.1) | 0.0(±0.0) | NS | <0.05 | <0.05 |
| **AM2** | 1.1(±0.2) | 0.8(±0.3) | 0.6(±0.3) | 0.5(±0.4) | NS | NS | NS |
| **AR1** | 0.0(±0.0) | 0.0(±0.0) | 0.0(±0.0) | 0.0(±0.0) | NS | NS | NS |
| **AR2** | 0.2(±0.1) | 0.1(±0.1) | 0.1(±0.0) | 0.1(±0.0) | <0.05 | <0.05 | <0.05 |

NS = Not statistically significantly different
<0.05= Statistically significantly different
Grey highlighting signifies that the reaction of the subgroup to the condition is different to the Sgroup when compared to the NAMgroup$_{ms}$
See pages 101-104 for a description of the abbreviations

**Subgroup 2**

The significant differences between SG2 and SG1 have already been discussed in the above section. This section will focus on the significant differences between SG2 and SG3. At the hip joint SG2 had a significantly increased maximum adduction angle of the hip in stance (H8) by 10.1 (±3.5)° and a significantly decreased maximum abduction angle of the hip during swing (H9) by 9.3 (±3.1)° in comparison to SG3 (Table 4.13). At the knee joint SG2 flexed their knee during stance (K3) to a significantly greater extent than SG3 by 12.3 (±3.4)° (Table 4.14). At the ankle joint SG2 had a significantly increased dorsiflexion angle of the ankle at heel strike (A1) by 7.7 (±2.5)°, a significantly decreased ankle plantarflexion angle during the loading response (A2) by 9.7 (±2.1)°, a significantly increased maximum dorsiflexion angle of the ankle during stance (A3) by 7.8 (±2.1)°, a significantly decreased plantarflexion angle of the ankle at toe-off (A4) by 7.8 (±2.7)°, and had a significantly decreased plantarflexion angle during swing (A7) than SG3 by 5.1 (±2.0)° (Table 4.15).

When the gait of the SG2 was compared to that of the $NAMgroup_{ms}$ significant differences were noted between the groups for most gait variables (Tables 3.18-3.23). Double support time (DS) was significantly decreased in the SG2 in comparison to that of the $NAMgroup_{ms}$ by 0.7 s (Table 4.18). The percentage at which foot off (FO) occurred was significantly increased in SG2 in comparison to that of the $NAMgroup_{ms}$ by 7.3% (Table 4.18). The limp index (LI) was significantly increased in the SG2 in comparison to that of the $NAMgroup_{ms}$ by 0.3 (Table 4.18). Single limb support time (SS) was significantly decreased in the SG2 in comparison to that of the $NAMgroup_{ms}$ by 0.8 s (Table 4.18). Step length (SL) was significantly decreased in the SG2 in comparison to that of the $NAMgroup_{ms}$ by 0.24 m (Table 4.18). Stride length (SRL) was significantly decreased in SG2 in comparison to that of the $NAMgroup_{ms}$ by 0.5m (Table 4.18). At the pelvis the SG2 had a significantly greater drop of the pelvis in the coronal plane during the stance phase of the gait cycle (HR2) by 7.6° and a significantly greater rise of the pelvis in the coronal plane during the swing phase (HR3) by 6.8° in comparison to that of the $NAMgroup_{ms}$ (Table 4.19). At the hip joint SG2 had significantly decreased extension of the hip during stance (H3) by 19.6° and during toe-off (H4) by 16.7° and its total sagittal plane angular excursion (H6) was decreased by 16.6° in comparison to that of the $NAMgroup_{ms}$ (Table 4.19). In the coronal plane the hip joint in the SG2 had significantly increased maximum hip adduction in stance (H8) by 9.9° and significantly decreased maximum hip abduction in swing (H9) by 9.7° in comparison to that of the $NAMgroup_{ms}$ (Table 4.19). When the knee angles of SG2 were compared to that of the $NAMgroup_{ms}$ the angle of the knee at heel strike (K1) was significantly increased by 9.7°, the knee angle during the loading response (K2) was significantly increased by 8.4°, the maximum knee extension angle during stance (K3) was significantly decreased by 5.4°, the maximum knee flexion angle obtained during the swing phase (K5) was significantly decreased by 23.2° and the total sagittal plane excursion angle of the knee (K6) was significantly decreased by 27.3° in the SG2 in comparison to that of the $NAMgroup_{ms}$ (Table 4.20). When the ankle angles of SG2 were compared to that of the $NAMgroup_{ms}$ the maximum dorsiflexion angle achieved during swing (A5) was significantly decreased by 10.6° in SG2 in comparison to that of the $NAMgroup_{ms}$ (Table 4.21). When the moments of the lower limbs in SG2 were compared to that of the $NAMgroup_{ms}$, the maximum ankle dorsiflexor moment (AM1) was significantly different and the maximum ankle internal rotation moment of the foot (AR2) was significantly decreased by 0.1 Nm/kg in the SG2 in comparison to that of the $NAMgroup_{ms}$ (Table 4.23).

When a comparison is made between the significant differences of the gait of SG2 and that of the $NAMgroup_{ms}$ and the gait of the affected limbs of the Sgroup and that of the $NAMgroup_{ms}$ differences can be noted. SG2 had an increased foot off percentage compared to the Sgroup by 7.3%, a lower step width (SW) by 0.01m and a decreased stride length (SRL) by 0.2 meters [Table 4.18]. At the knee joint SG2 had a greater flexion of the knee during the loading response (K2) by 2.7° and decreased extension of the knee during stance (K3) by 5.5° in comparison to the affected limb of the Sgroup (Table 4.20). At the ankle joint SG2 had decreased dorsiflexion of the ankle during stance (A3) by 0.9° in comparison to the affected limb of the Sgroup (Table 4.21). SG2 had different moments in the lower limb compared to the Sgroup, with a decreased hip flexor moment (HM1) by 0.2 Nm/kg, a increased ankle dorsiflexor moment (AM1) by 0.04 Nm/kg and a decreased ankle plantarflexor moment (AM2) by 0.1 Nm/kg [Table 4.23].

**Subgroup 3**

The significant differences between SG3 and SG1 and SG2 have been previously reported in the earlier two sections. When a comparison is made between the significant differences of the gait of SG3 and that of the $NAMgroup_{ms}$ and the gait of the affected limbs of the Sgroup and that of the $NAMgroup_{ms}$ differences can be noted. SG3 had an increased step width (SW) by 0.1 m and a different stride length (SRL) when compared to the affected limb of the Sgroup (Table 4.18). At the hip joint SG3 had increased extension of the hip during stance (H3) by 1.7° and increased flexion of the hip at toe-off (H4) by 7.6° in comparison to the affected limb of the Sgroup (Table 4.19). The SG3 also had decreased adduction of the hip in stance (H8) by 5.6° and increased abduction of the hip in swing (H9) by 4.8° [Table 4.20]. At the pelvis the SG3 had a decreased drop of the pelvis (HR2) during stance by 2.5° and an increased rise of the pelvis during swing (HR3) by 0.6° in comparison to the affected limb of the Sgroup (Table 4.20). The SG3 had decreased flexion of the knee at heel strike (K1) by 5.8° in comparison to the affected limb of the Sgroup (Table 4.20). At the ankle joint SG3 had decreased dorsiflexion of the ankle during swing (A5) by 4.9° in comparison to the affected limb of the Sgroup (Table 4.21). SG3 had different moments in the lower limb compared to that of the Sgroup, with a different hip flexor moment (HM1), ankle dorsiflexor moment (AM1) and a decreased ankle plantarflexor moment by 0.2 Nm/kg [Table 4.23].

When a comparison is made between the significant differences of the gait of SG3 and that of the $NAMgroup_{ms}$ and the Sgroup affected limbs gait and that of the $NAMgroup_{ms}$ many differences can be noted. SG3 had a greater step width (SW) by 0.03 and a decreased stride length (SRL) by 0.09 meters [Table 4.18]. At the hip joint SG3 had greater extension of the hip during stance (H3) by 1.7° and greater extension of the hip at toe-off (H4) by 2.2° in comparison to the affected limb of the Sgroup (Table 4.19). The SG3 also had increased abduction of the hip in stance (H8) by 10.2° and increased abduction of the hip in swing (H9) by 4.8° [Table 4.19]. SG3 also had a decreased pelvic obliquity by 2.5° in stance and 0.6° in swing (Table 4.19). The SG3 had decreased flexion of the knee at heel strike (K1) by 5.8° in comparison to the affected limb of the Sgroup (Table 4.20). At the ankle joint the SG3 had a decreased dorsiflexion angle during swing by 4.9° in comparison to the affected limb of the Sgroup (Table 4.21). The SG3 had different moments in the lower limb compared to the Sgroup, with a decreased ankle plantarflexor moment by 0.2Nm/kg and a significantly different interaction in ankle dorsiflexor and hip flexor moments with the $NAMgroup_{ms}$ than the Sgroup [Table 4.23].

**4.5 Discussion**

The results of this chapter support the hypotheses of this section of the study; (i) that stroke patients with equinus deformity of the foot have a significantly different gait pattern from that of normal subjects walking at a matched speed, (ii) that the gait pattern of stroke patients with equinus deformity of the foot is not homogenous, (iii) that the gait pattern of each subgroup identified within the stroke group differs from normal subjects and finally (iv) that the gait pattern of normal subjects is affected by walking speed.

The present study is the first to report the kinematic and kinetic parameters of the affected and unaffected limbs of stroke patients with equinus deformity of the foot and to compare the results to normal age matched subjects walking at a comparable walking speed. The results of the current study provide normative data of the kinematics and kinetics of these patients, which can be used in future research to aid in the treatment of this condition. The current work is also the first to provide normative data of the gait pattern of normal age matched healthy subjects walking at a comparable walking speed to stroke patients, while wearing similar footwear.

The values reported for temporal distance variables for the Sgroup in this study, were comparable to previously reported values (Lehmann et al., 1987; Cozean et al., 1988; Hesse et al., 1996; Hesse et al., 1999; Voigt and Sinkjaer, 2000; Gok et al., 2003). When the Sgroup was compared to the $\mathrm{NAMgroup}_{\mathrm{ms}}$, significant differences in temporal distance parameters were noted, suggesting that altered gait biomechanics were present in the stroke patients. Temporal distance parameters on the affected stroke limb unique to stroke patients, which could not be explained solely by reduced walking speed, were a significantly decreased step and stride length, a significantly increased step width, a significantly decreased length of time spent in double support and single limb support and an increased limp index. Two other studies (Lehmann et al., 1987; Chen et al., 2005) have compared the temporal distance parameters of stroke patients to an age and speed matched group, with the patients in the study by Lehamann et al. (1987) also having equinus deformity of the foot. Lehmann et al. (1987) reported no significant decrease in step length when the temporal distance parameters of their group of stroke patients, with equinus deformity of the foot, were compared to a normal age and speed matched group, in contrast with the present study. Chen et al. (2005) reported that when their chronic stroke patients were compared to an age and speed matched group while walking on a treadmill, only the variable of step width was significantly different, which is in agreement with the present study. Characteristics on the unaffected limb unique to stroke patients, which could not be explained solely by reduced walking speed, were a significantly decreased step and stride length and a significantly increased step width. The unaffected limb also had a significantly increased percentage of gait spent in stance, a significantly decreased length of time spent in double and single limb support, a significantly increased step time and a significantly increased limp index. On the unaffected limb, Chen et al. (2005) reported that when their chronic stroke patients walked on a treadmill, only the percentage of the gait cycle spent in stance was significantly increased. When the affected limb of the Sgroup was compared to the unaffected limb, the percentage of foot off, limp index, single limb support time, step time and step length were significantly different between the limbs, thus leading to asymmetries. Lehmann et al. (1987) reported similar findings with a significant difference in step length, stance phase duration and swing phase duration, between the affected and unaffected limbs. The above results would suggest that the Sgroup walk with shorter steps and strides, with a larger base of support and spend a shorter period of time in the stance phase of gait, on the affected limb in comparison to the unaffected limb, and spend a shorter length of time in double and single limb support. The short percentage of the gait cycle spent in stance, combined with the short period of time spent in double and single limb support on the affected leg, would suggest that the affected limb had difficulty accepting weight. The reluctance of the affected limb to accept the weight of the body may be due to a neuromuscular weakness or poor balance.

Many asymmetries were evident in the gait pattern of the stroke patients, with alterations in foot off percentages, single and double limb support time and step length between the affected and unaffected leg. The results of this study also indicate that the stroke patients had difficulty with limb clearance and difficultly maintaining dynamic balance during walking, due to the prolonged single support times on the unaffected limb, reduced step length on the affected limb and an abnormal stance/swing ratio (Sutherland, 1984). When the affected limb of the Sgroup was compared to the $NAMgroup_{ns}$, a greater number of significant differences were found in the temporal distance parameters. The affected limb of the stroke patients demonstrated a significantly lower cadence and foot off percentage, a significantly decreased step and stride time and a significantly decreased walking speed. The unaffected limb of the stroke patients demonstrated a significantly lower cadence and a significantly decreased stride time and walking speed. When either limb of the Sgroup were compared to that of the $NAMgroup_{ns}$ only the variables of double support time and step width were significantly different. When the unaffected limb of the Sgroup was compared to the $NAMgroup_{ns}$ only the variable of single support time was significantly different. Walking speed had a significant effect on many temporal distance variables in the NAMgroup. All temporal distance variables, with the exclusion of limp index, were significantly altered when walking speed was decreased. Lehmann (1987) noted similar findings for temporal distance variables, when walking speed was decreased in normal subjects.

Many angular displacement variables were significantly unique to the Sgroup, in comparison to the $NAMgroup_{ms}$. The present study is the first to report on the pelvic obliquity angle of stroke patients with equinus deformity of the foot, and found that a significantly increased pelvic obliquity angle was present. This may be due to neuromuscular weakness or spasticity of the gluteus medius, or perhaps a selective strategy by the body to improve the balance and stability of the limb. Hip extensors was significantly decreased in stance (by 11.5°) when compared to the $NAMgroup_{ms}$. This finding is in agreement with the study of Lehmann et al. (1987), who also reported a significant decrease in maximum hip extension during midstance (by 14°) when compared to an age and speed matched group. The hip joint also had a significantly decreased angle at toe off and a significantly decreased total excursion angle in the sagittal plane, a significantly increased adduction angle in stance and a significantly decreased abduction angle in swing, when compared to the $NAMgroup_{ms}$. A similar finding was noted by Chen et al. (2005), when their chronic stroke patients' hip angles were compared to a normal age and speed matched group. In contrast to the present study Lehmann et al. (1987) reported a decrease in hip adduction angle during stance (by 3.1°) however, the present study found a significant increase in hip adduction angle (by 5.2°). The presence of spasticity in the hip flexors and adductor muscles of the hip, may be the cause of the significant differences noted in the hip joint angles. Spasticity in theses muscles would prevent the hip from extending and abducting during the gait cycle thus decreasing the excursion of the joint. The results of the present study would also reject Lehmanns' hypothesis that, in stroke patients, a significant decrease in hip adduction was probably associated with a decreased pelvic obliquity and that this decreased pelvic obliquity was as a result of a lack of hip extension during stance. The lack of hip extension during the stance phase therefore, prevents the vertical trunk segment moving forward over the stance foot, and thus prevents the advancement of the opposite leg, resulting in a shorter step been taken. The knee joint had a significantly increased flexion angle at heel strike, a significantly decreased flexion angle at toe off and a significantly decreased total excursion angle in the sagittal plane. A significantly decreased maximum flexion angle during swing by 19.3° was also reported in the present study, which is comparable to the difference reported by Lehmann et al. (1987) of 17°, when the stroke group was compared to an age and speed matched group. A similar finding was noted by Chen et al. (2005) when their chronic stroke patients were compared to a normal age and speed matched group. The alteration in knee joint angles noted in the present study may be due to a decreased neuromuscular capability or spasticity of the quadriceps and hamstring muscles. It may also be a selective strategy by the body to improve balance by decreasing the excursion of the limb. The reduced knee flexion at toe off and during swing in the affected limb, may be indicative of poor propulsion of the limb during pre swing (Chen et al., 2005).

The ankle joint in the stroke patients had a significantly decreased dorsiflexion angle during midstance by 4.3°, which is comparable to the significant decrease of 3° noted by Lehmann et al. (1987), when compared to the $\mathrm{NAMgroup}_{ms}$. The current study was the first to report a significantly decreased total sagittal plane excursion angle of the ankle, when compared to the $\mathrm{NAMgroup}_{ms}$. The cause of the decreased dorsiflexion during midstance may be due to spasticity of the plantarflexor muscles, preventing passive dorsiflexion during midstance. This reduced excursion of dorsiflexion angle in midstance may also be the cause of the decreased total excursion of the ankle in the sagittal plane. The present study was the first to report on the angle of the ankle during the loading response and found the angle to be comparable to the $\mathrm{NAMgroup}_{ms}$. Lehmann et al. (1987) reported that the maximum dorsiflexion angle of the ankle in swing was significantly different from a normal age and speed matched group, however the present study did not find a significant difference. Three previous studies have examined the ankle displacement curves in the sagittal plane in patients with equinus deformity of the foot (Lehmann et al., 1987; Burdett et al., 1988; Gok et al., 2003). Burdett et al. (1988) and Gok et al. (2003) reported that the angle of the ankle at heel strike was -13 (±9)° and -16(±11)°, and the present study reported the angle to be 1.5 (±6.4)°. Both Lehmann's study and the present study reported that the angle of the ankle during midstance was 12.0°, however in contrast Burdett et al. (1988), reported the angle to be -2.0°. At toe off the angle of the ankle has previously been reported to be -5 (±11) ° and 11° (Lehmann et al., 1987; Burdett et al., 1988) whereas in this study the angle was intermediate to the two values [3.7 (±6.5)°]. The present study and the study by Lehmann et al. (1987) reported that the maximum dorsiflexion angle of the ankle during midswing to be 3° and 4.8 (±6.1)°, respectively in contrast Burdett et al. (1988) and Gok et al. (2003) reported the angle to be -6 (±6)° and -12 (±13)° respectively. From examination of the results of the present and previous studies it can be noted that there is a large variability in the pattern of the movement of the ankle during the gait cycle in stroke patients with equinus deformity of the foot. The stroke patients used in the present study are comparable to those of Lehmann et al. (1987). The results of the present and previous studies would suggest that stroke patients with equinus deformity of the foot use many different strategies to compensate for an inability to dorsiflex their foot and thus should be grouped according to the strategies they employ. Grouping of the patients based on gait characteristics will aid the development of specific treatment strategies for these patients.

No study to date has described the angular displacements of the Sgroup's unaffected limb. The unaffected limb, when compared to the $\text{NAMgroup}_{ms}$, demonstrated many significant differences, indicating that the hemiplegia on the affected limb significantly impacts on the movement pattern of the unaffected limb. The differences may be due to deconditioning, or possibly due to a selective compensatory strategy to maintain balance. In the unaffected limb, the pelvis had a significantly decreased pelvic obliquity when compared to the $\text{NAMgroup}_{ms}$. This decrease may be a selective strategy to compensate for the increased pelvic obliquity of the affected limb. In the sagittal plane the hip joint of the unaffected limb had a significantly increased flexion angle at heelstrike and during swing, and a significantly decreased extension angle at toe-off. The alterations in hip angles may be a deliberate strategy by the body to increase step length and maintain balance. In the coronal plane the hip joint of the unaffected limb had a significantly decreased adduction angle in stance and a significantly increased total coronal plane excursion angle when was compared to the $NAMgroup_{ms}$. The alterations in adduction angle and total coronal plane excursion, may be again a selected strategy employed by the body to increase step width, thus improving dynamic balance. At the knee joint, a significantly increased flexion angle at heel strike, during the loading response and at toe-off, and a significantly decreased extension angle of the knee during stance were noted. The increased knee flexion at heel strike and during the loading response, and the decreased knee extension angle during stance, may be a selected strategy employed by the limb to maintain balance due to the decreased step length of the affected limb. The increased knee flexion at toe off in the unaffected limb, may be a strategy to improve the propulsion of the limb into swing and increase step length. Chen et al. (2005) reported that increased limb propulsion is common in the unaffected limb to compensate for weakness or poor balance during single limb support on the affected leg (Chen et al., 2005). At the ankle joint, a significantly increased dorsiflexion angle during midstance, and a significantly decreased plantarflexion angle during the loading response, were noted. The decreased plantarflexion angle during the loading response and the increased dorsiflexion angle during midstance may be a selected strategy by the body to improve balance and to aid in the development of a greater propulsive force to move the limb into swing. When the Sgroups' affected limb was compared to the unaffected limb, many significant differences in angular displacement were noted. The affected limb had a significantly greater pelvic obliquity.

At the hip joint, the affected limb had a significantly decreased flexion angle at heelstrike and during swing, a significantly decreased extension angle at toe-off and a significantly decreased total sagittal plane excursion angle. In the coronal plane, the affected limb had a significantly increased total excursion angle of the hip, a significantly increased abduction angle of the hip in stance and a significantly increased abduction angle of the hip in swing. At the knee joint, the affected limb had a significantly decreased flexion angle during the loading response, at toe-off and during swing, a significantly decreased extension angle during stance, and a significantly decreased total sagittal plane excursion angle. At the ankle joint, the affected limb had a significantly decreased dorsiflexion angle at midstance, toe-off and during swing. In addition, there was a significantly increased plantarflexion angle during the loading response and during swing, and a significantly decreased total sagittal plane excursion. When the affected limb of the Sgroup was compared to the $NAMgroup_{ns}$ a greater number of significant differences in angular displacements were noted. However, when the unaffected limb of the Sgroup was compared to the $NAMgroup_{ns}$ a smaller number of significant differences in angular displacements were noted.

Many angular displacement curves varied significantly when the effect of walking speed was examined in the NAMgroup. At the pelvis the rise during the swing phase was significantly less as walking speed decreased. At the hip joint, the flexion angle at heel strike, the maximum extension angle in the stance phase and the hip flexion angle at toe off, all significantly decreased as walking speed decreased. The maximum hip adduction angle in the stance phase significantly decreased as walking speed decreased. Knee joint flexion at loading response, the maximum flexion angle during swing and the total excursion angle, all significantly decreased as walking speed decreased. Walking speed significantly affected some ankle joint measures. At heel strike, the ankle was significantly less dorsiflexed, while during midstance, toe off and during swing it was significantly more dorsiflexed, as the walking speed decreased. The total excursion of the ankle was also significantly decreased as walking speed decreased. The foot's rotation at heel strike was also significantly decreased as walking speed increased. Lehmann (1987) reported similar findings for the effects of walking speed on normal subjects, except for the variables of maximum adduction of the hip in the stance phase, flexion of the knee at loading response and knee flexion during midstance, which the present study found significantly differed when walking speed decreased. The results from the NAMgroup$_{ns}$ are generally comparable to previously reported gait data (Kadaba et al., 1989; Oberg et al., 1993; Benedetti et al., 1998; Titianova, 2003; Rose and Gamble, 1994; Winter et al., 1987; Whittle, 2000; Judge et al., 1996; Richards et al., 2003; Wade et al. 1987; Lehmann et al., 1987; Oberg et al., 1993; Kressig et al., 2003; Cho et al., 2004). Two variables are however, slightly different than reported previously. The NAMgroup$_{ns}$ had a greater mean hip extension angle (H3) by 3 degrees and greater mean ankle dorsiflexion angle at heelstrike (A1) by 3.9 degrees than previously reported for normal subjects walking barefoot (Kadaba et al., 1989; Oberg et al., 1993; Benedetti et al., 1998; Titianova, 2003; Rose and Gamble, 1994; Winter et al., 1987; Whittle, 2000; Judge et al., 1996; Richards et al., 2003; Wade et al. 1987; Lehmann et al., 1987; Oberg et al., 1993; Kressig et al., 2003; Cho et al., 2004). However, the subjects in this study were wearing trainers and not walking barefoot. Oeffinger et al. (1999) found that in children wearing shoes, ankle dorsiflexion increased significantly by 3° at toe off. Examination of their data also revealed, that by wearing shoes dorsiflexion at heel strike can increase by up to 7°. It is therefore feasible that the differences noted in the above gait parameters are due purely to the wearing of trainers.

Lower limb moments, which where significantly different between the Sgroup and $NAMgroup_{ms}$, were an increased hip flexor moment and a decreased plantarflexor and foot rotation moment. The ankle moments of stroke patients with equinus deformity of the foot have not been previously reported. The results of the present study would suggest that the hip flexors are stronger than normal, perhaps due to spasticity, which may pose a problem to the hip extensor muscles when they try to extend the hip. Lack of hip extension was a noted gait deviation in the Sgroup in the present study. A poor ankle plantarflexor moment in the Sgroup, would suggest that the plantarflexor muscles had difficulty generating maximum force during stance, perhaps due to spasticity or muscle stiffness or due to decreased dorsiflexion angle during midstance. The low foot internal and external rotator moments suggest that stroke patients may have difficulty controlling foot position during the stance phase however kinematically this was not noted. The values found in the present study for the hip flexor moment are higher and for the knee flexor moment lower, than that reported by Kim and Eng (2004) for chronic stroke patients. The ankle plantarflexor moment is of a similar value to that reported by Kim and Eng (2004) and Lamontagne et al. (2002). When the Sgroup was compared to the $NAMgroup_{ns}$, all lower limb moments were significantly lower in the Sgroup. When the lower limb moments of the NAMgroup were compared at different walking speeds, all lower limb moments were significantly decreased by reducing walking speed.

When the gait patterns of the stroke patients were compared to normal age and speed matched subjects, less gait abnormalities are present. This indicates that walking speed has a significant impact on the deviations in gait pattern noted between normal age matched and stroke patients. However, the gait patterns of the stroke group are still abnormal, with 42 out of 89 gait variables being significantly different between the groups. The results of the present work would suggest that future studies that examine the gait patterns of any group with a pathological gait causing a slow walking speed should be compared to a normal age matched group walking at a comparable speed. This form of assessment may provide insights concerning the impairment and related compensatory strategies, which are in addition to the observation of slow walking speed.

The present study was the first to identify that the gait pattern of stroke patients with equinus deformity of the foot was not homogenous, and successfully divided the stroke group into three subgroups based on their gait parameters. The gait variables which could divide the stroke patients into three subgroups were; walking speed, stride and step length, double support time, total excursion of the hip, maximum adduction of the hip in stance, maximum abduction of the hip in swing, maximum extension of the knee in stance, maximum dorsiflexion of the ankle at heel strike, maximum plantarflexion of the ankle during loading response, maximum dorsiflexion of the ankle in stance phase, plantarflexion of the ankle at toe-off, total sagittal excursion of the ankle and the maximum plantarflexion of the ankle in swing. The three gait patterns identified by the cluster analysis in the present study were similar, but not identical to, those documented by Mulroy et al. (2003) and Kramers de Quervain et al. (1996) for stroke patients. Both of these authors described four gait patterns, whereas this study noted only three gait patterns. The SG1 identified in this cluster analysis are similar to the group 1 reported by Mulroy (2003) and the intermediate gait velocity group of Kramers de Quevain et al. (1996). SG1 had the fastest walking velocity ($0.4 \pm 0.2 m.s^{-1}$), the greatest step length ($0.4 \pm 0.0$m), the greatest stride length ($0.7 \pm 0.2$ m) and the shortest double support time ($0.9 \pm 0.4$s). SG1 also had the greatest amount of total sagittal plane excursion of the hip ($29.3 \pm 6.1°$), the greatest amount of dorsiflexion in stance ($13.7 \pm 3.3°$) and the greatest excursion of the ankle ($17.2 \pm 3.5°$). The greater walking speed in SG1, incomparison to the other two subgroups, was probably due to SG1's larger step and stride length, which may have been a result of their greater excursion of the hip and the ability of the ankle to dorsiflex to a greater extent in stance. Overall the gait pattern of this subgroup would suggest that they had an enhanced dynamic balance in comparison to the other two subgroups and were more comfortable to move at a faster pace. When the gait pattern of SG1 was compared to the $NAMgroup_{ms}$, many gait deviations were still present, suggesting that the movement pattern of the hip, knee and ankle and the moments producing them were abnormal.

SG2 were similar to that reported by Mulroy (2003) for his group 3 and the buckling-knee pattern found by Kramers de Quervain et al. (1996). This group demonstrated a slow walking velocity ($0.1 \pm 0.1 m.s^{-1}$), the shortest step length ($0.2 \pm 0.1 m$), the shortest stride length ($0.3 \pm 0.2$ m), the least amount of hip abduction in swing ($-4.2 \pm 4.2°$), the greatest knee flexion during midstance ($2.3 \pm 4.0°$), the greatest dorsiflexion angle at heel strike ($3.8 \pm 2.8°$), the poorest plantarflexion angle during the loading response ($3.6 \pm 2.6°$) and the least amount of plantarflexion during swing ($3.3 \pm 2.8°$). The gait pattern of this subgroup was the most abnormal of the three groups. The slow walking speed in SG2 was probably due to their short step and stride length. The results of the present study would suggest that the affected limb of this subgroup did not hip hitch and they had difficulty controlling their dynamic balance. Higginson et al. (2005) suggested that the cause of the gait deformities noted may be due to inappropriate timing and level of muscle activation. Mulroy (2003) reported that the increased knee flexion in stance for this subgroup was caused by a greater weakness of the hip extensors than that present in the knee extensor muscles, which resulted in prolonged activity of the quadriceps muscles to support the flexed knee posture. The results from Higginson et al. (2005) supports this view. They also reported that a co-activation occurred between the biceps femoris and the quadriceps to aid in the stabilisation of the knee joint. Mulroy (2003) suggested that this patient group would benefit from an ankle foot orthosis with a dorsiflexion stop. When SG2 was compared to the NAMgroup$_{ms}$ this group had the greatest number of gait deviations present of the subgroups.

SG3 have gait pattern characteristics similar to Mulroy's (2003) group 2 and 4 and Kramers de Quervain et al. (1996) extension thrust pattern group. SG3 had the least amount of hip adduction in stance (-2.4±7.6°), the greatest amount of hip abduction in swing (5.1±7.3°), greater knee extension in midstance (-10.0±8.0°), greater amount of plantarflexion at heel strike (-3.9±5.5°) and during loading response (-6.2±4.3°), the least amount of dorsiflexion during midstance (3.6±5.5°), the greatest plantarflexion at toe off (-2.0±6.2°), the greatest plantarflexion during swing (-5.5±6.7°) and the lowest total excursion of the ankle (6.2±8.4°). The walking speed in SG3 was greater than that reported by Mulroy (2003) for their group 4 but similar to their group 2. The gait pattern of SG3 would suggest that they hip hitched in order to aid foot clearance, due to an increased plantarflexion during heel strike and swing. The noted hyperextension of the knee in stance in this group may be the result of a stronger hip extensor than knee extensor moment, and prolonged biceps femoris activity (Mulroy et al., 2003). Mulroy et al. (2003) noted that in their group 4, there was a decrease in adductor longus activity, which they believed resulted in poor hip flexion activation and poor knee flexion in swing. The results from the SG3 in the present study would in part support Mulroys' (2003) theory, as this group had the least amount of hip adduction in stance and poor knee flexion in swing. Mulroy (2003) suggested that an ankle foot orthosis would be appropriate for their group 4, in order to produce neutral ankle dorsiflexion in the swing phase. When the gait pattern of SG3 was compared to the NAMgroup$_{ms}$, many gait deviations were still present suggesting that the movement pattern of the hip, knee and ankle were abnormal and also the moments producing this movement.

The results of the present study provide clinicians with further insights into the gait pattern of stroke patients with equinus deformity of the foot. The present work has demonstrated gait deficits present in the stroke group that are not due simply to walking speed. The addition of the cluster analysis has divided the population with this deformity into significantly different gait profiles to aid in the development of appropriate intervention strategies. The causes of the key differences in the gait pattern noted in the present study need to be addressed in future studies. The gait deviations noted in the affected and unaffected limbs may have many causes. They may be a selected strategy used by the body to improve balance and to move at a quicker pace. They may however be an abnormal and unwanted pattern of movement due to a neuromuscular deficit or due to a decreased strength of the involved muscles. Future research may be able to identify the causes of the abnormal movement patterns during gait in the stroke patients with equinus deformity of the foot, with the use of EMG and by assessing muscle strength. By identifying the cause, an appropriate intervention strategy can be employed by the clinician.

The gait patterns of stroke patients with equinus deformity of the foot were not homogenous. It was possible to use a cluster analysis to successfully and objectively identify three different gait patterns within this group. The gait pattern of each subgroup showed significant differences in comparison to normal age and speed matched subjects. Speed matched comparisons allowed the identification of interrelationships between biomechanical variables and deviations in both the affected and unaffected limbs.

# Chapter 5: The effects of wearing ankle foot orthoses on the gait pattern of stroke patients

## 5.0 Introduction

This chapter examines the effect that wearing an AFO has on the gait biomechanics of the Sgroup, the NAMgroup$_{ms}$ and the Ngroup, with the primary focus on the Sgroup. Some leading names in stroke rehabilitative medicine have reported that wearing of an AFO in stroke patients should be avoided (Bobath, 1990; Davies, 1985). They feel that wearing of an AFO may have a negative impact on their gait; however they provide no quantitative evidence to support this view. This study was undertaken to provide evidence on what effect wear of an AFO may have on the gait pattern of stroke patients with equinus deformity of the foot. This research will examine the effects of wearing an AFO and different leaf forms of AFOs on gait biomechanics, in particular joint angular kinematics and kinetics, as this information is lacking in research to date. The effects of AFO wear on the cluster subgroups identified in Chapter 3 will be examined to identify whether the wear of an AFO is of benefit to all stroke patients with equinus deformity of the foot. The effects of wearing different leaf forms of AFOs in the cluster subgroups will also be examined to establish if the biomechanical characteristics of each AFO affects the gait pattern of each cluster subgroup differently. Such comparisons will be undertaken in an attempt to establish prescription criteria for AFOs and for each leaf form of AFO. In addition, comparisons will be made between the effects of AFO wear on the NAMgroup$_{ms}$ and the Ngroup to establish if different walking speeds altered the biomechanical effects of an AFO.

The following are the specific aims of the experimentation:

- To assess if wearing an ankle foot orthosis returns the gait of the stroke patients with equinus deformity of the foot to a more normal pattern.
- To assess if wearing an ankle foot orthosis has differing effects on the gait pattern of the subgroups of stroke patients with equinus deformity of the foot

— To assess if different leaf forms of ankle foot orthoses have differing effects on the gait pattern of stroke patients with equinus deformity of the foot.

— To assess if the gait pattern while wearing an AFO is altered by walking speed in normal subjects.

The following research hypotheses were formulated prior to experimentation:

— Wearing an ankle foot orthosis will have a significant effect on the gait pattern of stroke patients with equinus deformity of the foot.

— Wearing an ankle foot orthosis will have differing effects on the gait pattern of the subgroups of stroke patients with equinus deformity of the foot.

— Different leaf forms of ankle foot orthoses will have differing effects on the gait pattern of stroke patients with equinus deformity of the foot.

— The gait pattern while wearing an AFO in normal subjects will be altered when walking speed is decreased.

The methods and statistical analysis of this experimental work were previously presented in Chapter 2.

**Results**

In light of the aims of the study, each result section will in the first instance report on the effects of wearing AFOs on the affected limb of the Sgroup followed by the unaffected limb. This will be followed by a description of the effects of wearing AFOs on the gait pattern of the NAMgroup$_{ms}$ and the Ngroup. Only statistically significant effects will be reported in the text in the following sections. All differences are reported in Tables 5.1-5.29. Graphs of the angular displacement curves for kinematics and kinetics data for the Sgroup with the wearing of an AFO can be found in Appendix C.

**5.1 Temporal distance variables**

The means and standard deviations of the temporal distance parameters for the affected and unaffected limb of the Sgroup (n=23), the NAMgroup$_{ms}$ (n=10) and the Ngroup (n=30) with the wearing of AFOs are presented in Table 5.1 to 5.4. In the affected and unaffected limb of the Sgroup, wearing an AFO had a significant effect on some temporal variables (Table 5.1 and 5.2). In the affected limb AFO wear resulted in a significant increase in cadence (C) by 6.4 (±10.6)steps.min$^{-1}$ with the wear of the AAFO and by 5.2 (±8.7)steps.min$^{-1}$ with the wear of the PAFO and a significant decrease in step time (ST) by 0.1 (±0.3)s with the wear of the AAFO and 0.1 (±0.3)s with the wear of the PAFO. Walking speed (WS) significantly increased by 0.09 (±0.14)m.s$^{-1}$ with the wearing of an AAFO and by 0.07 (±0.14)m.s$^{-1}$ with the wearing of a PAFO. Wearing an AAFO resulted in a significant decrease by 2.9 (±5.9)% in the percentage at which foot off (FO) occurred on the affected limb and a significant decrease in stride time (SRT) in the affected limb by 0.2 (±0.4)s. Stride length (SL) increased significantly with the wearing of an AAFO in the affected limb by 0.11 (±0.21)m. Single limb support time was significantly increased by 0.02 (±0.04)s with the wearing of a PAFO. In the unaffected limb AFO wear resulted in a significant increase in cadence by 3.7 (±7.7) to 7.0 (±9.1)steps.min$^{-1}$ and step length by 0.06 (±0.13) and 0.11 (±0.25)m. Wear of either AFO also resulted in a significant decrease in the percentage at which foot off (FO) occurred on the unaffected limb by 2.7 (±3.4) to 3.5 (±5.7)% and a significant decrease in double support time by 0.1 (±0.6)s. Step time and stride time also significantly decreased with the wear of an AAFO in the unaffected limb by 0.1 (±±0.3) and 0.2

(±0.3)s. In the NAMgroup$_{ms}$, wearing an AFO had no significant effect on any temporal variables (Table 5.3). In the Ngroup, wearing an AFO had significant effects on some variables (Table 5.4). Wearing of either AFO resulted in a significant decrease in cadence by 2.0 (±4.0) to 2.5 (±4.2)steps.min$^{-1}$ and walking speed by 0.04 (±0.07) to 0.05 (±0.07)m.s$^{-1}$, a significant increase in step length by 0.31 (±0.52) to 0.34 (±0.47)m and a significant increase in double support time by 0.2 (±0.4) to 0.3 (±0.3)s. Foot off percentage and step time was also significantly increased with the wearing of the AAFO by 0.9 (±2.4)% and 0.2 (±0.3)s and its effects of wear was significantly different to that of the PAFO. Stride time also was significantly increased by 0.03 (±0.05)s with the wearing of a PAFO.

**Table 5.1:** Means (± standard deviation) of temporal distance parameters of the Sgroups affected limb while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO).

| | Sgroup (NAFO) | Sgroup (AAFO) | Sgroup (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOx AAFO | NAFOx PAFO | AAFOx PAFO |
| **C** (step.min$^{-1}$) | 61.8 (±18.7) | 68.2 (±17.3) | 67.0 (±17.9) | <0.05 | <0.05 | NS |
| **DS**(s) | 1.2 (±0.6) | 1.0 (±0.6) | 1.1 (±0.6) | NS | NS | NS |
| **FO**(%) | 69.8 (±8.0) | 66.9 (±8.2) | 68.9 (±8.8) | <0.05 | NS | NS |
| **LI** | 0.8 (±0.1) | 0.9 (±0.1) | 0.8 (±0.1) | NS | NS | NS |
| **SS**(s) | 0.4 (±0.1) | 0.4 (±0.1) | 0.4 (±0.1) | NS | <0.05 | NS |
| **SL**(m) | 0.34 (±0.16) | 0.39 (±0.16) | 0.37 (±0.15) | NS | NS | NS |
| **ST**(s) | 1.3 (±0.5) | 1.2 (±0.5) | 1.1 (±0.4) | <0.05 | <0.05 | NS |
| **SW**(m) | 0.25 (±0.06) | 0.24 (±0.06) | 0.24 (±0.05) | NS | NS | NS |
| **SRL**(m) | 0.54 (±0.29) | 0.66 (±0.28) | 0.63 (±0.28) | <0.05 | NS | NS |
| **SRT**(s) | 2.1 (±0.7) | 1.9 (±0.6) | 2.0 (±0.6) | <0.05 | NS | NS |
| **WS**(m.s$^{-1}$) | 0.31 (±0.25) | 0.40 (±0.23) | 0.38 (±0.24) | <0.05 | <0.05 | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different

C = Cadence
DS = Double support time
FO = Foot off percentage
LI = Limp Index
SS = single limb support time
SL = Step length
ST = Step time
SW = Step width
SRL = Stride length
SRT = Stride time
WS = Walking speed

**Table 5.2:** Means (± standard deviation) of temporal distance parameters of the Sgroups unaffected limb while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO).

| | Sgroup (NAFO) | Sgroup (AAFO) | Sgroup (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOx AAFO | NAFOx PAFO | AAFOx PAFO |
| C (step.min$^{-1}$) | 61.0 (±18.2) | 68.1 (±20.5) | 64.7 (±18.0) | <0.05 | NS | NS |
| DS(s) | 1.1 (±0.5) | 1.0 (±0.6) | 1.0 (±0.6) | <0.05 | <0.05 | NS |
| FO(%) | 81.3 (±6.5) | 78.6 (±7.8) | 77.8 (±9.4) | <0.05 | <0.05 | NS |
| LI | 1.2 (±0.1) | 1.1 (±0.2) | 1.1 (±0.2) | NS | NS | NS |
| SS(s) | 0.6 (±0.2) | 0.6 (±0.2) | 0.6 (±0.2) | NS | NS | NS |
| SL(m) | 0.24 (±0.17) | 0.35 (±0.28) | 0.30 (±0.18) | <0.05 | <0.05 | NS |
| ST(s) | 0.9 (±0.3) | 0.8 (±0.4) | 0.9 (±0.4) | <0.05 | NS | NS |
| SW(m) | 0.24 (±0.05) | 0.24 (±0.06) | 0.24 (±0.06) | NS | NS | NS |
| SRL(m) | 0.54 (±0.29) | 0.69 (±0.38) | 0.63 (±0.28) | NS | NS | NS |
| SRT(s) | 2.1 (±0.6) | 1.9 (±0.7) | 2.0 (±0.7) | <0.05 | NS | NS |
| WS(m.s$^{-1}$) | 0.33 (±0.26) | 0.42 (± 0.27) | 0.38 (±0.24) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 161 for a description of the abbreviations

**Table 5.3:** Means (± standard deviation) of temporal distance parameters of the NAMgroup$_{ms}$ while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO).

| | NAM$_{ms}$ group (NAFO) | NAM$_{ms}$ group (AAFO) | NAM$_{ms}$ group (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOx AAFO | NAFOx PAFO | AAFOx PAFO |
| C (step.min$^{-1}$) | 58.5 (±2.4) | 56.7 (±2.5) | 57.9 (±2.9) | NS | NS | NS |
| DS(s) | 2.2 (±0.3) | 2.2 (±0.8) | 2.2 (±0.6) | NS | NS | NS |
| FO(%) | 69.8 (±3.0) | 70.0 (±4.1) | 69.6 (±2.0) | NS | NS | NS |
| LI | 0.6 (±0.1) | 0.6 (±0.2) | 0.6 (±0.1) | NS | NS | NS |
| SS(s) | 1.2 (±0.1) | 1.2 (±0.3) | 1.2 (±0.2) | NS | NS | NS |
| SL(m) | 0.43 (±0.01) | 0.46 (±0.11) | 0.45 (±0.09) | NS | NS | NS |
| ST(s) | 1.4 (±0.3) | 1.4 (±0.8) | 1.4 (±0.5) | NS | NS | NS |
| SW(m) | 0.19 (±0.03) | 0.20 (±0.03) | 0.20 (±0.02) | NS | NS | NS |
| SRL(m) | 0.96 (±0.18) | 0.96 (±0.30) | 0.94 (±0.25) | NS | NS | NS |
| SRT(s) | 2.1 (±0.1) | 2.1 (±0.1) | 2.1 (±0.1) | NS | NS | NS |
| WS(m.s$^{-1}$) | 0.40 (±0.05) | 0.41 (±0.07) | 0.41 (±0.04) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 161 for a description of the abbreviations

**Table 5.4:** Means (± standard deviation) of temporal distance parameters of the Ngroup while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO).

| | Ngroup (NAFO) | Ngroup (AAFO) | Ngroup (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOx AAFO | NAFOx PAFO | AAFOx PAFO |
| **C**(step.min$^{-1}$) | 114.9 (±8.5) | 113.0 (±9.0) | 112.5 (±9.4) | <0.05 | <0.05 | NS |
| **DS**(s) | 0.4 (±0.2) | 0.7 (±0.3) | 0.6 (±0.3) | <0.05 | <0.05 | NS |
| **FO**(%) | 60.7 (±1.4) | 61.6 (±2.8) | 60.8 (±1.2) | <0.05 | NS | <0.05 |
| **SS**(s) | 0.4 (±0.1) | 0.4 (±0.4) | 0.4 (±0.3) | NS | NS | NS |
| **SL**(m) | 0.97 (±0.34) | 1.31 (±0.50) | 1.28 (±0.48) | <0.05 | <0.05 | NS |
| **ST**(s) | 0.2 (±0.2) | 0.4 (±0.3) | 0.5 (±5.9) | <0.05 | NS | NS |
| **SRL**(m) | 1.56 (±0.19) | 1.54 (±0.14) | 1.55 (±0.15) | NS | NS | NS |
| **SRT**(s) | 1.1 (±0.2) | 1.3 (±0.7) | 1.1 (±0.2) | NS | <0.05 | NS |
| **WS**(m.s$^{-1}$) | 1.51 (±0.18) | 1.46 (±0.15) | 1.46 (±0.16) | <0.05 | <0.05 | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 161 for a description of the abbreviations

## 5.2 Kinematic variables

### Pelvic Obliquity

The means and standard deviations for pelvic obliquity (HR2 and HR3) for the Sgroup's affected and unaffected limb, the NAMgroup$_{ms}$ and the Ngroup with the wearing of AFOs are presented in Table 5.5 to 5.8. In the Sgroup, NAMgroup$_{ms}$ and Ngroup wearing of an AFO had no significant effects on pelvic obliquity.

**Table 5.5:** Means (± standard deviation) of hip angles (°) in the sagittal and coronal planes and pelvic obliquity angles of the Sgroups affected limb while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO).

| | Sgroup (NAFO) | Sgroup (AAFO) | Sgroup (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOx AAFO | NAFOx PAFO | AAFOx PAFO |
| **H1** | 14.1(±15.5) | 14.5(±14.4) | 13.6(±14.6) | NS | NS | NS |
| **H2** | 22.8(±10.3) | 20.8(±-6.3) | 19.2(±10.3) | NS | NS | NS |
| **H3** | -4.1(±15.3) | -5.5(±14.2) | -6.3(±14.3) | NS | <0.05 | NS |
| **H4** | 5.0(±16.1) | 2.1(±14.1) | 1.5(±13.5) | <0.05 | <0.05 | NS |
| **H5** | 18.4(±16.9) | 17.8(±15.3) | 17.1(±14.6) | NS | NS | NS |
| **H6** | 23.4(±10.1) | 23.7(±10.4) | 24.4(±10.4) | NS | NS | NS |
| **H7** | 8.2(±4.4) | 7.9(±2.8) | 7.5(±3.1) | NS | NS | NS |
| **H8** | -7.8(±6.7) | -7.3(±6.3) | -7.6(±6.1) | NS | NS | NS |
| **H9** | 0.3(±6.1) | 0.3(±5.8) | -0.3(±5.8) | NS | NS | NS |
| **HR2** | -7.2(±5.5) | -6.9(±4.7) | -7.0(±5.2) | NS | NS | NS |
| **HR3** | 4.7(±5.5) | 4.6(±4.5) | 4.8(±5.1) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations.

**Table 5.6:** Means (± standard deviation) of hip angles (°) in the sagittal and coronal planes and pelvic obliquity angles of the Sgroups unaffected limb while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO).

| | Sgroup (NAFO) | Sgroup (AAFO) | Sgroup (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOx AAFO | NAFOx PAFO | AAFOx PAFO |
| **H1** | 26.3(±11.4) | 25.1(±12.6) | 26.2(±12.3) | NS | NS | NS |
| **H2** | 37.3(±8.1) | 26.8(±12.9) | 37.4(±10.7) | NS | NS | NS |
| **H3** | -7.9(±11.8) | -12.9(±14.9) | -10.2(±12.6) | NS | NS | NS |
| **H4** | 9.6(±8.9) | 3.4(±14.9) | 6.1(±12.0) | NS | NS | NS |
| **H5** | 28.8(±13.1) | 27.4(±13.1) | 27.7(±12.3) | NS | NS | NS |
| **H6** | 31.2(±15.4) | 31.7(±11.4) | 34.1(±11.0) | NS | NS | NS |
| **H7** | 3.7(±2.8) | 4.4(±2.4) | 3.9(±2.6) | NS | NS | NS |
| **H8** | 7.1(±5.5) | 7.4(±4.9) | 6.2(±5.2) | NS | NS | NS |
| **H9** | 3.4(±6.2) | 2.4(±5.4) | 2.6(±6.5) | NS | NS | NS |
| **HR2** | -7.9(±5.9) | -8.2(±4.5) | -7.2(±6.7) | NS | NS | NS |
| **HR3** | -1.7(±5.1) | -1.4(±4.2) | -1.2(±5.5) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 5.7:** Means (± standard deviation) of hip angles (°) in the sagittal and coronal planes and pelvic obliquity angles of the NAMgroup$_{ms}$ while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO).

| | NAM$_{ms}$ group (NAFO) | NAM$_{ms}$ group (AAFO) | NAM$_{ms}$ group (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOx AAFO | NAFOx PAFO | AAFOx PAFO |
| **H1** | 16.1(±8.2) | 17.5(±9.8) | 18.6(±9.0) | NS | NS | NS |
| **H2** | 28.3(±5.3) | 25.4(±5.4) | 25.3(±7.4) | NS | NS | NS |
| **H3** | -15.6(±8.7) | -16.0(±10.6) | -16.0(±9.4) | NS | NS | NS |
| **H4** | -5.0(±9.6) | -4.9(±8.7) | -5.8(±8.9) | NS | NS | NS |
| **H5** | 18.4(±8.3) | 20.2(±8.9) | 20.1(±8.5) | NS | NS | NS |
| **H6** | 33.0(±5.4) | 36.2(±3.3) | 36.2(±3.6) | <0.05 | <0.05 | NS |
| **H7** | 7.9(±3.2) | 7.9(±3.2) | 8.0(±2.2) | NS | NS | NS |
| **H8** | -2.6(±2.8) | -2.2(±2.2) | -2.4(±1.8) | NS | NS | NS |
| **H9** | 5.5(±3.0) | 5.6(±3.0) | 5.6(±2.6) | NS | NS | NS |
| **HR2** | 1.1(±2.1) | 0.4(±2.1) | 1.3(±3.0) | NS | NS | NS |
| **HR3** | -1.7(±2.4) | -1.8(±2.9) | -2.0(±2.7) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 5.8:** Means (± standard deviation) of hip angles (°) in the sagittal and coronal planes and pelvic obliquity angles of the Ngroup while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO).

| | Ngroup (NAFO) | Ngroup (AAFO) | Ngroup (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOx AAFO | NAFOx PAFO | AAFOx PAFO |
| **H1** | 22.0(±9.2) | 22.2(±8.7) | 23.0(±9.5) | NS | <0.05 | <0.05 |
| **H2** | 22.0(±9.4) | 22.4(±8.5) | 22.9(±9.8) | NS | NS | NS |
| **H3** | -26.0(±10.2) | -26.1(±9.7) | -25.9(±10.0) | NS | NS | NS |
| **H4** | -18.8(±10.5) | -18.3(±9.7) | -18.7(±10.1) | NS | NS | NS |
| **H5** | 21.4(±9.2) | 21.9(±8.0) | 22.4(±9.1) | NS | <0.05 | <0.05 |
| **H6** | 47.4(±4.9) | 48.0(±4.2) | 48.3(±4.2) | NS | NS | NS |
| **H7** | 15.5(±3.9) | 14.6(±3.9) | 14.9(±3.8) | NS | NS | NS |
| **H8** | -5.1(±3.2) | -4.7(±4.2) | -4.9(±3.0) | NS | NS | NS |
| **H9** | 10.5(±5.1) | 9.2(±4.6) | 9.6(±4.8) | NS | NS | NS |
| **HR2** | -5.1(±5.2) | -5.5(±4.8) | -4.8(±5.1) | NS | NS | NS |
| **HR3** | 4.0(±1.5) | 4.2(±1.6) | 3.8(±1.4) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Hip Angle**

The means and standard deviations for hip angles in the sagittal and coronal plane for the Sgroup's affected and unaffected limb, the $NAMgroup_{ms}$ and the Ngroup with the wearing of AFOs are presented in Table 5.5 to 5.8. In the Sgroup on the affected limb, wearing an AFO did not have many significant effects on the sagittal or coronal movements of the hip. In the sagittal plane wearing of either AFO resulted in a greater extension angle of the hip at foot off (H4) by 2.9 (±5.4)° with the wearing of the AAFO and by 3.5 (±5.4)° with the wearing of the PAFO. Wearing of a PAFO resulted in a significantly decreased maximum hip extension angle by 2.2 (±3.9)° during the stance phase (H3). Wearing of an AFO had no significant effect on hip sagittal or coronal plane movements on the unaffected limb in the Sgroup. In the $NAMgroup_{ms}$, the only variable, which significantly increased when either AFO was worn by 3.2 (± 4.8)° and 3.2 (±5.4)°, was the total sagittal plane excursion angle of the hip (H6). In the Ngroup, wearing of a PAFO significantly increased the flexion angle present at the hip during heel strike (H1) by 1.0 (±1.6)° and during the swing phase (H5) by 1.0 (±1.2)°. The angle of the hip at heel strike and during swing could significantly differentiate between the leaf forms of AFO worn in the Ngroup (Table 5.8).

**Knee angles**

The means and standard deviations for knee angles for the Sgroup's affected and unaffected limb, the $NAMgroup_{ms}$ and the Ngroup, with the wearing of AFOs, are presented in Table 5.9 to 5.12. For the affected limb of the Sgroup, wearing of an AAFO resulted in a significantly increased knee flexion angle at heel strike (K1) by 2.7 (±3.1)° and during the loading response (K2) by 3.1 (±3.5)°. The angle of the knee at heel strike could significantly differentiate between the leaf forms of AFO worn in the Sgroup with the difference being 1.6 (±2.9)° (Table 5.9). Wearing of an AFO had no significant effects on the knee joint on the unaffected limb in the Sgroup (Table 5 10) In the NAMgroup$_{ms}$, wearing either AFO resulted in a significant increase in the flexion angle of the knee at heel strike (K1) by 2 3 (±1 7) to 2 6 (±2 8)° and a significant increase in the flexion angle of the knee during midstance (K2) by 4 0 (±5 0)° to 4 5 (±4 6)°(Table 5 11) For the Ngroup wearing of either the AAFO or PAFO resulted in a significant increase in the angle of the knee at foot off (K4) by 0 4 (±1 5)° to 1 6 (±1 3)° and the maximum flexion angle of the knee during swing (K5) by 1 1 (±4 2)° to 1 9 (±4 2)° Wearing of a PAFO resulted in a significant increase in the knee flexion angle at heel strike (K1) by 1 2 (±3 3)° (Table 5 12) The angle of the knee at heel strike could significantly differentiate between the leaf forms of AFO worn in the Ngroup (Table 5 8)

**Table 5 9** Means (± standard deviation) of knee angles (°) in the sagittal plane of the Sgroups affected limb while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO)

| | **Sgroup (NAFO)** | **Sgroup (AAFO)** | **Sgroup (PAFO)** | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOx AAFO | NAFOx PAFO | AAFOx PAFO |
| **K1** | 6 2(±8 8) | 8 9(±8 2) | 7 3(±8 9) | <0 05 | NS | <0 05 |
| **K2** | 11 3(±9 1) | 14 4(±8 6) | 13 9(±9 6) | <0 05 | NS | NS |
| **K3** | -2 7(±9 7) | -1 8(±8 7) | -2 9(±10 0) | NS | NS | NS |
| **K4** | 18 2(±13 5) | 17 7(±12 3) | 17 5(±13 4) | NS | NS | NS |
| **K5** | 23 0(±14 6) | 22 9(±22 1) | 22 1(±13 7) | NS | NS | NS |
| **K6** | 24 5(±13 3) | 23 7(±10 7) | 23 8(±11 9) | NS | NS | NS |

NS = Not statistically significantly different
<0 05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 5.10:** Means (± standard deviation) of knee angles (°) in the sagittal plane of the Sgroups unaffected limb while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO).

| | Sgroup (NAFO) | Sgroup (AAFO) | Sgroup (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOx AAFO | NAFOx PAFO | AAFOx PAFO |
| K1 | 17.4(±10.6) | 13.8(±8.8) | 14.1(±9.4) | NS | NS | NS |
| K2 | 19.7(±8.8) | 16.8(±7.7) | 17.4(±8.0) | NS | NS | NS |
| K3 | 6.7(±6.6) | 5.7(±6.5) | 5.9(±6.2) | NS | NS | NS |
| K4 | 41.3(±10.9) | 39.6(±9.7) | 40.3(±8.8) | NS | NS | NS |
| K5 | 51.2(±13.1) | 50.9(±13.2) | 50.9(±11.2) | NS | NS | NS |
| K6 | 38.6(±16.7) | 37.5(±13.6) | 38.3(±14.5) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 5.11:** Means (± standard deviation) of knee angles (°) in the sagittal plane of the NAMgroup$_{ms}$ while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO).

| | NAM$_{ms}$ group (NAFO) | NAM$_{ms}$ group (AAFO) | NAM$_{ms}$ group (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOx AAFO | NAFOx PAFO | AAFOx PAFO |
| K1 | 0.0(±5.2) | 2.6(±5.8) | 2.3(±4.2) | <0.05 | <0.05 | NS |
| K2 | 4.1(±4.6) | 8.2(±5.3) | 8.6(±2.6) | <0.05 | <0.05 | NS |
| K3 | -3.1(±5.6) | -1.6(±7.2) | -0.8(±4.6) | NS | NS | NS |
| K4 | 31.3(±6.6) | 31.8(±7.3) | 30.8(±7.4) | NS | NS | NS |
| K5 | 43.7(±6.7) | 43.2(±7.3) | 42.2(±7.4) | NS | NS | NS |
| K6 | 45.4(±9.5) | 44.8(±10.9) | 42.9(±10.1) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 5.12:** Means (± standard deviation) of knee angles (°) in the sagittal plane of the Ngroup while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO).

| | Ngroup (NAFO) | Ngroup (AAFO) | Ngroup (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOx AAFO | NAFOx PAFO | AAFOx PAFO |
| **K1** | -1.6(±7.0) | -1.7(±7.0) | -0.3(±7.6) | NS | <0.05 | <0.05 |
| **K2** | 14.2(±9.4) | 13.7(±9.5) | 14.7(±10.3) | NS | NS | NS |
| **K3** | -2.4(±6.0) | -1.9(±5.8) | -2.1(±6.6) | NS | NS | NS |
| **K4** | 26.5(±10.4) | 28.1(±10.4) | 26.9(±11.3) | <0.05 | <0.05 | NS |
| **K5** | 53.8(±13.6) | 55.7(±12.3) | 54.9(±13.3) | <0.05 | <0.05 | NS |
| **K6** | 56.2(±10.1) | 57.6(±9.2) | 57.0(±9.4) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Ankle angles**

The means and standard deviations for ankle angles in the sagittal plane for the Sgroup's affected and unaffected limb, the NAMgroup$_{ms}$ and the Ngroup with the wearing of AFOs are presented in Tables 5.13 to 5.16. The angle of the ankle at toe off (A4) and the maximum dorsiflexion angle of the foot during swing (A5) could significantly differentiate between the leaf forms of AFO worn in the Sgroup (Table 5.13). At toe off wearing of the AAFO placed the ankle in a significantly greater dorsiflexion angle by 2.1 (±2.8)° when compared with wearing the PAFO. Wearing of the AAFO placed the ankle in an increased angle of dorsiflexion during swing (A5) by 2.1 (±3.0)° in comparison to wearing of the PAFO. Neither AFO resulted in a significant difference in the ankle angle at toe-off (A4) or the maximum ankle dorsiflexion angle during swing (A5) [Table 5.13]. For the affected limb of the Sgroup, the maximum plantarflexion angle of the ankle during swing (A7) was significantly decreased by 2.6 (±3.2)° by the wearing of a PAFO. The maximum plantarflexion angle of the ankle during swing (A7) could significantly differentiate between the leaf forms of AFO worn in the Sgroup (Table 5.13). Wearing of the AAFO resulted in a more plantarflexed angle ın comparıson to the wearıng of the PAFO by 1 5 (±3 0)° For the unaffected lımb of the Sgroup, wearıng an AFO had no sıgnıficant effect on the ankle joınt angles (Table 5 14) For the NAMgroup$_{ms}$, the total excursıon angle of the ankle ın the sagıttal plane (A6) was sıgnıficantly decreased by the wearıng of eıther AFO by 2 3 (±3 6)° to 2 9 (±3 1)° (Table 5 15) For the Ngroup, wearıng an AFO sıgnıficantly affected many varıables for the ankle joınt ın the sagıttal plane The angle of the ankle at toe off (A4) was sıgnıficantly ıncreased ınto dorsıflexıon wıth the wearıng of eıther AFO by 7 7 (±5 9)° to 9 0 (±5 6)° The total excursıon angle of the ankle ın the sagıttal plane (A6) was sıgnıficantly decreased by the wearıng of eıther AFO by 10 9 (±5 3)° to 11 8 (±5 6)° ın the Ngroup The maxımum plantarflexıon angle of the ankle durıng swıng (A7) was also sıgnıficantly decreased by the wearıng of eıther AFO by 12 5 (±6 7)° to 13 1 (±6 3)° ın the Ngroup (Table 5 16) Wearıng of a PAFO sıgnıficantly ıncreased the maxımum dorsıflexıon angle of the ankle durıng the stance phase (A3) by 2 2 (±6 7)° The maxımum dorsıflexıon angle of the ankle durıng stance (A3) could sıgnıficantly dıfferentıate between the leaf forms of AFO worn ın the Ngroup (Table 5 13) The total excursıon angle of the ankle ın the sagıttal plane (A6) could sıgnıficantly dıfferentıate between the leaf forms of AFOs worn ın the Ngroup (Table 5 16)

**Table 5.13** Means (± standard deviation) of ankle angles (°) in the sagittal plane of the Sgroups affected limb while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO)

| | Sgroup (NAFO) | Sgroup (AAFO) | Sgroup (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOxAAFO | NAFOxPAFO | AAFOxPAFO |
| **A1** | 1 5(±6 4) | 2 7(±5 6) | 1 6(±5 8) | NS | NS | NS |
| **A2** | -2 0(±5 0) | -1 2(±4 7) | -2 0(±4 7) | NS | NS | NS |
| **A3** | 12 3(±8 5) | 11 6(±6 8) | 11 1(±7 8) | NS | NS | NS |
| **A4** | 3 7(±6 5) | 4 8(±5 5) | 2 8(±6 5) | NS | NS | <0 05 |
| **A5** | 4 8(±6 1) | 5 8(±5 4) | 3 7(±6 4) | NS | NS | <0 05 |
| **A6** | 12 3(±7 0) | 11 8(±5 4) | 12 0(±6 1) | NS | NS | NS |
| **A7** | -0 2(±5 7) | 0 9(±6 0) | 2 4(±5 3) | NS | <0 05 | <0 05 |

NS = Not statistically significantly different
<0 05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 5 14** Means (± standard deviation) of ankle angles (°) in the sagittal plane of the Sgroups unaffected limb while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO)

| | Sgroup (NAFO) | Sgroup (AAFO) | Sgroup (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOxAAFO | NAFOxPAFO | AAFOxPAFO |
| **A1** | 9 5(±7 9) | 8 2(±5 5) | 8 4(±6 2) | NS | NS | NS |
| **A2** | 5 6(±7 4) | 4 1(±6 0) | 4 3(±6 8) | NS | NS | NS |
| **A3** | 22 2(±6 1) | 20 9(±4 8) | 21 3(±5 0) | NS | NS | NS |
| **A4** | 5 4(±8 1) | 3 9(±5 9) | 5 3(±5 7) | NS | NS | NS |
| **A5** | 14 3(±6 6) | 12 6(±5 5) | 13 3(±5 3) | NS | NS | NS |
| **A6** | 15 2(±6 5) | 15 6(±7 2) | 14 5(±6 3) | NS | NS | NS |
| **A7** | 3 9(±7 9) | 1 8(±6 0) | 2 9(±6 4) | NS | NS | NS |

NS = Not statistically significantly different
<0 05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 5.15:** Means (± standard deviation) of ankle angles (°) in the sagittal plane of the $NAMgroup_{ms}$ while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO).

| | $NAM_{ms}$ group (NAFO) | $NAM_{ms}$ group (AAFO) | $NAM_{ms}$ group (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOxAAFO | NAFOxPAFO | AAFOxPAFO |
| **A1** | 3.8(±7.8) | 4.1(±7.2) | 5.3(±8.0) | NS | NS | NS |
| **A2** | -2.2(±6.9) | -0.7(±7.0) | -1.2(±7.7) | NS | NS | NS |
| **A3** | 16.6(±7.0) | 15.4(±7.2) | 15.7(±7.6) | NS | NS | NS |
| **A4** | 1.4(±8.4) | 4.1(±8.3) | 3.5(±8.9) | NS | NS | NS |
| **A5** | 9.3(±7.2) | 8.7(±6.9) | 8.6(±7.6) | NS | NS | NS |
| **A6** | 18.9(±3.0) | 16.0(±2.2) | 16.6(±1.9) | <0.05 | <0.05 | NS |
| **A7** | -0.6(±7.6) | 2.9(±8.6) | 2.4(±8.7) | NS | NS | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 5.16:** Means (± standard deviation) of ankle angles (°) in the sagittal plane of the Ngroup while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO).

| | Ngroup (NAFO) | Ngroup (AAFO) | Ngroup (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOxAAFO | NAFOxPAFO | AAFOxPAFO |
| **A1** | 10.90(±8.6) | 10.7(±6.4) | 11.5(±7.2) | NS | NS | NS |
| **A2** | -0.6(±9.2) | 0.8(±7.7) | -0.2(±6.9) | NS | NS | NS |
| **A3** | 19.2(±10.2) | 19.8(±7.5) | 21.4(±7.9) | NS | <0.05 | <0.05 |
| **A4** | -3.8(±9.0) | 3.9(±7.5) | 5.2(±9.0) | <0.05 | <0.05 | NS |
| **A5** | -0.6(±7.6) | 2.9(±8.6) | 2.4(±8.7) | NS | NS | NS |
| **A6** | 29.7(±8.4) | 17.9(±4.6) | 18.8(±4.9) | <0.05 | <0.05 | <0.05 |
| **A7** | -10.6(±9.7) | 1.9(±7.8) | 2.5(±8.9) | <0.05 | <0.05 | NS |

NS = Not statistically significantly different
<0.05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Foot angles**

The means and standard deviations for foot angles in the transverse plane for the Sgroup's affected and unaffected limb, the $NAMgroup_{ms}$ and the Ngroup with the wearing of AFOs are presented in Tables 5.17 to 5.20. For the affected limb of the

Sgroup, wearing the PAFO placed the foot in a significantly increased external rotation angle at heel strike by 5 0 (±8 2)° (R1) The angle of the foot at heel strike (R1) could significantly differentiate between the different leaf forms of AFOs worn by 4 9 (±9 0)°(Table 5 17) The angle of the foot at toe off (R4) could significantly differentiate between the different leaf forms of AFOs worn (Table 5 17) For the unaffected limb of the Sgroup, wearing of an AFO had no significant effect on foot rotation angles (Table 5 18) For the NAMgroup$_{ms}$, wearing of either AFO significantly decreased the external rotation angle of the foot by 3 4 (±3 9) to 4 6 (±5 1)° during stance (R2) and wearing the AAFO had a similarly significant effect during swing (R5) decreasing the angle by 3 8 (±3 6)° [Table 5 19] For the Ngroup wearing of an AFO had no significant effect on foot rotation (Table 5 20)

**Table 5 17** Means (± standard deviation) of foot angles (°) in the transverse plane of the Sgroups affected limb while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO)

| | Sgroup (NAFO) | Sgroup (AAFO) | Sgroup (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOx AAFO | NAFOx PAFO | AAFOx PAFO |
| **R1** | -7 9(±19 1) | -7 9(±21 7) | -12 9(±19 2) | NS | <0 05 | <0 05 |
| **R2** | -18 0(±18 2) | -15 1(±22 0) | -18 1(±19 8) | NS | NS | NS |
| **R3** | -5 4(±18 9) | -4 7(±20 9) | -8 3(±18 7) | NS | NS | NS |
| **R4** | -12 2(±18 5) | -10 2(±21 5) | -15 5(±19 1) | NS | NS | <0 05 |
| **R5** | -15 8(±18 6) | -13 1(±21 2) | -17 1(±17 7) | NS | NS | NS |

NS = Not statistically significantly different
<0 05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 5.18** Means (± standard deviation) of foot angles (°) in the transverse plane of the Sgroups unaffected limb while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO)

| | Sgroup (NAFO) | Sgroup (AAFO) | Sgroup (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOx AAFO | NAFOx PAFO | AAFOx PAFO |
| **R1** | -4 4(±17 3) | -3 1(±17 9) | -3 8(±17 3) | NS | NS | NS |
| **R2** | -15 9(±14 3) | -15 5(±14 5) | -16 2(±13 7) | NS | NS | NS |
| **R3** | -3 6(±16 2) | -1 0(±17 7) | -2 3(±15 9) | NS | NS | NS |
| **R4** | -11 6(±13 8) | -11 8(±15 9) | -12 3(±13 7) | NS | NS | NS |
| **R5** | -14 1(±14 1) | -14 7(±14 4) | -14 8(±13 9) | NS | NS | NS |

NS = Not statistically significantly different
<0 05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 5 19** Means (± standard deviation) of foot angles (°) in the transverse plane of the NAMgroup$_{ms}$ while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO)

| | NAM$_{ms}$ group (NAFO) | NAM$_{ms}$ group (AAFO) | NAM$_{ms}$ group (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOx AAFO | NAFOx PAFO | AAFOx PAFO |
| **R1** | -0 1(±3 8) | 1 3(±6 5) | 1 1(±6 4) | NS | NS | NS |
| **R2** | -6 6(±9 9) | -3 2(±9 4) | -2 0(±6 6) | <0 05 | <0 05 | NS |
| **R3** | -4 7(±5 6) | -4 0(±7 5) | -2 7(±8 8) | NS | NS | NS |
| **R4** | -9 9(±4 4) | -6 2(±5 7) | -8 1(±5 7) | NS | NS | NS |
| **R5** | -12 3(±3 7) | -8 5(±5 0) | -9 2(±5 8) | <0 05 | NS | NS |

NS = Not statistically significantly different
<0 05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 5 20** Means (± standard deviation) of foot angles (°) in the transverse plane of the Ngroup while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO)

| | **Ngroup (NAFO)** | **Ngroup (AAFO)** | **Ngroup (PAFO)** | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOx AAFO | NAFOx PAFO | AAFOx PAFO |
| **R1** | 2 47(±5 60) | 1 57(±4 40) | 1 72(±4 70) | NS | NS | NS |
| **R2** | -8 19(±12 30) | -7 79(±11 15) | -7 90(±13 10) | NS | NS | NS |
| **R3** | -4 18(±11 50) | -3 78(±10 45) | -3 80(±10 50) | NS | NS | NS |
| **R4** | -6 90(±6 70) | -6 45(±6 30) | -6 96(±5 95) | NS | NS | NS |
| **R5** | -13 60(±6 80) | -12 70(±6 45) | -13 24(±5 90) | NS | NS | NS |

NS = Not statistically significantly different
<0 05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**5 3 Kinetic variables**

The means and standard deviations for lower limb moments in the sagittal plane and transverse planes for the Sgroup's affected limb, the $NAMgroup_{ms}$ and the Ngroup with the wearing of AFOs are presented in Tables 5 21 to 5 23 Wearing of an AFO had few significant effects on lower limb moments Wearing of an AFO had no significant effects on the affected limb of the Sgroup (Table 5 21) For the $NAMgroup_{ms}$ and the Ngroup wearing of an AFO significantly affected only the knee flexor moment (KM2) For the $NAMgroup_{ms}$ wearing of either the AAFO or the PAFO significantly increased the knee flexor moment by 0 02 (±0 11) and 0 05 (±0 05)Nm/kg (Table 5 22) For the Ngroup the wearing of the PAFO significantly increased the knee flexor moment by 0 1 (±0 2)Nm/kg, which was significantly different than the effects of the wearing of an AAFO (Table 5 23)

**Table 5 21** Means (± standard deviation) of the lower limb moments (Nm/kg) in the sagittal and transverse plane of the Sgroups affected limb while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO)

| | Sgroup (NAFO) | Sgroup (AAFO) | Sgroup (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOxAAFO | NAFOxPAFO | AAFOxPAFO |
| **HM1** | 0 5(±0 4) | 0 5(±0 3) | 0 5(±0 3) | NS | NS | NS |
| **HM2** | -0 3(±0 3) | -0 4(±0 2) | -0 4(±0 2) | NS | NS | NS |
| **KM1** | -0 2(±0 2) | -0 2(±0 2) | -0 2(±0 2) | NS | NS | NS |
| **KM2** | 0 0(±0 2) | 0 1(±0 3) | 0 0(±0 2) | NS | NS | NS |
| **KM3** | -0 4(±0 2) | -0 3(±0 2) | -0 3(±0 2) | NS | NS | NS |
| **AM1** | -0 0(±0 0) | -0 1(±0 1) | -0 1(±0 2) | NS | NS | NS |
| **AM2** | 0 7(±0 4) | 0 7(±0 4) | 0 7(±0 3) | NS | NS | NS |
| **AR1** | 0 0(±0 0) | -0 1(±0 3) | 0 0(±0 0) | NS | NS | NS |
| **AR2** | 0 1(±0 1) | 0 3(±0 8) | 0 2(±0 1) | NS | NS | NS |

NS = Not statistically significantly different
<0 05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 5 22** Means (± standard deviation) of the lower limb moments (Nm/kg) in the sagittal and transverse plane of the NAMgroup$_{ms}$ while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO)

| | NAM$_{ms}$ group (NAFO) | NAM$_{ms}$ group (AAFO) | NAM$_{ms}$ group (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOxAAFO | NAFOxPAFO | AAFOxPAFO |
| **HM1** | 0 3(±0 1) | 0 3(±0 2) | 0 3(±0 2) | NS | NS | NS |
| **HM2** | -0 3(±0 2) | -0 2(±0 2) | -0 3(±0 1) | NS | NS | NS |
| **KM1** | -0 1(±0 1) | -0 1(±0 1) | -0 1(±0 1) | NS | NS | NS |
| **KM2** | 0 0(± 0 1) | 0 0(±0 1) | 0 0(±0 1) | <0 05 | <0 05 | NS |
| **KM3** | -0 4(±0 1) | -0 4(±0 2) | -0 4(±0 1) | NS | NS | NS |
| **AM1** | 0 0(±0 0) | 0 0(±0 0) | -0 1(±0 1) | NS | NS | NS |
| **AM2** | 1 1(±0 2) | 1 1(±0 2) | 1 1(±0 2) | NS | NS | NS |
| **AR1** | 0 0(±0 0) | 0 0(±0 0) | 0 0(±0 1) | NS | NS | NS |
| **AR2** | 0 2(±0 1) | 0 2(±0 1) | 0 2(±0 1) | NS | NS | NS |

NS = Not statistically significantly different
<0 05= Statistically significantly different
See pages 101-104 for a description of the abbreviations

**Table 5.23:** Means (± standard deviation) of the lower limb moments (Nm/kg) in the sagittal and transverse plane of the Ngroup while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO).

| | Ngroup (NAFO) | Ngroup (AAFO) | Ngroup (PAFO) | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOxAAFO | NAFOxPAFO | AAFOxPAFO |
| **HM1** | 1.1(±0.3) | 1.1(±0.3) | -0.5(±0.4) | NS | NS | NS |
| **HM2** | -0.4(±0.2) | -0.4(±0.2) | -0.5(±0.4) | NS | NS | NS |
| **KM1** | -0.6(±0.2) | -0.5(±0.2) | -1.0(±0.3) | NS | NS | NS |
| **KM2** | -0.1(±0.3) | 0.0(±0.3) | 0.0(±0.3) | NS | <0.05 | <0.05 |
| **KM3** | -0.7(±0.3) | -0.8(±0.4) | -0.7(±0.4) | NS | NS | NS |
| **AM1** | 0.0(±0.2) | -0.1(±0.1) | -0.6(±0.8) | NS | NS | NS |
| **AM2** | 5.0(±1.7) | 2.0(±0.3) | 2.0(±0.6) | NS | NS | NS |
| **AR1** | -0.1(±0.0) | 0.0(±0.0) | -0.1(±0.1) | NS | NS | NS |
| **AR2** | 0.3(±0.1) | 0.3(±0.1) | 0.3(±0.1) | NS | NS | NS |

NS = Not statistically significantly different

<0.05= Statistically significantly different

See pages 101-104 for a description of the abbreviations

## 5.4 Effects of wearing AFOs on the stroke subgroups identified by cluster analysis

Wearing of an AFO resulted in significantly different effects on gait in each stroke subgroup. The significant results of each subgroup will be compared to the significant effects noted for the Sgroup with AFO wear to aid in understanding where the differences lie between the groups. Lastly the significant interactions between each subgroup on AFO wear will be discussed.

### Effects of wearing an AFO on SG1

The means and standard deviations for SG1s' gait parameters with the wearing of AFOs are presented in Tables 5.24 to 5.29. This subgroup had the greatest number of gait variables significantly affected by wearing of an AFO. Some temporal distance variables were significantly affected by the wearing of an AFO (Table 5.24). Walking speed was significantly increased with wearing either AFO by 0.06 (±0.08) and 0.12

(±0 10)m $s^{-1}$ and cadence was sıgnıficantly ıncreased by 8 7 (±10 6)steps mın $^{1}$ wıth the wearıng of an AAFO (Table 5 24) At the hıp joınt, wearıng the PAFO resulted ın a sıgnıficant decrease ın the coronal plane excursıon angle of the hıp by 1 7 (±1 3)° (Table 5 25) At the knee joınt, wearıng eıther AFO resulted ın a sıgnıficantly ıncreased knee angle at heel strıke (K1) by 2 8 (±4 0)° and 3 5 (±3 3)° and the wearıng of the PAFO sıgnıficantly ıncreased the knee angle durıng the loadıng response (K2) by 6 0 (±2 9)° [Table 5 26] At the ankle joınt wearıng the PAFO resulted ın a sıgnıficantly decreased plantarflexıon angle of the ankle durıng the swıng phase (A7) of the gaıt cycle by 4 0 (±3 1)° [Table 5 27]

In contrast to the findıngs for the Sgroup, ın SG1, wear of an AFO had no sıgnıficant effect on step and strıde tıme, sıngle lımb support percentage or strıde length Also ın contrast to the findıngs for the Sgroup, wear of the AAFO had no sıgnıficant effect on the angle of the hıp at toe-off (H4) [Tables 5 25 and 5 5] and the angle of the knee durıng the loadıng response (K2) [Tables 5 26 and 5 9] At the knee joınt, SG1 sıgnıficantly ıncreased the angle of the knee ınto flexıon at heel strıke (K1) wıth the wearıng of an AAFO, ın contrast to the findıngs of the Sgroup Also ın contrast to the findıngs for the Sgroup, wear of the PAFO had no sıgnıficant effect on the maxımum extensıon angle of the hıp ın stance (H3) [Tables 5 25 and 5 5] and on the angle of the foot at heelstrıke (R1) [Tables 5 17 and 5 28] The total excursıon angle of the hıp ın the coronal plane (H7) was sıgnıficantly decreased and the angle of the knee durıng the loadıng response (K2) was sıgnıficantly ıncreased ınto flexıon ın SG1 wıth the wearıng of a PAFO, ın contrast to the findıngs for the Sgroup (Tables 5 25 and 5 5)

**Table 5 24** Means (± standard deviation) of the temporal distance variables of the stroke subgroups while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO)

| | Orthosis | SG1 | SG2 | SG3 | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|---|
| | | | | | SG1x SG2 | SG1x SG3 | SG2x SG3 |
| **C** | NAFO | 67 5 (±16 2) | 52 6 (±14 1) | 59 2 (±18 0) | NS | NS | NS |
| | PAFO | 72 3 (±15 9) | 61 0 (±16 0) | 59 1 (±16 1) | | | |
| | AAFO | 76 2* (±13 9) | 62 5 (±15 0) | 60 6 (±15 9) | | | |
| **DS** | NAFO | 0 9 (±0 4) | 1 5 (±0 6) | 1 4 (±0 6) | NS | NS | NS |
| | PAFO | 0 7 (±0 2) | 1 3 (±0 7) | 1 4 (±0 7) | | | |
| | AAFO | 0 7 (±0 2) | 1 2 (±0 7) | 1 3 (±0 5) | | | |
| **FO** | NAFO | 65 0 (±5 6) | 84 5 (±5 1) | 69 5 (±7 6) | <0 05 | NS | NS |
| | PAFO | 63 8 (±4 2) | 80 4 (±11 2) | 72 2 (±6 8) | | | |
| | AAFO | 61 3 (±7 2) | 71 0* (±7 9) | 70 7 (±6 4) | | | |
| **LI** | NAFO | 0 8 (±0 1) | 0 9 (±0 1) | 0 9 (±0 1) | NS | NS | NS |
| | PAFO | 0 8 (±0 1) | 0 9 (±0 2) | 0 9 (±0 1) | | | |
| | AAFO | 0 8 (±0 1) | 0 9 (±0 1) | 0 9 (±0 1) | | | |
| **SS** | NAFO | 0 4 (±0 1) | 0 4 (±0 1) | 0 4 (±0 1) | NS | NS | NS |
| | PAFO | 0 4 (±0 1) | 0 4 (±0 1) | 0 4 (±0 1) | | | |
| | AAFO | 0 4 (±0 2) | 0 4 (±0 2) | 0 4 (±0 1) | | | |
| **SL** | NAFO | 0 45 (±0 12) | 0 19 (±0 10) | 0 30 (±0 12) | <0 05 | <0 05 | NS |
| | PAFO | 0 45 (±0 09) | 0 31 (±0 21) | 0 28 (±0 11) | | | |
| | AAFO | 0 50 (±0 13) | 0 30 (±0 11) | 0 30 (±0 13) | | | |
| **ST** | NAFO | 1 1 (±0 3) | 1 3 (±0 5) | 1 5 (±0 6) | NS | NS | NS |
| | PAFO | 1 1 (±0 2) | 1 2 (±0 5) | 1 3 (±0 4) | | | |
| | AAFO | 1 0 (±0 2) | 1 2 (±0 6) | 1 3 (±0 5) | | | |
| **SW** | NAFO | 0 24 (±0 03) | 0 23 (±0 09) | 0 29 (±0 09) | NS | NS | NS |
| | PAFO | 0 23 (±0 03) | 0 24 (±0 08) | 0 26 (±0 07) | | | |
| | AAFO | 0 23 (±0 04) | 0 24 (±0 09) | 0 26 (±0 05) | | | |
| **SRL** | NAFO | 0 73 (±0 17) | 0 27 (±0 19) | 0 45 (±0 13) | <0 05 | <0 05 | NS |
| | PAFO | 0 81 (±0 19) | 0 49 (±0 26) | 0 49 (±0 22) | | | |
| | AAFO | 0 82 (±0 24) | 0 53 (±0 24) | 0 47 (±0 14) | | | |
| **SRT** | NAFO | 1 9 (±0 5) | 2 4 (±0 7) | 2 3 (±0 9) | NS | NS | NS |
| | PAFO | 1 7 (±0 4) | 2 2 (±0 7) | 2 2 (±0 9) | | | |
| | AAFO | 1 6 (±0 3) | 2 1 (±0 7) | 2 2 (±0 8) | | | |
| **WS** | NAFO | 0 42 (±0 16) | 0 13 (±0 11) | 0 23 (±0 13) | <0 05 | <0 05 | NS |
| | PAFO | 0 49∝ (±0 18) | 0 27 (±0 19) | 0 25 (±0 15) | | | |
| | AAFO | 0 54* (±0 16) | 0 29 (±0 17) | 0 25 (±0 13) | | | |

* AAFO significantly different than NAFO
Δ AAFO significantly different than PAFO
∝ PAFO significantly different than NAFO
NS = Not statistically significantly different
<0 05= Statistically significantly different

Grey highlighting signifies that the reaction of the subgroup to the condition is different to the Sgroup
See pages 161 for a description of the abbreviations

**Table 5.25:** Means (± standard deviation) of the hip angles (°) in the sagittal and coronal planes and the pelvic obliquity angles of the stroke subgroups while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO).

| | Orthosis | SG1 | SG2 | SG3 | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|---|
| | | | | | SG1X SG2 | SG1 X SG3 | SG2X SG3 |
| H1 | NAFO | 11.8(±13.1) | 16.1(±17.0) | 11.0(±10.8) | NS | NS | NS |
| | PAFO | 13.4(±14.1) | 15.7(±16.7) | 7.2(±12.2) | | | |
| | AAFO | 14.4(±13.4) | 15.2(±15.9) | 6.7(±17.7) | | | |
| H2 | NAFO | 17.0(± 10.4) | 28.8(±4.6) | 30.2(± 7.1) | NS | NS | NS |
| | PAFO | 17.6(±9.7) | 32.0(±5.2) | 31.2(±9.9) | | | |
| | AAFO | 19.3(±10.7) | 28.0(± 7.8) | 29.4(±9.8) | | | |
| H3 | NAFO | -10.4(±12.6) | 4.0(±12.8) | -5.8(±21.4) | NS | NS | NS |
| | PAFO | -11.5(±11.8) | 2.0(±12.4) | -13.6(±14.0) | | | |
| | AAFO | -12.6(±11.9) | 2.4(±12.2) | -8.8(±19.4) | | | |
| H4 | NAFO | 0.7(±16.5) | 11.7(±13.8) | 2.7(±20.9) | NS | NS | NS |
| | PAFO | -1.9(±13.8) | 8.0(±13.2) | -5.7(±15.7) | | | |
| | AAFO | -3.1(±11.6) | 7.9(±13.1) | -0.2(±18.8) | | | |
| H5 | NAFO | 17.5(±17.2) | 20.4(±17.9) | 13.2(±20.2) | NS | NS | NS |
| | PAFO | 18.6(±16.3) | 17.9(±15.0) | 7.9(±15.0) | | | |
| | AAFO | 18.2(±14.2) | 18.3(±15.0) | 10.2(±17.6) | | | |
| H6 | NAFO | 29.3(±6.1) | 16.4(±12.1) | 20.8(±3.4) | <0.05 | NS | NS |
| | PAFO | 29.7(±7.5) | 16.8(±12.4) | 22.0(±2.8) | | | |
| | AAFO | 30.6(±6.4) | 16.6(±12.1) | 22.5(±6.3) | | | |
| H7 | NAFO | 8.7(±5.5) | 8.3(±4.9) | 7.5(±1.2) | NS | NS | NS |
| | PAFO | 7.0(±4.5)∝ | 7.8(±2.3) | 8.0(±1.3) | | | |
| | AAFO | 8.2(±3.7) | 7.9(±1.7) | 7.9(±2.2) | | | |
| H8 | NAFO | 6.8(±6.0) | 12.5(±4.6) | 2.4(±7.6) | NS | NS | NS |
| | PAFO | 4.9(±7.7) | 11.4(±3.0) | 7.1(±5.4) | | | |
| | AAFO | 5.5(±6.8) | 11.9(±4.0) | 4.6(±6.7) | | | |
| H9 | NAFO | -1.9(±4.7) | 4.2(±4.2) | -5.1(±7.3) | NS | NS | NS |
| | PAFO | 2.1(±6.1) | -3.7(±3.6) | 0.5(±7.1) | | | |
| | AAFO | 2.6(±5.1) | -3.9(±5.1) | 2.9(±5.6) | | | |
| HR2 | NAFO | -7.1(±3.9) | -8.7(±5.7) | -4.7(±8.6) | NS | NS | NS |
| | PAFO | -5.0(±4.7) | -8.6(±5.5) | -8.0(±6.3) | | | |
| | AAFO | -6.5(±4.3) | -8.3(±5.4) | -6.5(±5.8) | | | |
| HR3 | NAFO | 4.7(±4.6) | 5.1(±6.0) | 4.1(±7.8) | NS | NS | NS |
| | PAFO | 3.2(±4.7) | 5.4(±6.0) | 6.5(±5.7) | | | |
| | AAFO | 3.5(±4.1) | 5.3(±4.8) | 5.3(±6.1) | | | |

* AAFO significantly different than NAFO
Δ AAFO significantly different than PAFO
∝ PAFO significantly different than NAFO
NS = Not statistically significantly different
<0.05= Statistically significantly different
Grey highlighting signifies that the reaction of the subgroup to the condition is different to the Sgroup
See pages 101-104 for a description of the abbreviations.

**Table 5 26** Means (± standard deviation) of the knee angles (°) in the sagittal plane of the stroke subgroups while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO)

| | Orthosis | SG1 | SG2 | SG3 | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|---|
| | | | | | SG1xSG2 | SG1xSG3 | SG2xSG3 |
| **K1** | NAFO | 3 4(±7 8) | 9 6(±5 5) | 0 4(±4 7) | NS | NS | NS |
| | PAFO | 6 9(±8 1)∝ | 12 0(±6 9) | 3 7(±3 7)∝ | | | |
| | AAFO | 6 3(±9 3)* | 10 5(±5 7) | 0 4(±4 9) | | | |
| **K2** | NAFO | 4 8(±3 0) | 12 5(±3 2) | 4 2(±5 6) | NS | NS | NS |
| | PAFO | 13 6(±4 6)∝ | 15 6(±7 5) | 1 1(±9 7) | | | |
| | AAFO | 14 5(±8 5) | 14 2(±7 4) | 3 8(±15 0) | | | |
| **K3** | NAFO | -6 6(±5 7) | 2 3(±4 0) | -10 0(±8 0) | <0 05 | NS | <0 05 |
| | PAFO | -6 0(±6 7) | 2 8(±2 9) | -9 3(±4 8) | | | |
| | AAFO | -7 0(±7 3) | 2 7(±3 3) | -10 4(±8 4) | | | |
| **K4** | NAFO | 21 2(±15 7) | 19 3(±12 1) | 6 7(±6 2) | NS | NS | NS |
| | PAFO | 20 3(±13 4) | 20 1(±12 1) | 6 4(±6 0) | | | |
| | AAFO | 20 1(±13 7) | 18 4(±11 8) | 7 2(±7 7) | | | |
| **K5** | NAFO | 28 3(±17 7) | 20 5(±11 9) | 11 8(±3 9) | NS | NS | NS |
| | PAFO | 27 3(±14 0) | 21 5(±11 3) | 11 7(±6 3) | | | |
| | AAFO | 26 4(±16 1) | 20 4(±11 2) | 11 6(±7 4) | | | |
| **K6** | NAFO | 31 1(±15 5) | 18 1(±11 1) | 23 2(±6 9) | NS | NS | NS |
| | PAFO | 29 7(±9 6) | 18 7(±11 5) | 21 7(±2 8) | | | |
| | AAFO | 30 0(±12 3) | 17 8(±9 8) | 22 0(±9 4) | | | |

* AAFO significantly different than NAFO
Δ AAFO significantly different than PAFO
∝ PAFO significantly different than NAFO
NS = Not statistically significantly different
<0 05= Statistically significantly different
Grey highlighting signifies that the reaction of the subgroup to the condition is different to the Sgroup
See pages 101-104 for a description of the abbreviations

**Table 5 27.** Means (± standard deviation) of the ankle angles (°) in the sagittal plane of the stroke subgroups while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO)

| | **Orthosis** | **SG1** | **SG2** | **SG3** | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|---|
| | | | | | SG1xSG2 | SG1xSG3 | SG2xSG3 |
| **A1** | NAFO | -0 2(±4 4) | 3 8(±2 8) | -3 9(±5 5) | NS | NS | <0 05 |
| | PAFO | 1 7(±5 2) | 4 7(±3 1) | -2 3(±4 1) | | | |
| | AAFO | 0 7(±4 7) | 2 8(±3 7) | -2 9(±1 9) | | | |
| **A2** | NAFO | -3 5(±3 8) | 3 6(±2 6) | -6 2(±4 3) | <0 05 | NS | <0 05 |
| | PAFO | -2 1(±4 7) | 2 5(±3 3) | -5 2(±2 8) | | | |
| | AAFO | -2 9(±5 1) | 1 5(±3 7) | -5 4(±1 2) | | | |
| A3 | NAFO | 13 7(±3 3) | 11 4(±2 1) | 3 6(±5 3) | NS | <0 05 | <0 05 |
| | PAFO | 13 4(±4 8) | 11 0(±2 7) | 3 9(±3 9) | | | |
| | AAFO | 13 8(±5 0) | 9 7(±3 9) | 2 2(±2 2) | | | |
| **A4** | NAFO | 3 2(±4 6) | 5 8(±3 0) | -2 0(±6 2) | NS | <0 05 | <0 05 |
| | PAFO | 5 5(±4 4) | 5 5(±4 4) | -2 7(±5 5) | | | |
| | AAFO | 3 7(±6 5) | 3 4(±3 9) | -3 5(±2 5) | | | |
| **A5** | NAFO | 4 3(±5 2) | 5 7(±2 9) | -0 1(±5 4) | NS | NS | <0 05 |
| | PAFO | 6 0(±6 2) | 6 7(±2 5) | -1 3(±5 6) | | | |
| | AAFO | 3 9(±7 0) | 4 7(±2 8) | -2 3(±2 6) | | | |
| A6 | NAFO | 17 2(±3 5) | 8 3(±3 4) | 6 2(±8 4) | <0 05 | <0 05 | NS |
| | PAFO | 16 0(±4 1) | 7 9(±4 2) | 9 9(±4 5) | | | |
| | AAFO | 17 2(±3 2) | 8 1(±3 6) | 6 3(±5 2) | | | |
| A7 | NAFO | -1 8(±2 6) | 3 3(±2 8) | -5 5(±6 7) | NS | <0 05 | <0 05 |
| | PAFO | 2 2(±4 2)∝ | 4 0(±3 8) | -4 2(±5 1) | | | |
| | AAFO | 0 7(±5 5) | 2 5(±3 8) | -5 0(±2 8) | | | |

* AAFO significantly different than NAFO
Δ AAFO significantly different than PAFO
∝ PAFO significantly different than NAFO
NS = Not statistically significantly different
<0 05= Statistically significantly different
Grey highlighting signifies that the reaction of the subgroup to the condition is different to the Sgroup
See pages 101-104 for a description of the abbreviations

**Table 5 28** Means (± standard deviation) of the foot rotation angles (°) in the transverse plane of the stroke subgroups while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO)

| | Orthosis | SG1 | SG2 | SG3 | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|---|
| | | | | | SG1xSG2 | SG1XSG3 | SG2XSG3 |
| R1 | NAFO<br>PAFO<br>AAFO | -5 6(±24 7)<br>-12 0(±24 0)<br>-6 6(±30 1) | -10 2(±16 4)<br>-14 0(±19 2)<br>-7 8(±15 5) | -2 2(±11 1)<br>0 7(±28 0)<br>-4 5(±11 1) | NS | NS | NS |
| R2 | NAFO<br>PAFO<br>AAFO | -17 7(±21 5)<br>-17 8(±24 3)<br>-15 4(±16 4) | -17 3(±16 6)<br>-18 4(±19 5)<br>-15 4(±16 4) | -2 0(±9 4)<br>5 3(±26 3)<br>-1 2(±10 2) | NS | NS | NS |
| R3 | NAFO<br>PAFO<br>AAFO | -1 8(±23 9)<br>-6 2(±23 7)<br>-2 8(±28 9) | -6 9(±16 1)<br>-9 7(±17 3)<br>-4 3(±13 3) | 3 6(±5 3)<br>3 9(±3 9)<br>2 2(±2 2) | NS | NS | NS |
| R4 | NAFO<br>PAFO<br>AAFO | -10 9(±24 9)<br>-14 1(±24 2)<br>-9 3(±30 3) | -11 1(±14 6)<br>-15 6(±18 1)<br>-8 6(±13 2) | -7 7(±8 2)<br>-2 3(±29 7)<br>-6 6(±11 3) | NS | NS | NS |
| R5 | NAFO<br>PAFO<br>AAFO | -15 6(±25 7)<br>-14 9(±21 4)<br>-12 8(±30 4) | -14 6(±14 7)<br>-5 4(±29 2)<br>-11 3(±13 7) | -11 4(±7 8)<br>-5 4(±29 2)<br>-9 7(±9 4) | NS | NS | NS |

* AAFO significantly different than NAFO
Δ AAFO significantly different than PAFO
∝ PAFO significantly different than NAFO
NS = Not statistically significantly different
<0 05= Statistically significantly different
Grey highlighting signifies that the reaction of the subgroup to the condition is different to the Sgroup
See pages 101-104 for a description of the abbreviations

**Table 5 29** Means (± standard deviation) of the lower limb moments in the sagittal and transverse planes of the stroke subgroups while wearing no ankle foot orthosis (NAFO), a posterior leaf ankle foot orthosis (PAFO) and an anterior leaf ankle foot orthosis (AAFO)

| | Orthosis | SG1 | SG2 | SG3 | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|---|
| | | | | | SG1xSG2 | SG1xSG3 | SG2xSG3 |
| **HM1** | NAFO | 0 6(±0 4) | 0 3(±0 3) | 0 5(±0 5) | NS | NS | NS |
| | PAFO | 0 5(±0 2) | 0 6(±0 6) | 0 4(±0 2) | | | |
| | AAFO | 0 5(±0 3) | 0 4(±0 3) | 0 6(±0 4) | | | |
| **HM2** | NAFO | -0 5(±0 2) | -0 2(±0 1) | -0 2(±0 4) | NS | NS | NS |
| | PAFO | -0 4(±0 3) | -0 3(±0 1) | -0 5(±0 1) | | | |
| | AAFO | -0 5(±0 3) | -0 2(±0 1) | -0 4(±0 2) | | | |
| **KM1** | NAFO | -0 3(±0 2) | -0 2(±0 3) | -0 3(±0 3) | NS | NS | NS |
| | PAFO | -0 2(±0 1) | -0 1(±0 1) | -0 3(±0 2) | | | |
| | AAFO | -0 3(±0 3) | -0 1(±0 2) | -0 3(±0 3) | | | |
| **KM2** | NAFO | 0 0(±0 3) | -0 1(±0 3) | 0 0(±0 1) | NS | NS | NS |
| | PAFO | 0 0(±0 3) | 0 1(±0 1) | 0 0(±0 2) | | | |
| | AAFO | 0 1(±0 4) | -0 1(±0 2) | 0 0(±0 1) | | | |
| **KM3** | NAFO | -0 5(±0 2) | -0 3(±0 3) | -0 5(±0 2) | NS | NS | NS |
| | PAFO | -0 4(±0 3) | -0 3(±0 3) | -0 3(±0 2) | | | |
| | AAFO | -0 4(±0 3) | -0 3(±0 3) | -0 4(±0 2) | | | |
| **AM1** | NAFO | -0 04(±0 0) | -0 0(±0 1) | 0 0(±0 0) | NS | NS | NS |
| | PAFO | -0 06(±0 1) | -0 1(±0 1) | 0 0(±0 0) | | | |
| | AAFO | -0 07(±0 1) | -0 1(±0 0) | -0 1(±0 1) | | | |
| **AM2** | NAFO | 0 8(±0 3) | 0 6(±0 3) | 0 5(±0 4) | NS | NS | NS |
| | PAFO | 0 8(±0 2) | 0 7(±0 2) | 0 5(±0 2) | | | |
| | AAFO | 0 8(±0 3) | 0 5(±0 4) | 0 5(±0 3) | | | |
| **AR1** | NAFO | 0 0(±0 0) | 0 0(±0 0) | 0 0(±0 0) | NS | NS | NS |
| | PAFO | 0 0(±0 0) | 0 0(±0 0) | 0 0(±0 0) | | | |
| | AAFO | -0 2(±0 4) | 0 0(±0 0) | 0 0(±0 0) | | | |
| **AR2** | NAFO | 0 1(±0 1) | 0 1(±0 0) | 0 1(±0 0) | NS | NS | NS |
| | PAFO | 0 2(±0 3) | 0 1(±0 0) | 0 1(±0 0) | | | |
| | AAFO | 0 6(±1 3) | 0 1(±0 0) | 0 1(±0 0) | | | |

* AAFO significantly different than NAFO
Δ AAFO significantly different than PAFO
∝ PAFO significantly different than NAFO
NS = Not statistically significantly different
<0 05= Statistically significantly different
Grey highlighting signifies that the reaction of the subgroup to the condition is different to the Sgroup
See pages 101-104 for a description of the abbreviations

**Effects of wearing an AFO on SG2**

The means and standard deviations for SG2s' gait parameters with the wearing of AFOs are presented in Tables 5 24 to 5 29 This subgroup had only one gait parameter significantly altered when an AFO was worn Wearing of an AAFO resulted in a significant decrease in the percentage at which foot off percentage occurred by 13 4 (±8 2)% (Table 5 24)

In contrast, to the findings for the Sgroup, in SG2, wear of an AFO had no significant effect on cadence, step and stride time, single limb support percentage, stride length and walking speed Also wear of the AAFO had no significant effect on the angle of the hip at toe-off (H4) [Tables 5 25 and 5 5] and the angle of the knee during the loading response (K2) [Tables 5 26 and 5 9] The wear of the PAFO also had no significant effect on the maximum extension angle of the hip in stance (H3) [Table 5 25 and 5 5], on the knee flexion angle at heel strike (K1) [Tables 5 26 and 5 9], on the maximum ankle plantarflexion angle during swing (A7) [Tables 5 17 and 5 27] and on the angle of the foot at heelstrike (R1) [Tables 5 17 and 5 28]

**Effects of wearing an AFO on SG3**

The means and standard deviations for SG3s' gait parameters with the wearing of AFOs are presented in Tables 5 24 to 5 29 This subgroup had only one significant alteration in their gait with the wearing of an AFO Wearing of a PAFO resulted in a significant increase in knee flexion at heel strike (K1) by 3 3 (±1 6)° (Table 5 26)

In contrast to the findings for the Sgroup, in SG3 wear of an AFO had no significant effect on cadence, step and stride time, single limb support percentage, stride length and walkıng speed Also wear of the AAFO had no sıgnıficant effect on the angle of the hıp at toe-off (H4) [Tables 5 25 and 5 5] and the angle of the knee durıng the loadıng response (K2) [Tables 5 26 and 5 9] Addıtıonally wear of the PAFO had no sıgnıficant effect on the maxımum extensıon angle of the hıp ın stance (H3) [Tables 5 25 and 5 5], on the maxımum ankle plantarflexıon angle durıng swıng (A7) [Tables 5 17 and 5 27] and on the angle of the foot at heelstrıke (R1) [Tables 5 17 and 5 28]

**Dıfferences ın the effects of wearıng an AFO between the cluster groups**

There were some sıgnıficant dıfferences noted between the subgroups for the effects of wearıng an AFO on temporal dıstance varıables The percentage at whıch foot off occurred was sıgnıficantly dıfferent between SG1 and SG2 wıth the wearıng of an AFO (Table 5 24) Wearıng eıther AFO resulted ın a greater decrease ın foot off percentage ın SG2 ıncomparıson to SG1 The effects of wearıng an AFO on step length were also sıgnıficantly dıfferent between SG1 and SG2 and between SG1 and SG3 (Table 5 24) In SG2, wearıng eıther AFO resulted ın an ıncrease ın step length and ın SG3 ıt resulted ın a decrease ın step length In SG1 wear of an AFO dıd not alter step length whıle strıde length was also sıgnıficantly dıfferent between SG1 and SG3 and between SG1 and SG2 wıth the wearıng of an AFO (Table 5 24) SG2 obtaıned a greater ımprovement ın strıde length compared to SG1, whıle SG3 had a smaller ıncrease ın strıde length when compared to SG1 (Table 5 24) The effects of wearıng AFOs on walkıng speed were also sıgnıficantly dıfferent between subgroups (Table 5 24) SG2 obtaıned a greater ımprovement ın walkıng speed over SG1, and SG3 receıved a smaller ıncrease ın walkıng speed when compared to SG1 (Table 5 24)

At the hip joint the effects of wearing an AFO on the total excursion of the hip was significantly different between SG1 and SG2 (Table 5 25) Wearing of either AFO resulted in an increase in hip total excursion angle in the sagittal plane (H6) in both subgroups, however, the increase in excursion angle was greater in SG1 with the wear of a PAFO (Table 5 25) Wearing of an AFO also had a significantly different effect between SG1 and SG2 and between SG2 and SG3 on the extension angle of the knee during midstance [K3] (Table 5 26) In SG3 and SG1 wearing an AAFO resulted in a greater extension angle of the knee during midstance whereas in SG2 it resulted in an increase in knee flexion angle (Table 5 26) There were many significant differences between how the subgroups reacted to wearing of an AFO at the ankle joint At heel strike at the ankle joint wearing an AFO resulted in significantly different changes between SG2 and SG3 (Table 5 27) In SG2 wearing an AAFO resulted in a decreased dorsiflexion angle of the ankle at heel strike [A1] whereas in SG3 it resulted in an increased dorsiflexion angle The maximum plantarflexion angle during the loading response (A2) with the wearing of an AFO was also significantly different between SG1 and SG2 and between SG2 and SG3 (Table 5 27) In SG1 and SG3 wearing of the AAFO and PAFO resulted in a more dorsiflexed angle during the loading response whereas it had the opposite effect in SG2 resulting in increased plantarflexion angle Wearing an AFO also resulted in significant differences between SG1 and SG3, and between SG2 and SG3 on the maximum dorsiflexion angle achieved during midstance (A3) [Table 5 27] Wearing an AAFO resulted in a decreased dorsiflexion angle in SG3 whereas in SG1 it resulted in an increased dorsiflexion angle Wearing a PAFO had an opposite effect between the subgroups, with SG1 obtaining a decreased dorsiflexion angle and SG3 obtaining an increased dorsiflexion angle with its wear In SG2, wearing the PAFO resulted in a decreased dorsiflexion angle and in SG3 an increased dorsiflexion angle The angle of the foot at toe off (A4) was significantly different between SG1 and SG3 and between SG2 and SG3 with the wearing of an AFO (Table 5 27) In SG1, wearing of either AFO resulted in an increased dorsiflexion angle at toe off whereas in SG2 and SG3 it had the opposite effect resulting in an increased plantarflexion angle In SG2, wearing an AAFO resulted in a greater increased plantarflexion angle at toe off compared to SG3 The angle of the ankle during swing was also affected differently by the wearing of AFOs (Table 5 27) The maximum dorsiflexion angle of the ankle during swing (A5) was significantly different between SG2 and SG3 with the wearing of AFOs In SG2, wearing the PAFO resulted in an increased dorsiflexion angle and in contrast in SG3 the same AFO placed the ankle in an increased plantarflexion angle The maximum plantarflexion angle of the ankle during swing (A7) was also significantly different between SG1 and SG3 and between SG2 and SG3 In SG1, wearing the PAFO placed the ankle in a greater dorsiflexed angle in comparison to SG3 which obtained an increased dorsiflexion angle with the wearing of the same AFO In SG2, wearing an AAFO placed the ankle in an increased plantarflexion angle whereas in SG3 the same AFO placed the ankle in an increased dorsiflexion angle The total excursion angle of the ankle in the sagittal plane (A6) was also significantly different between SG1 and SG2 and between SG2 and SG3 with the wearing of an AFO (Table 5 27) Wearing the PAFO in SG2 resulted in a greater decrease in the total excursion angle in comparison to SG2 In SG3, wearing the PAFO resulted in an increased total excursion of the ankle whereas in SG1 it had the opposite effect

## 5 5 Discussion

The results of this chapter support the acceptance of the hypotheses of this section of the study, (i) that wearing an ankle foot orthosis has a significant effect on the gait pattern of stroke patients with equinus deformity of the foot, (ii) that wearing an ankle foot orthosis has differing effects on the gait pattern of the subgroups of stroke patients with equinus deformity of the foot, (iii) that different leaf forms of ankle foot orthoses (AFOs) have differing effects on the gait pattern of stroke patients with equinus deformity of the foot and (iv) the gait pattern while wearing an AFO in normal subjects was altered when walking speed was decreased

The overall results of the present study would suggest that wearing of an AFO was beneficial to the Sgroup by making the gait pattern more normalized and symmetrical and allowing the patients to walk at a faster pace However, when the Sgroup was divided into subgroups, only one group (SG1) received a beneficial effect from the wearing of an AFO Different leaf forms of AFOs had different effects on the gait pattern of the Sgroup and its subgroups The effects of wearing an AFO on the stroke subgroups will be discussed later

On the affected limb in the Sgroup, wearing of either AFO resulted in a significantly increased cadence and walking speed and a significantly decreased step time The alterations in these global temporal distance parameters were of a magnitude which would suggest that they may be of clinical and statistical significance Perry et al (1995) postulated that a 0 2m $s^{-1}$ difference should be seen in walking speed before it can be classified as clinically significant The increase of 0 1 m $s^{-1}$ noted in the present study would be classified clinically insignificant based on this criterion, however, they correspond to a 23% and 29% improvement in walking speed, which could have a meaningful effect, particularly for a person with a slow gait speed. However the significant increase noted in walking speed was of a low statistical power of 0.22.

Comparable improvements in walking speed have been previously observed (Burdett et al., 1988; Diamond and Ottenbacher; 1990; Gok et al., 2003; Wang et al. 2005). The increase in cadence of the affected limb by 8% (PAFO) and 10% (AAFO) is also comparable to the reported results of other authors (Mojica et al., 1988; Hesse et al., 1996; Gok et al., 2003). Previous research has not examined the effects of wearing an AFO on step time and in the present study step time was significantly decreased by 8%. The improvements in walking speed, cadence and step time would suggest that when the stroke patient walked with an AFO they had improved balance and joint stability on their affected limb, thus allowing them to move at a quicker speed. This is supported by Wang et al. (2005) who found that wearing an AFO significantly improved dynamic standing and balance. The improvement in walking speed was probably as a result of the significantly increased cadence, decreased step time and increased hip extension angle at toe off (H4).

When the anterior AFO was worn, there were additional significant improvements in temporal distance parameters. The affected limb spent a significantly shorter length of time in stance, as the patients possibly had a greater ability to transfer their weight between the two limbs more smoothly, making the percentage of the gait cycle spent in stance more normalised. The present study is the first to report that wearing of an AFO significantly decreases (4%) the length of time the limb spends in stance, although Hesse et al. (1988) reported a similar but not significant change. The differences in the reported results may be due to the different types of AFOs worn in the two studies, with

Hesse et al (1988) wearing a valens calliper and the present study using a plastic AFO Giuliani (1990) suggested that an improvement in the stance to swing ratio with the wearing of an AFO may be indicative of an improved limb clearance However, he also suggested that this should be accompanied by a more symmetrical single support period and an improved plantarflexor moment at toe off These latter changes were not observed in the present study In addition to the AFOs' effects on the percentage of the gait cycle spent in stance, wear of the AAFO resulted in a significantly improved stride length and time The improvements were of a magnitude that would suggest that they were of clinical as well as statistical relevance Stride length increased by 20%, which is similar to the reported results of Mojica et al (1988), Hesse et al (1996) and Tyson and Thornton (2001), and stride time significantly decreased by 10%, which previously has not been reported in the literature The increase in stride length may be due to the significantly increased hip extension at toe off, which would result in an increased excursion of the limb during the stride The decrease in stride time would suggest that the affected limb's ability to weight bear and transfer weight improved, allowing the stride to occur at a quicker pace Wearing of the PAFO only significantly altered one additional variable, single support time The increase in this variable was only in the order of 5%, which may not be clinically relevant However, this increase in single support time would support the hypothesis that with the wear of a PAFO an improved weight bearing capacity and balance occurred on the affected limb, allowing the limb to support the body's weight for a longer period of time

The present study was the second to examine the effects of wearing an AFO on the unaffected limbs' temporal distance parameters In the Sgroup, wearing either AFO resulted in the unaffected limb spending significantly less time (3-4%) in stance, which would suggest that the affected lımb had a greater abılıty to weıght bear Hesse et al (1999) reported sımılar findıngs Addıtıonally, wear of the AFO ın the present study sıgnıficantly ıncreased cadence by 3 and 4% and sıgnıficantly decreased double support tıme by 0 1s, whıch would suggest that the unaffected lımb moved at a quıcker pace Wearıng an AFO also resulted ın a sıgnıficantly ıncreased step length (0 1m), whıch may be due to the sıgnıficantly ıncreased hıp extensıon at toe off ın the affected lımb In addıtıon, wear of an AAFO also resulted ın a sıgnıficant decrease ın step and strıde tıme by 0 1 to 0 2s Prevıous studıes have not examıned ıf wearıng an AFO can sıgnıficantly affect cadence, step length, double support tıme or step and strıde tıme on the unaffected lımb However, the sıgnıficant changes noted ın thıs study are possıbly too mınor to be classıfied as clınıcally sıgnıficant

In the Ngroup, wearıng eıther AFO made the temporal dıstance parameters more abnormal In contrast, ın the NAMgroup$_{ms}$ wearıng an AFO had no effect on temporal dıstance varıables The results of the present study would suggest that walkıng speed sıgnıficantly ımpacts on the effects an AFO has on temporal dıstance varıables Two prevıous studıes have examıned the effects of AFO wear on temporal dıstance parameters ın normal subjects (Opara et al , 1985, Lehmann et al , 1982) Opara et al (1985) reported that wearıng an AFO sıgnıficantly decreased step and strıde length whıch ıs contradıctory to the findıngs of the present study The present study also found a sıgnıficant ıncrease ın step and strıde tıme wıth the wear of an AFO, whıch ıs ın contrast to the reported findıngs of Lehmann et al (1982) The dıfferent bıomechanıcal characterıstıcs of the AFO used between the studıes may account for the changes noted, wıth the present study usıng a posterıor and an anterıor leaf AFO whereas Lehmann et al (1987) used a double-stopped AFO and Opara et al (1985) used a double upright

AFO

In the Sgroup, wearing of an AFO had few significant effects on hip or pelvic movements Wear of either AFO significantly increased the hip extension angle at toe off by 2 9 (±5 4)° and 3 5 (±5 4)° and wearing the PAFO also significantly increased the maximum hip extension angle during stance by 2 2 (±3 9)°[H3] The magnitude of the increased extension angles are very small and therefore may not be viewed as clinically relevant However, the significantly increased hip extension angle during stance possibly resulted in the significantly increased step length noted on the unaffected limb The ability of the hip to increase its extension angle may be due to an improved joint stability and balance thus allowing a greater excursion These alterations in hip angles with the wear of an AFO have not been previously noted (Burdett et al , 1988, Gok et al , 2003)

No study to date has assessed the effects of wearing AFOs on hip coronal plane movements or pelvic obliquity in stroke patients with equinus deformity of the foot The results of the present study demonstrated that wearing an AFO did not significantly alter hip abduction or adduction, or pelvic obliquity

In the Sgroup and the NAMgroup, the effects of wearing an AFO on knee joint angles were similar In the Sgroup and the NAMgroup$_{ms}$, the knee joint angle at heelstrike (K1) and during the loading response (K2) significantly increases into flexion [ranging from 2 3 (±1 7)° to 3 1 (±3 5)°] In the Sgroup, this effect was only observed with the wear of the AAFO, whereas in the NAMgroup$_{ms}$, it was noted when both AFOs were worn It has been suggested by Lehmann et al (1979) that increased knee flexion at heel strike and during the loading response may be due to an abnormal knee flexor moment Lehmann suggested that at heel strike the foot cannot move into plantarflexion due to the resistance of the AFO, so the foot rocks over the posterior portion of the heel until the foot becomes flat This would result in the ground reaction force moving posterior to the ankle creating a flexor moment at the knee (Lehmann et al , 1979) Normally the knee flexor moment will only result in increased knee flexion during this phase of the gait cycle in patients with weak knee extensor muscles (Lehmann et al , 1979) Since there was no significant alteration in the knee moment during this phase of the gait cycle in either the Sgroup or in the $NAMgroup_{ms}$, and due to the fact that the $NAMgroup_{ms}$ do not have weak knee extensor muscles, this potential cause of increased knee flexion may not apply in this case It is therefore unclear the cause of these changes The effects of wearing an AFO on knee angular displacement noted in the present study have not been reported elsewhere (Burdett et al , 1988, Gok et al , 2003)

In the Sgroup, only the biomechanical characteristics of the PAFO had any significant effects on ankle joint kinematics Wearing of the PAFO significantly decreased [2 6 (±3 2)°] the maximum plantarflexion angle during swing This change in angle may have a profound impact on foot clearance The effects that wear of either AFO had on the angle of the ankle at toe off (A4) or its maximum dorsiflexion (A5) and plantarflexion angle (A7) during swing were significantly different in the Sgroup Wear of the AAFO tended to place the stroke patients in a more plantarflexed position during these gait events, whereas the wear of the PAFO tended to place the patients in a more dorsiflexed position The resultant dorsiflexor moment produced by the AFOs during swing may account for the noted differences in the effects of the AFOs at the ankle joint The results would suggest that the resultant dorsiflexor moment produced by the

PAFO at toe-off and during swing may be larger than the AAFO, thus positioning the ankle in greater dorsiflexion. The results of the present study would suggest that the AAFO was not flexible into dorsiflexion and that the PAFO was not flexible into plantarflexion. An increased dorsiflexed position during swing may improve the clearance of the foot. Thirteen millimetres is the normal displacement of the foot from the ground during foot clearance (Winter, 1992). With the wearing of the PAFO the Sgroup would achieve approximately 6mm more foot clearance than when they wear an AAFO. This improved foot clearance may be of great clinical significance, by decreasing the likelihood of toe-catch. The results of the present study on the effects of AFOs on ankle kinematics in stroke patients are different from previous reports. Both Burdett et al. (1988) and Gok et al. (2003) reported that the angle at heel strike and during midswing were significantly altered into greater dorsiflexion with the wear of an AFO. The differences in the effects of wearing an AFO on ankle angles between the present and previous studies may be due to the different types of AFOs worn and also due to the different ankle angular displacement curves (Chapter 3) without the wearing of an AFO between the groups.

The rotational position of the foot was also affected by the wearing of an AFO. No other study to date has reported on the ability of an AFO to control the foot rotation angle during the gait cycle. In the Sgroup, wearing of the PAFO placed the foot in a significantly greater external rotation angle at heel strike (R1). The increased external rotation at heel strike may be due to a variety of reasons. Firstly, it may be due to a lack of neuromuscular control, which may be caused by spasticity or due to an increased level of spasticity in the external rotators of the foot as a result of wearing an AFO. Previous research has not examined the effects of wearing AFOs on the spasticity of the external rotators of the foot however there is anecdotal evidence to suggest the wearing of an AFO may alter the level of spasticity of muscles surrounding the ankle joint. Secondly, it may be a selected strategy by the patient to increase the area of their base of support thus improving their static and dynamic balance. The ability of an AFO to control the amount of rotation present in the foot at heel strike and at toe off could determine whether an AAFO or PAFO was worn. This would suggest that the biomechanical design of the AFO may have a possible impact on the rotational position of the foot during these gait events.

No study to date has assessed the effects of wearing an AFO on the unaffected limbs joint angular displacements. In the Sgroup, wearing an AFO had no significant effect on the unaffected limbs angular displacements.

In the $NAMgroup_{ms.}$ wear of an AFO had some different effects from that of the Ngroup. Different angular displacement variables were affected in the Ngroup in comparison to the $NAMgroup_{ms}$ with the wearing of an AFO, suggesting that walking speed has an impact on the biomechanical performance of an AFO.

Surprisingly, wearing an AFO had few effects on lower limb kinetics. In the Sgroup, wearing of an AFO had no significant effect on any lower limb moments. Lehmann et al. (1987) reported a significant increase in knee flexor moment with the wearing of one type of AFO, which was placed in a 5° dorsiflexion stop, while Gok et al. (2003) reported a significant decrease in this variable with the wearing of an AFO. In the $NAMgroup_{ms}$ wearing either AFO, and in the Ngroup wearing of a PAFO, significantly increased the knee flexor moment (KM2). However it did not result in an increased knee flexion angle at the same point in the gait cycle (K3) The cause of this increased knee flexor moment is unknown

Wearing of an AFO had significantly different effects on each subgroup as identified by the cluster analysis (Chapter 4) Each of these are outlined below In addition, each leaf form of AFO also had significantly different effects on each subgroup This would suggest that when AFOs are being prescribed, the biomechanical design of the orthotic must be appropriately matched to the gait pattern of the stroke patient Functionally SG1 obtained the greatest improvement in gait with the wearing of an AFO With its wear patients had an improved joint stability and balance, allowing them to move at a greater speed The results of this present study would suggest that the prescription of an AFO should be aimed at stroke patients who display a gait pattern similar to SG1 The biomechanical characteristics of the AFOs had differing effects of this subgroup, however both were beneficial to the stroke patients With the wear of a PAFO, SG1 obtained (i) significantly increased walking speed, (ii) significantly decreased hip excursion in the coronal plane, suggesting that less hip hitching was occurring, (iii) significantly increased flexion of the knee during heelstrike and the loading response and, (iv) significantly decreased maximum plantarflexion during the swing The decreased plantarflexion during swing may indicate that an improved foot clearance occurred with AFO wear The wear of an AAFO also resulted in additional beneficial effects, causing a significant increase in cadence and walking speed

In comparison to SG1 the wearing of an AFO in SG2 had a significant effect on only one variable, the percentage of the gait cycle spent in stance Wear of an AAFO resulted in the affected limb spending a reduced percentage of the gait cycle in stance This may be a negative effect as it would suggest that with the wear of an AAFO that patients were more reluctant to weight bear on the affected limb and thus transferred their weight earlier onto the unaffected limb Therefore wearing of an anterior leaf AFO resulted in an undesirable effect on the gait pattern of this subgroup Mulroy (2003) suggested that two of their four gait patterns (Group 3 and Group 4) as identified by cluster analysis in stroke patients would benefit from the wearing an AFO However, she provided no experimental evidence to support this statement Their Group 3 gait pattern was similar to that of SG2 in the present study (Chapter 3) The results of the present study would reject the hypothesis that Group 3 would benefit from the wear of an AFO

SG3 also did not obtain any functionally beneficial effects from the wearing of an AFO Mulroy (2003) had also suggested that their stroke patients (Group 4) with a gait pattern similar to that of SG3 would benefit from wearing an AFO as it would hold the ankle at 90° during swing The results of the present study would reject this hypothesis, as wear of an AFO did not significantly alter ankle displacement in SG3 In SG3, only one gait variable was significantly altered with the wearing of an AFO Wearing of a PAFO resulted in greater flexion of the knee at heel strike

The present study provides scientific evidence to physiotherapists and stroke rehabilitators that the wearing of an AFO can be beneficial to some patients with stroke If a clinician wishes to assess if wearing an AFO is of benefit to their patients it is suggested that they assess temporal distance variables such as walking speed, step and stride length and the percentage of the gait cycle spent in stance A beneficial effect will be indicated by a faster walking speed and longer step and stride lengths

Accompanying this will be a reduced percentage of the gait cycle spent in stance and an improved symmetry in step and stride lengths  Only stroke patients with a gait pattern similar to that of SG1 obtained a beneficial improvement in gait with the wearing of an AFO  SG1 had the most normalised gait pattern of the subgroups without wear of an AFO, therefore the results of the present study would suggest that stroke patients with a more abnormal gait pattern may not obtain any benefit from the wearing of an AFO  Clinicians should attempt to identify if a gait pattern similar to SG1 is present in their stroke patients upon initial assessment for suitability of AFO wear  Initial examination should include the temporal distance parameters discussed above and also assessment of the dorsiflexion angle during stance and the total sagittal plane excursion of the hip and ankle  The patients who will benefit from the wearing of an AFO will have characteristics similar to SG1 which were, a fast walking speed (0 4±0 2m $s^{-1}$), a relatively long stride length (0 73±0 17m), a relatively long step length (0 45±0 12m), a more normalised stance percentage (65 0±5 6%), a large total sagittal plane excursion of the hip (29 3±6 1°), a large dorsiflexion angle in stance (13 7±3 3°) and a large total excursion angle of the ankle (17 2±3 5°)

It is recognised by the author that there were a number of limitations to the present study  The number of stroke patients with equinus deformity of the foot which took part in the present study was small  This number was again decreased when the patients were divided into their subgroups  However, this clinical trial was one of the largest to date  The results of the present study are also limited in that they assessed only two different leaf forms of AFOs  It was not feasible to study all AFOs on the market  The two AFOs examined in the present study were chosen as they had frequently been used clinically by the author  Another limitation in the present study was that the AFOs used were new Over time the flexibility of an AFO may vary with wear This may alter the effects that the AFO may have on the gait pattern

There are a number of further avenues that could be pursued in this research area Future studies may wish to directly assess if wear of different leaf forms of AFOs have a significant effect on toe clearance The results of the present study would suggest that they do, due to the altered plantarflexion angle during swing with the wear of only the PAFO The cause of the significantly increased external rotation angles of the foot at heel strike with AFO wear is another area of possible research It is unclear if it is a selected strategy by the stroke patient, an unwanted strategy due to a lack of neuromuscular control (due to spasticity or increased spasticity with the wearing of an AFO) of the muscles around the ankle joint, or due to the effect of wearing an AFO Further research may be able to establish which of these factors are responsible by examining the EMG pattern of the muscles around the ankle joint, with and without the wear of AFOs and also by directly assessing the strength of the ankle joint muscles Assessment of the forward and lateral trajectory of the centre of pressure at heelstrike (or during perturbed standing) may also help to establish if the cause of this alteration in foot angle is a selected strategy to improve balance The results of the present study have suggested that walking speed had a significant influence on the biomechanical effects of AFOs with a speed of approximately 0 4m s$^{-1}$ producing the best biomechanical performance of the AFOs Further research could examine if the biomechanical design of the AFO can be altered to generate an improved performance at slower speeds, which are frequently employed

In conclusion, different leaf forms of AFOs had differing effects. These effects were dependent upon the gait pattern of the stroke patients. Only SG1 had significantly improved gait with the wearing of an AFO. SG1 was the subgroup with the most normalised gait pattern prior to wear of an AFO. The results of the present study would therefore suggest that the wearing of an AFO may not be beneficial to stroke patients who demonstrate a more abnormal gait pattern. Clinicians, when assessing stroke patients for suitability of AFO wear should try to identify if the gait pattern of their patients are similar to that of SG1. Finally, walking speed had a significant effect on the biomechanical performance of an AFO.

# Chapter 6: The effects of wearing ankle foot orthoses on oxygen uptake, energy cost and COM displacement during gait

## 6.0 Introduction

This chapter focuses on the effects of wearing an AFO on energy expenditure during gait in the Ngroup and the Sgroup. In previous studies energy expenditure during gait has been indirectly assessed by measuring the oxygen uptake during walking and also by calculating the energy cost of gait [dividing the oxygen uptake by the walking speed] (Danielsson and Sunnerhagen, 2004). Energy expenditure during gait is believed to be high in stroke patients and it would be advantageous to decrease their energy expenditure through improved gait economy. A reduction in energy expenditure would allow them to walk for a longer period of time without becoming fatigued. A number of studies have examined the effects of wearing an AFO on energy expenditure in patients with stroke and they have found contrasting results (Corcoran et al., 1970; Franceschini et al., 2003; Danielsson and Sunnerhagen, 2004). These contrasting findings may in part be due to the lack of control of walking speed employed in certain studies and also due to the length of time by which oxygen uptake was measured. The energy cost of walking in stroke patients depends on the speed (Zamparo et al. 1995). Unless walking speed is controlled across the conditions of wearing and not wearing an AFO, the true alteration in energy expenditure with the wear of an AFO cannot be determined (Franceschini et al., 2003). The present study controlled walking speed across conditions. Previous studies have collected oxygen over various different time periods normally at the commencement of walking without allowing for a steady state to be achieved. A steady state reflects a balance between energy required by the working muscles and ATP production in aerobic metabolism (Katch et al., 2001). During exercise, subjects should have steady state oxygen consumption for the reliable determination of energy expenditure (Danielsson and Sunnerhagen, 2000). The present study allowed all subjects to achieve a steady state condition prior to oxygen uptake being measured

If energy expenditure is reduced with wear of an AFO, assessment of energy expenditure may be a key criterion employed by physiotherapists to select the most appropriate AFO Assessment of energy expenditure during gait in stroke patients is normally examined by a relatively invasive method which involves the stroke patient wearing a facemask An alterative and a more user friendly method however, may be the measurement of the vertical displacement of the COM, which may be assessed by whole body kinematics or by examining the displacement of a sacral marker The second method would be easily assessed in a physiotherapy setting The vertical displacement of the COM is related to the potential energy changes that occur in the body during walking (Tesio et al , 1985) In normal walking, the vertical displacement of the COM is minimized thereby requiring the least energy input however, in patients with gait disorders it is much higher than normal (Saunders et al , 1953, Johnson, 1977, Kerrigan et al , 1995) Consequently, the vertical COM displacement has been proposed as a method to assess the energy requirement of walking Kerrigan et al (1995) found a significant relationship between oxygen uptake and the vertical displacement of the COM by sacral marker method Kerrigan's study however was undertaken on healthy individuals The present study also examined whether a correlation exists between oxygen uptake and measures of the displacement of the COM in the Ngroup and Sgroup This comparison was undertaken to examine the viability of the displacement of the COM as an alternative method to assessing oxygen uptake, as direct measurement of oxygen uptake in these stroke patients can be invasive and distressing

The following are the specific aims of the experimentation

1. To investigate the effects of walking with an AFO on the energy expenditure of gait in stroke patients with equinus deformity of the foot.
2. To investigate if the measurement of the COM, assessed using whole body kinematics and sacral displacement, is a viable method of determining the energy expenditure of gait in stroke patients with equinus deformity of the foot.

The following are the specific hypotheses of the experimentation

1. Wearing of an AFO will decrease the energy expenditure of gait in stroke patients with equinus deformity of the foot.
2. Measurement of the COM, assessed using whole body kinematics and sacral displacement, is a viable method to determine the energy expenditure of gait in stroke patients with equinus deformity of the foot.

The methods and statistical analysis of this experimental work were previously presented in Chapter 3. In summary, six stroke patients [age: 53.8(±12.3) years] and thirty normal subjects [age: 41.4(±16.9) years] took part in the present section of this study. Subjects walked on a treadmill at a speed of $1km.hr^{-1}$(Sgroup) and $3.5km.hr^{-1}$(Ngroup) with walking speed being controlled across the test conditions. Subjects were randomly assessed under the conditions of wearing no AFO, wearing a PAFO and wearing an AAFO. Energy expenditure was indirectly assessed through the measures of oxygen uptake and the energy cost of gait (oxygen uptake/walking speed). COM vertical displacement was measured using two methods: i) whole body kinematics and ii) sacral displacement.

Energy expenditure is normally assessed by measuring the volume of $O_2$ in litres per minute multiplied by the kilo calories used per litre of $O_2$ (Wilmore and Costill, 2004). Energy expenditure can be converted into an energy cost of an activity by determining the average oxygen uptake per unit of time and then calculating the kilocalories of energy used per minute during the activity (Wilmore and Costill, 2004). In the present experiment and in previous experiential work in this area (Corcoran et al., 1970; Franceschini et al., 2003; Danielsson and Sunnerhagen, 2004) energy expenditure has been assessed indirectly by examining the average oxygen uptake during gait. This oxygen uptake can be converted into an energy cost by dividing the oxygen uptake by walking speed (Danielsson and Sunnerhagen, 2004).

The effect of an AFO on oxygen uptake, the energy cost of gait and COM displacement were assessed through an ANOVA. The relationship between COM measures (whole body kinematics and sacral displacement) and oxygen uptake were assessed using a correlation analysis.

**Results**

**6.1 Oxygen uptake and energy cost of gait**

As expected oxygen uptake was significantly higher and energy cost (oxygen uptake/walking speed) was significantly lower in the Ngroup in comparison to the Sgroup when they walked without an AFO at their customary walking speed ($p<0.05$). No significant changes were found in either absolute oxygen uptake or energy cost in either the Ngroup or the Sgroup when an AFO was worn ($p>0.05$) [Tables 6.1 & 6.2]. Therefore wearing an AFO did not significantly affect the energy expenditure of stroke patients with equinus deformity of the foot as measured by either of these variables.

Examination of each of the Sgroup patients on an individual level indicates that some patients reacted differently to the wearing of an AFO (Figures 6.1 and 6.2). Patient 6 experienced a decrease in oxygen uptake by 5 $ml.kg^{-1}.min^{-1}$ and a decreased energy cost of gait by 0.3 $ml.kg^{-1}.m^{-1}$, whereas Patient 2, experienced increased oxygen uptake by 1.5 $ml.kg^{-1}.min^{-1}$ and an increased energy cost of gait by 0.1 $ml.kg^{-1}.m^{-1}$

**Table 6.1:** Ngroup changes in absolute $O_2$ uptake, energy cost of gait and sacral and center of mass displacement

| | NAFO | PAFO | AAFO | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOx PAFO | NAFOx AAFO | AAFOx PAFO |
| $0_2$ absolute change ($ml.kg^{-1}.min^{-1}$) | 5.4(±1.5) | 5.7(±1.5) | 5.5(±1.7) | NS | NS | NS |
| Energy cost ($ml.kg^{-1}.m^{-1}$) | 0.14(±0.02) | 0.15(±0.02) | 0.14(±0.03) | NS | NS | NS |
| COM - Sacral displacement method (mm) | 59.5(±13.3) | 62.2(±12.3) | 60.5(±15.4) | NS | NS | NS |
| COM – Segmental methods (mm) | 56.3(±11.1) | 57.8(±10.8) | 57.8(±12.2) | NS | NS | NS |

NS = Not significantly different
<0.05 = Significantly different

**Table 6.2:** Sgroup changes in absolute $O_2$ uptake, energy cost of gait and sacral and center of mass displacement

| | NAFO | PAFO | AAFO | Statistical Analysis (p) | | |
|---|---|---|---|---|---|---|
| | | | | NAFOx PAFO | NAFOx AAFO | AAFOx PAFO |
| $0_2$ absolute change ($ml.kg^{-1}.min^{-1}$) | 4.9(±2.6) | 4.1(±1.7) | 4.3(±1.4) | NS | NS | NS |
| Energy cost ($ml.kg^{-1}.m^{-1}$) | 0.4(±0.1) | 0.4(±0.1) | 0.4(±0.1) | NS | NS | NS |
| COM - Sacral displacement method (mm) | 66.6(±40.9) | 54.0(±23.1) | 55.1(±21.0) | NS | NS | NS |
| COM – Segmental methods(mm) | 68.0(±39.3) | 48.4(±29.6) | 42.6(±25.7) | NS | NS | NS |

NS = Not significantly different
<0.05 = Significantly different

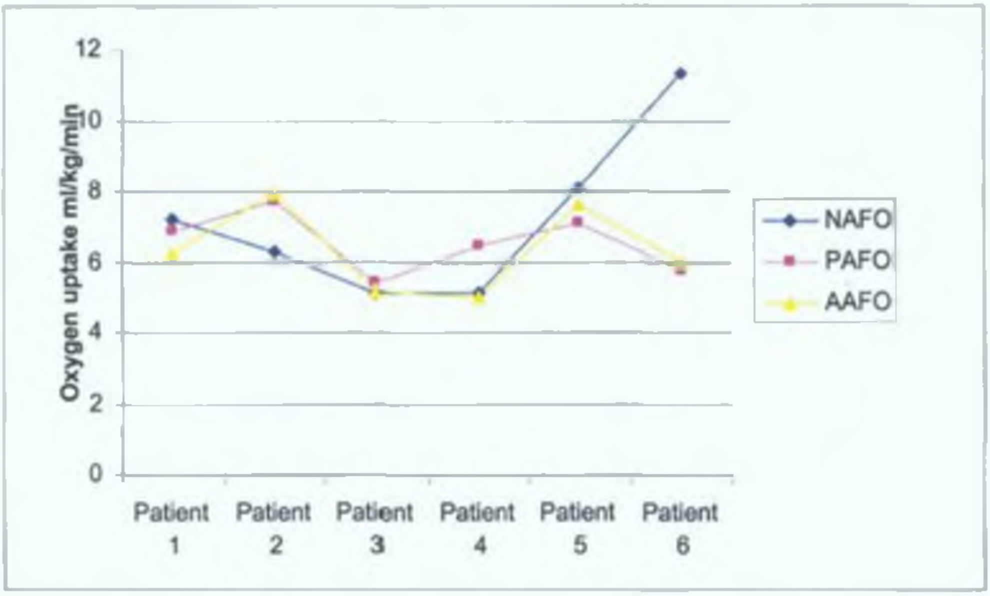



**Figure 6.1:** Oxygen uptake of Sgroup under the conditions of wearing no AFO, a PAFO and an AAFO.

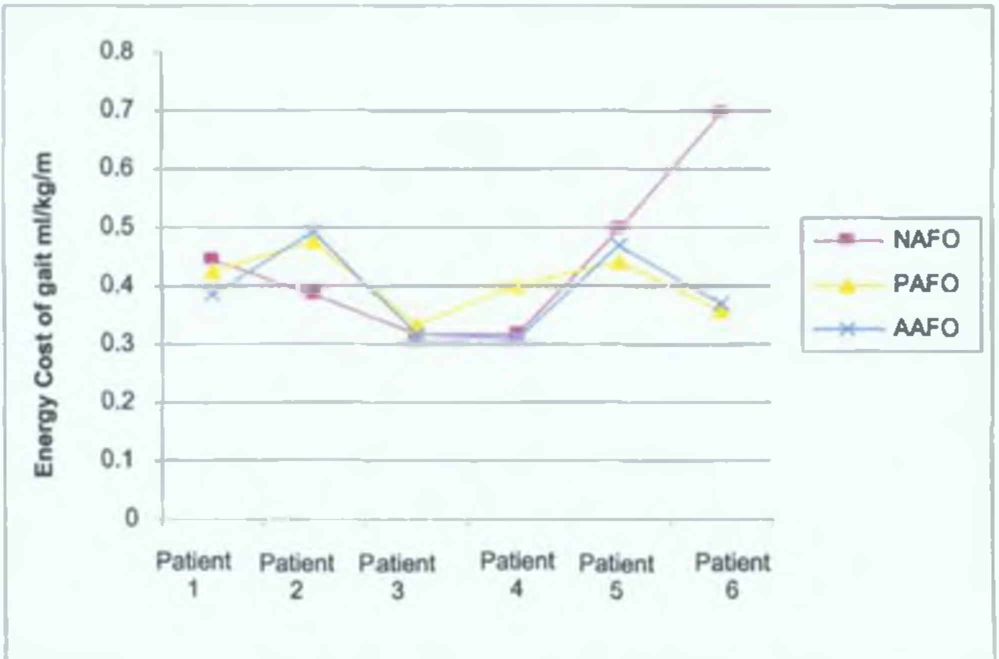



**Figure 6.2:** Energy cost of gait of Sgroup under the conditions of wearing no AFO, a PAFO and an AAFO.

### 6.2 Correlation of variables

In the Ngroup and the Sgroup the two COM methods correlated reasonable well to one another [r=0.7, p<0.05; Table 6.3&6.4]. When the two COM methods were examined relative to oxygen uptake only the sacral displacement method correlated significantly to oxygen uptake [r=0.2, p<0.05 (Ngroup); r=0.5, p<0.05 (Sgroup); Table 6.3&6.4]. However, the correlation was extremely weak with the sacral displacement method being able to account for only 4% (Ngroup) and 25% (Sgroup) of the variation in oxygen uptake.

**Table 6.3:** Matrix of correlation measures for stroke patients

| | $VO_2$ | $COM_{seg}$ | $COM_{sd}$ |
|---|---|---|---|
| $VO_2$ | | 0.2 | 0.5* |
| $COM_{seg}$ | 0.2 | | 0.7* |
| $COM_{sd}$ | 0.5* | 0.7* | |

$VO_2$ = Oxygen uptake
$COM_{seg}$ = Centre of mass by segmental method
$COM_{sd}$ = Centre of mass by sacral displacement method
* = statistically significant ($p<0.05$)

**Table 6.4:** Matrix of correlation measures for normal subjects

| | $VO_2$ | $COM_{seg}$ | $COM_{sd}$ |
|---|---|---|---|
| $VO_2$ | | -0.06 | 0.2* |
| $COM_{seg}$ | -0.06 | | 0.7* |
| $COM_{sd}$ | 0.2* | 0.7* | |

$VO_2$ = Oxygen uptake
$COM_{seg}$ = Centre of mass by segmental method
$COM_{sd}$ = Centre of mass by sacral displacement method
* = statistically significant ($p<0.05$)

### 6.3 COM displacement

Neither COM displacement by segmental analysis or by the sacral marker method was significantly affected by the wear of an AFO ($p>0.05$) [Table 6.2].

### 6.4 Discussion

If a reduction in energy expenditure was found with the wearing of an AFO it would be beneficial to a patient with stroke, as it would allow them to walk for a longer period of time without becoming fatigued. The key finding of the present study was that the wearing of an AFO did not significantly reduce the energy expenditure of gait as measured by the variables of oxygen uptake, energy cost of gait and displacement of the body's COM. Secondly, while the measure of the displacement of the COM using the sacral displacement method correlated significantly with oxygen uptake it only accounted for 4% and 25% of the variance in oxygen uptake in the Ngroup and the Sgroup, respectively. Therefore the hypotheses of this section of the present study were not supported, and it is not a reasonable method to determine oxygen uptake (or energy expenditure) from the COM vertical displacement.

The oxygen uptake during gait was significantly lower in the Sgroup compared to the Ngroup while walking at their customary speed. This is due to the marked reduction in the customary walking speeds of the two groups (Waters and Mulroy, 1999) and has been a consistent finding in many studies (Corcoran et al., 1970; Hash, 1978; Zamparo et al. 1995). In the Sgroup the oxygen uptake values ($7.2\pm2.3 ml.kg^{-1}.min^{-1}$) were lower than those reported by Franceschini et al. (2003) [$9.9\pm1.9 ml.kg^{-1}.min^{-1}$] and Danielsson and Sunnerhagen (2004) [$8.6\pm0.4$ $ml.kg^{-1}.min^{-1}$]. This would suggest that the cohort used in the present study may have had a greater aerobic capacity compared to patients in those two other studies. In the Sgroup, oxygen uptake did not significantly alter with the wearing of either AFO. This result is in agreement with the findings of Franceschini et al. (2003) and Danielsson and Sunnerhagen (2004) but is in contrast to the results of Corcoran et al. (1970).

In contrast to the findings for oxygen uptake the energy cost of gait (oxygen uptake/walking speed) was significantly higher in the Sgroup in comparison to the Ngroup while walking at their customary speed. Zamparo et al. (1995) reported a similar finding and believed that the differences in energy cost between normal and stroke patients were due to biomechanical causes. The values for energy cost of gait reported in the present study for the Sgroup [$0.4\pm0.1 ml.kg^{-1}.m^{-1}$] were slightly lower than those reported by Danielsson and Sunnerhagen (2004) [$0.6\pm0.1 ml.kg^{-1}.m^{-1}$] and considerably lower than those reported by Franceschini et al. (2003) [$0.8\pm0.4 ml.kg^{-1}.m^{-1}$]. This would again suggest that the cohort used in the present study may have had a greater aerobic capacity. The difference in aerobic capacity observed between the present and previous studies may be a factor in the differing outcomes. In contrast to the results of Danielsson and Sunnerhagen (2004) and Franceschini et al. (2003), the present study found no significant difference in the energy cost of gait with the wearing of an AFO. The possible causes of the difference in findings for oxygen uptake and the energy cost of gait between the present and previous studies are discussed below.

There were many methodological differences between the present study and previous studies that may explain the contrasting results. Methodological differences include the control of walking speed, the length of collection of oxygen uptake and randomisation of conditions. Firstly, walking speed in the majority of previous studies (Danielsson and Sunnerhagen, 2004; Franceschini et al., 2003) was not controlled across the conditions of wearing and not wearing an AFO. Zampora et al. (1995) demonstrated that in both stroke and normal subjects, the energy cost of gait is dependent on the walking speed of the subject, with a greater walking speed requiring a lower energy cost (Zampora et al., 1995). Any reduction in energy cost of gait with the wear of an AFO may stem from the biomechanical effectiveness of the AFO or from the speed increase as a result of wearing an AFO. Only a comparison of the energy cost of gait with and without the wearing of an AFO at the same walking speed can provide a definite answer. In the present study when walking speed was controlled across testing conditions no significant alteration in oxygen uptake or the energy cost of gait occurred. Secondly, the timing of collection of oxygen uptake varied across studies. In the current study oxygen uptake was measured over a six-minute period to ensure that steady state was achieved and no further increase in oxygen uptake occurred. Oxygen uptake was then collected for the last two minutes of the test. This is necessary for a valid measure of oxygen uptake (Danielsson and Sunnerhagen, 2000). In the previous studies (Corcoran et al., 1970; Danielsson and Sunnerhagen, 2004; Franceschini et al., 2003) patients walked for a maximum of two to three minutes prior to any oxygen uptake being collected, therefore steady state may not have been reached (Mc Ardle et al., 2001). Finally, in previous studies (Corcoran et al., 1970; Danielsson and Sunnerhagen, 2004; Franceschini et al., 2003) the order of collection of oxygen uptake between the wearing and not wearing of an AFO was not randomised. This may have resulted in greater fatigue in the second test condition, were an AFO was worn which may possibly increased the energy expenditure of the gait.

If energy expenditure had been reduced by the wearing of an AFO, it would have been beneficial to be able to measure energy expenditure less invasively. Measurement of energy expenditure by the assessment of oxygen uptake is a relatively invasive method as it involves the stroke patients wearing a face mask. Examination of the vertical displacement of the COM has been purported to be an indirect measure of energy expenditure, as it has been reported to correlate to oxygen uptake in normal subjects (Kerrigan et al., 1995). However, the COM displacement in stroke patients with equinus deformity of the foot has not been previously measured. COM displacement was not significantly different in the Sgroup in comparison to the Ngroup without the wearing of an AFO or significantly altered when an AFO was worn. Kerrigan et al. (1996) reported similar findings in neurological patients. The results of the present study indicated that wearing an AFO does not significantly affect the displacement of the COM and this outcome is not surprising given that the results of Chapter 5 indicated that wear of an AFO had no significant impact on the determinants (Croce et al., 2001) of the displacement of the COM (eg. ipsi and contra-lateral knee flexion, pelvic obliquity). In the Sgroup and the Ngroup the two measures of COM displacement (whole body kinematics and sacral displacement methods) significantly correlated to one another, as previously reported (Saini et al., 1998). However, only the sacral displacement method significantly correlated to oxygen uptake, but the correlation was low (Ngroup: $r=0.2$; Sgroup: $r=0.5$). Kerrigan et al. (1995) previously reported a significant correlation between COM displacement using the sacral marker method and oxygen uptake however, their reported correlation was much higher ($r=0.90$). The lower correlation in the present study may have been due to methodological differences. In Kerrigan's study oxygen uptake was measured prior to a steady state condition being achieved, whereas in the present study oxygen uptake was more appropriately assessed during steady state.

In light of the low correlation between oxygen uptake and COM displacement using the sacral method the hypothesis that measurement of the COM was a viable method of determining energy expenditure must be rejected.

The results of the current work demonstrated that the effect of AFO wear on energy expenditure was variable in stroke patients, with some stroke patients demonstrating an increase and others a reduction in energy expenditure. The results of the current work would suggest that future studies assessing energy expenditure in stroke patients should employ a cluster analysis. A limitation of this present study was the small number of stroke patients. This prevented a cluster analysis being employed. In conclusion, wearing of an AFO did not significantly decrease the energy expenditure of gait in stroke patients with equinus deformity of the foot as assessed indirectly via oxygen uptake, energy cost of gait and the displacement of the COM. The measurement of the COM using whole body kinematics and sacral displacement methods were not viable to assess the energy expenditure of gait in stroke patients with equinus deformity of the foot. AFO wear did not significant affect energy expenditure at a group level. Given that their may be individual based responses to AFO wear (Chapter 5) which are dependent on the degree of gait disability, it maybe more appropriate for future studies to employ a cluster analyses or a single subject design when examining the effects of wearing an AFO on energy expenditure.

# Chapter 7: Conclusion

This thesis has added to the body of knowledge of the gait pattern of stroke patients with equinus deformity of the foot. It has provided reference data for future researchers on the true gait deformities present in this stroke group, independent of the variables of walking speed and footwear. The current work is the first to provide joint angular kinematics and kinetics data for the affected and unaffected limbs for stroke patients with equinus deformity of the foot in both the sagittal, coronal and transverse planes.

The gait patterns of stroke patients with equinus deformity of the foot were not homogenous. It was possible to use a cluster analysis to successfully and objectively identify three different gait patterns within this group. The gait pattern of each subgroup showed significant differences in comparison to normal, age and speed matched subjects. Speed matched comparisons allowed the identification of the true gait deviations present in both the affected and unaffected limbs in the stroke patients.

This thesis was the first to examine the effects of wearing different leaf forms of AFOs on the gait pattern of stroke patients with equinus deformity of the foot in both the sagittal, coronal and transverse planes. The present study, was also the first to examine the affect that wearing an AFO on the affected limb had on the unaffected limbs, joint kinematics and kinetics. The results of the present study concluded that wearing an AFO had no detrimental effects on the unaffected limb.

The present study provides scientific evidence for physiotherapists and stroke rehabilitators that the wear of an AFO can be beneficial to some patients with stroke, with different leaf forms of AFOs having differing effects. These effects were dependent upon the gait pattern of the stroke patients. Only stroke patients with a gait pattern similar to that of SG1 obtained a beneficial improvement in gait with the wearing of an AFO. SG1 had the most normalised gait pattern of the subgroups without wear of an AFO, therefore the results of the present study would suggest that stroke patients with a more abnormal gait pattern may not obtain any benefit from the wearing of an AFO. Clinicians, when assessing if their stroke patients are suitable to wear an AFO should try to identify if the gait pattern of their patient is similar to that of SG1. Finally, walking speed had a significant effect on the biomechanical performance of an AFO in the normal subject group.

Stroke patients with equinus deformity of the foot were found to have a significantly higher energy cost of gait when compared to normal subjects walking at their customary walking speed. When walking speed was controlled across testing conditions, wearing of an AFO did not significantly decrease energy expenditure as assessed indirectly via oxygen uptake, energy cost of gait and the displacement of the COM. The measurement of the COM using whole body kinematics and sacral displacement methods were not viable to determine the energy expenditure of gait in stroke patients with equinus deformity of the foot.

**Key messages of the thesis**

- Stroke patients with equinus deformity of the foot have many different gait strategies to deal with this deformity but can be categorised into subgroups.
- AFOs are beneficial to stroke patients with a gait pattern similar to SG1.
- The design characteristics of AFOs have differing effects on hemiplegic gait (as demonstrated in the present study in the comparison of an anterior to a posterior leaf AFO)
- Energy expenditure is not significantly different when AFOs are worn.
- Displacement of the COM is not a viable method of assessing energy expenditure in stroke patients with equinus deformity of the foot

**Future studies**

- Future research should examine if the biomechanical design of an AFO can be altered to result in an improved gait pattern for stroke patients presenting with a very slow walking speed lower than $0.4 m.s^{-1}$.
- Future studies should employ a cluster analysis to identify the gait characteristics of stroke patients who received improved energy expenditure with the wear of an AFO.
- Future research should be undertaken to identify the causes of the abnormal movement patterns observed during gait in the stroke patients with equinus deformity of the foot. The use of EMG and the assessment of muscle strength may help in this aim. By identifying the cause, appropriate intervention strategies may be employed by the clinician.
- Different footwear may have a significant effect of the gait pattern of patients with equinus deformity of the foot. Future research should examine if this is the case.
- Future studies should directly assess if wear of different leaf forms of AFOs have a significant affect on toe clearance.
- Future research should examine the cause of the significantly increased external rotation angles of the foot in the present study at heel strike with AFO wear.

# Appendices

## **Appendix A** – Application for research ethics for normal subjects

DCU RESEARCH INVOLVING HUMAN SUBJECTS RESEARCH SUBMISSION

Name of student submitting research proposal: Sharon Kinsella

Thesis advisor: Dr. Kieran Moran

Medical Consultant: Dr. Noel Mc Caffrey

Project Title: 'The effect of an Ankle Foot Orthoses on hemiplegic gait'

**Describe the basic purposes of the research proposed.**

To determine if wear of an ankle foot orthosis has an effect on hemiplegic gait

To determine what type of orthotic is of greatest effect on hemiplegic gait

**Outline the design and methodology of the project.**

The proposed research will evaluate the wear of the AFO using an AB design, with subjects gait being assessed with and without the wear of the AFO. The kinematic, kinetic and energy cost of the gait of hemiplegic subjects will be assessed with and without the wear of the AFO.

A convenience sample of 20 hemiplegic subjects will be referred to the study from local physiotherapists, consultants and orthotists. Inclusion criteria will be that the subjects have suffered a stroke which has resulted in an inability to control there ankle joint in the normal manner during gait. The subjects will have experienced their stroke a minimum of 6 months previous to testing. Subjects will be excluded if: they cannot walk 10 metres independently, if they have any other condition other than as the result of the hemiplegia which may affect they gait such as hip or knee replacement or severe osteoarthritis of the hip or knee. Subjects will be also excluded if they suffer from any of the following illnesses: Diabetes Mellitus, Renal and Kidney dysfunction, Hypertension, Metabolic disorders, Acute MI, Uncontrollable cardiac arrhythmias, Active endocarditis, Symptomatic severe aortic stenosis, Acute Pulmonary. Written informed consent will be sought from each subject prior to his or her involvement in the study (Appendix II).
Subjects will be assessed during free walking and in a subgroup during walking on a treadmill.

**Freewalking**
During freewalking the kinematics and kinetics of the subjects gait will be assessed under three conditions, 1) walking without an AFO, 2) walking with a prefabricated anterior leaf AFO and 3) walking with a prefabricated posterior leaf AFO. The subjects will be required to wear tight fitting clothing, preferably shorts and thirty-six reflective markers with a diameter of 16mm will be placed on their body. Subjects will then be asked to walk in a pre designated area at their comfortable speed under the three conditions, during which data will be collected from forceplates and motion cameras. Subjects will be given amble time to rest during the conditions as necessary.

Treadmill walking
In a subgroup of the hemiplegic subjects they will be assessed during treadmill walking. The inclusion criteria for hemiplegic subjects into this group will be a willingness to walk on a treadmill, a walking speed greater than one kilometre per hour and a demonstrated exercise capacity to walk for 6 minutes continuously.
During the treadmill walking trial the kinematics and oxygen cost of walking on a treadmill will be assessed under three conditions, 1) walking on the treadmill without an AFO 2) walking on the treadmill with a prefabricated AFO 3) and walking on a treadmill with a prefabricated posterior leaf AFO. The subjects will be required to wear tight fitting clothing, preferably shorts and thirty-six reflective markers and an accelerometer will be placed on their body. A gas mask will be placed on the subjects during the trial for each condition to measure oxygen uptake. Subjects will then be asked to walk on the treadmill at their preferred walking speed for six minutes per condition. Kinematic and accelerometric data will be measured for the last 30 seconds of each condition and oxygen uptake will be measured for six minutes for each condition. Subjects will be given ample time to rest during the conditions as necessary.

**Describe the research procedures as they affect the research subject and any other parties involved.**

Subjects will visit the Biomechanics Laboratory, DCU, on one occasion. The following will take place in the Laboratory:
The experiment will be explained to the subject and they will be shown all the equipment involved in the study
They will sign an informed consent form (Appendix II.)
The subjects will be asked to change into shorts
Weight and height will be measured
36 reflective markers will be placed on the subject for gait analysis. Subjects will be asked to walk 6m, nine times; 3 times wearing no AFO, 3 times wearing a posterior leaf AFO and 3 times wearing an anterior leaf AFO. Kinematic and kinetic data will be acquired as the subject walks in a pre-determined area. The order of wear of the AFO will be randomised. Subjects will be given ample time to rest during each test.
Oxygen uptake in the subgroup of the hemiplegic subjects will be measured as the subject walks on a treadmill using standard open circuit spirometry. Subject will be fitted with a mouthpiece and nose-clip prior to the test. Flexible tubing to the electronic instruments will connect the mouthpiece. These will monitor the amount of oxygen consumed and carbon dioxide produced. A baseline value for resting oxygen consumption will firstly be measured as the subject sits still for 5 minutes. The subject will then be placed on the treadmill. The

speed of the treadmill will be predetermined from the normal cadence of the subject without the wear of an orthotic. While on the treadmill an accelerometer will be attached to their back at the level of second sacral veterbra with tape. The subject will walk on the treadmill for 6 minutes, 3 times, without an AFO, with a posterior leaf AFO and with an anterior leaf AFO. The speed of the treadmill will remain constant throughout the test. During the last 30 seconds of each test kinematic and accelerometer data will be collected. Subjects will be given ample time to rest during each test.

**What in your opinion are the ethical considerations involved in this proposal?** (You may wish for example to comment on issues to do with consent, confidentiality, risk to subjects, etc.)

Hemiplegic subjects with an inability to control their ankles during gait are at a higher risk of falling then other non-disabled subjects. There is therefore a risk associated with them walking in the laboratory. However this risk is no greater than if they were walking outside the laboratory or in their own homes. Prior to the commencement of the study, subjects will be asked to walk with supervision without the wear of an orthotic to ensure the subjects are safe to do so unaided.

Hemiplegic subjects due to the nature of their condition may have a decreased exercise capacity and feel breathless with activities that they are not familiar with. Therefore subjects taking part in this study will be advised to rest in-between trials if they wish to do so or if the investigator notices breathlessness

The subjects will not be used to wearing a facemask while walking for the collection of oxygen. Subjects during the test will be given 5 minutes at the start of the test to accommodate to this sensation. If the subject still feels uncomfortable wearing this mask the test will be terminated. The subjects will be free to with draw from the test at any time.

The reflective markers, used, will be attached with hypoallergenic double-sided sticky tape. In some patients this may result in redding of the skin when markers are removed.

The subjects will be unfamiliar with walking on a treadmill and may feel uncomfortable walking on it. As with walking on a normal surface are at risk of falling. The treadmill will have support bars on the front of the machine so that the subjects can hold onto to them if they feel unstable. Subjects will be given ample time to familiarise themselves with walking on a treadmill prior to the commencement of the test. Subjects who feel uncomfortable walking on the treadmill will be free to withdraw from the study.

The subjects will be required to wear shorts to participate in the study. Some subjects may feel uncomfortable wearing such clothing. If so they will be free to withdraw from the study.

Subjects' anonymity and confidentiality will be protected at all times. Subjects' personal identity and personal information will not be revealed, published or used in further studies. Subjects will be assigned an ID number under which all personal information will be stored in a secure computer. The study results may be used as part of a series of studies and study findings may be presented at scientific meetings and published at scientific journals.

Subjects may withdraw from the study at any point. There will be no penalty if withdrawal occurs before testing is complete. If subjects are students at DCU their withdrawal with not affect their relationship with the school or University.

**Outline the reasons, which lead you to be satisfied that the possible benefits to be gained from the project justify any risks or discomforts involved.**
With all studies there are a number of risks associated with participation, as with this study (Section 4). However the continued use of AFOs in clinical practice without evidence based research strongly indicates the need for this present study.
The effects of the wear of an orthotic on hemiplegic gait are unclear. Research has yet to examine the effects of wear of an AFO on the kinetic and kinetics of gait. AFO's may or may not improve the gait of hemiplegic subjects. If AFO's are beneficial there is a need to determine which AFO is of greater benefit an anterior or posterior leaf.
Subjects following there participation in this study will be able to know firstly if wear of an AFO is beneficial for them and if so which type of orthotic suits them best

**Who are the investigators (including assistants) who will conduct the research and what are their qualifications and experience?**

Sharon Kinsella Proposer of research; Physiotherapist and Lecturer in Rehabilitation Medicine.

Dr. Kieran Moran: Lecturer and researcher

Dr. Noel McCaffrey: Medical Consultant; Sports Medicine Physician with an adjunct position in the Faculty of Science & Health

Xavier Monedero: technician

**Are arrangements for the provision of clinical facilities to handle emergencies necessary? If so, briefly describe the arrangements made.**

Yes. There is an emergency crash cart and defibrillator, a fully equipped first-aid kit and an 'on-call' physician in the Laboratory block, Dr. Noel Mc Caffrey.
The main researcher of this study is qualified in advance first aid and has years of experience dealing with neurological patients in a rehabilitation setting.

In cases where subjects are identified from information held by another party (for example, a doctor or hospital) describe the arrangements whereby you gain access to this information.
Patients will be accessed by referral from the patients' consultant, orthotist or physiotherapist. Upon referral the patient will be contacted directly by the investigator to assess whether the patient is interested in partaking in the study and to assess whether they will fulfil the inclusion criteria. Following initial contact the patient will be met to assess their functional ability and to explain fully the experimental protocol and to give them the research informed consent form to read. If the patient is still interested in participating in the study and they fulfil the inclusion criteria a date will be arranged for them to attend the biomechanics laboratory

**Specify whether subjects will include students or others in a dependent relationship.**

No

Specify whether the research will include children or those with mental illness, disability or handicap. If so, please explain the necessity of using these subjects.
Hemiplegic subjects are the subjects of choice for this experiment as they have specific gait disturbances due to the effects of their stroke on their ankle joint. The main treatment method for their resultant gait disturbances is the wear of an AFO. As yet there is no clear evidence based research to support the use of AFO in the treatment of their gait disturbances. This research project aims to investigate if the wear of AFO's improve the gait of these subjects.

**Will payment be made to any research subject?**

No

**Describe the procedures to be used in obtaining a valid consent from the subject. Please supply a copy of the information sheet provided to the individual subject.**

A one-page summary of the study will be provided during the recruitment phase (Appendix I). Also a copy of the research information consent form will be given to the subject to read prior to the arrival to the Biomechanics Laboratory
Upon arrival at the Biomechanics Laboratory, subjects will be given a detailed description of the investigation and asked to provide their written consent to participate. The informed consent sheet will be provided at the initial screening and any questions or queries will be answered (Appendix II).

**Comment on any cultural, social or gender-based characteristics of the subject which have affected the design of the project or which may affect its conduct.**

None

**Give details of the measures, which will be adopted to maintain the confidentiality of the research subject.**

Subjects will be assigned an ID number that will be used to identify data. Data will be stored in a secure filling cabinet within the Centre of Sports Science and Health. Only staff within the Centre of Sport Science and Health will have access to the cabinet.

Data will only be viewed by the main investigator and co-investigators, Dr. Kieran Moran and Dr. Noel Mc Caffrey.

**Will the information gained be anonymized? If not, please justify.**

Yes, no names or details will be mentioned in dissemination of the study. See question 14.

**Will the intended group of research subjects, to your knowledge, be involved in other research? If so, please justify.**

No

**Date on which the project will begin:** April 2003 and end: Dec 2003

**Please state location(s) where the project will be carried out.**

The study will take place in the Biomechanics Laboratory at the Sport Science and Health Department, Dublin City University, Dublin 9.

Signed________________________ Date __________________
(Proposer of research)

Signed ____________________ Date __________________
(Supervisor of student)

COMMENT FROM HEAD OF DEPARTMENT/GROUP/INSTITUTE/CENTRE

Signed ____________________ Date __________________
(Head of Department/Group/Institute/Centre)

## Appendix I

This is what will happen to me during the research study

I will be asked to attend the Biomechanics Laboratory in Dublin City University, for an appointment to be assessed of the quality of my walking. The project will require me to attend the Laboratory for approximately one and a half hours.
During this time I will be firstly introduced to the chartered physiotherapist who will be involved in this study. I will be examined by the physiotherapist to ensure that I am suitable to take part in this study and I will be asked to answer questions regarding my health. The study will then be explained to me and the equipment, which will be used to assess my walking, will be shown to me. Any questions that I may have about the study will be answered.

I will then be asked to read and sign a consent form, verifying that I understand the study and my role within the study. I will be able to stop the test at any time that I wish.
I will then be asked to wear tight clothing preferably shorts. 36 small balls will be placed on me, using sticky tape.

I will be asked to walk at my normal walking speed across a pre set area in the room. I will be assessed firstly to ensure that I am safe to walk without my orthotic. I will then be asked to walk at a comfortable speed across a preset area in the room without an orthosis and then with 2 different types of orthoses. An orthosis is a device, which holds a foot in a fixed position when one walks. I will be given ample time to rest during these trials as necessary.
I maybe asked to take part in the second part of the study, which involves walking on a treadmill. I will only be allowed to take part in this part of the experiment if I am happy to walk on a treadmill and if I am used to walking for continuously for six minutes. In this part of the experiment I will be asked to walk on a treadmill at my own walking speed. A small accelerometer, which measures the movement of walking, will be placed on the small of my back with sticky tape. I will be given time to get used to walking on the treadmill prior to starting the test. When I am on the treadmill I will be asked to place a mouthpiece into my mouth, which will measure the amount of oxygen I use when walking. I will be given time to get used to the sensation of having a mouthpiece in my mouth. This part of the experiment will be repeated while I walk with both types of orthoses. I will be asked to walk on the treadmill for 6 minutes. This part of the experiment will then be repeated as I walk with the 2 different types of orthoses. I will again be given ample time to rest during these tests.
When the tests are finished the markers will be removed from my body and I will be free to get dressed and leave the laboratory. I will be contacted in due course for the results of the tests to be explained to me.

**Appendix II**

RESEARCH - INFORMED CONSENT FORM

I. Project Title: The effect of wear of Ankle Foot Orthosis on hemiplegic gait

II. Introduction to the study: Many people following stroke have difficulty walking. One of the more common problems that people experience after stroke is that they have difficulty lifting their foot as they walk. Ankle foot orthosis are normally given to people with this walking problem to help them walk better. However very little research has been done to show that wearing the ankle foot orthosis will improve walking pattern. The aim of this study is to examine if walking with an ankle foot orthosis is better than walking without it.

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I am being asked to take part in this research study. The purpose of the study is to determine the effects of wear of an ankle foot orthosis(AFO) on walking patterns

This research study will take place at Dublin City University and will require me to attend the University on one occasion for a period of approximately one and a half hours

This is what will happen during the research day
I will be examined by a chartered physiotherapist to ensure my suitability for the study and asked to fill out a questionnaire on my health
I will be asked to read and sign this consent form, verifying that I understand the study and my role within the study
I will be asked to wear tight clothing preferably shorts
36 small balls will be placed on my skin, with sticky tape
A small accelerometer which measures speed will be placed on the bottom of my back with sticky tape
I will then be asked to walk at a comfortable speed across a pre set area in a room without an orthosis and then with 2 different types of orthoses.
I will be given ample time to rest between walks
I may then be asked to walk on a treadmill at a self-selected comfortable walking speed for 6 minutes without an orthosis and then with 2 different types of orthoses.
I will be given time to get used to walking on the treadmill prior to starting the test
When on the treadmill I will be asked to place a mouthpiece in my mouth which will measure the amount of oxygen I am using
Again I will be given ample time to rest during these tests

VI. Sometimes there are problems associated with this type of study. These are:
1. Having suffered a stroke I will be at a higher risk of sustaining a second stroke. It should be noted however that with the experimental protocol been employed the likelihood of this occurring is minimal. In the event of an emergency the research laboratory is equipped with an emergency crash car and defibrillator. An individual trained in resuscitation will be present in the building during each test.

2. I may find that wearing a mouthpiece and a nose clip uncomfortable especially when walking, but I understand that it will only be for a short time, and that they will not interfere with my ability to walk.
3. I may find walking on a treadmill unusual, especially if I have never walked on one before. However I will be given ample time to get used to this and I can withdraw from this study at any time if I wish to do so.
4. If there are any adverse effects during my visit, I will be monitored until the effects pass.

VII. Their maybe benefits to me from this treatment. These are:
1. I will reserve a report on my walking action, which I can forward to my doctor /consultant that may help him/her in treating me for problems with the way I walk.
2. I will receive information on whether an AFO will benefit me and which type of AFO would be best for me.

VIII. My confidentiality will be guarded. Dublin City University will make reasonable efforts to protect the information about me and my part in this study and no identifying data will be published. This will be achieved by assigning me an ID number against which all data will be stored. Details linking my ID number and name will not be stored with the data. The results of the study maybe published and used in further studies.

IX. If you have any questions about the study, I am free to call Dr. Kieran Moran at (01) 7008011 or Sharon Kinsella at (0503) 76218

Taking part in this study is my decision. If I do agree to take part, I may withdraw at any point including during the exercise test. There will be no penalty if I withdraw before I have completed all stages of the study.
XI. Signature:
I have read and understood the information in this form. My questions and concerns have been answered by the researchers, and I have a copy of this consent form. Therefore, I consent to take part in this research project entitled 'The effect of wear of Ankle Foot Orthosis on hemiplegic gait'

Signature:______________________ Date: ______________________

Witness:______________________ Witness:______________________
Signiture Printed name

**Health questionnaire form (PR01)**

Name____________________________

Date:______________________________

Age_____________________________

Time since stroke:____________________

Weight__________________________

Please tick the appropriate boxes, answering all questions

| | YES | NO |
|---|---|---|
| I am currently in good health | ☐ | ☐ |
| I can walk 10 meters at my own pace, with the help of aids | ☐ | ☐ |
| I have had my knee or hip joints replaced | ☐ | ☐ |
| I do have severe arthritis in my knees or hips | ☐ | ☐ |
| I do get excessively short of breath when I walk short distances | ☐ | ☐ |
| I presently have injuries to my legs that might affect my ability to walk | ☐ | ☐ |
| I do suffer from Diabetes Mellitus | ☐ | ☐ |
| I do suffer from Renal and Kidney dysfunction | ☐ | ☐ |
| I do suffer from High blood pressure | ☐ | ☐ |
| I do suffer from any of the following | ☐ | ☐ |

1. Metabolic disorders
2. Acute Myocardiac infarction
3. Uncontrollable cardiac arrhythmias,
4. Active endocarditis,
5. Symptomatic severe aortic stenosis,
6. Acute Pulmonary

Signature:______________________________

## **Appendix B** – Application for research ethics for normal subjects

DCU RESEARCH INVOLVING HUMAN SUBJECTS RESEARCH SUBMISSION

| | |
|---|---|
| Name of student submitting research proposal: | Sharon Kinsella |
| Thesis advisor: | Dr. Kieran Moran |
| Medical Consultant: | Dr. Noel Mc Caffrey |

Project Title: 'The effect of Ankle Foot Orthosis on gait'

**Describe the basic purposes of the research proposed.**

To determine if wear of an ankle foot orthosis has an effect on normal gait

To determine what type of orthotic is of greatest effect on normal gait

**Outline the design and methodology of the project.**

The proposed research will evaluate the wear of the AFO using an AB design, with subjects gait being assessed with and without the wear of the AFO. The kinematic, kinetic and energy cost of the gait of normal subjects will be assessed with and without the wear of the AFO.

A convenience sample of 30 normal subjects, 15 between the ages of 18-30 and 15 between the ages of 45-60 will be recruited to the study from local students and staff of DCU. Inclusion criteria will be that the subjects are healthy, can walk independently, have no condition that may affect their gait such as hip or knee replacement or severe osteoarthritis of the hip or knee or recent sports injury. Subjects will be also excluded if they suffer from any of the following illnesses: Diabetes Mellitus, Renal and Kidney dysfunction, Hypertension, Metabolic disorders, Acute MI, Uncontrollable cardiac arrhythmias, Active endocarditis, Symptomatic severe aortic stenosis, Acute Pulmonary. Written informed consent will be sought from each subject prior to his or her involvement in the study (Appendix 2). A subgroup of the same will be recruited to undertake a reliability study. They will be requested to return to the laboratory to repeat the tests the next day or in a week's time.

Subjects will be assessed during free walking and walking on a treadmill.

Free walking
During free walking the kinematics and kinetics of the subjects gait will be assessed under three conditions, 1) walking without an AFO, 2) walking with a prefabricated anterior leaf AFO and 3) walking with a prefabricated posterior leaf AFO. The subjects will be required to wear tight fitting clothing, preferably shorts and thirty-six reflective markers with a diameter of 16mm will be placed on their body. Subjects will then be asked to walk in a pre designated area at their comfortable speed under the three conditions, during which data will be collected from forceplates and motion cameras. Subjects will be given amble time to rest during the conditions as necessary.

Treadmill walking
During the treadmill walking trial the kinematics and oxygen cost of walking on a treadmill will be assessed under three conditions, 1) walking on the treadmill without an AFO 2) walking on the treadmill with a prefabricated AFO 3) and walking on a treadmill with a prefabricated posterior leaf AFO. The subjects will be required to wear tight fitting clothing, preferably shorts and thirty-six reflective markers and an accelerometer will be placed on their body. A gas mask will be placed on the subjects during the trial for each condition to measure oxygen uptake. Subjects will then be asked to walk on the treadmill at their preferred walking speed for six minutes per condition. Kinematic and accelerometric data will be measured for the last 30 seconds of each condition and oxygen uptake will be measured for six minutes for each condition. Subjects will be given ample time to rest during the conditions as necessary.

**Describe the research procedures as they affect the research subject and any other parties involved.**

Subjects will visit the Biomechanics Laboratory, DCU, on one occasion. The following will take place in the Laboratory:
The experiment will be explained to the subject and they will be shown all the equipment involved in the study
They will sign an informed consent form (Appendix II.)
The subjects will be asked to change into shorts
Weight and height will be measured
36 reflective markers will be placed on the subject for gait analysis. Subjects will be asked to walk 6m, nine times; 3 times wearing no AFO, 3 times wearing a posterior leaf AFO and 3 times wearing an anterior leaf AFO. Kinematic and kinetic data will be acquired as the subject walks in a pre-determined area. The order of wear of the AFO will be randomised. Subjects will be given ample time to rest during each test.
Oxygen uptake will be measured as the subject walks on a treadmill using standard open circuit spirometry. Subject will be fitted with a mouthpiece and nose-clip prior to the test. Flexible tubing to the electronic instruments will connect the mouthpiece. These will monitor the amount of oxygen consumed and carbon dioxide produced. A baseline value for resting oxygen consumption will firstly be measured as the subject sits still for 5 minutes. The subject will then be placed on the treadmill. The speed of the treadmill will be predetermined from the normal cadence of the subject without the wear of an orthotic. While on the treadmill an accelerometer will be attached to their back at the level of second sacral veterbra with tape. The subject will walk on the treadmill for 6 minutes, 3 times, without an AFO, with a

posterior leaf AFO and with an anterior leaf AFO. The speed of the treadmill will remain constant throughout the test. During the last 30 seconds of each test kinematic and accelerometer data will be collected. Subjects will be given ample time to rest during each test.

**What in your opinion are the ethical considerations involved in this proposal?** (You may wish for example to comment on issues to do with consent, confidentiality, risk to subjects, etc.)

The subjects will not be used to wearing a facemask while walking for the collection of oxygen. Subjects during the test will be given 5 minutes at the start of the test to accommodate to this sensation. If the subject still feels uncomfortable wearing this mask the test will be terminated. The subjects will be free to with draw from the test at any time.
The reflective markers, used, will be attached with hypoallergenic double-sided sticky tape. In some patients this may result in redding of the skin when markers are removed.
The subjects will be unfamiliar with walking on a treadmill and may feel uncomfortable walking on it. Subjects will be given ample time to familiarise themselves with walking on a treadmill prior to the commencement of the test. Subjects who feel uncomfortable walking on the treadmill will be free to withdraw from the study.
The subjects will be required to wear shorts to participate in the study. Some subjects may feel uncomfortable wearing such clothing. If so they will be free to withdraw from the study.
Subjects' anonymity and confidentiality will be protected at all times. Subjects' personal identity and personal information will not be revealed, published or used in further studies. Subjects will be assigned an ID number under which all personal information will be stored in a secure computer. The study results may be used as part of a series of studies and study findings may be presented at scientific meetings and published at scientific journals.
Subjects may withdraw from the study at any point. There will be no penalty if withdrawal occurs before testing is complete. If subjects are students at DCU their withdrawal with not affect their relationship with the school or University. However once testing is finished subjects' personal information and results may not be removed form the database.

**Outline the reasons, which lead you to be satisfied that the possible benefits to be gained from the project justify any risks or discomforts involved.**
With all studies there are a number of risks associated with participation, as with this study (Section 4). However the continued use of AFOs in clinical practice without evidence based research strongly indicates the need for this present study.
The effects that the AFOs may have on normal gait patterns will significantly help in interpreting results on the wear of orthotics by hemiplegic subjects. The study will also help to indicate what kinematic or kinetic variables might demonstrate the greatest change with wear of the orthotic which will be beneficial for a later studies examining the effects of wear of AFO on hemiplegic subjects.

**Who are the investigators (including assistants) who will conduct the research and what are their qualifications and experience?**

Sharon Kinsella Proposer of research; Physiotherapist and Lecturer in Rehabilitation Medicine.

Dr. Kieran Moran: Lecturer and researcher

Dr. Noel McCaffrey: Medical Consultant; Sports Medicine Physician with an adjunct position in the Faculty of Science & Health

Xavier Monedero: technician

**Are arrangements for the provision of clinical facilities to handle emergencies necessary?** If so, briefly describe the arrangements made.

Yes. There is an emergency crash cart and defibrillator, a fully equipped first-aid kit and an 'on-call' physician in the Laboratory block, Dr. Noel Mc Caffrey.
The main researcher of this study is qualified in advance first aid and has years of experience dealing with neurological patients in a rehabilitation setting.

**In cases where subjects are identified from information held by another party (for example, a doctor or hospital) describe the arrangements whereby you gain access to this information.**
Patients will be accessed by referral from the patients' consultant, orthotist or physiotherapist. Upon referral the patient will be contacted directly by the investigator to assess whether the patient is interested in partaking in the study and to assess whether they will fulfil the inclusion criteria. Following initial contact the patient will be met to assess their functional ability and to explain fully the experimental protocol and to give them the research informed consent form to read. If the patient is still interested in participating in the study and they fulfil the inclusion criteria a date will be arranged for them to attend the biomechanics laboratory

**Specify whether subjects will include students or others in a dependent relationship.**

Yes students will be included in this study

**Specify whether the research will include children or those with mental illness, disability or handicap. If so, please explain the necessity of using these subjects.**
Not applicable

**Will payment be made to any research subject?**

No

**Describe the procedures to be used in obtaining a valid consent from the subject. Please supply a copy of the information sheet provided to the individual subject.**

A one-page summary of the study will be provided during the recruitment phase (Appendix I).
Upon arrival at the Biomechanics Laboratory, subjects will be given a detailed description of the investigation and asked to provide their written consent to participate. The informed consent sheet will be provided at the initial screening and any questions or queries will be answered (Appendix II).

**Comment on any cultural, social or gender-based characteristics of the subject which have affected the design of the project or which may affect its conduct.**

None

**Give details of the measures which will be adopted to maintain the confidentiality of the research subject.**

Subjects will be assigned an ID number that will be used to identify data. Data will be stored in a secure filling cabinet within the Centre of Sports Science and Health. Only staff within the centre of Sport Science and Health will have access to the cabinet.

Data will only be viewed by the main investigator and co-investigators, Dr. Kieran Moran and Dr. Noel Mc Caffrey.

**Will the information gained be anonymized? If not, please justify.**

Yes, no names or details will be mentioned in dissemination of the study. See question 14.

**Will the intended group of research subjects, to your knowledge, be involved in other research? If so, please justify.**

No

**Date on which the project will begin**: April 2003 and end: Dec 2003

**Please state location(s) where the project will be carried out.**

The study will take place in the Biomechanics Laboratory at the Sport Science and Health Department, Dublin City University, Dublin 9.

Signed ____________________ Date ________________
(Proposer of research)

Signed ____________________ Date ________________
(Supervisor of student)

COMMENT FROM HEAD OF DEPARTMENT/GROUP/INSTITUTE/CENTRE

Signed ____________________ Date ________________
(Head of Department/Group/Institute/Centre)

## Appendix I

This is what will happen to me during the research study

I will be asked to attend the Biomechanics Laboratory in Dublin City University, for an appointment to be assessed of the quality of my walking. The project will require me to attend the Laboratory for approximately one hour.
During this time I will be firstly introduced to the chartered physiotherapist who will be involved in this study. I will be examined by the physiotherapist to ensure that I am suitable to take part in this study and I will be asked to answer questions regarding my health. The study will then be explained to me and the equipment, which will be used to assess my walking, will be shown to me. Any questions that I may have about the study will be answered.

I will then be asked to read and sign a consent form, verifying that I understand the study and my role within the study. I will be able to stop the test at any time that I wish.
I will then be asked to wear tight clothing preferably shorts. 36 small balls will be placed on me, using sticky tape.

I will be asked to walk at my normal walking speed across a pre set area in the room. I will then be asked to walk at a comfortable speed across a preset area in the room without an orthosis and then with 2 different types of orthoses. An orthosis is a device, which holds a foot in a fixed position when one walks. I will be given ample time to rest during these trials as necessary.
In the second part of the experiment I will be asked to walk on a treadmill at my own walking speed. A small accelerometer, which measures the movement of walking, will be placed on the small of my back with sticky tape. I will be given time to get used to walking on the treadmill prior to starting the test. When I am on the treadmill I will be asked to place a mouthpiece into my mouth, which will measure the amount of oxygen I use when walking. I will be given time to get used to the sensation of having a mouthpiece in my mouth. This part of the experiment will be repeated while I walk with both types of orthoses. I will be asked to walk on the treadmill for 6 minutes. This part of the experiment will then be repeated as I walk with the 2 different types of orthoses. I will again be given ample time to rest during these tests.
When the tests are finished the markers will be removed from my body and I will be free to get dressed and leave the laboratory.

# Appendix II

RESEARCH - INFORMED CONSENT FORM

I. Project Title: The effect of wear of Ankle Foot Orthosis on gait

II. Introduction to the study: Many people following stroke have difficulty walking. One of the more common problems that people experience after stroke is that they have difficulty lifting their foot as they walk. Ankle foot orthosis are normally given to people with this walking problem to help them walk better. However very little research has been done to show that wearing the ankle foot orthosis will improve walking pattern. The aim of this study is to examine if walking with an ankle foot orthosis effects the walking pattern of normal healthy subjects.

---

I am being asked to take part in this research study. The purpose of the study is to determine the effects of wear of an ankle foot orthosis (AFO) on walking patterns

This research study will take place at Dublin City University and will require me to attend the University on one occasion for a period of approximately one hour

This is what will happen during the research day
I will be examined by a chartered physiotherapist to ensure my suitability for the study and asked to fill out a questionnaire on my health
I will be asked to read and sign this consent form, verifying that I understand the study and my role within the study
I will be asked to wear tight clothing preferably shorts
36 small balls will be placed on my skin, with sticky tape
A small accelerometer which measures speed will be placed on the lower part of my back with sticky tape
I will then be asked to walk at a comfortable speed across a pre set area in a room without an orthosis and then with 2 different types of orthoses.
I will be given ample time to rest between walks
I will then be asked to walk on a treadmill at a self-selected comfortable walking speed for 6 minutes without an orthosis and then with 2 different types of orthoses.
I will be given time to get used to walking on the treadmill prior to starting the test
When on the treadmill I will be asked to place a mouthpiece in my mouth which will measure the amount of oxygen I am using
Again I will be given ample time to rest during these tests

VI. Sometimes there are problems associated with this type of study. These are:
1. I may find that wearing a mouthpiece and a nose clip uncomfortable especially when walking, but I understand that it will only be for a short time, and that they will not interfere with my ability to walk.
2. I may find walking on a treadmill unusual, especially if I have never walked on one before. However I will be given ample time to get used to this and I can withdraw from this study at any time if I wish to do so.
4. If there are any adverse effects during my visit, I will be monitored until the effects pass.

VII. Their maybe benefits to me from this treatment. These are:
1. I will receive a report on my walking action, which I can keep as a baseline value of my normal walking pattern.

VIII. My confidentiality will be guarded. Dublin City University will make reasonable efforts to protect the information about me and my part in this study and no data will be published. This will be achieved by assigning me an ID number against which all data will be stored. Details linking my ID number and name will not be stored with the data. The results of the study maybe published and used in further studies.
IX. If you have any questions about the study, I am free to call Dr. Kieran Moran at (01) 7008011 or Sharon Kinsella at (0503) 76218

Taking part in this study is my decision. If I do agree to take part, I may withdraw at any point including during the exercise test. There will be no penalty if I withdraw before I have completed all stages of the study.

XI. Signature:
I have read and understood the information in this form. My questions and concerns have been answered by the researchers, and I have a copy of this consent form. Therefore, I consent to take part in this research project entitled 'The effect of wear of Ankle Foot Orthosis on gait'

Signature:________________________________ Date:_____________

Witness:___________________________ Witness:_______________________
Signiture Printed name

## Health questionnaire form (PR01)

Name____________________________

Date:____________________________

Age____________________________

Weight____________________________

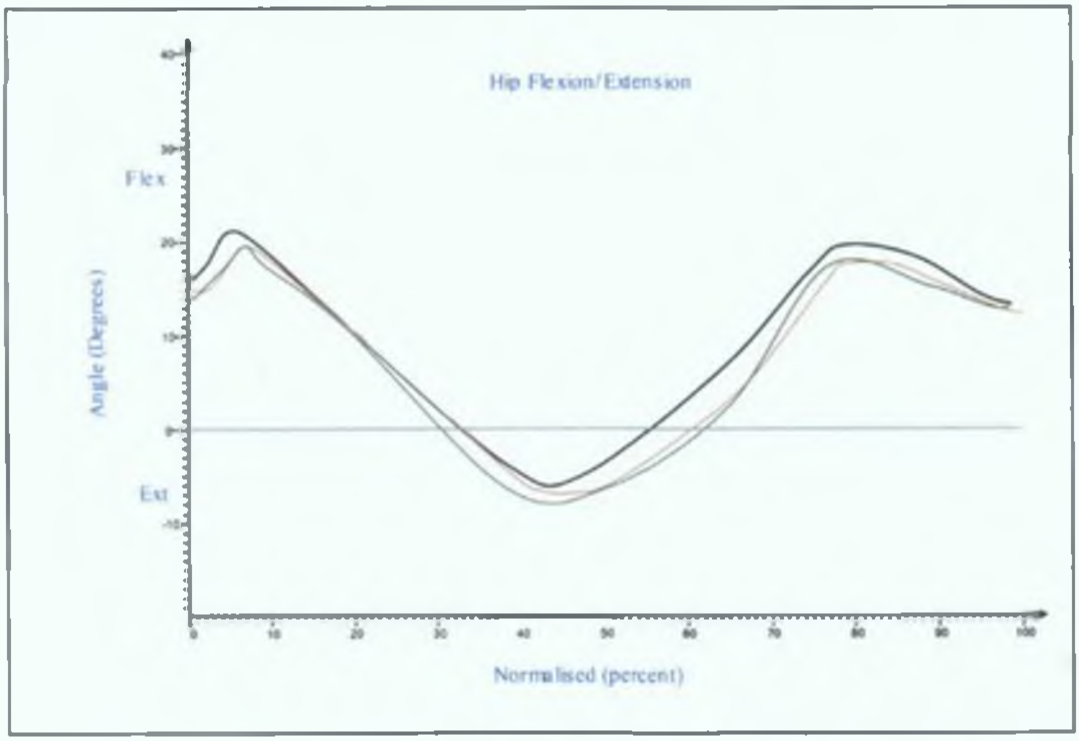
Please tick the appropriate boxes, answering all questions

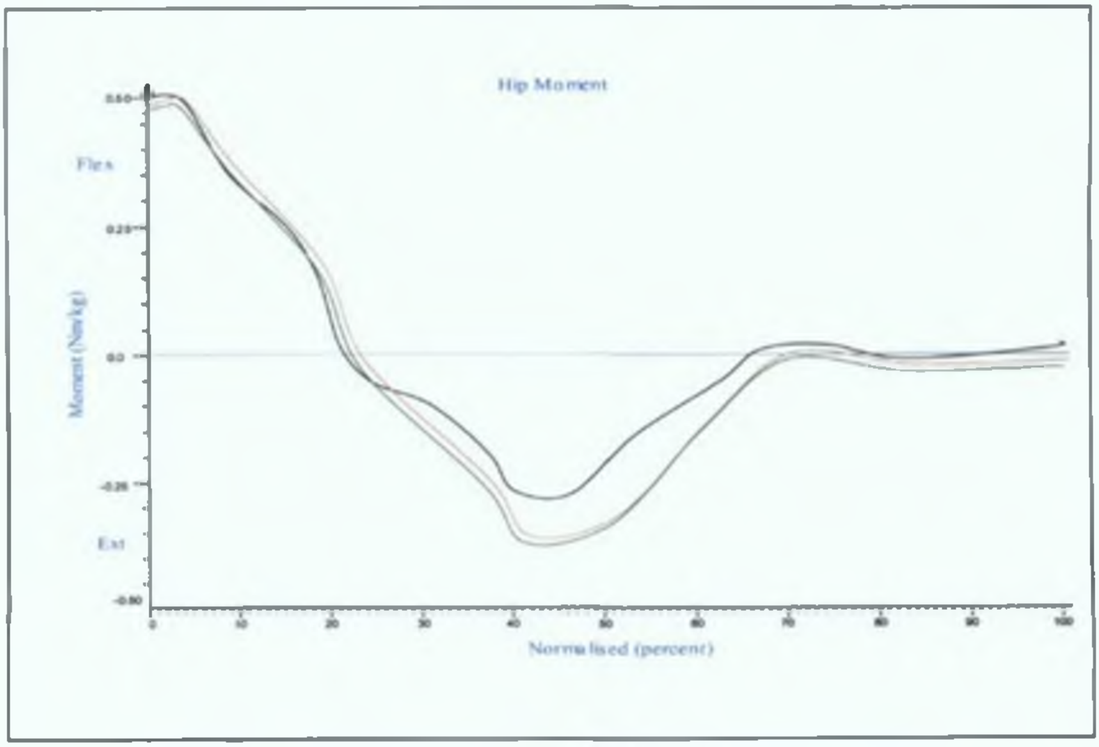
| | YES | NO |
|---|---|---|
| I am currently in good health | ☐ | ☐ |
| I can walk 10 meters at my own pace, with the help of aids | ☐ | ☐ |
| I have had my knee or hip joints replaced | ☐ | ☐ |
| I do have severe arthritis in my knees or hips | ☐ | ☐ |
| I do get excessively short of breath when I walk short distances | ☐ | ☐ |
| I presently have injuries to my legs that might affect my ability to walk | ☐ | ☐ |
| I do suffer from Diabetes Mellitus | ☐ | ☐ |
| I do suffer from Renal and Kidney dysfunction | ☐ | ☐ |
| I do suffer from High blood pressure | ☐ | ☐ |
| I do suffer from any of the following | ☐ | ☐ |

1. Metabolic disorders
2. Acute Myocardiac infarction
3. Uncontrollable cardiac arrhythmias,
4. Active endocarditis,
5. Symptomatic severe aortic stenosis,
6. Acute Pulmonary

Signature:____________________________

**Appendix C** – Kinematic and kinetic graphs of Sgroups wearing AFOs

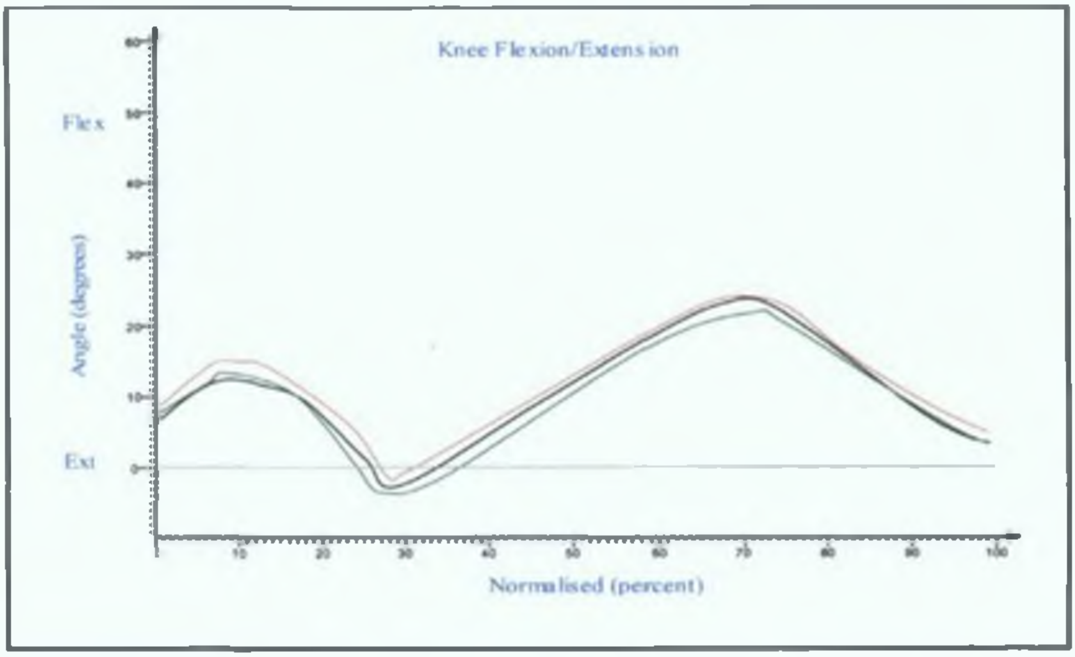


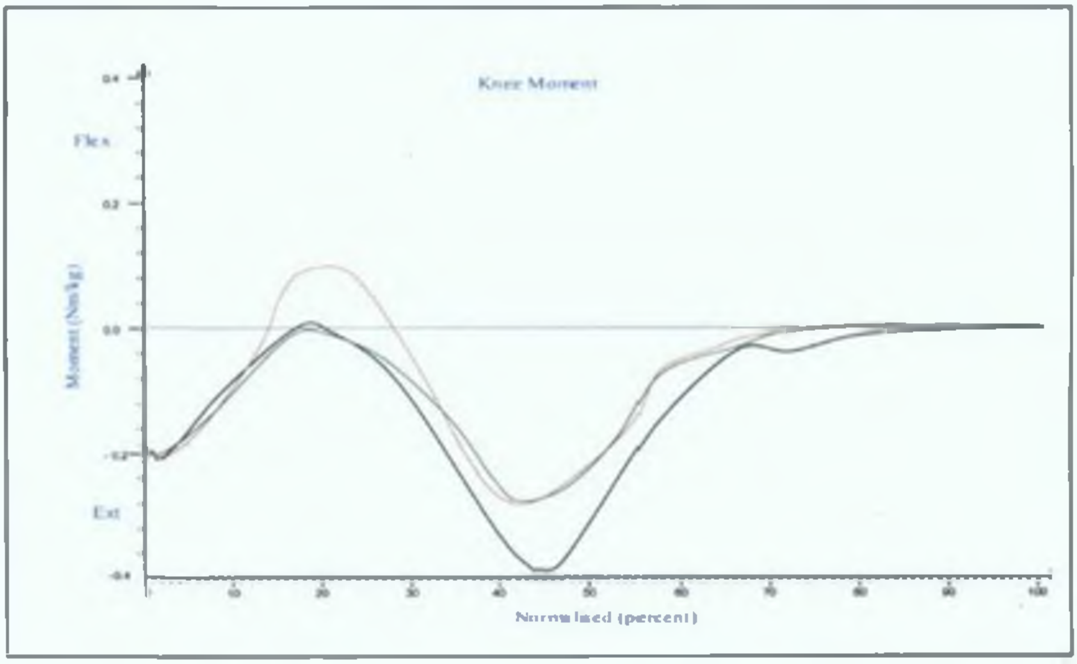



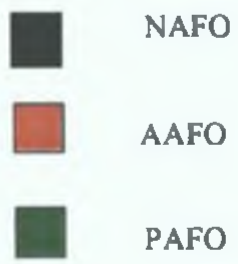

NAFO

AAFO

PAFO


Knee Flexion/Extension
Flex
Ext
Angle (degrees)
Normalised (percent)


Knee Moment
Flex
Ext
Moment (Nm/kg)
Normalised (percent)


NAFO
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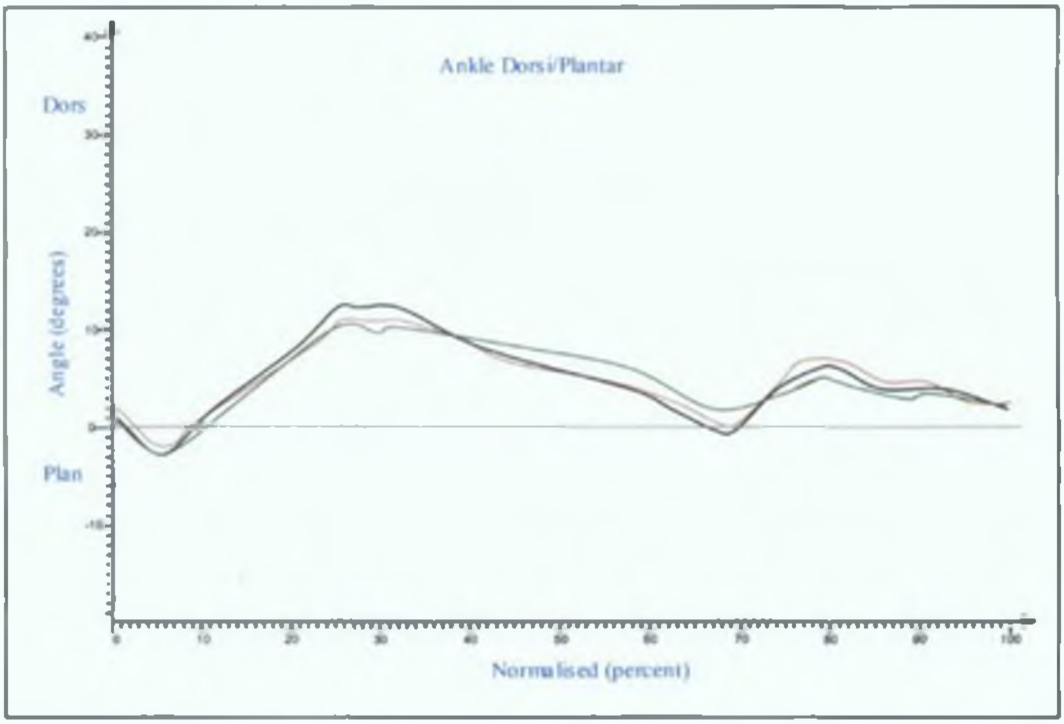
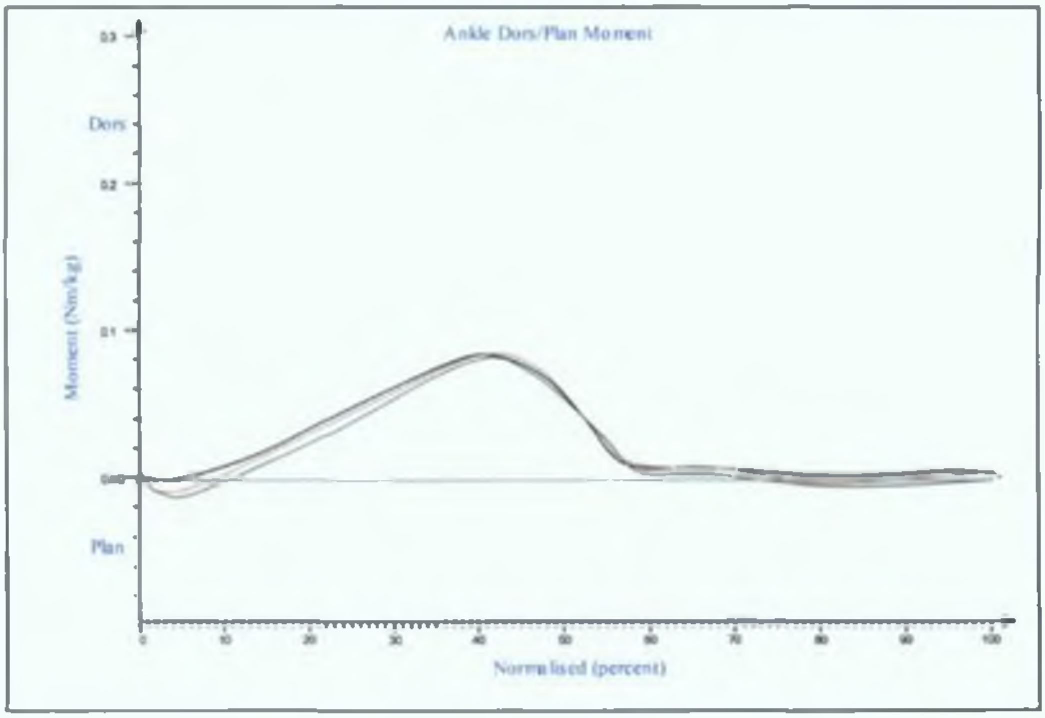
PAFO

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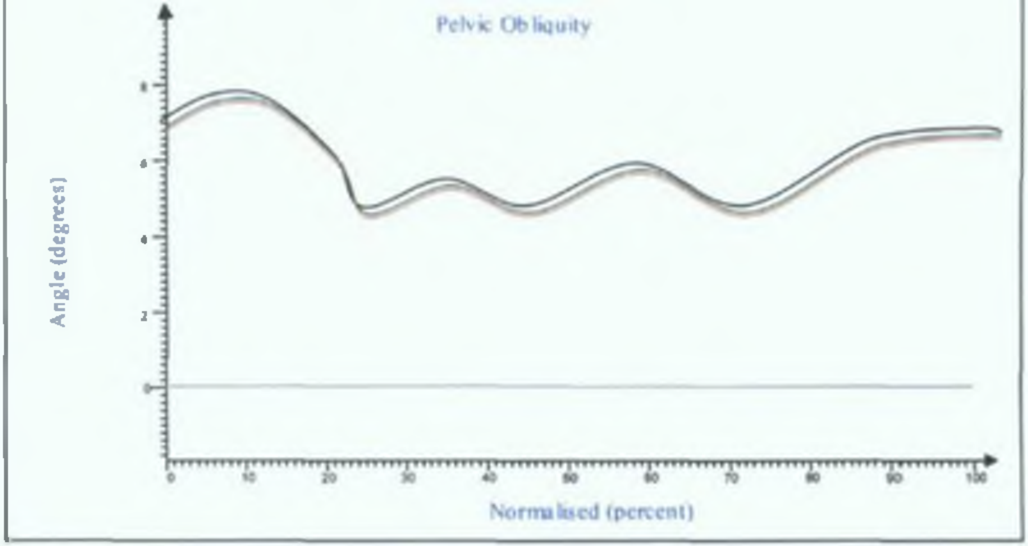


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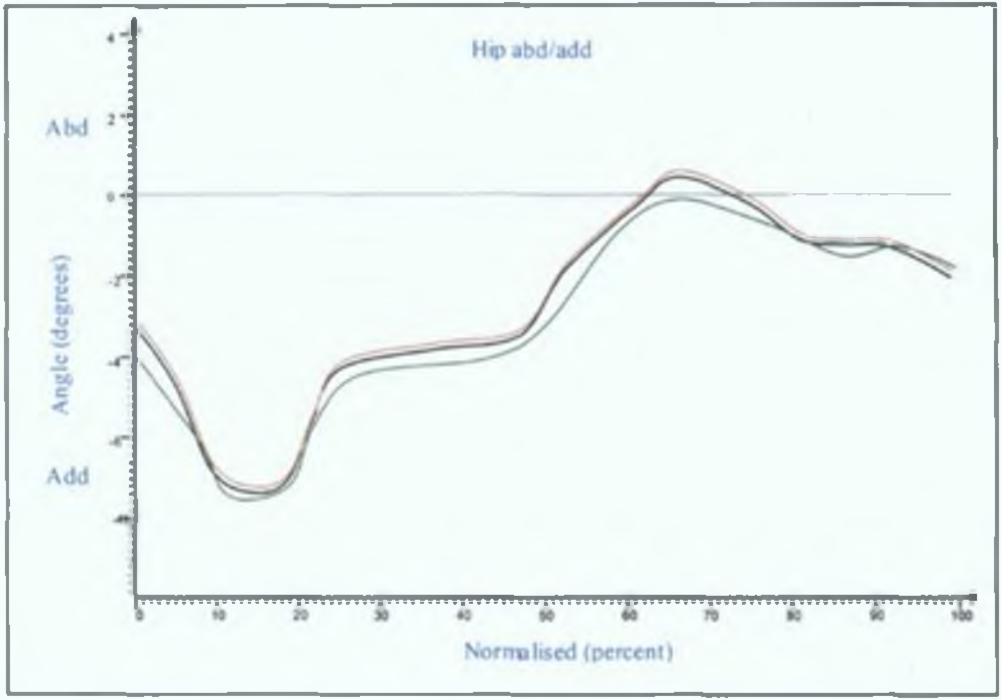
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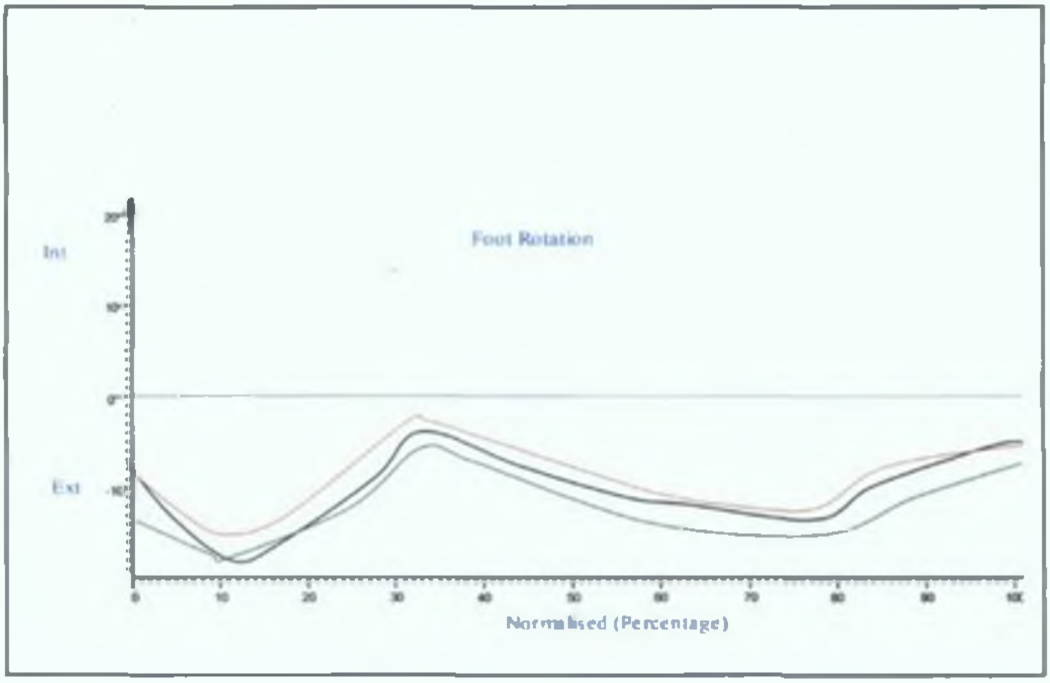


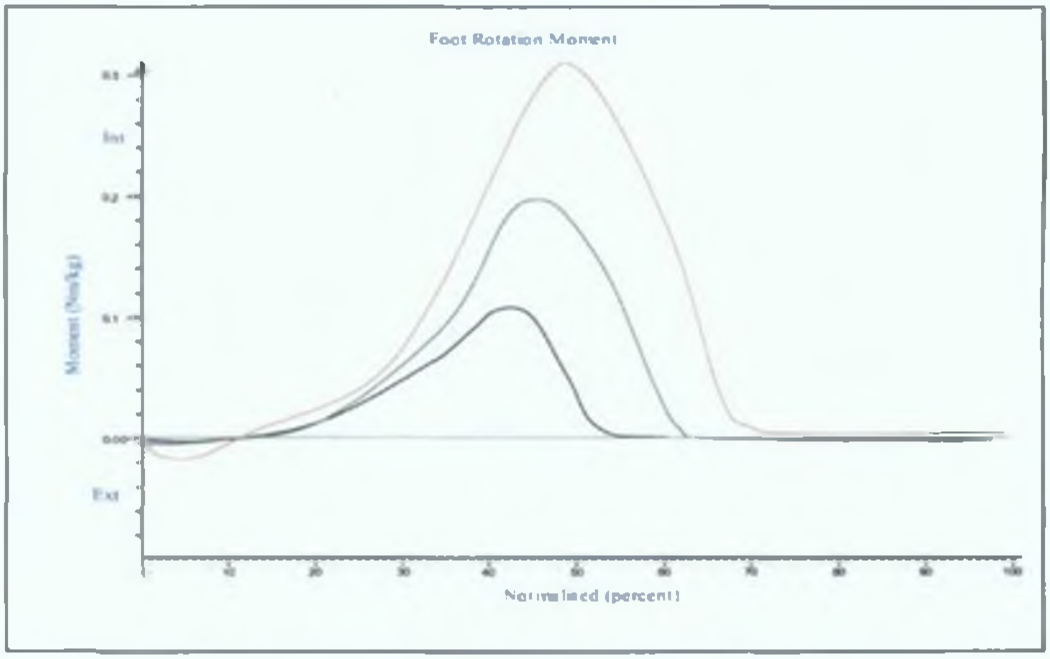


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## **Appendix D** –Publications directly related to the work from this thesis

**Poster presentations**

Kinsella, S., Moran, K. A case study examining the effects of different types of AFO on an adult hemiplegic subject. ISEK Congress, June 18-21, 2004, Boston, MA, USA.

Kinsella, S., Moran, K. The effects of different types of AFO on hemiplegic gait. ISEK Congress, June 18-21, 2004, Boston, MA, USA.

**Oral presentation**

Kinsella, S., Moran, K. The effects of different types of AFO on oxygen consumption. ISEK Congress, June 18-21, 2004, Boston, MA, USA.

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