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THE EFFECT OF ANKLE-FOOT ORTHOSIS ON GAIT AND THE PROCEDURES USED TO ASSESS BI-ARTICULAR MUSCLE LENGTH AND SPASTICITY IN SUBJECTS WITH DIPLEGIC CEREBRAL PALSY

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Abstract

Ankle-foot orthoses are commonly prescribed amongst subjects with diplegic cerebral palsy as a conservative orthotic intervention. Multi-articular and bi-articular muscles are more severely involved in subjects with diplegic cerebral palsy. Appropriate clinical assessment including assessment of these muscles enables optimum ankle-foot orthosis footwear combination (AFO-FC) prescription and any required adjunct therapy.

The overarching aim of this thesis was to investigate the effect of optimisation of the temporal midstance shank kinematics with the use of an AFO- FC on several variables including shank kinematics, thigh kinematics, vertical component of ground reaction force (FZ2) and ground reaction force (GRF) alignment in relation to hip and knee joint in temporal midstance to late stance. The understanding of the implication of, and the tools used to conduct assessment procedures and measurement processes were tested in the assessment of bi-articular muscles (specifically rectus femoris), to ascertain if results were predictive of presenting gait deviations and hence provide information to optimise treatment.

The results of initial study suggested that a positive influence on the shank kinematics, thigh kinematics, FZ2 and GRF alignment in relation to hip and knee joint in temporal midstance to late stance was observed with the use of an AFO-FC. The results of a further study illustrated that a dominance relationship of the catch angle/length of rectus femoris (RF) measured using the Duncan-Ely with the knee or the hip joints varies at different gait points/periods. Additionally, the effect of dynamic shortness of

the RF differed on gait from the effect of true shortness. The results of series of studies testing the feasibility of using a 2-dimensional analysis system (*PnO Clinical Movement Data)* for hip, knee and ankle sagittal plane passive joint range of motion measurement during physical assessment confirmed the reliability and accuracy of this system.

Summary

Cerebral palsy (CP) is the most common physical disability in childhood.[1] This non-progressive neuromuscular disorder leads to the development of abnormal muscle tone, diminished selective motor control and impaired proprioception affecting gait.[2] One of the frequent manifestations of abnormal muscle tone amongst subjects with CP is spasticity.[3]

Spastic diplegia is a form of CP in which the most prominent motor impairment is observed in the lower limbs. It may also include to a lesser extent the trunk, face and upper limbs.[4] Determining intervention in subjects with spastic diplegia may be challenging due to the heterogeneous nature of this impairment and the asymmetrical involvement in both lower limbs.[5]

Ankle-foot orthoses (AFOs) are commonly prescribed amongst subjects with diplegic CP as a conservative orthotic intervention. An AFO is an externally applied device that encompasses the ankle joint and the whole or part of the foot.[6] The orthotic device works in conjunction with the footwear (ankle-foot orthosis footwear combination) in an attempt to exert control over and optimise shank kinematics. Appropriate shank kinematics/inclination during temporal midstance is vital to produce required temporal midstance stability which leads to energy conservation and reduces the demand on musculoskeletal system (Figure 2-1).[7-10] Additionally, optimising shank kinematics during temporal midstance is deemed to increase the possibility of improved knee/hip kinetics and thigh/trunk kinematics.[11] As the shank inclines further and the thigh reaches its maximum inclination during terminal stance, the ground reaction force

(GRF) lever arm at the hip and knee increases, which in combination with the large magnitude of the GRF produces a strong stabilising external moment at the hip and the knee joints (Figure 2-3).[8, 12] Additionally, during terminal stance the second peak of magnitude of vertical component of the GRF (FZ2) occurs exceeding the bodyweight which allows the body to support its own weight and produce the required propulsion forces (Figure 1-2 & 2-2).[12, 13] Successful orthotic intervention and control of shank kinematics is considered to be enhanced by an intimately fitted and optimally aligned ankle-foot orthosis footwear combination (AFO-FC). While AFO-FC is frequently used amongst subjects with diplegic CP for various applications, the evidence supporting their efficacy especially in affecting the hip and knee joint is found inconclusive.[14]

Detailed clinical assessment including physical assessment enhances the development of the optimum intervention plan. Orthotic intervention should be designed in a way which considers all presenting impairments including length, strength and stiffness of the muscles. Multi-articular and bi-articular muscles, specifically hamstring, rectus femoris (RF) and gastrocnemius, are more severely involved in subjects with diplegic CP.[4, 15, 16] These muscles are commonly truly or dynamically/functionally short. Functional shortness is caused by spasticity, which is a velocity-dependent increase in muscle tone observed as resistance (spastic catch) to passive movement.[17] Spastic catch is measured in degrees (catch angle) by measuring the available joint range of motion (ROM). Appropriate assessment of these muscles enables optimum AFO-FC prescription and any required adjunct therapy.

The overarching aim of this thesis was to investigate the effect of optimisation of the temporal midstance shank kinematics with the use of an AFO- FC on several variables including shank kinematics, thigh kinematics, FZ2 and GRF alignment in relation to hip and knee joint in temporal midstance to late stance. The understanding of the implication of, and the tools used to conduct assessment procedures and measurement processes were tested in the assessment of bi-articular muscles (specifically rectus femoris), to ascertain if results were predictive of presenting gait deviations and hence provide information to optimise treatment. A series of studies were completed in this thesis to fulfil the overall aim.

The initial study investigated the effect of optimising temporal midstance shank kinematics/shank inclination to vertical (SVA) with the use of an AFO-FC on thigh kinematics in temporal midstance and on thigh and shank kinematics at FZ2 and at maximum thigh inclination to vertical (TVA). The effect of optimisation of temporal midstance shank kinematics on FZ2 and on the alignment of the GRF in relation to the hip and knee joints in temporal midstance, at FZ2 and at maximum TVA was also examined.

The results of this study suggested that a positive influence on the shank kinematics (at FZ2 and maximum TVA) and thigh kinematics (in temporal midstance, at FZ2 and at the maximum TVA) was observed with the use of an AFO-FC (Table 2-3). Additionally, a positive increase in FZ2 was found in 9 participants from a total of 16 participants (Table 2-9). The use of an AFO-FC caused reduction in the degree of Ben Lomonding in 9 participants (Table 2-9). The term “Ben Lomonding” is used to

describe the situation where the second peak of magnitude of vertical component of GRF (FZ2) is smaller than the first peak of magnitude of the vertical component of GRF (FZ1).[18] With AFO-FC use, FZ2 was found to be greater than or equal to the body weight in 5 out of 9 participants (Table 2-9). Some effect was observed on improving the GRF alignment in relation to the hip and knee joints in temporal midstance, at FZ2 and at maximum TVA following optimisation of temporal midstance shank kinematics (Table 2-9) (Figures 2-10 to 2-12).

A further analysis of this study aimed to investigate if the examined gait variables in temporal midstance, at FZ2 and at maximum TVA, improved when the temporal midstance shank kinematics are optimised using an AFO-FC. The correlation values in limbs with barefoot temporal midstance shank inclination less inclined than normal and normal (SVA≤12°) indicated that amongst all the variables investigated, the alignment of the GRF at the knee joint in temporal midstance and at the hip joint at FZ2 may optimise simultaneously when the temporal midstance shank kinematics are optimised (Table 2-7). The correlation values in limbs with barefoot temporal midstance shank inclination more inclined than normal (SVA>12°) suggested that amongst all the examined variables, FZ2, thigh inclination and GRF alignment in relation to the knee joint at the maximum TVA may optimise simultaneously when the temporal midstance shank kinematics are optimised (Table 2-8). Additionally, correlation values suggest that the initial barefoot temporal midstance shank inclination affects the results found using an AFO-FC.

Furthermore, agreement across the variables investigated including the shank kinematics, thigh kinematics and FZ2, was found in 2 participants. Another, two participants had agreement of 5 out of 6 variables (Table 2-10).

A major limitation found in this study was that 9 participants exhibited an increase in SVA during the gait which may explain some of the negative results found, e.g. GRF alignment in relation to the hip and the knee joints (Table 2-12). Possible causes for further shank inclination include AFO being insufficiently stiff/too flexible, footwear characteristics and un-accommodated gastrocnemius length. The results of this study emphasised the importance of choosing the appropriate AFO-FC characteristics followed by a tuning process to achieve the optimum temporal midstance shank kinematics.

Although this thesis focused on a single muscle (RF) to highlight the key findings of the assessment, it provides a research methodology that may be adopted for other bi-articular muscles. The RF muscle was selected for the assessment after the literature review, which showed that the exact timing reported of muscle activation of RF during gait is inconsistent.[19-23] Additionally, the exact effect of the shortness and spasticity of the RF muscle on gait is vague.[24-30] The Duncan-Ely test is commonly used to measure the length and spasticity of the RF.[31, 32] However, limited studies were found evaluating the Duncan-Ely test and explaining how results obtained from this test link to the deviations observed in gait. Furthermore, the hip position, which is vital, was not considered in the studies found. Further research is required to understand the influence of the catch angle/length measured using the Duncan-Ely test on gait.

This thesis evaluated the current assessment method, the Duncan-Ely test, and the tools, the goniometers used to measure RF length and spasticity. A further study of this thesis aimed to examine the relationship between the catch angle/length measured using the Duncan-Ely test and the hip or the knee joint position at selected points/periods of gait as follows: maximum hip extension in stance, early swing, peak knee flexion in swing and peak hip flexion in swing. Additionally, this study examined if a dominance effect was found with the hip or the knee joint at the selected points/periods of gait. The relationship between the catch angle/length of the RF measured using the Duncan-Ely test and the timing of peak knee flexion in swing was also examined.

In limbs with catch angle/length<60°, the regression models between the catch angle/length and the knee joint at maximum hip extension, peak knee flexion and peak hip flexion were found to be weak but significant (Table 3-2 & 3-3). This may indicate that a dominance relationship of the catch angle/length was found with the knee joint at these points. In contrast, in early swing a dominance relationship was found between the catch angle/length and the hip joint in limbs with catch angle/length<60°. In limbs with catch angle/length≥60°, the regression model between the catch angle/length and the knee at peak hip flexion was found to be significant, suggesting that the knee joint exhibited a dominance relationship with the catch angle/length at this point (Table 3-2 & 3-3). Furthermore, the regression models for the timing of peak knee flexion with the catch angle/length were found to be weak and not significant (Table 3-9).

A further aim of this study was to investigate if the effect of dynamic shortness of the RF on gait differed from the effect of true shortness of the RF, and if the dominant relationship was found with the hip or the knee joint. In limbs with catch angle<60°, a dominance relationship of the catch angle was found with the hip joint in early swing and with the knee joint at peak knee flexion and peak hip flexion (Table 3-11 & 3-12). In contrast, in limbs with length<60° a dominance relationship was found with the hip joint at peak knee flexion (Table 3-19 & 3-20). In limbs with catch angle≥60°, a dominance relationship was found with the knee joint in early swing, with the regression value being good and significant (Table 3-11 & 3-12). In limbs with length≥60°, all the regression models were found to be weak and not significant (Table 3-19 & 3-20).

This study illustrated that a dominance relationship of the catch angle/length with the knee or the hip joint varies at different gait points/periods. This suggests that the RF can influence the knee or the hip joint. Additionally, the effect on gait of dynamic shortness of the RF differed from the effect of true shortness of the RF.

Currently, different designs of goniometer are used for measuring passive joint ROM and bi-articular muscle length. These designs include the universal goniometer (UG), electrical goniometer (EG) and inclinometer. A literature review was performed to evaluate the reliability of different designs of goniometer. This review aimed to investigate the intratester and intertester reliability of the UG, EG and inclinometer to measure hip, knee and ankle joint ROM. In addition, the review aimed to examine how different factors influence measurement reliability. The review highlighted that the

evidence for the reliability of these tools is largely inconclusive, especially amongst subjects with diplegic CP (Tables 4-1 to 4-4).[33] Main identified causes of measurement error using the goniometer amongst subjects with diplegic CP included the presence of spasticity, defining the end ROM of the joint and stabilising the limb especially while measuring bi-articular and multi-articular muscle length and spasticity. Additionally, it has been recommended that the measurements obtained using the UG should be applied with caution for assistance in clinical decision making.[33]

The feasibility of using a 2-dimensional analysis system (*PnO Clinical Movement Data (PnO CMD))* for passive joint ROM measurement during physical assessment was therefore investigated. Initially, a study investigating the reliability of the *PnO CMD* and the UG in the measurement of sagittal plane hip, knee and ankle joints passive ROM amongst healthy subjects was performed. The highest intratester and intertester ICC values were found for the *PnO CMD* with markers, which confirmed the reliability of this system and the measurement procedure used (Table 4-5). Following that, the accuracy of the *PnO CMD* with markers was tested in comparison to Vicon (benchmark) amongst healthy subjects. High ICC values were obtained using this system for measurement of hip, knee and ankle sagittal plane dynamic ROM measured at 5 predefined points of gait (Table 4-8). Finally, the reliability of the *PnO CMD* for measurement of joint ROM amongst subjects with diplegic CP was investigated. Intratester and intertester ICC values for hip, knee, and ankle joint sagittal plane passive joint ROM with markers were found to be high (Table 4-9). The results of this study confirm that reliable measurements can be obtained using this system amongst subjects with diplegic CP.

Abbreviations

2-dimensional	2D
3-dimensional	3D
Adjustable response ankle-foot orthosis	ADR-AFO
Ankle angle of the ankle-foot orthosis	AAAFO
Ankle-foot orthosis	AFO
Ankle-foot orthosis footwear combination	AFO-FC
Body weight	BW
Botulinum toxin type A	BoNT-A
Cerebral palsy	CP
Confidence interval	CI
Dynamic ankle-foot orthosis	DAFO
Electrical goniometer	EG
Footwear	FW
Gross Motor Function Classification System	GMFCS
Ground reaction ankle-foot orthosis	GRAFO
Ground reaction force	GRF
Ground reaction/floor reaction ankle-foot orthosis	GRAFO/FRAFO
Heel sole differential	HSD
Hinged ankle-foot orthosis	HAFO
Hinged/articulated ankle-foot orthosis	HAFO
Hip at peak knee flexion	HPKF
Intraclass correlation coefficient	ICC
knee at maximum hip extension	KMHE
Knee at peak hip flexion	KPHF
Leg length discrepancy	LLD
Maximum hip extension in stance	MHE
Peak hip flexion in swing	PHF

Peak knee flexion in swing	PKF
PnO Clinical Movement Data	*PnO CMD*
Posterior leaf spring	PLS
Range of hip flexion in early swing	RHFES
Range of knee flexion in early swing	RKFES
Range of motion	ROM
Rectus femoris	RF
Scottish Intercollegiate Guideline Network	SIGN
Selective dorsal rhizotomy	SDR
Shank to vertical angle	SVA
Solid ankle-foot orthosis	SAFO
Solid/rigid ankle-foot orthosis	SAFO
Standard deviation	SD
Standard error of measurement	SEM
Stiff-knee gait	SKG
Surveillance of Cerebral Palsy in Europe	SCPE
Thigh to vertical angle	TVA
Timing of the peak knee flexion in swing	TPKF
Trochanteric prominence angle test	TPAT
Universal goniometer	UG

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Chapter 1 Introduction and background

1.1 Thesis overview

Initially, Chapter One will introduce the definition, incidence and classification of cerebral palsy (CP), concentrating on diplegic CP. Following that, Chapter One will also provide an insight to the normal and pathological biomechanics of the gait. Different designs and biomechanical effects of ankle-foot orthoses (AFOs) will then be discussed. Additionally, the importance of clinical assessment and several elements of prescription criteria will be reviewed. Finally, a summary of the evidence supporting the biomechanical efficiency of an ankle-foot orthosis footwear combination (AFO-FC) on lower limb joints in this population will be provided.

Chapter 2 focuses on the effect of an AFO-FC on gait. Initially, the importance of optimising shank kinematics on gait will be discussed. This is followed by a presentation of results of a study investigating the effect of optimisation of temporal midstance shank kinematics with the use of an AFO-FC on several variables including shank kinematics, thigh kinematics, FZ2 and ground reaction force (GRF) alignment in relation to hip and knee joint in temporal midstance to late stance.

In Chapter Three, the role of the rectus femoris (RF) during normal gait will be discussed. This chapter will then examine literature related to the efficiency of the Duncan-Ely test. Finally, the results of research investigating the relationship between the dynamic/true shortness of the RF and the hip or the knee joint at selected points/periods of gait will be presented.

Chapter Four aims to evaluate the commonly used measurement tools for joint range of motion (ROM) measurements and bi-articular muscle length. Additionally, this chapter aims to test the feasibility of a 2-dimensional (2D) video analysis system *PnO Clinical Movement Data* (*PnO CMD*) for passive joint ROM measurement during physical assessment. Initially, the results of a literature review evaluating the reliability of different designs of goniometer will be presented. Furthermore, in this review different factors influencing the reliability of goniometers will be discussed. Following that, the results of studies investigating the reliability and accuracy of the *PnO CMD* amongst healthy subjects will be demonstrated. Finally, the results of a reliability study of the *PnO CMD* amongst subjects with diplegic CP will be presented.

The final chapter provides an overview of the results and implications of the findings. Additionally, the limitations of this research will be presented. Future avenues of research will be discussed, followed by the overall conclusions of the thesis.

1.2 Definition and incidence

CP refers to a group of heterogeneous and permanent disorders of motor function caused by either a static lesion or abnormality to the brain.[34] CP is an upper motor neuron lesion which affects approximately 2-2.5 per 1000 live births, with the ratio of affected males to females being 1.4:1.[1, 35, 36] CP is considered to be the most common cause of significant physical impairment in childhood[1] and is defined by Rosenbaum et al. as "a group of permanent disorders of the development of movement and posture, causing activity limitation, that is attributed to non-progressive disturbances that occurred in the developing fetal or infant brain. The motor disorders of cerebral palsy

are often accompanied by disturbances of sensation, perception, cognition, communication, and behaviour, by epilepsy, and by secondary musculoskeletal problems."[37]

Due to the complex nature of CP, a wide variety of classification systems have been developed and implemented. The categories for classification systems include the type of motor impairment, topography and function. The system suggested by the Surveillance of Cerebral Palsy in Europe (SCPE) classifies the subjects according to the type of motor impairment presented as spastic bilateral (53.9%) and unilateral (31%), dyskinetic (6.6%) and ataxic (4.1%).[38, 39] In contrast, the topographical classification categorises subjects with CP according to the distribution of limb involvement, which includes hemiplegia (unilateral), diplegia (upper limbs less affected than lower limbs) and quadriplegia (all limbs).[5]

The Gross Motor Function Classification System (GMFCS), which evolved from The Gross Motor Function Measure, is used to measure the functional ability in subjects with CP.[40, 41] This classification consists of five functional levels in five different age bands (<2y, 2–4y, 4–6y, 6–12y, 12–18y). Sellier et al.[42] included 12 subjects with CP (2 to 6 years) and 31 testers (clinicians and professionals) to investigate the intertester reliability of the GMFCS. The study found excellent intertester reliability (intraclass correlation coefficient (ICC=0.80-0.88)). Another study by Nordmark et al.[43] included 15 testers (physiotherapists) and reported high intratester (ICC=0.68) and intertester (ICC=0.77-0.88) reliability of the GMFCS amongst subjects with CP. It has been suggested that the use of this system has a positive influence on the management of

the condition.[44] This system is used to enhance communication amongst clinicians, increase focus on function, aid in realistic goal setting and intervention planning and enable evidence-based practice and resource allocation.[44]

1.3 Spastic diplegic cerebral palsy

CP is considered to be a progressive neuromuscular impairment which causes changes in soft tissues (muscles, tendons and ligaments) with growth.[17, 34, 45] Spasticity is the most common tonal abnormality associated with CP.[3] Spasticity can be defined as 'a motor disorder characterised by a velocity-dependant increase in tonic stretch reflex, with exaggerated tendon jerks resulting from hyperexcitability of the stretch reflex, as one component of the upper motor neuron syndrome'.[46] The clinical presentation of spasticity is the velocity-dependant increase in muscle resistance to the passive movement (spastic catch).[17] As the rate of muscle stretch increases, the resistance to movement felt in the muscles increases.[47] Spasticity can affect function in several ways as it acts as a 'brake' during motion, thus increasing the energy consumption. It inhibits voluntary control of the muscles, it prevents normal muscle lengthening during activity, affecting muscle growth, leading to development of contractures, and causing excessive torques on long bones, leading to bone deformities in the growing skeleton.[4] Additionally, spasticity can cause pain due to over-activity of the muscles and the abnormal forces imposed. Different scales are used to assess the severity of the spasticity including: the Ashworth scale,[48] the Modified Ashworth scale (MAS)[49, 50] and Tardieu scale.[51]

Diplegic CP refers to the bilateral involvement of the limbs where the lower limbs are more affected than the upper limbs. In many subjects with diplegic CP, the involvement is not symmetrical, which means different presentation on one side in comparison to the other, hence making this a complex type.[5] Due to the nature of the lesion in subjects with spastic diplegic CP, the selective control impairment is less severe in the proximal portion when compared to the distal portion of the limb.[4] Selective control is fairly good at the hip and poor at the ankle and foot. Additionally, it has been demonstrated that the distal bi-articular and multi-articular muscles are affected more severely and primarily in comparison to the mono-articular and more proximal muscles.[4, 15, 16] This highlights the importance about investigating lower limb bi-articular and multi-articular muscles.

1.4 Gait classifications for spastic diplegic cerebral palsy

When treating gait abnormalities in subjects with diplegic CP, it is essential to distinguish between the primary, secondary and tertiary abnormalities. The primary gait abnormalities are due to the damage of the nervous system causing deficient selective motor control, abnormalities of balance and abnormal tone.[2] This leads to the occurrence of secondary anomalies of insufficient muscle growth and bone deformities. Tertiary anomalies occur as a compensation mechanism to allow the individual to cope with the primary and secondary abnormalities. The separation of the true pathology from the coping mechanism is mandatory to optimise gait. This is because when the true pathology is treated, the compensation responses will no longer be needed and will disappear.[2] However, some coping responses turn into habits, which may be harder to change and would require gait training.

Using classifications for gait is mandatory in the clinical setting to allow optimum communication between health professionals. Additionally, these classifications enhance communication with the patients and their families. The use of these classifications aids in determining the prognosis and treatment options. Being able to define homogeneous groups is useful in research because it helps to effectively and clearly discuss the research results.[52]

Gait classification system developed by Sutherland and Davids[53] is frequently cited and widely used for diplegic CP. This classification focuses on sagittal plane knee kinematics, identifying 4 patterns which include jump knee, crouch knee, stiff knee and recurvatum knee. This work was built on by Rodda et al.[54, 55] In the modified classification by Rodda et al.,[54, 55] ankle kinematics were used to differentiate between the groups, and all the lower limb joints in the sagittal plane were considered, i.e. hip, knee, ankle and pelvic joints. This gait classification system follows the commonly observed changes with age and intervention and highlights the dynamic relationship between joints (hip, knee, ankle and pelvic) and segments (shank and thigh).[54, 55] Although these gait classifications are useful tools, it has been suggested that due to the heterogeneous nature of CP each subject should be treated individually.[53] These classifications help clinicians recognise the most common pathological patterns, which, in combination with understanding subject's gait, should lead to the optimum treatment plan.[53]

1.5 Biomechanics of normal and pathological gait

1.5.1 Normal gait

Normal gait has been defined as ‘a highly controlled, coordinated, repetitive series of limb movements whose function is to advance the body safely from place to place with a minimum expenditure of energy’.[12, 56] The term ‘gait cycle’ describes the interval between the occurrence of any two identical events of the cycle, e.g. initial contact of the right foot to the initial contact of the same foot.[57, 58] Each cycle is divided into two phases: stance phase (60% of gait) and swing phase (40% of gait). The stance phase is further divided into five subphases: initial contact, loading response, midstance, terminal stance and preswing. The swing phase is further divided into three subphases: initial swing, midswing and terminal swing (Figure 1-1).[57, 59] Another method for describing gait is using the rockers of stance phase, which are based on the movement of the ankle and metatarsophalangeal joints.[13] Movement at the ankle for the first and second rockers and at the metatarsophalangeal joint for the third rocker facilitate the forward progression of the shank.[12, 59]

Three important functional tasks are accomplished during gait, which are weight acceptance in early stance, single limb support in mid-to-late stance and limb advancement during swing.[57, 59]Six variables have been outlined which influence the energy expenditure in continued walking, including pelvic rotation, pelvic tilt, lateral pelvic displacement, knee flexion in stance phase, foot interaction with the knee and ankle interference with the knee.[9, 60] In order to design an optimum intervention plan, it is essential to understand the kinematics and kinetics of normal gait. During gait, movements occur in all three anatomical planes: sagittal, coronal and transverse.

However, the description of gait below will focus on the sagittal plane motion in lower limb joints. This is because the largest movements during gait occur in the sagittal plane, in comparison with the transverse and coronal planes. Additionally, the information included in this thesis focuses on the sagittal plane motion and assessment in lower limb joints.

1.5.1.1 Kinematics of normal gait

Ankle

Initial contact is achieved by the heel, and the ankle is held in neutral dorsiflexion/plantarflexion position. Following that, the ankle starts to plantarflex, bringing the forefoot down onto the ground during loading response. From this position, the tibia advances forward over the foot, moving the ankle to approximately 10º dorsiflexion. In terminal stance, the ankle is held relatively rigid. Following that, the ankle plantarflexes rapidly to approximately 20º in preswing. During the swing phase, the ankle moves back to neutral position, and this position is maintained until the next initial contact.

Knee

From a position of a full/nearly full extension at initial contact, the knee starts to flex rapidly to approximately 18º during the loading response and early part of midstance. Following that, the knee starts to extend, reaching maximum extension at 40% of gait. Then, the knee starts to flex again, reaching peak knee flexion in early swing phase. From this position, the knee then extends again prior to the next initial contact.

Hip

From a flexed position at the initial contact, the hip starts to extend during stance, reaching a position of maximum 10º extension at 50% of gait. From this position, the hip starts to flex, reaching maximum flexion (35º) in the middle of the swing phase. This position is held constant until the next initial contact.

Segment kinematics

During the stance phase, the shank and the thigh move from a reclined position (leaning backward from the vertical) to an inclined position (leaning forward from the vertical). The position of these segments can be determined at any point of gait by measuring the angle of the segment relative to the vertical. The shank to vertical angle (SVA) is the angle of the shank relative to the vertical, and the thigh to vertical angle (TVA) is the angle of the thigh relative to the vertical. The angular velocity (the rate of change of angular position) of segmental movement differs between gait rockers and between the thigh and the shank.[12] As the limb approaches midstance, the angular velocity of the shank slows down as it moves into forward inclination, while the thigh moves from a reclined to an inclined position at a faster rate of angular velocity.[9, 61, 62] Midstance and terminal stance are challenging phases as they are single support phases, and many kinematic and kinetic changes occur during these phases. The slowing angular velocity of the shank with the optimum inclination of the shank (10-12º inclination) in temporal midstance is important for several reasons.[8] It aids in achieving the required stability in stance, it enhances smooth/ballistic movement of the thigh, pelvis and trunk, it determines thigh, pelvis, trunk and head kinematics, it

facilitates appropriate GRF alignment to the knee and hip, and it contributes to conservation of energy.[11, 12]

1.5.1.2 Kinetics of normal gait

Appropriate segmental and joint kinematics are vital for producing the required normal kinetics. The ground reaction force (GRF) has three important defining features , which are the point of application, magnitude and line of action.[12] The GRF produces external moments at each of the lower limb joints. The moments increase as the distance between the line of action of the GRF and joint centre increases (moment=force X distance).[63] The magnitude of the vertical component of the GRF has two peaks (FZ1 and FZ2) during gait when its exceeds bodyweight.[12] This enables the body to support its own weight and generate the required propulsion forces during early stance.[12, 13] The FZ1 is the initial peak of the vertical component of the GRF, which occurs during the loading response. This peak is followed by a reduction in downward force in midstance (FZ0). Then, the second peak (FZ2) occurs in terminal stance (Figure 1-2). The 'butterfly diagram' described by Pedotti[64] represents the magnitude, direction and point of application of the GRF at uniform time intervals (10 ms) of gait in the sagittal plane (in the direction of progression). This diagram is known as the 'butterfly diagram' because it resembles the shape of the wings of a butterfly (Figure 1-3).

Appropriate segmental alignment during gait places the GRF close to the joint centers, causing a reduction in the external moments created by the GRF. This decreases the opposing internal muscle moments required to stabilise the joints during gait.

Additionally, it lowers the biomechanical demand on the neuromuscular system, making gait efficient. In late stance, the external extension moments create a desirable effect at the hip and knee joints, minimising/removing the need for opposing internal moments. This joint stabilisation achieved by the external moments results in energy conservation. The important alignment of the GRF anterior to the knee joint and posterior to the hip joint in late stance is only achieved with the optimum shank and thigh inclination. Furthermore, only a narrow range of shank inclination with an optimum of 10-12º inclination during the temporal midstance will allow the required kinematics and kinetics changes to take place.[8, 12] During terminal stance, the ankle is held in a rigid position. This position of the ankle ensures that the point of application of the GRF is anteriorly located on the foot, aligning the GRF anterior to the knee. Additionally, this position allows heel raise, which causes further thigh and shank inclination, aligning the GRF posterior to the hip. This situation leads to the production of large external extension moments at the hip and the knee, creating the required stability during terminal stance. Furthermore, as the heel raises, the body centre of mass raises as well, causing the occurrence of FZ2.

Muscles

Muscles play an important role during gait in producing the required internal moments. In gait, these internal moments may be required to be greater or lower than the external moments to produce and control the angular movements of the segments. It is mandatory that these muscles are at the appropriate length, strength and stiffness to fulfil their function. These muscles may contract concentrically, eccentrically or isometrically and are neurologically active during appropriate times throughout the

gait. The muscle can be mono-articular (crossing one joint), bi-articular (crossing two joints) or mutli-articular (crossing more than two joints). Each muscle has a mandatory involvement during the gait, as outlined below.

1.5.1.3 Tasks of normal gait

Weight acceptance

This task is achieved during initial contact and loading response.[65]

Initial contact: initial contact by the heel is a crucial event. This heel contact is achieved by isometric contraction of the pretibial muscles, which establishes the heel rocker. The GRF is at the heel and passes anterior to the hip and the knee. The ankle is in neutral dorsiflexion/plantarflexion, and the knee is nearly fully extended, while the hip is flexed at approximately 30º.

Loading response: the hip begins to extend at this phase. The GRF is anterior to the hip, causing an external flexor moment, which is controlled by the opposing internal moment created by the concentric contraction of the gluteus maximus, hip extensor and hamstrings. The knee flexes during this phase. The GRF passes posterior to the knee and the ankle joints. These external moments are controlled by the opposing internal moments produced by the eccentric contraction of the quadriceps at the knee joint and the tibialis anterior at the ankle joint.

Stance limb progression

In the two phases associated with stance limb progression, the basic function achieved is the advancement of the limb over the supporting foot.

Midstance: the hip continues to extend. The GRF moves behind the hip and the muscle action (gluteus maximum and hamstring) ceases. Minimum concentric contraction is required from the quadriceps because the GRF is closer to the knee joint centre. The GRF is anterior to the ankle, and the external dorsiflexion moment is restrained by the action of the soleus muscle contracting eccentrically. As the limb progresses, the GRF moves anterior to the knee joint, and no quadriceps action is required.

Terminal stance: as the limb progresses, the GRF advances across the forefoot. The eccentric contraction of the soleus and the gastrocnemius virtually locks the ankle in a slightly dorsiflexed position. As the heel raises, it aligns the GRF anterior to the knee and posterior to the hip, causing stabilising external extension moments.

Limb advancement

The aim of the four remaining phases is to flex the limb for floor clearance and limb advancement.

Preswing: as the weight transfers to the other limb, the magnitude of the GRF reduces. Concentric contraction of the soleus and gastrocnemius muscles causes a rapid arc of ankle plantarflexion to 20º. This is followed by passive knee flexion in response to the posteriorly aligned GRF, which is restrained by the eccentric contraction of the rectus femoris if required.[66] The action of the flexor component of the adductor longus and

rectus femoris opposes the posterior hip external moment caused by the alignment of the GRF.

Initial swing: this phase involves total limb flexion. The concentric action of the iliacus muscle causes active flexion at the hip, while the active flexion at the knee is caused by the concentric action of the biceps femoris. Furthermore, the concentric contraction of the tibialis anterior initiates ankle dorsiflexion.

Midswing: further hip flexion occurs. The knee starts to extend passively while the persistent ankle dorsiflexion to neutral is achieved by the eccentric contraction of the pretibial muscles.

Terminal swing: at this phase, the hip flexion is controlled/stopped by a quick and intense eccentric contraction of hamstring muscles. The knee starts to extend passively while the hamstring intensity reduces to allow controlled knee extension and to prevent hyperextension. The ankle remains in a dorsiflexed position through the concentric contraction of the pretibial muscles.

1.5.2 Pathological gait

In pathological gait, one or more of the five prerequisites which are required for normal gait are commonly absent. These are stability in stance, sufficient foot clearance during swing, adequate step length, appropriate swing phase pre-position of the foot and energy conservation.[56, 67]

1.5.2.1 Kinematics of pathological gait

As discussed earlier, gait classification system initially developed by Sutherland and Davids,[53] and later modified by Rodda et al.,[54, 55] is commonly referenced for subjects with diplegic CP. Five patterns/groups are identified that describe the joint kinematics as follows:

Group I, true equinus: this pattern is classified by excessive ankle plantarflexion over the majority of the gait. Additionally, increased knee flexion is observed in terminal swing and at initial contact. In midstance, the knee is in full extension/hyperextension. The hip achieves the required range of extension in terminal stance.

Group II, jump gait: excessive ankle plantarflexion is observed over most of gait. Additionally, in this pattern increased knee flexion in terminal swing and at initial contact is found. The knee and hip are excessively flexed in early stance and then extend to a variable degree in late stance but never achieve the required range of extension.

Group III, apparent equinus: in this pattern, the ankle ROM over the entire gait is within normal limits. The pattern of the knee and hip are like the jump gait.

Group IV, crouch gait: this pattern is characterised by excessive ankle dorsiflexion and excessive hip and knee flexion over the entire gait.

Group V, asymmetric gait: this gait pattern represents asymmetric gait with a combination of issues; for example, the right limb is classified as apparent equinus while the left lower limb is classified as jump gait.

Segments

Inappropriate segment kinematics or angular velocity of movement are frequently present in a pathological gait.[12, 68] The segments may fail to accelerate or decelerate during the stance phase. In addition, segments may be excessively or insufficiently inclined.[7, 69, 70] If the shank kinematics are altered, it will compromise the stability required during mid-to-late stance, causing secondary compensating abnormal kinematics of the proximal segment, i.e. the thigh.[12] Inability to achieve an inclined position is the most common deviation found affecting the thigh.[12] Owen[71] classified gait deviation associated with neurological conditions such as CP into 3 categories. These categories include insufficient tibial/shank inclination, excessive tibial/shank inclination and absence of suitable external hip extension moment coupled with excessive or insufficient tibial/shank inclination.[71, 72]

1.4.2.2 Kinetics of pathological gait

When looking at the kinetics of the pathological gait, one or all components of the GRF can be abnormal. For example, a common deviation observed in CP is impaired plantarflexion–knee extension couple.[73] In normal gait, the eccentric action of plantarflexion muscles controls the progression of the GRF.[73, 74] This is achieved by regulating the rate and degree of dorsiflexion and shank inclination. In normal gait, the GRF progresses until it aligns anterior to the knee, causing an appropriate external

moment and knee extension, which is described as plantarflexion–knee extension couple.[73, 74] Reduced plantarflexion–knee extension couple is frequently seen with crouch gait.[75] This is caused by increased tibial inclination, which leads to the GRF vector moving from anterior to the knee to behind the knee, causing knee flexion.[73]

1.6 Clinical assessment

Comprehensive clinical assessment is the primary step in the preparation of treatment plans. Additionally, it is mandatory to accurately assess outcomes of treatment. The medical history is the initial step of the clinical assessment where information is collected about the birth history, developmental milestones, medical problems, surgical history, current and past physiotherapy treatment, the past and current orthotic treatment and medication.[76] This is followed by the physical and neurological examination, which aims to determine the strength and selective motor control of isolated muscle groups, assess the degree and nature of muscle tone, estimate the extent of static deformity and/or muscle contracture, examine bone deformities, assess fixed and mobile foot deformities and evaluate balance, equilibrium responses and standing posture.[47] The clinical assessment, including physical examination and gait analysis, helps to form a full picture of the orthopaedic and neurological challenges presented in the subject. The following section will discuss different parts of the physical examination.

1.6.1 Muscle strength

As discussed above, muscles have an important role during gait. Muscles produce internal moments required to stabilise the joints. Evaluation of muscle strength is

mandatory to ensure optimum clinical outcomes.[47] Muscle strength is defined as the greatest amount of force a muscle can exert in a single contraction.[77] The most common method of muscle strength measurement in the clinical setting is manual muscle testing.[77] The Medical Research Council Scale is an ordinal scale used to grade muscle strength.[78] The scale contains 6 points for muscle grading as follows: 0: no contraction, 1: flicker or trace of contraction, 2: active movement with gravity eliminated, 3: active movement against gravity, 4: active against gravity and resistance, 5: normal muscle strength.[78] This scale provides an easy and quick way to assess for significant muscle weakness or imbalance. Additionally, this scale is very practical to use as it only requires an assessment table and standardised positioning. However, this scale is very subjective as it relies heavily on the amount of force exerted by the examiner, their experience and their ability to accurately position the subject.[47] Currently, the use of hand-held dynamometers has increased in clinics and research to better quantify strength variation.[47] This method has been found to be a valid and reliable tool to measure isometric strength amongst subjects with CP.[47, 79]

1.6.2 Selective motor control

Selective motor control has been defined as 'the ability to isolate the activation of muscles in a selected pattern in response to demands of a voluntary movement or posture'.[80] Selective motor control involves isolating movements upon request, suitable timing and maximal voluntary contraction without overflow movement.[47] Selective motor control impairment can affect the ambulatory and function. There has been growing support for the assessment of selective motor control as a predictive factor of the functional ability.[81-84] A typical scale includes three grades of control as

follows: 0: no ability to isolate movement, 1: partial ability to isolate movement and 2: complete ability to isolate movement.[85, 86]

1.6.3 Muscle tone assessment

As discussed above, abnormal muscle tone is common amongst subjects with CP. When carrying out physical examination it is important to determine the nature and the extent of the abnormal tone. Tone is defined as the resistance to passive stretch.[47] Abnormally increased resistance to an externally imposed movement is known as hypertonia, while hypotonia is abnormally decreased resistance.[87] Hypertonia can be caused by spasticity, dystonia, rigidity or a combination of these.[88] As discussed in Section 1.2, spasticity is an increase in resistance felt when testing at higher speeds.[17, 46] On the other hand, dystonia shows an increase in muscle activity when at rest. For optimum examination results, standardisation for testing positions and the use of a grading scale are important.[47, 88] The Hypertonia Assessment Tool-Discriminant (HAT-D) is a tool developed and used to distinguish between spasticity, dystonia and rigidity.[89] The reliability and validity of this tool for dystonia and mixed tone is moderate, while good for spasticity and rigidity.[89] Mixed tone represents the presence of both types of hypertonicity in the same subject. Mixed tone is harder to diagnose and quantify in comparison with pure spasticity.[47]

1.6.4 Range of motion and contracture

Muscle shortening and contracture can be static or dynamic. In order to differentiate between static and dynamic impairments, it is important to measure the joint ROM at slow and fast speeds and with varying amounts of force.[90] The slow joint ROM

measurement potentially measures the muscle length and static contracture, while the fast joint ROM measurement possibly measures the dynamic/functional shortness or contracture.[17] Furthermore, it is important to distinguish between contracted mono-articular and bi-articular and multi-articular muscles. Bi-articular and multi-articular muscles cross two or more joints, and shortening in these muscles can limit the ROM available at these joints. The Silverskiöld test, Duncan-Ely test and Popliteal angle are used to measure spasticity and contracture of the gastrocnemius, rectus femoris and hamstring muscles, respectively.[47] To perform the Silverskiöld test, the subject is positioned supine and the knee is flexed to 90º with the hindfoot positioned in varus and ankle in maximal dorsiflexion. Then, the knee is extended; if the ankle plantarflexes, it confirms contracture of the gastrocnemius. In the Duncan-Ely test, the subject is positioned prone and the knee is flexed. Flexion of the ipsilateral hip indicates rectus femoris contracture. The subject is positioned in supine to perform the Popliteal angle test. The ipsilateral hip is flexed to 90º while the contralateral hip is in full extension. Then, the knee is extended until the first endpoint of resistance is felt. Measurement of the degrees lacking from full extension will indicate hamstring contracture. However, it has been observed with the use of fine wire electromyography that these tests do not distinguish between the mono-articular and bi-articular muscles crossing the joint, as both muscles were found to contract.[91] Spasticity and contracture in the bi-articular and multi-articular muscles limits the available joint ROM at the crossed joints. This affects joint and segment kinematics. Additionally, it disrupts the progression and optimum alignment of the GRF in relation to lower limb joints during the gait.

1.6.5 Bone deformity

1.6.5.1 Anteversion

Excessive femoral anteversion is common amongst subjects with CP. Femoral anteversion is the angle made by the femoral neck with the femoral condyles. It is the degree of forward projection of the femoral neck from the coronal plane of the shaft.[31] The trochanteric prominence angle test (TPAT) is used to measure femoral anteversion.[31] With the subject prone, the clinician palpates the point of maximal trochanteric prominence. Then, the clinician measures the angle of the tibia from the vertical, which represents the angle of femoral anteversion.[47] High correlation of the TPAT with X-rays (within 4º) has been reported in patients.[92] However, two studies contradicted the findings of the study by Ruwe et al.[92], reporting that the TPAT was found to be unreliable and inaccurate.[93, 94] These studies stated that this test should be used as a screening technique rather than as a definitive measurement method for femoral anteversion.[93, 94] The common clinical presentation seen with the femoral anteversion is increased medial hip rotation and decreased lateral rotation in extension. Additionally, the common compensation method seen for femoral anteversion is internal rotation of the femur and/or increased pelvic tilt. This deformity leads to squinting patellae, lumber lordosis and toeing-in during gait.

1.6.5.2 Patella alta

This deformity is common amongst subjects who walk with excessive knee flexion.[47, 95] To assess for patella alta, the subject is positioned supine with the knees in extension. Then, the top of the patella is palpated. The adductor tubercle is typically one finger distal to the superior edge of the patella in subjects with patella alta.[47] This condition

is associated with anterior knee pain, which may limit the distance the subject can walk.[47] Additionally, it can lead to further biomechanical and quadriceps lever arm dysfunction.[96-98] Complications may occur with this condition such as stress fractures at the inferior pole of the patella with palpable tenderness and subluxation and dislocation of the patella.[99, 100] Furthermore, extensor lag is suggestive of patella alta and quadricep insufficiency. Extensor lag is defined as a condition in which the active ROM of knee extension is less than the passive ROM.[101]

1.6.5.3 Tibial torsion

Tibial torsion is a rotational deformity around the long axis of the tibia.[102] This deformity is common in subjects with CP but is acquired. It has been shown that tibial torsion may be a significant contributor to crouch gait, especially when the tibial torsion angle is approximately 30º above normal.[102-106] Three different measurements/tests are used to measure tibial torsion as follows: measurement of the thigh-foot angle, measurement of the bimalleolar axis and the 2nd toe test. The thigh-foot angle can be measured as follows: the subject is positioned prone, the knee is flexed to 90º and the hindfoot is positioned vertically, the ankle is dorsiflexed to 90º, and the angle between the posterior axis of the femur and the axis of the hindfoot with the point between the 2nd and 3rd metatarsals is measured. Measurement of the bimalleolar axis starts with marking the midpoints of the medial and lateral malleolus. Then, the subject is positioned supine with the knee in extension, and the thigh segment is rotated until the medial and lateral femoral condyles are parallel in the frontal plane. The angle between the malleolar axis and the condylar axis represents the bimalleolar axis. To perform the 2nd toe test, the subject is positioned prone with the knee extended.

Then, the leg is rotated to positon the 2nd toe pointing directly towards the floor. Following that, the knee is flexed while maintaining the position of the thigh, and the angle from the vertical is measured. A study be Lee et al.[107] examined the reliability and the validity of these three methods in comparison with the 2D computed tomographic. The study found that the measurement of the thigh-foot angle had the highest reliability and validity.[107]

1.6.5.4 Leg length

Two different methods are used to measure the leg length discrepancy (LLD) which include true LLD and apparent/functional LLD. A true LLD, which is measured from the anterior superior iliac supine to the apex of the medial malleolus with the subjects in supine position, indicates an actual difference in length, which may be a result of femoral fracture and/or reduced bone growth due to a pathology.[31] A common compensation method for the true LLD is pelvic tilt and scoliosis, which may become a fixed deformity. The apparent LLD, which is measured from the xiphoid to the apex of the medial malleolus with the subject in supine position, identifies a functional difference where the limb segment is equal in length but other factors, such as scoliosis, hip subluxation, pelvic obliquity, unilateral contracture of the hip and knee and other biomechanical abnormalities, lead to functional LLD.[31, 47] Osteoarthritic hips can cause both true and apparent LLD. True LLD is measured with the subject supine, while the apparent LLD can be measured with the subject supine or standing.[31]

1.6.6 Posture and balance

It has been stated that 'of all the motor problems in cerebral palsy, deficient equilibrium reactions interfere the most with functional walking'.[108] This means that assessing posterior, anterior, medial and lateral equilibrium responses is mandatory. Delayed and deficient posterior equilibrium is common amongst subjects with CP.[47] Balance can be checked in a sitting or standing position. To test the sitting balance, the subject is moved off-balance and his or her ability to right himself or herself is observed. To test the standing balance, the subject is asked to stand on one limb at a time with the eyes first open then closed. This test gives an indication of the balance and proprioception abilities of the subject.[31] Additionally, assessment of the posture of the trunk, pelvis and lower extremity during standing and gait in sagittal and coronal planes gives an indication of areas of weakness, poor motor control and compensation strategies that subjects adapt to overcome these abnormalities.[47]

1.7 Interventions

When planning an intervention, it is mandatory to understand the interaction between the subject, the environment and the social factors. 'Disability', according to the World Health Organisation and the International Classification of Functioning, Disability and Health, is defined now as an umbrella term demonstrating the active interference between the subject and the environment (Figure 1-4).[109] This aids clinicians in seeing the bigger picture and promoting subjects' participation in their social environments. Currently, two approaches are used in treatment of subjects with CP, which include context focused therapy and child-focused therapy. Context-focused therapy aims to enable the subject to carry out the function according to their functional ability by

altering the activity or the environment.[110, 111] Child-focused therapy aims to modify the performance by treating the functional limitations, e.g. muscle weakness and limited joint ROM.[112] When these two therapies were compared, it was found that both approaches were equally effective in improving activities of daily living.[112] Common interventions used amongst subjects with CP include non-operative interventions and operative interventions. Conservative/non-operative interventions are important amongst children with CP until the motor patterns are matured (usually between 8 to 10 years).[113] This is because the results of early operative interventions are less predictable and have a higher risk of failure and relapse.

1.7.2 Operative interventions

Different surgeries/operative interventions are performed for spasticity reduction and treatment of muscle contractures, long bone torsions and joint deformities. Selective dorsal rhizotomy (SDR) is a common procedure used for spasticity reduction. Evidence has shown that SDR has become a standard neurosurgical procedure amongst subjects with CP.[114, 115] SDR procedures were found to reduce spasticity, improve ROM and function, enhance self-care activities, increase comfort and normalise gait pattern.[116]

Bi-articular and multi-articular muscles are more commonly affected and contracted in subjects with CP in comparison to the mono-articular muscles.[4, 15, 16, 117] Accurate estimation of the bi-articular and multi-articular muscle length and spasticity is mandatory to guide clinical decision-making for muscle lengthening or transferring.[117] In general, the goal of lengthening the muscle contracture is to increase joint ROM,

which will allow the joints and segments to achieve the desired positions during gait. This will reduce the secondary compensatory movement which may be presented due to insufficient joint ROM.[117]

The goals of treatment of long bone torsions and joint deformities are comfort, function, cosmesis, restoration of normal lever arms, improvement of joint moments, elimination of gait compensation, improvement of foot shape and improvement of gait efficiency.[118-120] In subjects with spastic diplegic CP, the typical long bone torsions present are femoral anteversion and tibial torsion. Malrotated levers caused by tibial torsion and femoral anteversion cause two effects: decrease of the magnitude of the primary or intended moment and production of secondary moments. Additionally, lever arm dysfunction affects the external moments produced by the GRF and internal moments produced by muscles.[4, 118] Derotational osteotomy procedures are performed to correct excessive femoral anteversion and tibial torsion.[102, 118]

Furthermore, common hip problems presented amongst subjects with CP include increased hip adduction, increased hip flexion, increased internal rotation and hip subluxation and dislocation.[119] Operative treatments for hip problems include muscle lengthening and osteotomy.[119]

Dynamic muscle imbalance presented amongst subjects with CP leads to the development of foot deformities such as equinus, equinoplanovalgus and equinocavovarus.[120] Mild foot deformities are best treated with soft-tissue (i.e. muscle-tendon unit) surgeries, which include release, lengthening, partial or split transfer and

complete transfer.[121-124] Moderate to severe foot deformities are best treated with a combination of soft-tissue and skeletal surgeries. Skeletal surgeries include arthrodesis and osteotomy.[120]

1.7.1 Non-operative interventions

1.7.1.1 Physiotherapy

In the past, the main aims of physiotherapy were to normalise movement patterns, reduce neurological signs and minimise the development of secondary impairments.[125] However, in recent years (as stated above) the concept of promoting function the natural environment has received major attention.[109] This has increased the awareness of the physiotherapists and allowed them to focus on enabling subjects with neurological disorders to master tasks and participate in their activities of daily living.[125] Currently, the three main goals of physiotherapy are to maintain ROM, improve strength and facilitate mobility.[126] Customised combination of interventions is decided according to the subject's ability/requirement. Treatment interventions include stretching and casting to maintain ROM, strengthening to increase power and endurance, practice of functional activities, gait training, and electrical stimulation (in combination with functional activities).[126] Physiotherapy is often one of the first interventions recommended following the diagnosis. Physiotherapy intervention is periodic and will continue occur over a lifetime. The planned intervention is unique to each subject and depends on the age and the presenting impairment.[126] Infants and young children are involved in an early physiotherapy intervention plan.[126] A child's needs, motivation, interests and family's goals are evaluated by the physiotherapist to determine the intervention plan.[126] At this stage, intervention is provided in the child's

home to enable his participation in the most natural environment. Following that, the intervention is provided in the educational environment (schools) and in an outpatient setting to enable practice of functional activities across several settings. Periodic physiotherapy intervention in an outpatient setting is required for adolescents.[126] This is because an adolescent may only have a reviewing service in middle and high school. The common physiotherapy interventions used amongst adolescents include stretching, strengthening, cardiovascular endurance activities, functional activities, motor training and facilitation of balance control and coordination.[126]

1.7.1.2 Pharmacology

There are several ways of administering pharmacological interventions, including orally, locally by injections and intrathecally. A variety of oral medications are available for treatment of abnormal muscle tone and movement disorders in subjects with CP, including Benzodiazepines, Baclofen, Dantrolene Sodium, Alpha-2 Adrenergic Agonists, Levodopa and Trihexyphenidyl.[127] However, clinicians must be cautious about the potential side effects and risks associated with each medication, and should weigh the efficiency of the medication against the side effects. Oral medication can decrease spasticity, the frequency of uncomfortable muscle spasm and the frequency of uncomfortable pain.[127] One side effect associated with these medications is sedation, which leads to cognitive diminishing, may cause hepatotoxicity and may produce physical dependency.[127] These oral medications are used to treat tone that is affecting the daily care and comfort of subjects with CP. These medications are less commonly used in subjects with hemiplegic or diplegic CP unless it is severe.[127]

Botulinum toxin type A (BoNT-A) and phenol injections are other forms of non-operative pharmacological treatments commonly used for abnormal tone. BoNT-A and phenol are injected intramuscularly and lead to temporary muscular denervation. The effect of BoNT-A injection deteriorates, while the effect of phenol injection is permanent. Indications for this treatment include lack of progression in motor development, development of muscle contractures, intolerance of day and night splinting and/or decrease in functionality and increased muscle tone.[128] The general goals of this treatment include improving function/gait and thus influencing the pathological process, improving balance and control of sitting, positioning and facilitating hygiene care and bracing for a non-ambulatory subject.[113, 129] Examples of specific goals include facilitation of orthotic management, relief of pain postoperatively and simulation of orthopaedic surgery.[130-133] It has been recommended that this treatment should start at an early stage, between 2 to 6 years.[134] This is because gait patterns and motor function are still flexible, and larger results can be achieved. Adequate follow-up treatments, such as orthotic management, serial casting and intensive physiotherapy, are mandatory for optimising/maximising the results of BoNT-A and phenol injections.[128]

Intrathecal administration of baclofen or phenol is another pharmacological approach used for spasticity management. In this approach, a system is surgically implanted which infuses baclofen or phenol into the spinal canal and around the spinal cord.[135] The benefit of this approach is that the side effects associated with the oral medications are minimised due to the dose being delivered directly to the spinal receptors.[136, 137] This approach is not typically used in the ambulant child. Although this approach is

effective in spasticity management, there are several associated risks, which include infection, hardware problems and drug-related adverse events.[137] Several studies investigated the effect of this approach and reported improvements in positioning, activities of daily living, oral motor skills, hand use, sleep and comfort.[138-140]

1.7.1.3 Orthosis

An orthosis is defined as 'an externally applied device used to modify the structural and functional characteristics of neuromuscular and skeletal systems'.[141] Lower limb orthoses are biomechanical devices which are widely prescribed to aid individuals with neuromuscular impairments.[142] An AFO is 'an orthosis which encompasses the ankle joint and the whole or part of the foot'.[6] AFOs are commonly used as a conservative management amongst subjects with CP. AFOs may be prefabricated (off-the-shelf) or custom-made. Materials such as high temperature thermoplastic, thermosetting, silicone and carbon fibre are used to manufacture AFOs. In the presence of triplanar foot deformity and hip and knee problems, custom-made AFOs should be considered.[143, 144] Because these AFOs are individually fabricated over a positive model of the subject's own limb, they provide an intimate fit and specific control to meet the neurobiomechanical needs of each subject.[143] The term 'neurobiomechanics' is often used to describe the interaction between biomechanics and the neuromuscular system and its application to the management of subjects with neurological impairments.[12] Several designs of custom-made AFOs are available, which include: solid/rigid AFO (SAFO), ground reaction/floor reaction AFO (GRAFO/FRAFO), posterior leaf spring (PLS) and hinged/articulated AFO (HAFO).[145] Further information about different designs of AFOs is provided below.

Posterior leaf spring ankle-foot orthosis

This AFO design is flexible (Figure 1-5). The flexibility is affected by the material choice, material thickness and the radius of the curvature of the posterior leaf.[146-148] The primary indication for the PLS is isolated dorsiflexion weakness, which leads to drop foot in swing. This design should not be used if other problems exist, such as high tone or spasticity, significant mediolateral instability or the need to influence the knee and the hip joints.[68] Most of off-the-shelf AFOs are of this design.

Solid ankle-foot orthosis

This AFO design prevents all movement at the foot and ankle (Figure 1-6). Inadequate tibial progression caused by high tone, spasticity and contracture of the plantarflexors can be altered using the SAFO. Additionally, excessive tibial inclination caused by plantarflexor weakness and/or hip and knee joint problems can be controlled using the SAFO. Other benefits of the SAFO include maximum triplanar control and maximum effect on sagittal plane kinetics and kinematics of the knee and hip joints.[68] The success of this device relies solely on the ability to prevent ankle motion by the appropriate selection of the material type, material thickness, location of trimlines, intimacy of fit and use of ankle reinforcement.[68, 145]

Ground reaction ankle-foot orthosis

GRAFO is a type of solid AFO with rear entry (Figure 1-7). This device is designed to restore the plantarflexion/knee extension couple commonly absent in subjects with excessive tibial inclination.[68, 149, 150] One characteristic of the GRAFO is a plastic pretibial shell close to the knee. For optimum control of excessive tibial progression,

the GRAFO should be very stiff and optimally aligned to maintain the GRF in front of the knee in mid-to-late stance. As above, the success of the GRAFO depends on the ability to prevent movement by the appropriate selection of the material type, material thickness, location of trimlines, intimacy of fit and use of ankle reinforcement.[68, 145] Presence of a fixed deformity in any anatomical plane, dynamic contracture of the knee and/or hip, tibial torsions and insufficient stiffness will limit the effectiveness of the GRAFO.[68, 150-152]

Hinged ankle-foot orthosis

This AFO design incorporates ankle joints to allow or assist motion in one direction, while preventing limiting or resisting motion in another direction (Figure 1-8). A variety of mechanical joints with different design characteristics are available. The HAFO, which blocks plantarflexion at 90º and allows unrestricted dorsiflexion, is commonly prescribed. This design should only be considered when a sufficient length of gastrocnemius muscle is available. This will allow adequate range of dorsiflexion without compromising the ROM at the knee joint. For the success of this design, dorsiflexion of approximately 10º should be achievable with the knee in full extension without any spastic catch and/or tone in the plantarflexors.[12, 90, 143] However, the HAFO may not be suitable in the presence of moderate to severe mediolateral instability of the foot. The space required for ankle joints compromises the fit and control of the HAFO.[68, 150]

1.8 Ankle-foot orthosis prescription criteria

The initial step in forming the prescription of an AFO is to understand subject's disease process and natural history of movement disorder. This is followed by the other parts of clinical assessment, including physical examination to clearly outline the neuro-biomechanical impairment and functional objectives. Additionally, defining the functional loss and deficits in gait enables robust prescription criteria. Functional problems could be present at joints (ankle, knee and hip) or segments (shank and thigh). Several functions/goals can be achieved using an AFO, which can be summarised as preventing/correcting deformity, promoting a base of support, enabling training of skills and improving the efficiency of movement, e.g. standing and walking.[153] The AFO design (motion allowed or blocked) will depend solely on the functional objectives required. Furthermore, the AFO features, including materials, use of reinforcements, straps, sole plate length and wedges, should be decided.[68] During the casting process, the ankle angle and the foot position should be considered. This is followed by cast modification where appropriate corrective forces are applied. Proper footwear (FW) selection and FW modifications, including heel height, sole profile and stiffness and heel profile, are also mandatory.[8] Finally, during the fitting process, tuning of the AFO and footwear should be performed to optimise the function.[7, 8, 154] In addition, appropriate referrals for physiotherapy or pharmaceutical or surgical intervention may be required. These interventions may be considered before orthotic treatment in order to facilitate appropriate fitting of the AFO and optimise its function.[68] Some of the important prescription criteria are discussed below.

1.8.1 Ankle-foot orthosis material and features

After appropriate selection of the AFO design, careful consideration should be given to the choice of material and any additional features of the AFO. This includes material type, material thickness, trimlines, ankle reinforcement, use of straps and length of foot/sole plate. As stated previously, the stiffness of the AFO can be controlled by the material choice, material thickness, trimlines and ankle reinforcement (inserts).[68] Polypropylene (thermoplastic) is the most commonly used material for AFO fabrication. This material has the benefit of quick and cheap fabrication.[145, 155] Two types of polypropylene are frequently used: homopolymer and copolymer. Homopolymer offers a high strength-to-weight ratio and is firmer and stronger than copolymer, and is used for AFOs where rigidity is required, such as the SAFO and GRAFO. The rigidity of the AFO can be further increased by the appropriate selection of the thickness of the material. It has been reported that inappropriate selection of the material type and thickness can allow bending of the material anterior to the ankle (medial and lateral), influencing the function.[156, 157] To ensure maximum rigidity and bending resistance, the location of the trimlines should be located anterior to the malleoli.[154, 158, 159] Varying the trimlines will assist in modifying the pressure for triplanar deformity correction and mediolateral stability. It has been reported that careful selection of trimlines is the most important factor affecting the stiffness of the AFO.[160, 161] A study by Sumiya et al.[162] investigated quantitatively the change in PLS stiffness resulting from altering the ankle trimlines. This study concluded that the PLS ability to resist ankle movements decreased nearly in proportion to the reduction of posterior upright width.[162] Furthermore, the use of ankle reinforcement/inserts, such as carbon fibre inserts can increase the stiffness of the AFO. It has been reported that

the use of carbon fibre inserts can provide complete ankle stiffness without needing to alter the overall material thickness.[143, 163]

The use of ankle straps aids in plantarflexion control, especially in subjects with increased tone or spasticity. Additionally, straps may help to maintain the location of the heel within the AFO and to control the position of the calcaneus and subtalar joint when FW does not provide adequate support.[68] The ankle strap should be positioned to apply the force at approximately 45º to the dorsum of the foot near the ankle joint.[68] For effective force application and transmission, a non-elasticated material should be used for the ankle strap. Special care must be taken for proper design and placing of the ankle strap with respect to upper limb function availability[154, 164] The design of the ankle strap can be a single strap or a 'figure 8' crossover strap.[68]

The length of the foot/sole plate may be full, ¾ length or sulcus. The full sole plate is designed to hold the toes in extension to stretch the long toe flexors, which aid in easy exit from stance phase and may help reduce the abnormal toe grasp reflex.[68, 145] The sole plate may be left flexible or rigid to allow or block the normal exit from the stance phase.[145]

1.8.2 Ankle-foot orthosis angle and modification

The ankle angle of the AFO (AAAFO) is the sagittal angle of the foot relative to the shank within the AFO. The gastrocnemius muscle is a multi-articular (tri-articular) muscle which crosses the knee, ankle and subtalar joints. Hence, shortness and spasticity in the gastrocnemius muscle will affect the available dorsiflexion ROM at

the ankle joint when the knee joint is fully extended or knee extension when the ankle is dorsiflexed. As discussed earlier, during late midstance of normal gait, extension at the knee joint and hip joint is essential in order to achieve stance stability, which reduces the energy cost.[165-167] Hence, caution must be taken in determining the AAAFO in the presence of shortness and spasticity of the gastrocnemius muscle. An AFO which does not take into account gastrocnemius shortness will limit the necessary knee extension during late stance and will therefore adversely affect knee and hip kinetics.[12, 168] The ratio of the lever arms of the muscle pull at the knee and at the ankle is approximately 1:2 at 40%.[169] This means that increasing the ankle dorsiflexion by 5º with the presence of gastrocnemius muscle spasticity and shortness will decrease the knee extension by 10º. Other considerations for the appropriate AAAFO choice include stiffness of the calf muscles, the length at which the gastrocnemius and soleus muscles can produce maximal muscle power, the length and the stiffness of dorsiflexors, the triplanar requirements of the bones and joints of the foot, and gait pattern.[168] An algorithm for determining the optimum AAAFO has been proposed by Owen.[168] In this algorithm, consideration is given to all the factors for deciding the optimum AAAFO and any risks associated with using chosen alignments.

The foot alignment and the ankle angle should be considered while casting. If the foot deformity is flexible, then correction of the triplanar deformity must be applied during casting. For example, for correction of pronation initially the ankle is plantarflexed fully to release the tension on the Achilles tendon. Then, the subtalar joint is supinated to the neutral position. This is followed by pronation and adduction of the forefoot. Finally, the ankle can be dorsiflexed. While modifying the cast, corrective forces

should be applied. Cast shaping must be done in a way that matches the contours of the underlying skeletal structure. Furthermore, the forces should be applied far apart and over large areas to maximise the lever arms and reduce the pressure. These steps aid in increasing the comfort of the AFO.

1.8.3 Footwear and footwear modification

The footwear used with the AFO has an integral role in the determination of the overall biomechanical control achieved.[7, 71, 170, 171] Because the footwear is considered to be an important part of the overall orthotics prescription, the term AFO-FC is sometimes used. Condie et al.[171] described the AFO-FC as a powerful and relatively easy adjustable tool which can alter hip and knee kinematics. Additionally, they reported an increase in energy expenditure with the use of an inappropriate AFO-FC which can be greater than walking barefoot.[171] Footwear modifications may be useful to compensate for the fixed or limited ankle movement available with the use of an AFO. The use of an AFO has an impact on the three rockers of the gait. In a solid AFO, the ankle is held in a fixed position throughout the stance phase. This limits the required movement at the ankle during the first and second rockers. The length alignment and stiffness of the foot plate will determine the effect of the AFO on the third rocker. Immobilising the metatarsophalangeal joint prevents the normal anatomical third rocker.

When considering the AFO-FC, it is useful to divide the stance phase into three subphases rather than five.[12] This includes entry to temporal midstance, temporal midstance and exit from temporal midstance. Changes to the design of the footwear

have been reported to affect the GRF point of application and location throughout the gait.[71, 172] Heel profile and stiffness affect entry to temporal midstance, while the sole design and stiffness affect the exit from temporal midstance. The effective heel height (the difference in thickness between the heel and sole, heel sole differential (HSD)) affects temporal midstance by influencing the SVA.[7, 8, 12, 71, 168, 173] Tuning is a three-step process of making adjustments to the SVA by altering the HSD, the sole design and stiffness and the heel profile and stiffness of the FW.[71, 172] Different heel and sole designs, thickness, contour and width can be used to influence the entry and exit for temporal midstance.[68] Examples of heel designs include plain heel, positive heel, negative heel and cushioned heel. Examples of sole design include flexible sole with flat or rounded profile and stiff sole with a rounded profile or point loading rocker.

1.9 Biomechanical effects of ankle-foot orthoses

1.9.1 Direct biomechanical effects of ankle-foot orthoses

The AFO provides direct control of motion in all 3 anatomical planes by application of systems of 3 corrective forces.[68] Movement such as plantarflexion, dorsiflexion, pronation and supination can be controlled by using AFOs.[68] For correction of plantarflexion and dorsiflexion, a single 3 corrective force system is applied, while two 3 corrective force systems are applied in the coronal and transverse planes for control of pronation and supination.[145] For example, to prevent plantarflexion of the foot in swing, the AFO should apply a system of 3 corrective forces to the posterior calf (F1), the plantar surface of the foot near the metatarsal heads (F2) and the dorsum of the foot near the ankle joint (F3) (Figure 1-9).[12, 145] For maximum control, intimate fit of the AFO with optimum force application is required. Poor or inadequate control

of triplanar deformity (supination and pronation) causes immediate effects which include pain, contracture and inadequate stretch on the Achilles tendon.[68] Additionally, it can cause long term effects which include increasing deformity and change to skeletal deformity.[68] For optimum force application, the lever arms and the area of force application must be maximised. Additionally, these forces should be applied in a way that respects the underlying anatomy and matches the contour of the skeletal structure.[12]

1.9.2 Indirect biomechanical effects of ankle-foot orthoses

The AFO can affect the GRF components (point of application, magnitude and line of action) by controlling the alignment and motion of the ankle joint.[12, 174] Additionally, the AFO can modify the kinematics and kinetics of the segments to become closer to normal. This will lead to the GRF and the moments generated at the lower limb joint to also approximate to normal.[7, 8, 154] In subjects with excessive ankle plantarflexion (equinus), controlling the motion and alignment of the ankle with the use of an AFO aids in allowing the entire plantar surface of the foot to bear weight.[68] This helps to move the point of application of the GRF posteriorly along the foot.[68] The use of an AFO controls the shank alignment, which aids in optimising the shank kinematics. The optimum forward inclination of the shank with the posterior placement of the point of application of the GRF along the foot realigns the GRF at the knee joint.[12, 174] This reduces the knee flexion moment or, ideally, may even replace it by an external extension moment. Furthermore, the appropriate forward inclination of the thigh aligns the GRF posterior to the hip joint, which may decrease the external flexion moment or provide a suitable external extension moment.

1.10 Studies supporting the biomechanical influence of the ankle-foot orthosis

Several systematic reviews have been published investigating the efficacy of the AFO-FC in achieving the outlined biomechanical goals amongst subjects with CP.[68, 175-178] The most recent systematic review was conducted by Bowers and Ross.[68] In this review, a number of research questions were formulated and answered, including the influence of the AFO-FC on lower limb kinematics and kinetics. The results of the studies were divided per the limb involvement of the subjects included, i.e. hemiplegic, diplegic and mixed. Improvement in ankle kinematics was reported in the review, but reduction in power generation and absorption was observed. A positive indirect impact of the AFO-FC on hip and knee kinematics and kinetics was found; however, results were inconsistent. This is because of the variation in the methodologies implemented in the studies, and the variety of different designs of AFOs examined amongst heterogeneous groups of subjects with CP. Generally, limited information about the subjects, AFO-FC design and features and the tuning process was provided by the studies.

Studies investigating the biomechanical effects of the AFO-FC amongst subjects with diplegic CP published following the systematic review by Bowers and Ross to present were reviewed (2008-2016). Observations from the recent studies are discussed below (Table 1-1). Rogonzinski et al.[151] investigated the biomechanical impact of GRAFO amongst subjects with diplegic, triplegic (involvement of 3 limbs) and quadriplegic CP. A significant reduction in peak dorsiflexion in stance was found. Additionally, a significant increase in peak knee extension during stance and non-significant improvement in knee moment at midstance were reported. Possible reasons for this

may include hip and knee flexion contracture and a short gastrocnemius muscle, which were not accommodated. The author reported better results in the presence of knee and hip flexion contracture≤10º when compared with contracture≥15º. It should be noted that the study failed to report sufficient information about the AFO-FC features, for example material thickness, FW characteristics and SVA.

Kerkum et al.[179] investigated the biomechanical influence of the GRAFO with ankle joint/hinge in comparison to shoes only amongst subjects with spastic CP. The joints used allowed the stiffness of the GRAFO to be varied using different pre-compressed springs with different mechanical properties. Three different stiffness configurations were investigated, which were classified as follows: rigid, stiff and flexible by varying the spring properties in the hinged used. Ankle ROM was significantly reduced and the peak internal plantarflexion moment in stance was significantly improved by the 3-different stiffnesses, although the ankle ROM was less reduced with the stiff and flexible configurations in comparison to the rigid configuration. Peak ankle power generation was reduced significantly by rigid configuration, while it was preserved by the stiff and flexible configurations. A significant increase in the knee extension angle was observed at contralateral toe-off and in peak knee extension during stance, while a non-significant increase in knee extension in midstance was reported. Furthermore, significant improvement in knee moment at midstance and at peak knee extension during stance was found. At the hip joint, a non-significant reduction in the hip flexion angle was reported. Although this study concluded some positive results, the AFO design used in the study is questionable. The AFO used had a single hinge on the lateral side and only covered the lateral and proximal anterior part of the shank. As stated

above, for maximum control two 3 corrective force systems are used and applied by enclosing the foot and the shank. This may be compromised by the design used in this study. The author stated that the AFO was tuned, but no information about the SVA and FW modification was provided. All 3 stiffness configurations allowed ankle motion, as illustrated in the graphs supplied in the paper. These reasons may explain the limited effect of the AFO on the hip and knee joint kinematics and kinetics.

Another study examined the influence of the PLS and HAFO amongst subjects with hemiplegic and diplegic CP.[180] The joints used for the HAFO had an elastomer component that produced variable resistance to dorsiflexion and plantarflexion, and the stiffness to resist each motion could be adjusted separately. The degree of ankle motion for each design of the AFO was adjusted per the requirement of each subject. For ankle kinematics, the following was reported: significant improvement in dorsiflexion during swing for the PLS and HAFO, significant improvement in dorsiflexion during stance with the PLS and non-significant improvement with the HAFO. Significant reduction in push-off power was observed with the PLS and HAFO. This was found to be significantly greater in the HAFO. Additionally, a significant increase in peak plantarflexion moment in late stance with the PLS and a non-significant increase with the HAFO were stated. The effect of the PLS and HAFO on the improvement of peak knee extension in stance was reported to be not significant. Better knee extension was achieved with the use of the HAFO in comparison to the PLS. This may be explained by better control of the ankle movement allowing better shank alignment. However, non-significant improvement in knee moment in peak knee extension during stance was reported with the PLS and HAFO.

At the hip joint, significant improvement in hip extension in stance with the PLS and HAFO was observed. Incomplete information about the subjects and the design criteria of both AFOs was presented. Additionally, the SVA of the HAFO was not optimum as the shank in standing was vertical and not inclined forward. This may explain why optimum results were not achieved at the hip and the knee.

Observations made from reviewing the recent studies agree with the findings of the recent systematic review.[14] While the results on the ankle kinematics and kinetics are consistent, the influence of the AFO on the biomechanics of the hip and knee is inconclusive. This is due to the differences in the AFO designs used, inappropriate design of AFO used and the heterogeneity of subjects with CP. Additionally, it was suggested by the recent review that the difference found in the SVA used in the studies may have led to the variation found in the influence of the AFO on the kinematics and kinetics of the lower limb joints.[14]

In this chapter, the biomechanical effects of AFOs were discussed. Additionally, the importance of clinical assessment and several elements of prescription criteria were presented. The evidence related to supporting the biomechanical efficiency of an AFO-FC on lower limb joints amongst subjects with diplegic CP was reviewed. The findings of this review showed that the effect of an AFO-FC on the biomechanics of the hip and knee is inconclusive. The following chapter will examine the effect of an AFO-FC on gait. Initially, the role of optimising shank kinematics using an AFO-FC on thigh kinematics and on kinetics of lower limb joints will be discussed. Additionally, this chapter will discuss the importance of the temporal midstance to late stance phase

of gait for the examination. Following this, the results of a study investigating the effect of optimisation of temporal midstance shank kinematics with the use of an AFO-FC on several gait variables will be presented.

1.11 Figures

Figure 1-1: Phases of gait reproduced from Perry.[13]

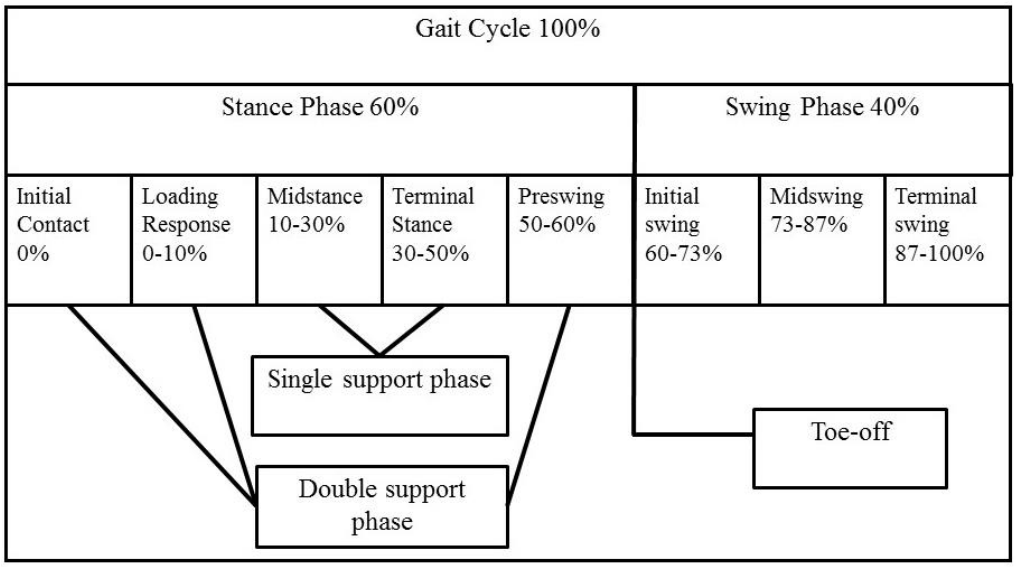


Figure 1-2: Normal vertical component of ground reaction force in gait.[18]

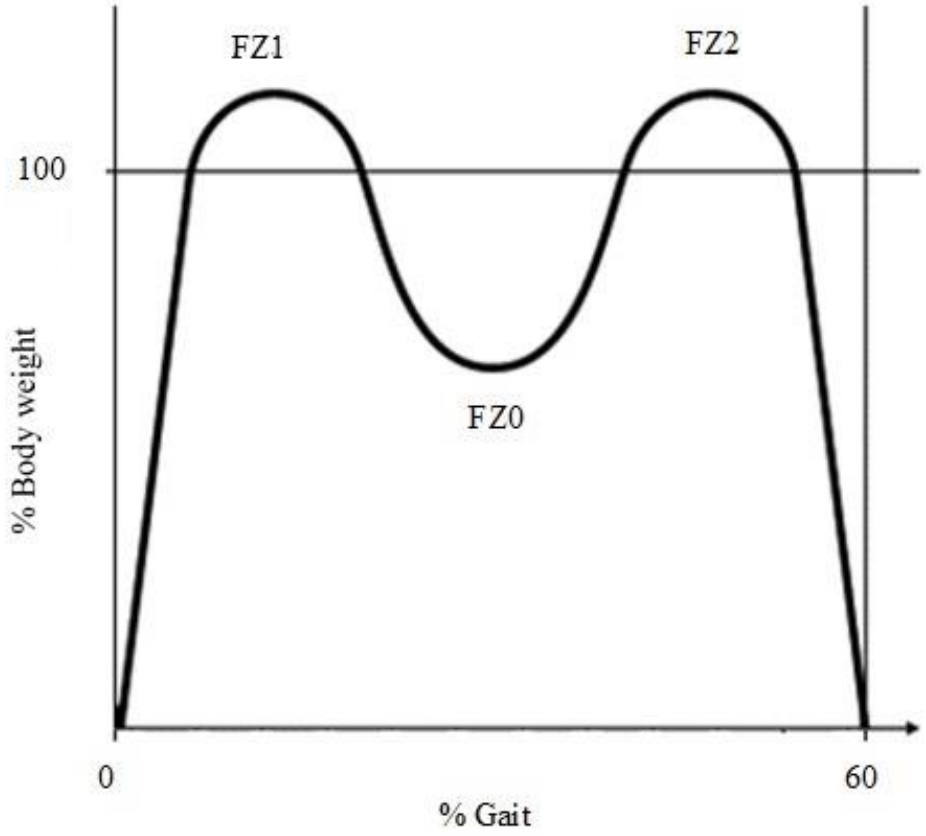


Figure 1-3: ‘Butterfly diagram’ represents the magnitude, direction and point of application of the GRF at uniform time intervals (10 ms) of gait in the sagittal plane.[64, 181]

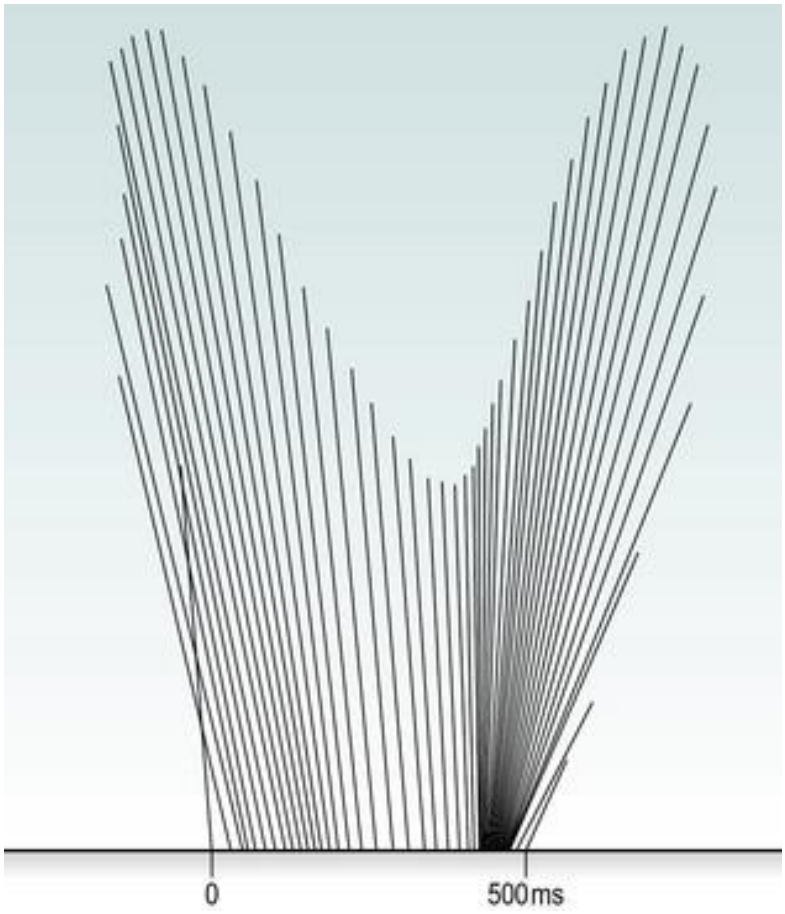


Figure 1-4: Representation of the model of disability that is the basis of the International Classification of Functioning, Disability and Health reproduced from the World Health Organisation.[109]

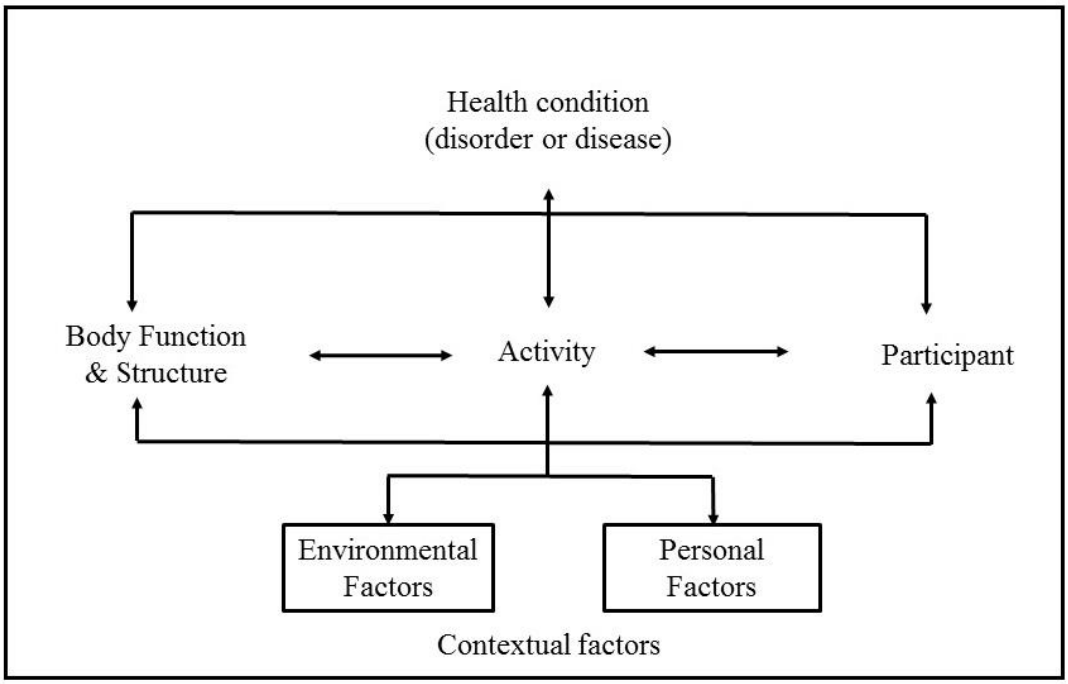


Figure 1-5: Posterior leaf spring ankle-foot orthosis.

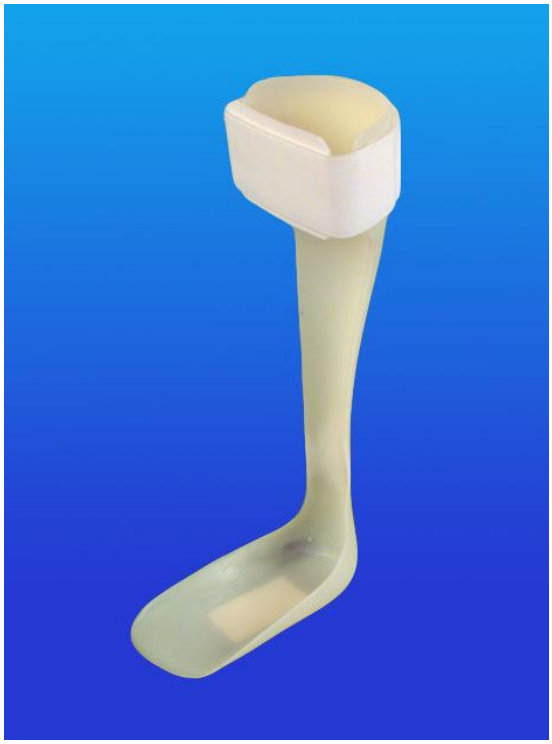
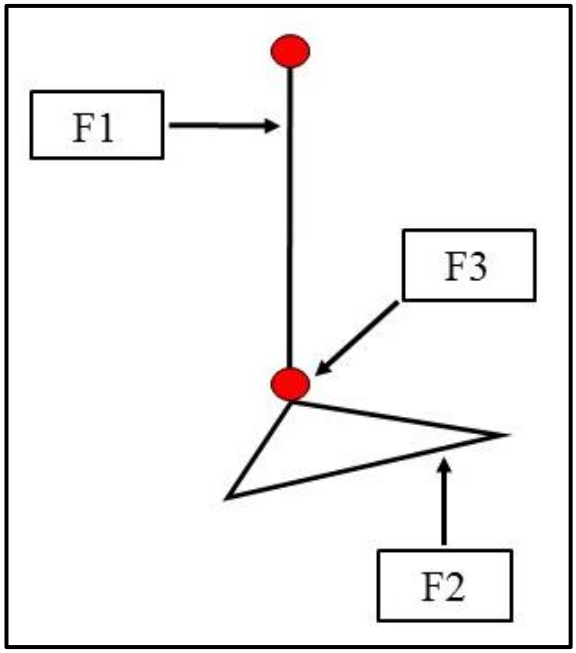


Figure 1-6: Solid ankle-foot orthosis.

Figure 1-7: Ground reaction ankle-foot orthosis.

Figure 1-8: Hinged ankle-foot orthosis.

Figure 1-9: 3 corrective force system applied to prevent plantarflexion of the foot



1.12 Tables

Table 1-1: Summary of the papers which studied the kinematic and kinetic influence of ankle-foot orthoses on lower limb joints amongst subjects with cerebral palsy.

Author	Research design	Number and age range of subjects	Cerebral palsy type	AFO type	Key findings
Rogozinski et al. (2009)[151]	Retrospective within subject comparison	n=27 Range=7-16 years	Mixed (25 diplegic, 1 triplegic, 2 quadripelgic)	GRAFO	**Ankle: 1**-significant reduction in peak dorsiflexion in stance **Knee: 1**-significant increase in peak knee extension in stance, **2**-non-significant improvement in knee moment in midstance (n=9)
Kerkum et al. (2015)[179]	Prospective Crossover	n=15 Mean±SD=10±2	Mixed	GRAFO with spring-hinged (comparison between 3 stiffness configurations, rigid, stiff and flexible, vs shoes)	**Ankle: 1**-ankle ROM was significantly reduced by the 3-different stiffness (maximum with rigid), **2**-peak ankle power generation was reduced significantly by rigid stiffness and preserved by the stiff and flexible, **3**-significant improvement in peak internal plantarflexion moment in stance **Knee: 1**-significant increase in knee extension angle at contralateral toe-off and at peak knee extension in stance, **2**-non-significant increase in knee extension in midstance, **3**-significant improvement in knee moment at midstance and at peak knee extension in stance **Hip: 1**-non-significant reduction in hip flexion angle at contralateral initial contact**, 2**-no change in hip moment at contralateral initial contact
Wren et al. (2015)[180]	Prospective Crossover	n=10 Kinematics (n=15) Kinetics (n=7) Range=4-12 years	Mixed (5 diplegic and 5 hemiplegic)	PLS, HAFO	**Ankle: 1**-significant improvement in dorsiflexion in swing with PLS and HAFO, **2**-significant improvement in dorsiflexion in stance with PLS and non-significant improvement with HAFO, **3**-significant reduction in push-off power with PLS and HAFO and was significantly higher in HAFO, **4**-significant increase in peak plantarflexion moment in late stance with PLS and non-significant increase with HAFO **Knee: 1**-non-significant improvement in peak knee extension in stance with HAFO and PLS, **2**-knee extension was better with HAFO than PLS, **2**-non-significant improvement in knee moment in peak knee extension in stance with PLS and HAFO **Hip: 1**-significant improvement in hip extension in stance with PLS and HAFO
Danino et al. (2016)[182]	Retrospective within subject comparison	n=53 mean=8.3 years	Diplegic	SAFO, PLS, HAFO	**Ankle: 1**-significant increase in ankle dorsiflexion in stance and swing **Knee: 1**-significant increase in knee extension in stance and decrease in swing

AFO: ankle-foot orthosis, GRAFO: ground reaction ankle-foot orthosis, SAFO: solid ankle-foot orthosis, DAFO: dynamic ankle-foot orthosis, HAFO: hinged ankle-foot orthosis, ADR-AFO: adjustable response ankle-foot orthosis.

Chapter 2 The effect of ankle-foot orthoses on shank and thigh kinematics during gait of subjects with spastic diplegic cerebral palsy

2.1 Introduction

As discussed in Chapter One, optimal alignment of the GRF during gait relative to the joints of the lower limb is considered essential to produce a controlled and energy-efficient gait. AFOs have been widely prescribed for use in subjects with CP in an attempt to influence the kinetics and kinematics of the gait. The use of an AFO provides both a direct and an indirect biomechanical effect on the body. By enclosing the ankle and foot, the AFO controls the alignment and movement of the joints. This, in turn, can manipulate the GRF, placing it closer to lower limb joint centres, which aids in the reduction of the energy expenditure. This is achieved by optimising the shank alignment with the use of an AFO-FC during temporal midstance to an optimum position of approximately 10-12º inclination, measured from the vertical (Figure 2-1).[7, 8] As discussed in Chapter One, in normal gait this alignment of the shank, combined with a controlled angular motion is vital to produce the required temporal midstance stability.[8, 69] Temporal midstance stability aids in energy conservation, lowers the vertical excursion of the centre of mass and reduces the demand on the musculoskeletal system.[9, 165-167] In addition, this shank alignment allows the required normal kinematic and kinetic changes to take place. As the GRF progresses along the foot, it aligns anterior to the knee joint and posterior to the hip joint during terminal stance by further inclination of the shank and thigh. This position creates an external

extension moment at both the hip and knee joints, causing passive stability. This passive stability, accompanied with heel raise during terminal stance, aids in energy conservation due to the rise of the body centre of mass.[12] As the thigh reaches its maximum inclination, the GRF lever arm at the hip and knee increases. This increase in lever arm in combination with the large magnitude of the GRF produces a strong stabilising external moment at the hip and the knee joints during terminal stance (Figure 2-2 & 2-3).[7, 8] Furthermore, the position of maximum thigh inclination in terminal stance aids in lengthening the hip flexors and gastrocnemius muscles in the stance limb and the hamstrings and gastrocnemius muscles in the swing limb, causing 'therapy while walking'.[71, 183]

Inability to support the body weight (BW) and to decelerate the downward velocity of the centre of mass in terminal stance is a common problem observed in subjects with CP. This is seen as a decrease in the GRF, with FZ2 being smaller than the BW (FZ2<BW).[18] The reduction of FZ2 has been described as "Ben Lomonding"[7, 18] due to the second peak (FZ2) being smaller than the first peak (FZ1), resembling the shape of the Scottish mountain, Ben Lomond.[18] The reduction in FZ2 is often counteracted by the actions of the contralateral limb, resulting in excessive loads during the loading response (FZ1). Williams et al.[18] reported that 87% of subjects with CP exhibited some degree of "Ben Lomonding".[18] Improved terminal stance stability provided by the use of an AFO-FC may help to achieve an adequate second peak of the vertical component of the GRF (FZ2) greater than body weight.[7] This aids in the reduction of the excessive contralateral first peak (FZ1) observed in subjects with CP.

Optimising shank kinematics with the use of an AFO-FC is an initial step for successful orthotic prescription and problem solving.[7, 8, 184, 185] In addition, optimising shank kinematics increases the opportunity to optimise knee and hip kinetics and thigh and trunk kinematics.[7, 184] To be able to achieve optimum alignment, consideration must be given to the design specification of the AFO-FC. One of the specifications that should be taken into consideration is the AAAFO, as discussed in Chapter One. Following the appropriate selection of the AAAFO along with the design and material of the FW, the initial SVA can be further adjusted by fine-tuning of the AFO-FC. Fine-tuning of the AFO-FC is considered mandatory for gait optimisation and was first described by Meadows.[7] The process includes adding or removing the heel or sole height to alter the HSD or altering the characteristics of the FW.[7, 8, 154] It is recognised that temporal midstance is a fundamental phase of gait for AFO-FC tuning.[69, 71, 154]

Tuning of the AFO-FC amongst subjects with CP has been previously addressed by a number of studies.[7, 186-191] The results of these studies demonstrate the positive influence of tuning the AFO-FC. Improved FZ2,[7] knee kinematics and kinetics[7, 186, 188-191] and hip kinematics and kinetics[7] have been reported. However, it has been established that in studies examining the effect of AFO-FC tuning on gait, insufficient details about participants, AFO-FC details and the tuning process were provided, which prevented drawing clear conclusions for clinical application.[14, 192]

As discussed earlier, altering the alignment of the AFO-FC primarily affects the temporal midstance, while altering the heel and sole profile influences early stance and terminal stance, respectively. However, 4 studies found that the effect of modifying

the SVA on knee kinetics and kinematics extends to include the terminal stance.[178, 186, 190, 193] Additionally, one of the studies suggested that altering the sole of the FW further improves the knee kinetics in terminal stance and preswing.[193] As discussed in Chapter One, thigh kinematics during gait are important for manipulating the GRF for stance stability and this has been well-documented.[11, 12] However, the effect of optimising the temporal midstance shank kinematics on thigh kinematics has not been considered in the studies found.

To the best of the author's knowledge, the present study is the first to investigate the effect of the use of an AFO-FC on the shank and thigh kinematics in temporal midstance to late stance. In this study, temporal midstance to late stance phase is defined as the phase starting from the point when the contralateral medial malleolus is in line with the dominant side to the point of the maximum inclination of the thigh. Additionally, this study investigated if optimisation of the temporal midstance shank kinematics leads to systematic changes in a range of variables of gait, i.e. do these variables optimise simultaneously when the temporal midstance shank kinematics is optimised? The variables and the points chosen for the examination are as follows:

- The thigh kinematics and alignment of the GRF in relation to hip and knee joints in temporal midstance
- The shank kinematics, thigh kinematics, alignment of the GRF in relation to hip and knee joints and FZ2 when FZ2 occurs during gait
- The shank kinematics, thigh kinematics and alignment of the GRF in relation to hip and knee joints when the maximum TVA occurs during gait

2.2 Methods

2.2.1 Participants

This retrospective study received appropriate ethical approval from the West of Scotland Research Ethics Service, the Clinical Research & Development Office of Greater Glasgow & Clyde Health Board and Caldicott Guardian. All the data were collected from the Neurobiomechanics Department, West of Scotland Mobility and Rehabilitation Centre, Queen Elizabeth University Hospital, Glasgow, UK.

The inclusion criteria for this study required that each subject:

- was diagnosed with diplegic CP
- was aged≥7
- did not suffer from any conditions resulting in lower limb sensory deficit
- was ambulatory (with/without walking aids)
- underwent related physical assessment
- underwent lower limb kinematic and kinetic gait data collection

Participants were excluded if they had fixed static lower limb flexion contracture deformity of the hip and/or knee joint. Participants from age 7 were included because it is accepted that an adult gait pattern is reached by that age.[194-198]

Sixteen participants were identified from a pre-existing patient database in the West of Scotland Mobility and Rehabilitation Centre, Queen Elizabeth University Hospital. A total of 16 limbs were included (dominant side was included). If the dominant limb did not meet the inclusion criteria, the contralateral limb was included. Involved

participants were aged between 7 and 17 years (11 years±3 years). All participants were ambulatory (14 independent, 2 with assistive devices). Further information about the participants is presented in Table 2-1, following the best practice reporting guidelines for AFO intervention in subjects with CP.[14, 199]

2.2.2 Study design

A full clinical examination, including a detailed physical examination and clinical gait analysis, was carried out for each subject as part of routine clinical procedure at the Neurobiomechanics Department by the same group of examiners.[200] All the physical assessments were carried out by the same two physiotherapists who are specialised in paediatric neurological conditions. 2D video vector data (sagittal and coronal plane) and kinetic data were collected for each subject using the Vicon Giganet, with 10 x Vicon Bonita B10 2 x Bonita 720c digital video cameras in conjunction with 2 x AMTI 2416-1000 force plates (A & B). A plug-in gait model was applied, and all the subjects were asked to walk barefoot at their own chosen speed. Trials with clean foot strike on either of the force platforms (A or B) were used for analysis and to construct the GRF graphs (normalised to BW). Nexus software was used to process and analyse the kinetics data collected. All the TVAs and SVAs in barefoot and with an AFO-FC was measured from the 2D video data. The sagittal plane video was played back several times to clearly identify the 3 investigated instances; then, the video was paused at each phase, and the SVA and TVA were measured by placing the UG on the screen. For the SVA, the stable arm of the UG was set to follow the vertical, and the movable arm was set to follow the line bisecting the shank. For the TVA, the stable arm of the UG was set to follow the vertical, and the movable arm was set to follow the line

bisecting the thigh (Figure 2-4). The alignment of the GRF was categorised according to its distance from the joint centres following a modified scale developed by the clinical team of the Neurobiomechanics department.[201] This scale is colour and number coded, as indicated in Figure 2-3.

Three instances of gait were identified for examination: temporal midstance, when the contralateral medial malleolus (swinging side) is in line with the dominant side (stance side); FZ2, identified as maximum height of the GRF by the clinical team from the vertical component of ground reaction force graphs; and the maximum TVA (the maximum inclination of the thigh) which is identified by measuring the TVA from the motion video captured. Initially, the maximum TVA was identified visually from the video; then, it was confirmed by measuring the TVA using the UG, as described above.

2.2.3 Statistical Analysis

Using similar data from previous work about CP barefoot gait parameters, the estimated standard deviation (SD) of 13.3º was found.[202] A clinically relevant change of 10º of gait parameters with the AFO has been identified within the research team. Thus, it should be possible to see such a clinical change at the 5% level of significance with a power of 80% if 16 participants are recruited.

A paired sample test (paired t-test/paired Wilcoxon test) was employed to establish the effect of the AFO-FC on the SVA, TVA and FZ2 data in comparison to barefoot walking. This was achieved using one tailed t-test with a significant level set at 0.05.

Initially, data was tested for normality using the Kolmogorov-Smirnov test. If normality was achieved, then the paired t-test was applied. If normality was not achieved, the equivalent nonparametric test was used. To compensate for the fact that a 4º change above the normal (12º inclination) is the same as a 4º change below normal, absolute values were used for the SVA and TVA. McNemar's test[203] was applied for analysis of GRF alignment. The ideal GRF location was rated as (1) if it was within normal boundaries (green colour) and (0) if it was not within normal boundaries (amber, red and brown colours) (Figure 2-3). These boundaries are a modification of a scale used and developed by the Neurobiomechanics Department.[201]

The normalised GRF vertical component was plotted with the AFO-FC and barefoot to illustrate the effect of the AFO-FC on optimising the GRF vertical force. Adaptations of the classification of 'Ben Lomonding' described by Williams et al.[18] and Gibbs[204] was applied to categorise the barefoot and AFO-FC graphs, as illustrated in Table 2-2.

Variability within normal subjects exists for the SVA and TVA; however, for statistical analysis purposes an approximate single value for the TVA and SVA was required to be identified (Table 2-3). These approximate control values were obtained from 10 healthy subjects (30.8±4.6 years) and collected by the same department (unpublished data) using 3D analysis system. For the SVA and TVA, the sequence of rotation used to calculate the 3D segment orientation angles was tilt-rotation-obliquity, as recommend by the International Society of Biomechanics.[205] The SVA was measured following the line from the ankle joint centre to the knee joint centre, against

the vertical. The TVA was measured following the line from the knee joint centre to the hip joint centre, against the vertical. The reported mean SVA value by the department (11.8º±3.5º SD) was found to be approximately similar to the value reported by Pratt et al.[206] (11.4º±3.4º SD), although a different measuring method was implemented. These values are approximately similar to the values reported previously, which stated an optimum SVA of 10-12º (11.86º±2º SD) with the use of an AFO-FC amongst subjects with CP.[8] The TVA values obtained from the department differed from the values reported by Perry and Burnfield.[59] However, the values reported by the department were used because insufficient information about the participants and method of measuring were provided by Perry and Burnfield.[59] In order to identify the 3 different instances amongst normal subjects, an approximate percentage of 30% of gait for temporal midstance, 45% for FZ2 and 50% for the maximum TVA was chosen according to a number of gait references.[59, 181, 207-209] However, it is acknowledged that variation in the values of gait parameters within normal subjects does exist.

The temporal midstance SVA was correlated with each examined variable barefoot and with an AFO-FC. This was done to examine how optimisation of the temporal midstance shank kinematics using an AFO-FC affects these variables and whether these variables optimise systematically (i.e. if the correlation increases using an AFO-FC). For correlations, participants were divided into two groups: Group One: participants with barefoot SVA less inclined than normal and normal (n=7) and Group Two: participants with barefoot SVA more inclined than normal (n=9) (Figure 2-6). The correlation values are graded as follows: 0.00-0.25 is a low correlation, 0.25-0.50

is a fair correlation, 0.50–0.75 is a good correlation and a value above 0.75 is excellent correlation.[210]

2.3 Results

2.3.1 T-test

A paired t-test was applied as normality was achieved for all the elements investigated. Table 2-4 illustrates all the values achieved using the t-test. As shown in Table 2-4, some t-values were large, e.g. 8.31, whilst all the p-values were significant, other than FZ2's. The narrowest CI range was found to be for the TVA at temporal midstance with the smallest SD value (Table 2-4).

2.3.2 McNemar's test

McNemar's test was applied to investigate the effect of an AFO-FC on optimising the GRF alignment in relation to the hip and knee joints in each phase examined. None of the p-values were significant (Table 2-5). This indicates that the proportion of limbs with abnormal alignment of the GRF in relation to the hip and knee joints with and without AFO intervention is not significantly different. Table 2-6 represent the distribution of limbs according to the effect of an AFO-FC on the alignment of the GRF. An AFO-FC had the greatest effect on the hip joint at the maximum TVA as 7 limbs moved from having abnormal to normal alignment (Table 2-6). However, 5 limbs were affected negatively, moving from normal to abnormal alignment (Table 2-6). On the other hand, an AFO-FC had the least effect on the hip joint at temporal midstance as 11 limbs started with abnormal alignment and stayed abnormal with an AFO-FC (Table 2-6).

2.3.3 Correlation

The SVA was correlated with each investigated element barefoot and with an AFO-FC. Pearson's correlation was used for the SVA, TVA and FZ2 at the 3 studied instances. Meanwhile, Spearman's correlation was used for the GRF location. The correlation values (barefoot and with an AFO-FC) differed between both groups, indicating the influence the initial barefoot SVA has on the outcome. The results for each group are stated below.

2.3.3.1 Group 1

Barefoot: only FZ2 resulted in excellent and significant correlation value. The rest of the correlation values were low and not significant (Table 2-7). P-values for the TVA and GRF alignment at the knee at FZ2 were found to be close to significance with good values of correlation.

AFO-FC: two non-significant values became significant with good correlation (Table 2-7). The GRF alignment at the hip joint at FZ2 correlated negatively with the SVA. On the other hand, the GRF alignment at the knee joint at temporal midstance correlated positively with the SVA at temporal midstance. It should be noted that FZ2 correlation value changed from being a positive and excellent correlation value to a negative, non-significant and fair correlation value with the use of an AFO-FC. Furthermore, the TVA and GRF alignment at the knee at FZ2 moved from being close to significance with good values of correlation to a low/fair correlation and non-significant p-value.

2.3.3.2 Group 2

Barefoot: only the GRF alignment at the knee joint at the maximum TVA resulted in a good and significant correlation value with the SVA at temporal midstance (Table 2-8). Meanwhile, the alignment of the GRF at the knee joint at FZ2 was found to be close to significance with a good value of correlation.

AFO-FC: As seen in Table 2-8, FZ2 and the TVA at the maximum TVA moved from low correlation and non-significant values to good correlation and significant values. The GRF alignment at the knee joint at the maximum TVA moved from good to excellent correlation value with an AFO-FC in comparison to barefoot. The alignment of the GRF at the knee joint at FZ2 moved from being close to significance with a good value of correlation to low and non-significant correlation.

2.3.4 Vertical component of ground reaction force (Ben Lomonding)

The use of an AFO-FC decreased the degree of Ben Lomonding in 9 participants (Table 2-2 & 2-9). Amongst these 9 participants, the use of AFO-FC assisted 5 participants in supporting their BW in late stance (FZ2≥BW). For example, Figures 2-7 to 2-9 illustrate the graphs of vertical component of GRF in barefoot and with AFO-FC of P1, P2 and P3. In 5 participants, the use of an AFO-FC worsened the degree of Ben Lomonding. In the remaining 2 participants, the type of Ben Lomonding did not change with the use of AFO-FC (Table 2-2).

2.4 Discussion

This study examined the effect of optimisation of shank kinematics in temporal midstance with the use of an AFO-FC on several gait variables which included thigh

kinematics in temporal midstance, shank kinematics, thigh kinematics and FZ2 at FZ2 and shank and thigh kinematics at maximum TVA. Additionally, this study investigated the influence of an AFO-FC on the GRF alignment in relation to the hip and knee in temporal midstance to late stance. Overall, the use of an AFO-FC had a positive significant influence on the thigh kinematics in temporal midstance and on shank and the thigh kinematics at FZ2 and at maximum TVA (Table 2-4). In contrast, a non-significant improvement in the alignment of GRF in relation to the hip and knee joint in temporal midstance to late stance was found (Table 2-5). The use of an AFO-FC helped to increase FZ2 and decreased the degree of Ben Lomonding in 9 participants (Table 2-2 & 2-9). Amongst these 9 participants, 5 participants showed an increase in FZ2≥BW (Figures 2-7 to 2-8). For example, Figures 2-10 to 2-12 illustrate the effect of AFO-FC on shank inclination, thigh inclination and alignment of GRF in relation to the hip and knee joint in temporal midstance for P6, P8 and P11.

When analysing the results of the t-test and McNemar's test clinically, the significance/meaning of the results may differ. The application of the t-test does not consider the measurement error. Clinically, a change of within ±5° might reflect an error in the measurements rather than an actual change in the angle. When accounting for this potential error, improvement in temporal midstance TVA was only found in 4 participants (Table 2-10). At FZ2, the SVA improved in 7 participants, while the TVA was better in 8 participants. Only 5 participants showed an improvement in the SVA at the maximum TVA with the use of an AFO-FC. Additionally, the TVA at the maximum TVA was the variable most positively affected using an AFO-FC. The SVA at the maximum TVA was the variable most negatively affected using an AFO-FC.

Furthermore, the application of the McNemar's test does not account for change in alignment of GRF from worse to bad, i.e. red to amber in relation to the joints. This change is important clinically as moving from worse to bad is a positive effect. This will enable clinicians to monitor the progression seen in subjects with the AFO-FCs. Additionally, this observed change means that the participants are moving from an alignment of a serious cause of concern to an alignment of a moderate cause of concern. Table 2-11, which illustrates the specific change in the GRF alignment, may indicate that a positive effect was found on the alignment of the GRF in relation to the hip joint at maximum TVA, followed by temporal midstance with the use of an AFO-FC. However, in contrast 5 and 4 participants showed a negative effect with the use an AFO-FC on the alignment of the GRF in relation to the hip joint at maximum TVA and in temporal midstance respectively. In general, the number of participants with unchanged alignment was high. However, it should be noted that amongst these participants there were a number that started from the optimum position and stayed the same with the use of an AFO-FC (Table 2-11).

A possible explanation for the unchanged/negative results found above is that the AFO-FCs in some participants allowed further shank inclination than expected during gait. The SVA measured in temporal midstance ($15^{o}\pm4^{o}$) was found to be different than the SVA measured while standing ($9^{o}\pm3^{o}$) (Table 2-12). With the use of an optimum AFO-FC prescription, both angles should be equal. Information about the AFO-FC characteristics, such as material thickness and trimlines, was not available. This was because each subject attended the Neurobiomechanics department with their AFO-FC having been prescribed by their local health service. If required, fine-tuning was

carried out by the clinical team at the Neurobiomechanics Department. To provide further information regarding the AFO stiffness, the SVA in standing was compared to the temporal midstance SVA. (Table 2-12). A difference of greater than ±5° (measurement error) between the SVA measured when standing and the temporal midstance SVA may suggest that the AFO is insufficiently stiff.[211] In six participants the difference in the temporal midstance SVA and SVA when standing was within ±5°, which may indicate that the AFOs were adequately stiff. Meanwhile, in 9 participants the temporal midstance SVA was greater than (>±5°) the SVA when standing, suggesting that the AFOs were insufficiently stiff and allowed further shank inclination during gait (Table 2-12). Additionally, early heel raise may cause further shank inclination. However, this cause was eliminated because none of the participants exhibited early heel raise as observed from the gait videos. The value of the SVA while standing in the remaining participant was not recorded. For analysing purposes, the SVA measured during temporal midstance of gait was used as this value was more representative.

The importance of accommodating the gastrocnemius muscle length has been discussed in Chapter One. The AAAFO of P1 and P14 did not accommodate the gastrocnemius muscle length (2-13). Furthermore, further shank inclination in temporal midstance was observed in these two participants leading to a total shortness in gastrocnemius length of 13° in P1 and 9° in P 14. If the AFO-FC allows further shank inclination during temporal midstance, this may limit the available length of the gastrocnemius even if the AFO is held at an appropriate angle. As seen in Table 2-13, the amount of the change in the temporal midstance SVA from the standing SVA

limited the available length of the gastrocnemius for full knee extension in P12 and P16. For example, in P16 the AAAFO was set at 8° plantarflexion and the SVA in temporal midstance was further inclined by 16°, which led to shortness of 3° in the available length of the gastrocnemius as 5° dorsiflexion was achievable with the knee extended. The shortness in the available length of the gastrocnemius may restrict the knee extension in temporal midstance, terminal stance and terminal swing.[10, 166, 168] This may have caused the negative/no change effect with the use of an AFO-FC on the TVA, SVA, FZ2 and the alignment of the GRF in relation to the hip and the knee joints in these participants. For instance, in P1 and P14 the alignment of the GRF in relation to the knee joint worsened or did not change with the use of an AFO-FC at the 3 selected points (Table 2-14). Additionally, FZ2 decreased and the degree of the Ben Lomonding worsened in P14 and P16 with the use of an AFO-FC (Table 2-9).

When observing the correlation between the barefoot temporal midstance SVA with the investigated elements, the correlation values differed between the groups, indicating the influence of the barefoot SVA. In Group One, the use of an AFO-FC moved two values from fair correlation and non-significant to good correlation and significant values (Table 2-7). The two values were the GRF alignment in relation to the hip joint at FZ2 and the GRF in relation to the knee joint in temporal midstance. This indicates that optimisation of the temporal midstance shank kinematics with the use of an AFO-FC may systematically optimise the hip and knee kinetics at different points of gait. However, FZ2 moved from being excellent correlation to negative and fair correlation. Additionally, the TVA and GRF alignment in relation to the knee joint at FZ2 moved from good correlation and borderline significant values to very low/fair

correlation and non-significant values (Table 2-7). In contrast, FZ2 moved from being low correlation to good correlation and significant in Group Two (Table 2-8). Another difference between the two groups is the GRF alignment in relation to the knee joint at the maximum TVA. In Group One, values for barefoot and with the use of an AFO-FC had good correlation with non-significant p-values (Table 2-7). However, in Group Two the value moved from good correlation to excellent correlation with the use of an AFO-FC (Table 2-8). This indicates that optimisation of the temporal midstance shank kinematics with the use of an AFO-FC may systematically optimise FZ2, knee joint kinetics and the thigh kinematics at the maximum TVA. It should be noted that the results may have been adversely affected because the AFO-FC allowed further shank inclination. Additionally, the small sample size in each group may account for these differences in interpretation; hence, caution is required in any interpretation or generalisation. A study by Ridgewell[178] reported that hip kinematics, hip kinetics and femur projection angles did not change systematically when the AFO-FC alignment was optimised amongst subjects with CP. In contrast, optimising the AFO-FC alignment had a systematic effect on gait variables relating to the knee kinematics and kinetics and ankle kinetics.[178]

Further analysis was carried out to investigate the degree of the agreement between the examined variables. Variables with significant results were used, which included the TVA in temporal midstance, the SVA, TVA and FZ2 at FZ2 and the SVA and TVA at the maximum TVA. A value of (1) was used to indicate that the use of an AFO-FC led to a positive change of>±5° (measurement error). Meanwhile, the value (0) was used to indicate that the use of an AFO-FC led to a negative change of>±5°. If

the AFO-FC caused no effect or a positive or negative change of within ±5°, no value was given to the variable, and the letter N was used. Optimising the temporal midstance shank kinematics caused a positive effect on shank kinematics, thigh kinematics and FZ2 in 2 participants (Table 2-10). Two participants had agreement of 5 out of 6 variables. No influence of the barefoot temporal midstance SVA on the agreement between the variables was observed (Table 2-10).

Meadows[7] included subjects with CP and reported an increase in FZ2 with the use of the optimum prescription of an AFO-FC and fine-tuning, which was similar to the results found in the current study. Additionally, Meadows' study reported improved hip and knee external moments at FZ2 and improved GRF alignment, which does not agree with the results of the current study. This could be due to the characteristics of the AFO-FCs in this study not being optimal. Stallard and Woollam[212] reported improved GRF alignment in the majority of subjects with neurological conditions. However, this study only presented qualitative results, and insufficient details about the participants and the AFO-FC characteristics, e.g. material thickness, were provided. Butler et al.[186] included 5 subjects with CP to examine the effect of a tuned AFO-FC on the knee extension moment during the stance phase. The study reported a significant reduction in knee extension moment and moment arm with the use of the AFO-FC ($p<0.01$). The study also reported improvement in the alignment of the GRF in relation to the knee joint and increase in the magnitude of the GRF during temporal midstance However, no numerical values were reported, and the study failed to provide sufficient information about the AFO-FC characteristics. Furthermore, each subject underwent adjunct therapy; thus, the improvement cannot be related only to the use of an AFO-FC.[186]

All the results of this study were analysed using a single normal value. This limits the representation of normal, as variation does exist in the values considered normal. Another limitation of this study is that the age of the control group differed from the age of subjects with diplegic CP. Another source of variation in this study was the difference in measurement method used to measure the TVA and SVA between the control group and subjects with diplegic CP. In control group, TVA and SVA was measured using 3D analysis system, while the UG was used to measure the TVA and SVA amongst subjects with diplegic CP. Using advanced 2D or 3D measuring systems to measure SVA and TVA may produce more accurate and reliable results in comparison to the measuring method followed in this study. For analysis purposes, an assumption of the timing of temporal midstance, FZ2 and maximum TVA was made. This also limited the generalisability of the results. Insufficient information was provided about the AFO-FC specification, which may have affected the conclusions drawn. Additionally, the GRF alignment was described using a qualitative method. Using quantitative methods, e.g. measuring the perpendicular distance from the centre of the joint to the vector, may be more accurate. The number of participants included in this study met the required power calculation, however CP is not a homogenous disorder and subjects present with a variable degree of disability. This may indicate the need to include more subjects and subdivide them according to similar presentation, e.g. gait patterns, in future studies. Furthermore, being more earnest about the AFO-FC prescription is likely to alter the overall results as further inclination of the shank was observed in all participants. Another important element which was not considered in this study is the effect of the speed on these variables. Some of these

variables may automatically improve due to the increased speed rather than optimisation of the temporal midstance shank kinematics with the use of an AFO-FC.

All physical assessment was carried out by the same clinical team at the Neurobiomechanics Department prior to gait analysis. It should be acknowledged that due to the referral period, the results of the physical examination may have differed from the results of the physical examination carried out by the local health service when prescribing the AFO-FC. Additionally, the methods of carrying out the clinical assessment, prescription criteria and AFO-FC tuning may differ. This may explain the reason for the variability seen in the AFO-FCs prescribed to the participants included in this study. Eddison et al.[213] reported a lack of understanding of the AFO-FC tuning process amongst UK orthotists. In majority of the participants included, the FZ2 had a clear peak (identified by the clinical team). However, in some of the participants, the FZ2 did not exhibit a clear second peak and was identified based on the clinical judgement; hence, caution must be applied when using results related to FZ2.

It has been demonstrated that optimising shank kinematics in temporal midstance using AFO-FC in subjects with spastic diplegic CP may have a positive influence on optimising thigh kinematics in temporal midstance and thigh and shank kinematics when FZ2 and the maximum TVA occur. This, in turn, aids in increasing FZ2 and, in some cases, in supporting the BW and reducing the degree of Ben Lomonding. Some effect of optimising the temporal midstance shank kinematics was observed on improving the GRF alignment in relation to the hip and knee joints in this study.. However, these results should be applied with an understanding of their clinical

implications/significance. The results of this study suggest that some variables of gait may optimise when the temporal midstance shank kinematics is optimised. Additionally, this effect was found to differ based on the barefoot SVA. To the best of the author's knowledge, this is the first study to investigate the effect of optimisation of the temporal midstance shank kinematics on the thigh kinematics during gait. Additionally, the variables involved in paired t-test and McNemar's test were subsequently used in correlation analysis; hence, Bonferroni correction factor should be used on p-values. However, given the general lack of significance, very few results would have been affected if significant was reduced by the Bonferroni correction factor to $p<0.02$. The overall implication determined from the p-values (Table 2-7 & 2-8) did not change.

This study highlighted the influence of the AFO-FC characteristics on gait. Further shank inclination was observed during gait in 9 subjects, which may have caused the negative/no change effect of the AFO-FC in the investigated variables. Further shank inclination may be caused by the AFO being insufficiently stiff, undesirable FW characteristics and un-accommodated gastrocnemius muscle length. As emphasised in Chapter One, detailed physical assessment is a mandatory step in forming an appropriate AFO-FC prescription. The neuropathology of CP selectively targets the bi-articular and multi-articular muscles. Additionally, unlike the bi-articular and multi-articular muscles, the mono-articular muscles rarely become shortened due to the growth pattern and biomechanics of movement in subjects with CP.[117] This highlights the importance of examining length and spasticity in bi-articular and multi-articular muscles including the RF, gastrocnemius and hamstrings. Careful assessment of these

muscles will enhance understanding of the functional limitations presented. Additionally, this will allow determinination of the optimum intervention plan and the AFO-FC prescription, e.g. AAAFO. The following chapter will discuss the normal function of one of the commonly affected bi-articular muscle, the RF, during gait. The literature related to the common assessment techniques used to assess the length and spasticity of the RF and the effect of RF spasticity on gait is presented in the next chapter. The chapter will also discuss the results of a study investigating the relationship between the catch angle/length of the RF measured during physical assessment and the hip or the knee joint during gait.

2.5 Figures

Figure 2-1: Normal shank inclination, thigh inclination and alignment of GRF in relation to the hip and knee joint in temporal midstance.

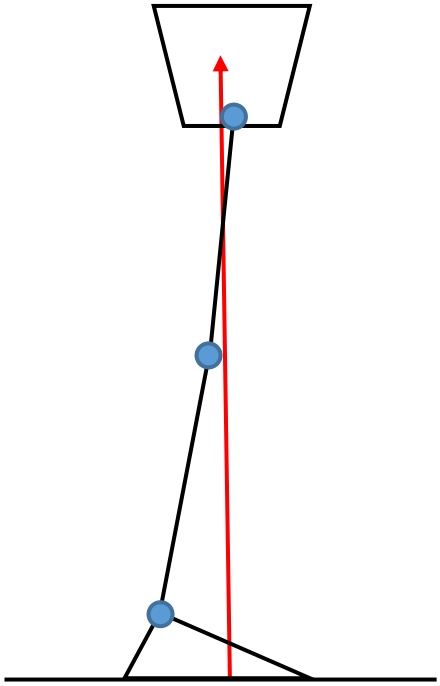


Figure 2-2: Normal shank inclination, thigh inclination and alignment of GRF in relation to the hip and knee joint at FZ2.

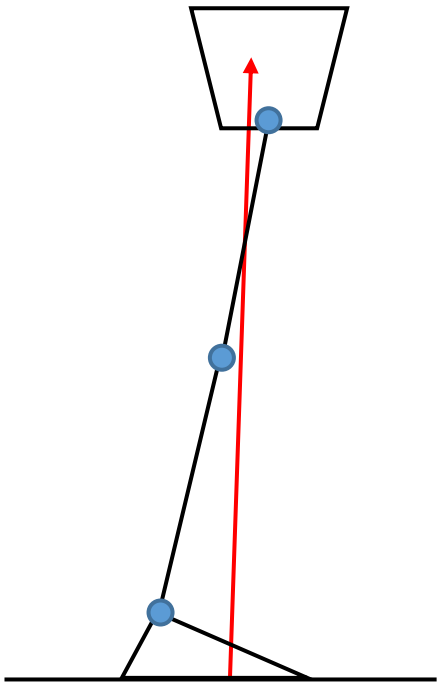


Figure 2-3: Normal shank inclination, thigh inclination and alignment of GRF in relation to the hip and knee joint at maximum TVA.

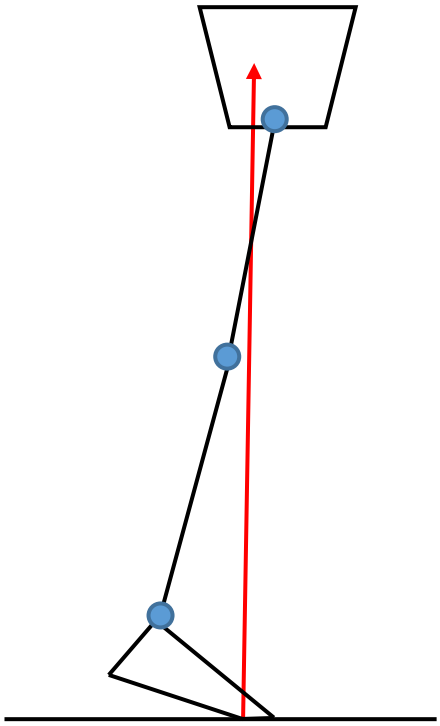


Figure 2-4:Measurement method followed to determine the position of the segment relative to the vertical

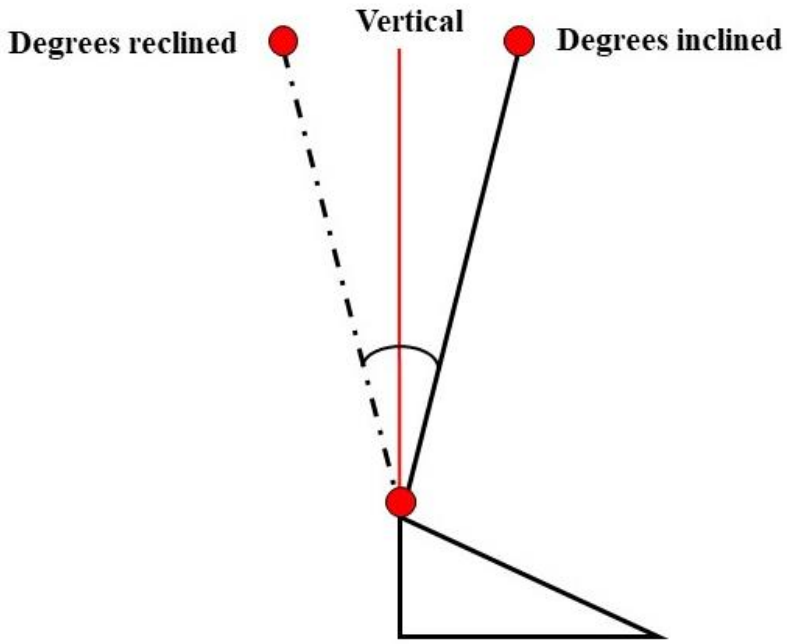


Figure 2-5: Modification of a scale used and developed by the Neurobiomechanics Department to classify the ground reaction force alignment in relation to the hip and knee joints in temporal midstance, at FZ2 and at the maximum thigh to vertical angle.[201]

At temporal midstance

Knee joint

2	1	1	0	0	1	2	3	3
Ext GOSM	Ext OSM	Ext ASM	Ext WSM	KJC	Flex WSM	Flex ASM	Flex OSM	Flex GOSM

Hip joint

2	2	1	0	0	1	2	3	3
Ext GOSM	Ext OSM	Ext ASM	Ext WSM	HJC	Flex WSM	Flex ASM	Flex OSM	Flex GOSM

At the FZ and at the maximum thigh to vertical angle

Knee joint

2	1	0	0	0	1	2	2	3
Ext GOSM	Ext OSM	Ext ASM	Ext WSM	KJC	Flex WSM	Flex ASM	Flex OSM	Flex GOSM

Hip joint

2	1	0	0	1	1	2	3	3
Ext GOSM	Ext OSM	Ext ASM	Ext WSM	HJC	Flex WSM	Flex ASM	Flex OSM	Flex GOSM

GRF alignment: Flex=the GRF causes joint flexion, Ext=the GRF causes joint extension, KJC=the GRF passes through the knee joint centre, HJC=the GRF passes through the hip joint centre, ASM=at skin margins, OSM=outside skin margins, WSM=within skin margins, GOSM=grossly outside skin margins.

Descriptions above in green are optimal (0), amber indicates a moderate cause for concern (1), and red and brown indicate a serious cause for concern (2 & 3).

Figure 2-6: Participants division into Group One and Group Two based on shank inclination.

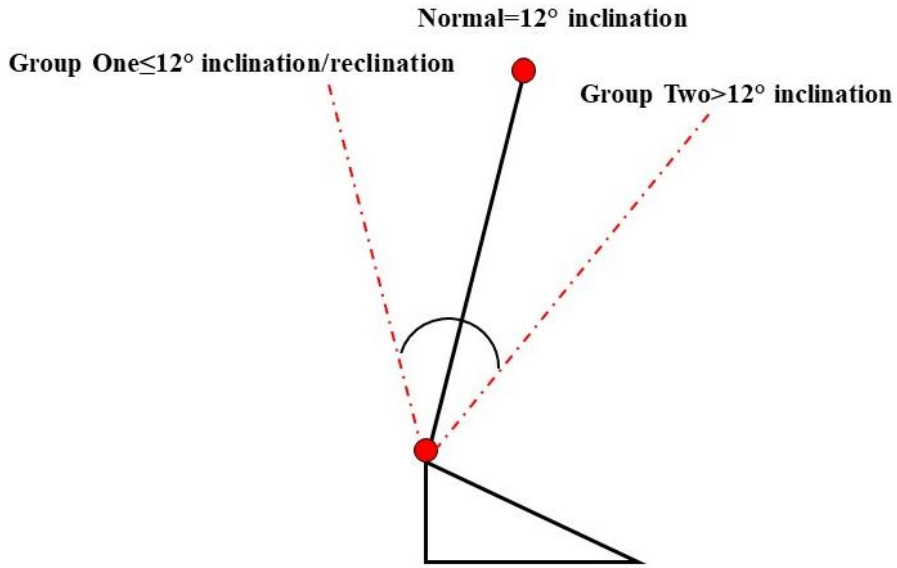
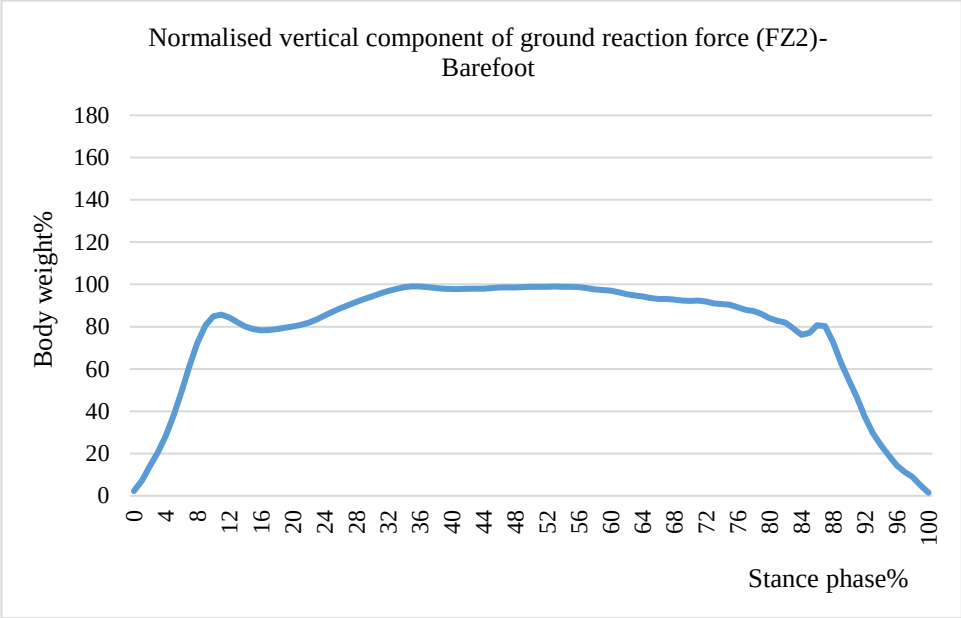


Figure 2-7: Participant 1's normalised vertical component of ground reaction force (FZ2) barefoot and with ankle-foot orthosis-footwear combination (AFO-FC).



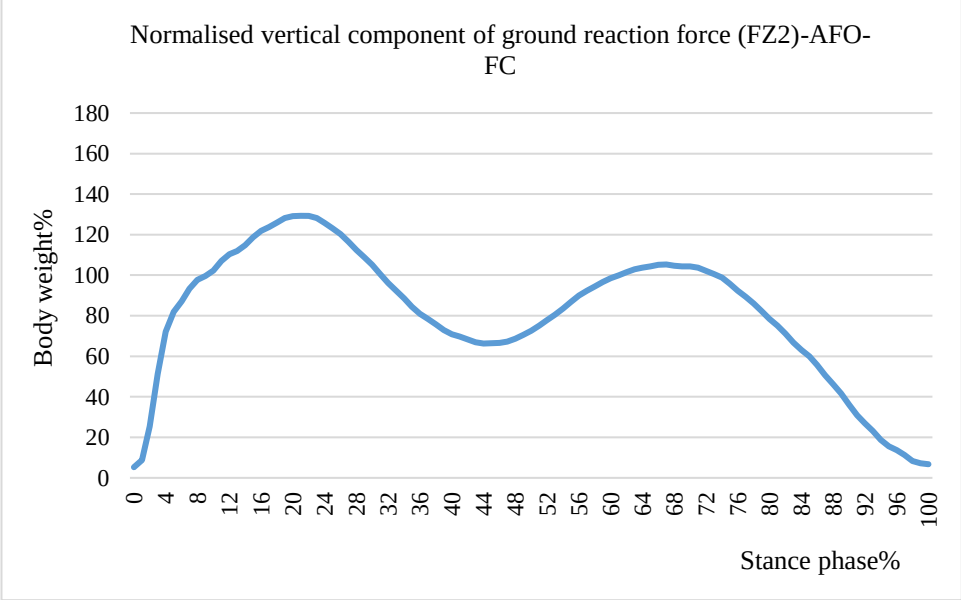
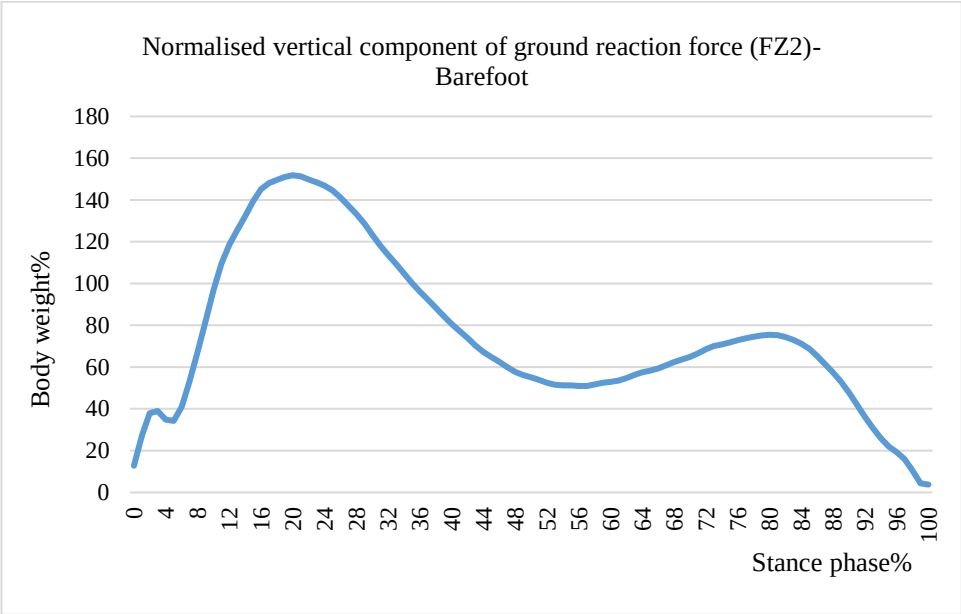


Figure 2-8: Participant 2's normalised vertical component of ground reaction force (FZ2) barefoot and with ankle-foot orthosis-footwear combination (AFO-FC).



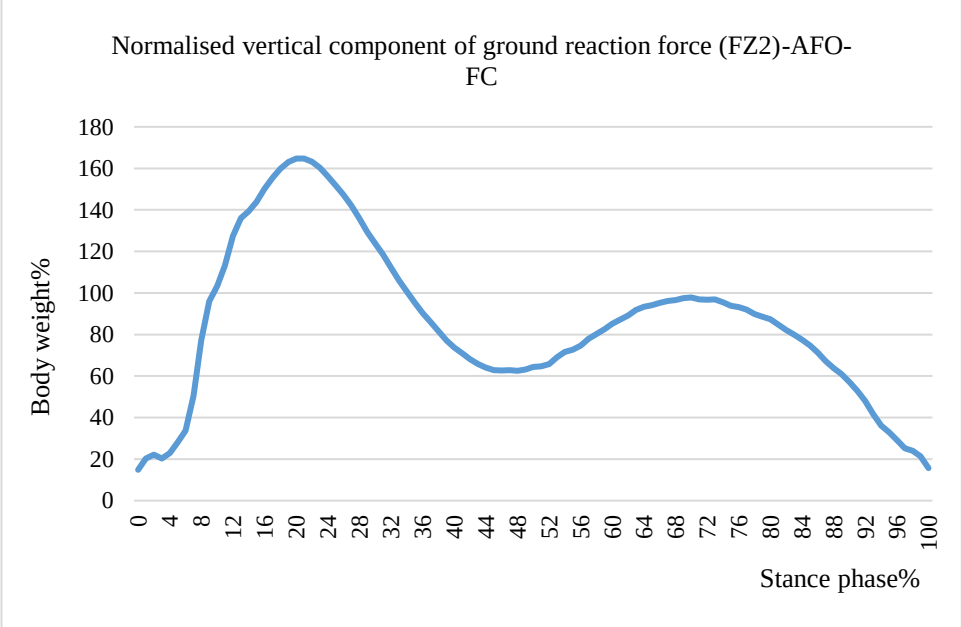
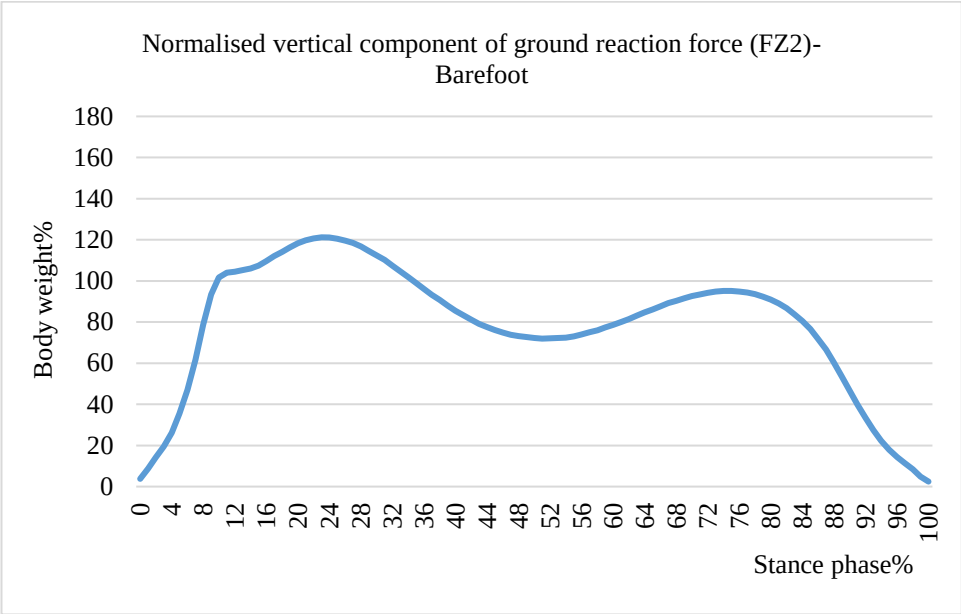


Figure 2-9: Participant 3's normalised vertical component of ground reaction force (FZ2) barefoot and with ankle-foot orthosis-footwear combination (AFO-FC).



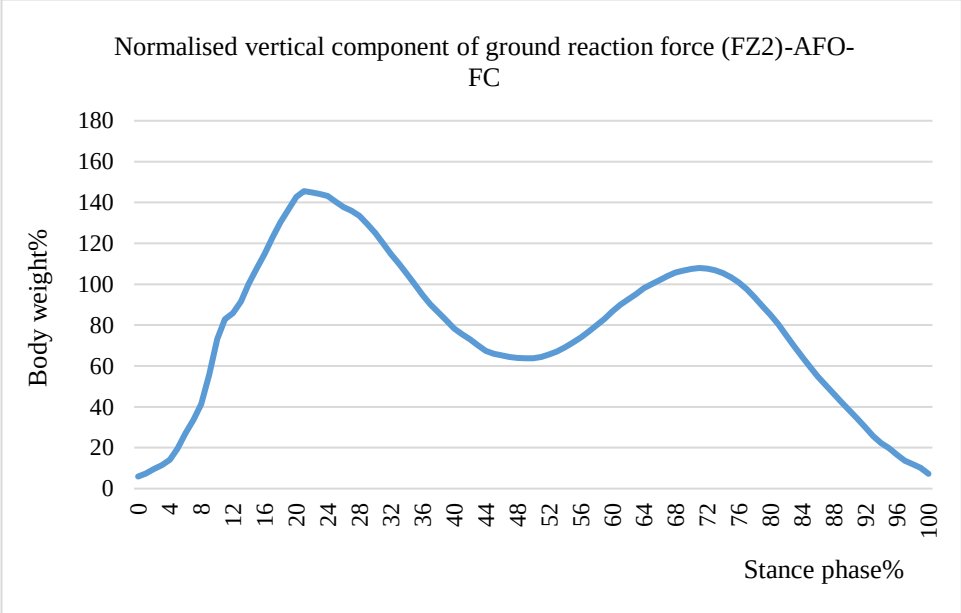
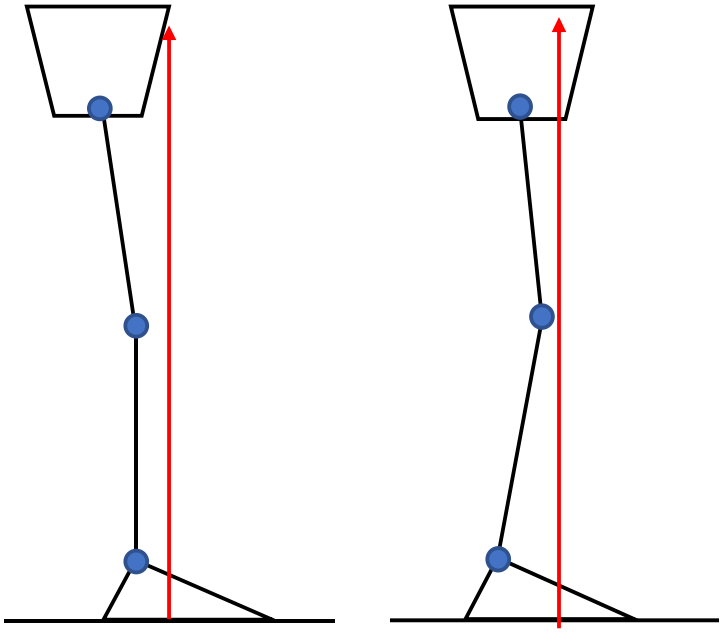


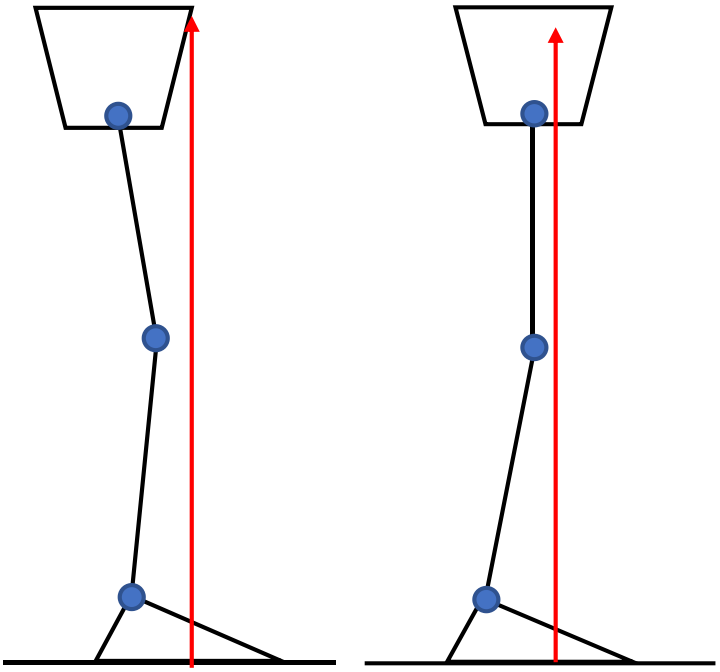
Figure 2-10: Effect of AFO-FC on shank inclination, thigh inclination and alignment of GRF in relation to the hip and knee joint in temporal midstance for Participant 6



Barefoot | With AFO-FC

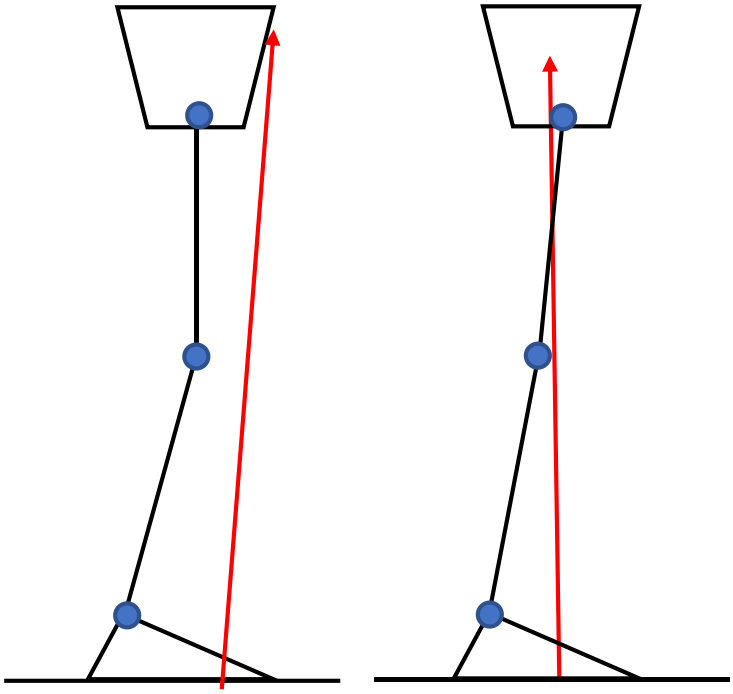
Shank	Vertical	12º inclination
Thigh	10º reclination	4º reclination
Alignment of GRF at knee	Extension at skin margin	Extension within skin margin
Alignment of GRF at hip	Flexion at skin margin	Flexion within skin margin

Figure 2-11: Effect of AFO-FC on shank inclination, thigh inclination and alignment of GRF in relation to the hip and knee joint in temporal midstance for Participant 8



	Barefoot	With AFO-FC
Shank	7º inclination	12º inclination
Thigh	8º reclination	Vertical
Alignment of GRF at knee	Extension at skin margin	Extension within skin margin
Alignment of GRF at hip	Flexion at skin margin	Flexion within skin margin

Figure 2-12: Effect of AFO-FC on shank inclination, thigh inclination and alignment of GRF in relation to the hip and knee joint in temporal midstance for Participant 11



Barefoot With AFO-FC

Shank	18º inclination	12º inclination
Thigh	Vertical	6º inclination
Alignment of GRF at knee	Extension at skin margin	Extension within skin margin
Alignment of GRF at hip	Flexion at skin margin	Extension within skin margin

2.6 Tables

Table 2-1: Information about the participants included: age, GMFCS, side included, hip and knee extension ROM, ankle ROM with knee extended (slow and fast), and AAAFO.

Code	Age	GMFCS Level	Side included	Knee extension ROM	Hip extension ROM	Ankle ROM with knee extended (slow)	Ankle ROM with knee extended (fast)	AAAFO
P1	12	2	L	3H	4	6PF	17PF	0
P2	10	2	R	0	10	3DF	9PF	10PF
P3	13	2	L	7H	2	4PF	10PF	No info
P4	14	2	L	3H	4	10DF	4PF	10DF
P5	16	3	R	0	4	9DF	10PF	0
P6	9	3	L	2H	0	4PF	18PF	10PF
P7	9	2	L	2H	9	6 DF	12 PF	10 PF
P8	12	3	R	0	9	2 DF	No catch	0
P9	7	3	L	4H	7	0	18 PF	14 PF
P10	15	3	L	0	11	3DF	12PF	0
P11	7	2	R	0	6	4DF	10PF	No info
P12	17	3	R	0	10	9 DF	10 PF	0
P13	10	2/3	R	5 H	2	0	10PF	0
P14	8	2/3	L	2 H	6	2 PF	9 PF	0
P15	10	1	R	3 H	6	3 DF	2 PF	No info
P16	11	3	L	0	18	5 DF	10 PF	8 PF

SVA: shank to vertical angle, H: hyperextension, PF: plantarflexion, DF: dorsiflexion, GMFCS: Gross Motor Function Classification System, AAAFO: angle of the ankle of the ankle-foot orthosis, ROM: range of motion.

Table 2-2: Degree of Ben Lomonding, adaptation of a scale by Williams et al.[11] and Gibbs.[204]

Type		Definition
0	FZ2	≥105% BW
1	FZ2	100 to 104% BW
2	FZ2	95 to 99% BW
3	FZ2	90 to 94% BW
4	FZ2	80 to 89% BW
5	FZ2	≤79% BW

BW: body weight, FZ2: second peak of the vertical component of the ground reaction force.

Table 2-3: The normal values used for SVA and TVA in temporal midstance to late stance.

Normal values (°of inclination)	Temporal midstance		FZ2		Maximum TVA	
	Range (SD)	Used value	Range (SD)	Used value	Range (SD)	Used value
SVA	8.5-15.5° (3.5)	12°	22-28° (3)	25°	30-36° (3)	33°
TVA	4-8° (2)	6°	19-23° (2)	21°	19-25° (3)	22°

SVA: shank to vertical angle, TVA: thigh to vertical angle, SD: standard deviation, FZ2: second peak of the vertical component of the ground reaction force.

Table 2-4: T-test values for SVA, TVA and FZ2 in temporal midstance to late stance.

Phase of gait	Elements	Absolute Mean	SD	t-value	Df	P-value	95% CI	
							Lower	Upper
Temporal midstance	TVA	3.62	3.07	4.71	15	<0.001	1.99	5.26
At FZ2	SVA	8.31	6.70	4.96	15	<0.001	4.74	11.88
	TVA	5.37	3.81	5.64	15	<0.001	3.34	7.40
	FZ2	6.37	16.95	1.50	15	0.076	-2.66	15.41
At maximum TVA	SVA	7.43	5.45	5.45	15	<0.001	4.53	10.34
	TVA	8.18	4.46	7.34	15	<0.001	5.81	10.56

SVA: shank to vertical angle, TVA: thigh to vertical angle, FZ2: second peak of the vertical component of the ground reaction force, SD: standard deviation, df: degrees of freedom, P-values in red: significant values at 0.05 level.

Table 2-5: P-values for McNemar's test for GRF alignment in relation to the hip and knee joint in temporal midstance to late stance.

Phase/joint	Knee	Hip
Temporal midstance	0.5	0.5
At FZ2	0.5	0.5
At maximum TVA	0.34	0.38

TVA: thigh to vertical angle, FZ2: second peak of the vertical component of the ground reaction force, GRF: ground reaction force.

Table 2-6: The effect of an AFO-FC on the alignment of GRF in relation to the hip and knee joint in temporal midstance to late stance.

Temporal midstance: Knee

Barefoot	With AFO-FC	
	Abnormal	Normal
Abnormal	8	3
Normal	3	2

Hip

Barefoot	With AFO-FC	
	Abnormal	Normal
Abnormal	11	1
Normal	2	2

At FZ2: knee

Barefoot	With AFO-FC	
	Abnormal	Normal
Abnormal	7	2
Normal	1	6

Hip

Barefoot	With AFO-FC	
	Abnormal	Normal
Abnormal	8	2
Normal	1	5

At maximum thigh to vertical: knee

Barefoot	With AFO-FC	
	Abnormal	Normal
Abnormal	6	2
Normal	4	4

Hip

Barefoot	With AFO-FC	
	Abnormal	Normal
Abnormal	3	7
Normal	5	1

TVA: thigh to vertical angle, FZ2: second peak of the vertical component of the ground reaction force, GRF: ground reaction force, AFO-FC: ankle-foot orthosis-footwear combination.

Table 2-7: Temporal midstance correlation results and p-values with each investigated element in Group One.

Temporal midstance SVA		Temporal midstance			FZ2					Maximum TVA			
		TVA	GRF alignment		SVA	TVA	GRF alignment		FZ2	SVA	TVA	GRF alignment	
			Knee	Hip			Knee	Hip				Knee	Hip
Barefoot	Pearson and Spearman Correlation	0.46	-0.42	0.02	0.50	0.65	-0.65	0.50	0.79	0.50	-0.42	0.53	-0.22
	P-value	0.14	0.16	0.48	0.12	0.055	0.056	0.12	0.01	0.12	0.16	0.10	0.31
With AFO-FC	Pearson and Spearman Correlation	-0.44	0.73	0.18	-0.57	0.19	-0.34	-0.72	-0.47	-0.29	0.18	-0.52	-0.24
	P-value	0.16	0.03	0.34	0.09	0.33	0.22	0.03	0.14	0.26	0.34	0.11	0.30

SVA: shank to vertical angle, TVA: thigh to vertical angle, FZ2: second peak of the vertical component of the ground reaction force, GRF: ground reaction force, AFO-FC: ankle-foot orthosis-footwear combination, P-values in red: significant values at 0.05 level, p-values in blue: borderline significance at 0.05 level.

Table 2-8: Temporal midstance correlation results and p-values with each investigated element in Group Two.

Temporal midstance SVA		Temporal midstance			FZ2					Maximum TVA			
		TVA	GRF alignment		SVA	TVA	GRF alignment		FZ2	SVA	TVA	GRF alignment	
			Knee	Hip			Knee	Hip				Knee	Hip
Barefoot	Pearson and Spearman Correlation	-0.19	-0.03	-0.48	0.35	-0.22	0.57	-0.45	0.14	-0.42	-0.08	0.67	-0.19
	P-value	0.31	0.46	0.09	0.17	0.27	0.054	0.11	0.35	0.12	0.41	0.02	0.31
With AFO-FC	Pearson and Spearman Correlation	0.34	0.08	-0.51	0.11	0.20	0.20	0.095	0.69	0.02	0.61	0.79	-0.04
	P-value	0.18	0.41	0.07	0.38	0.30	0.29	0.40	0.01	0.47	0.03	0.006	0.45

SVA: shank to vertical angle, TVA: thigh to vertical angle, FZ2: second peak of the vertical component of the ground reaction force, GRF: ground reaction force, AFO-FC: ankle-foot orthosis-footwear combination, P-values in red: significant values at 0.05 level, p-values in blue: borderline significance at 0.05 level.

Table 2-9: FZ2 values and Ben Lomonding type in barefoot and with AFO-FC.

P	FZ2 in barefoot%	Type	FZ2 with AFO-FC%	Type
P1	92	2	105	0
P2	76	5	97	2
P3	95	2	108	0
P4	99	2	84	4
P5	92	3	103	1
P6	72	5	100	1
P7	89	4	115	0
P8	80	4	93	3
P9	71	5	98	2
P10	85	4	89	4
P11	80	4	99	2
P12	107	0	108	0
P13	96	2	81	4
P14	95	2	85	4
P15	106	0	99	2
P16	92	3	65	5

P: participant, AFO-FC: ankle-foot orthosis-footwear combination, FZ2: second peak of the vertical component of the ground reaction force.

Table 2-10: The agreement between the investigated variables with positive significant influence with the use of an AFO-FC.

Participants	Temporal midstance	FZ2			Maximum TVA		Overall influence		
	TVA	SVA	TVA	FZ2	SVA	TVA	Better	Same	Worse
P1	N	1	N	1	1	1	4	2	-
P2	N	1	N	1	0	N	2	3	1
P3	N	N	1	1	N	1	3	3	-
P4	N	1	N	0	1	1	3	2	1
P5	N	N	1	1	N	1	3	3	-
P6	1	1	1	1	1	1	6	-	
P7	1	1	1	1	0	1	5	-	1
P8	1	1	1	1	1	1	6	-	-
P9	0	N	N	1	0	0	1	2	3
P10	N	1	1	1	1	1	5	1	-
P11	1	1	N	1	0	N	3	2	1
P12	N	N	1	1	0	1	3	2	1
P13	N	0	N	0	0	0	-	2	4
P14	N	N	N	0	0	1	1	3	2
P15	N	0	N	0	N	0	-	3	3
P16	N	N	N	0	N	N	-	5	1
Overall positive influence	4	8	7	11	5	10			

P: participant, SVA: shank to vertical angle, TVA: thigh to vertical angle, FZ2: second peak of the vertical component of the ground reaction force, AFO-FC: ankle-foot orthosis-footwear combination, (0): indicates negative influence, (1): indicates positive influence, N: indicates no change.

Table 2-11: The effect of an AFO-FC on the alignment of the GRF in relation to the hip and knee joints in temporal midstance to late stance.

	Temporal midstance		FZ2		Maximum TVA	
	Knee	Hip	Knee	Hip	Knee	Hip
Better	3	6	4	3	3	7
Worse	5	4	1	1	5	5
Same	8	6	11	12	8	4
Ideal position	2*	2*	6*	5*	4*	1*

(*): indicates the number of participants with optimum alignments before and after an AFO-FC, TVA: thigh to vertical angle, FZ2: second peak of the vertical component of the ground reaction force, GRF: ground reaction force, AFO-FC: ankle-foot orthosis-footwear combination.

Table 2-12: AFO characteristics for each participant, comparing the SVA measured while standing and in temporal midstance.

Participants	SVA measured at temporal midstance	SVA measured while standing	AFO characteristics
P1	+17	+10	Insufficiently stiff
P2	+18	+7	Insufficiently stiff
P3	+15	+7	Insufficiently stiff
P4	+14	+11	Stiff
P5	+12	+10	Insufficiently stiff
P6	+13	No info	-
P7	+12	+11	Stiff
P8	+12	+10	Stiff
P9	+18	+10	Insufficiently stiff
P10	+10	+6	Stiff
P11	+12	+12	Stiff
P12	+18	+5	Insufficiently stiff
P13	+12	+8	Stiff
P14	+16	+9	Insufficiently stiff
P15	+15	+8	Insufficiently stiff
P16	+26	+10	Insufficiently stiff

SVA: shank to vertical angle, AFO: ankle-foot orthosis, (+): indicates shank inclination, Stiff: the difference is within±5°, insufficiently stiff, the difference is above±5°.

Table 2-13: Difference between the SVA in temporal midstance and when standing, gastrocnemius length with the knee extended, AAAFO and the active/available length of the gastrocnemius muscle in participants with insufficiently stiff AFO.

P	Difference in SVA	Gastrocnemius length	AAAFO	Gastrocnemius active/available length
P1	7	6PF	0	-13
P2	11	3DF	10PF	+2
P3	8	4PF	No info	-
P9	8	0	14 PF	+6
P12	13	9 DF	0	-4
P14	7	2 PF	0	-9
P15	7	3 DF	No info	-
P16	16	5 DF	8 PF	-3

P: participant, PF: plantarflexion, DF: dorsiflexion, SVA: shank to vertical angle, AAAFO: angle of the ankle of the ankle-foot orthosis, AFO: ankle-foot orthosis, (-): indicates shortness in the available gastrocnemius muscle length, (+): indicates the available length of the gastrocnemius.

Table 2-14: GRF alignment in relation to the hip and knee joint in P1 and P14 in barefoot and with an AFO-FC.

P	Temporal midstance				At FZ2				At maximum TVA			
	Knee		Hip		Knee		Hip		Knee		Hip	
	Barefoot	AFO-FC	Barefoot	AFO-FC	Barefoot	AFO-FC	Barefoot	AFO-FC	Barefoot	AFO-FC	Barefoot	AFO-FC
P1	Ext WSM	Ext ASM	Flex ASM	Ext ASM	KJC	Flex WSM	Flex WSM	HJC	KJC	Flex WSM	HJC	Ext WSM
P14	KJC	Ext OSM	HJC	Flex OSM	Flex OSM	Flex OSM	Ext ASM	Ext OSM	Flex ASM	Flex OSM	Ext ASM	Ext OSM

P: participant, GRF: ground reaction force, FZ2: second peak of the vertical component of the ground reaction force, TVA: thigh to vertical angle, Flex=the GRF causes joint flexion, Ext=the GRF causes joint extension, KJC=the GRF passes through the knee joint centre, HJC=the GRF passes through the hip joint centre, ASM=at skin margins, OSM=outside skin margins, WSM=within skin margins.

Chapter 3 Relationship between the catch angle/length of rectus femoris measured using the Duncan-Ely test and the hip and the knee joint during gait

3.1 Introduction

Spasticity (dynamic shortness) and true shortness of the RF is common amongst subjects with diplegic CP. The RF is one of the four muscles which form the quadriceps femoris muscle group. It is a bi-articular muscle which crosses the hip and knee joints. Its action flexes the hip joint and extends the knee joint.

Contradictory evidence was observed in the literature about the exact timing of the RF muscle activation during normal gait. Perry et al.[19, 59] and Nene et al.[20] reported a single burst of RF muscle activity at the transition from stance to swing phase. This single-burst activity was found to increase with increased walking velocity to limit heel rise. Meanwhile, other authors have reported two bursts of activity for the RF during gait.[21, 22, 66, 214-219] The first burst of RF activity occurred at the transition from swing-to-stance phase, while the second burst of activity occurred at the transition from stance-to-swing phase.[22, 66, 214-220] Conrad et al.[21] explain the occurrence of the second burst of RF activity as actively participating in rapid hip flexion and/or controlling passive knee flexion. The first burst of activity of the RF was proposed by a number of studies to help the vastii muscles to control weight acceptance, assist in knee extension and develop the muscle tension required during the loading response to control knee flexion.[22, 214, 215] On the other hand, Perry et al.[19] reported that the RF did not assist the

vastii muscles during the loading response and early midstance. Additionally, they stated that no RF activity or brief bursts of activity in the stance-to-swing transition were found in subjects walking at a self-selected speed.[19] In contrast, a study by Csongradi et al.[23] reported that the RF activity during the transition from stance-to-swing is smaller and non-significant compared with the RF activity in the transition from swing-to-stance phase. It was reported that this inconsistency may be justified due to the different methods of EMG (Electromyography) recording used.[20] Studies that reported two bursts of activity of the RF used surface electrodes, whereas studies that reported a single burst of activity of the RF used fine wire electrodes. The use of fine wire electrodes diminishes any cross talk activity produced from the surrounding muscles, such as the vastus intermedius muscle.[23, 66, 221, 222] Furthermore, Perry et al.[25] classified four patterns of RF muscle activity in subjects with diplegic CP. A total of 45 patients were included, and the following 4 patterns for RF muscle activity were identified using fine wire EMG: activity in swing phase only (n=26), continuous and intense activity throughout gait (n=9), low level stance phase activity and strong swing phase activity (n=5), and isolated stance phase activity prior to initial swing (n=5).[25]

True shortness and spasticity (dynamic shortness) of the RF cause restriction of hip extension when the knee is flexed or restriction of knee flexion when the hip is extended. In the 1980s, it was found that inappropriate activity of the RF muscle during swing phase is the primary cause of stiff-knee gait (SKG) due to generation of excessive internal knee extension moments in swing. The inappropriate activity of the RF in SKG, which causes diminished and/or delayed peak swing phase knee flexion, has been considered an indication for RF transfer surgery.[24-26] Furthermore, it was

found that SKG is not only associated with reduced and/or delayed peak knee flexion but is also associated with reduced total knee ROM during gait[24, 223] and reduced range of knee flexion in early swing phase.[27] In 1996, Piazza and Delp[224] carried out a study using a muscle-actuated dynamic simulation module to examine how muscle action affects the peak knee flexion in swing in normal gait. They reported that, in addition to the abnormal activity of the RF during swing phase, altered toe-off knee flexion velocity, altered hip flexion velocity and decreased internal hip flexion moment in swing may reduce the peak knee flexion in swing.[224] This was further confirmed by Goldberg et al.,[27] where kinematic conditions at toe-off and internal joint moments in early swing phase of subjects with SKG and spastic RF muscle (CP) were compared with normal subjects. Goldberg et al.[27] suggested that the reduced peak knee flexion observed in SKG may be caused by low knee flexion velocity at toe-off observed in subjects with CP.[27, 225] When simulated models were applied, increased peak knee flexion in swing was achieved by increasing knee flexion velocity at toe-off to the average normal values. However, the timing of the peak knee flexion remained delayed. Additionally, this study found that most subjects exhibited normal or below normal internal knee extension moment in swing, thus contradicting the proposed link commonly made between altered RF muscle function, altered swing phase internal knee extension moments and reduced peak knee flexion. Another observation reported by this study was that most of the limbs did not exhibit reduced internal hip flexion moment, contradicting the findings of the study by Piazza and Delp.[224] Two further studies confirmed the importance of investigating the RF muscle activity during preswing as it may influence knee flexion velocity at toe-off, which in turn reduces peak knee flexion.[30, 226]

A further study used a forward simulation model of normal gait and found that abnormal activity of the RF muscle during double support has the potential to decrease knee flexion velocity at toe-off, which causes reduction of peak knee flexion in swing.[29] This was taken forward and tested amongst subjects with CP in a study comparing internal hip and knee joint moments during double support and swing phase and knee flexion velocity during toe-off before and after RF transfer surgery.[28] This study confirmed that none of the subjects showed reduction in internal hip flexion moment or excessive internal knee extension moment during the swing phase. Subjects with SKG tended to walk with reduced knee flexion velocity at toe-off and high internal knee extension moment in the double support phase of gait.[28] This indicates that investigating the internal moments during double support is important as these moments may affect the peak knee flexion by reducing the toe-off knee flexion velocity. However, vague and insufficient information about related physical assessment (muscle spasticity and joint contractures) was provided in both studies by Goldberg et al.,[27, 28] the study by Reinbolt et al.[30] and the study by Knuppe et al.[226]

Currently, the Duncan-Ely test is used to measure the length and the spasticity of the RF.[31, 32] With the subject prone and the hip in anatomical neutral position, the knee is slowly flexed to test the length of the RF (R2) and rapidly flexed to test the spasticity of the RF (R1). The test is positive if the ipsilateral hip rises and is negative if it does not rise. If the test is positive, the angle at which the ipsilateral hip raise occurs is measured. The Duncan-Ely test has been frequently referenced in several orthopaedic physical assessment books and manuals and used as a research tool to test RF flexibility.[31, 227-231] This suggests the Duncan-Ely's test face validity. Additionally, this

test is found to be a helpful predictor of outcomes for RF transfer surgery. Kay et al.[232] investigated the prognostic significance of the Duncan-Ely test for distal RF transfer surgery amongst subjects with CP. This study established that the subjects who exhibited a positive Duncan-Ely test preoperatively seemed to achieve better results following the surgery in comparison to the subjects who had negative test results preoperatively.[232] A further study investigated the sensitivity (the ability to detect RF dysfunction resulting in SKG), specificity (the ability to detect the absence of RF dysfunction resulting in SKG) and the predictive value of the Duncan-Ely test (R1) amongst subjects with CP.[32] The positive predictive value indicates that the subjects who test positively would have SKG with spastic RF. In contrast, the negative predictive value indicates that subjects who test negatively would not exhibit SKG with spastic RF. This study concluded that the sensitivity, specificity and negative predictive values were found to be low. The specificity was affected by the high number of subjects who had a negative Duncan-Ely test during the examination but still exhibited some features of SKG, including decreased knee ROM, abnormal RF EMG in swing and delayed timing of peak knee flexion in swing. The author justified this by the fact that passive examination may fail to provide information about dynamic activity and that RF spasticity is not the only cause of the presence of SKG. The test proved to have a good positive predictive value, indicating that a positive test value predicts the presence of SKG with spastic RF. However, it would have been useful to know the catch angles of the RF to evaluate if there was a threshold value for the catch, i.e. if the catch angle crosses the threshold value, no or minimal features of RF dysfunction will be presented.[32] Another study examined the sensitivity, specificity and intertester reliability of the Duncan-Ely test amongst subjects with CP.[233] In this

study, the knee joint was moved using three different velocities (slow, gravity and fast) according to the Tardieu scale.[51, 234] The ICC values were found to be good (ICC=0.62-0.80), with the highest ICC values obtained using the fast velocity test (ICC=0.80). The highest sensitivity and specificity values were reported with the fast velocity test and were found to be higher than the values reported by Marks et al.[32] Additionally, this study found that 63% of subjects during the gravity velocity test and 66.7% of subjects during the fast velocity test exhibited decreased peak knee flexion in swing when the knee angle measured using the Duncan-Ely test was less than 78.3º during the gravity velocity test and less than 65º during the fast velocity test.[233] In contrast, Peeler and Anderson[235] found lower intertester reliability (ICC=0.66) for the Duncan-Ely test amongst healthy subjects. The intratester reliability ranged from weak to excellent reliability (0.50-0.83), with the mean having good reliability (0.69).[235]

It can be observed from the literature that there is inconsistent evidence about the activation of the RF during gait and the effect of RF dysfunction on gait. Additionally, none of the studies examining the effect of RF dysfunction and the sensitivity and specificity of the Duncan-Ely test considered the position of the hip joint while the knee joint motion was measured during the gait. As mentioned above, the RF is a bi-articular muscle which crosses both hip and knee joints. It will be useful to observe if the knee joint flexion is achieved at the expense of the hip joint, i.e. if the knee joint 'borrowing' the RF length causes the hip joint to flex further.

It is noted in the literature that the Duncan-Ely test is frequently used as an assessment tool for measuring RF spasticity and length. However, vague information is found in

the literature about the test method and the interpretation of the results obtained. The test method is ambiguous in terms of the knee flexion velocity, and the results are based on the examiner's judgment, making this test subjective. Studies provided valuable awareness of RF dysfunction contribution to SKG, but the significance of the catch angle and muscle length is not well defined, i.e. how does a catch angle/length measured at 45º affect the gait? What is the difference between a catch angle/length measured at 40º or 70º on the walking pattern? What, if any, is the threshold value for the catch angle/length that, if exceeded, has no effect on the gait?

In a clinical environment, 3D gait analysis is not always available; hence, it will be useful if the results of the physical examinations are found to link directly to the deviations found in the gait. Based on the literature and theoretical basis, dynamic or true shortness of the RF can affect the hip and/or the knee joint at maximum hip extension in stance, in early swing[27], at peak knee flexion in swing[26, 223] and at peak hip flexion in swing. Additionally, true and dynamic shortness of the RF can influence the timing of peak knee flexion in swing.[26, 223]

This study was carried out to investigate gaps found in the literature about the understanding of results obtained using the Duncan-Ely test. The developed research questions are as follows:

1. What is the relationship between the catch angle/length of the RF measured using the Duncan-Ely test and the hip and/or the knee joint:

a. at maximum hip extension in stance (MHE)?

b. in early swing?

c. at peak knee flexion in swing (PKF)?

d. at peak hip flexion in swing (PHF)?

2. Is a dominance relationship between the catch angle/length of the RF found with the hip or the knee joint?

3. What is the relationship between the catch angle/length of the RF measured using the Duncan-Ely test and the timing of peak knee flexion in swing (TPKF)?

4. Is the effect of dynamic shortness of the RF on gait different from the effect of true shortness of the RF, and is a dominant relationship found with the hip or the knee joint?

3.2 Methods

3.2.1 Participants

This retrospective study received appropriate ethical approval from the West of Scotland Research Ethics Service, Clinical Research & Development Office of Greater Glasgow & Clyde Health Board and Caldicott Guardian. All the data were collected from the Neurobiomechanics Department, West of Scotland Mobility and Rehabilitation Centre, Queen Elizabeth University Hospital, Glasgow, UK.

The inclusion criteria for this study required that each subject:

- was diagnosed with diplegic CP
- was aged≥7
- did not suffer from any conditions resulting in lower limb sensory deficit
- was ambulatory (with/without walking aids)
- underwent the required physical assessment (Duncan-Ely test, hip and knee ROM)
- had a positive Duncan-Ely test (dynamic and/or true shortness)

- underwent lower limb kinematic and kinetic gait data collection

Participants were excluded if they had fixed static lower limb flexion contracture deformity mono-articular muscles or if any of the physical assessment information was not reported. Participants from age 7 were included because it is accepted that an adult gait pattern is reached by that age.[194]

Twenty participants were identified from a pre-existing patient database in the Neurobiomechanics Department, West of Scotland Mobility and Rehabilitation Centre, Queen Elizabeth University Hospital, Glasgow. A total of 37 limbs were included (3 limbs were excluded due to the presence of fixed flexion contracture and/or insufficient physical assessment information provided). Involved participants were aged between 7 and 19 years (11 years±3.4 years). All participants were ambulatory (17 independent, 3 with assistive devices). Physical examination and gait analysis (3D kinematic analysis) were carried out for each subject as part of the routine clinical procedure by the same group of examiners. Vicon Polygon software and Nexus software were used to process and analyse the data collected using the Vicon motion analysis system. A plug-in gait model was applied, and all the subjects were asked to walk barefoot at their own chosen speed. All the physical assessments were carried out by two physiotherapists specialised in paediatric with neurological conditions. The UG was used to measure the catch angle/length of the RF using the Duncan-Ely test. Control gait data were collected from the same department using the same methods from 10 healthy subjects who were aged between 5 to 8 years (6 years±0.82 years) (Table 3-1).

In normal gait, the maximum knee flexion angle is approximately 60º; hence, a catch or shortness greater than or equal to 60º should have minimal or no effect on gait.[65] Limbs were divided into two main groups per the catch angle/length: Group A: limbs with catch angle/length<60° (n=19) and Group B: limbs with catch angle/length≥60° (n=18). To study Question 4, all limbs were divided based on the presence of the dynamic or true shortening: Group C: limbs with dynamic shortness (n=20) and Group D: limbs with true shortness (n=17). Within Group C, limbs were further divided: Group C1: limbs with catch angle<60º (n=13) and Group C2: limbs with catch angle≥60º (n=7). Within Group D, limbs were further divided: Group D1: limbs with length<60º (n=6) and Group D2: limbs with length≥60º (n=11).

3.2.2 Study design

The investigated gait variables were measured from the average values of the 3D gait kinematics data collected by the department and were identified as follows:

- Knee at MHE (KMHE): the measured position of the knee at MHE
- MHE: the maximum extension/minimum flexion of the hip in stance
- RKFES: the range of knee flexion measured from toe-off to PKF
- RHFES: the range of hip flexion measured from toe-off to PKF
- PKF: the peak/maximum knee flexion observed in swing
- Hip at PKF (HPKF): the measured position of the hip at the point of PKF
- Knee at PHF (KPHF): the measured position of the knee at the point of PHF
- PHF: the peak/maximum hip flexion observed in swing
- TPKF: the timing when PKF occurs, calculated as a percentage of the entire gait

3.2.3 Statistical analysis

Linear regression models were employed to predict the relationship between the catch angle/length found during the Duncan-Ely test and the hip or the knee joint at the selected points/periods of gait. For a power of 80% at the 5% level of significance, an R^2 of 39% could be detected from a sample of 20 limbs in any sub-sample when 2 input variables were used.[236] All calculations were performed using *SPSS*™ software. The right limb was given the value 1, and the left limb was given the value 0. Additionally, the interaction between the side of the limb and the input variables was calculated. To establish a dominance effect of the RF, catch angle/length on the hip and the knee joints at the chosen points/periods of gait, several steps were followed. Initially, the knee was set as the output which included KMHE, RKFES, PKF and KPHF. The following inputs were tested: the limb (left or right), catch angle/length, hip position and interaction of the limb with catch angle/length and hip. Then, the hip was set as the output including MHE, RHFES, HPKF and PHF. The tested inputs included the limb (left or right), catch angle/length, knee position and interaction of the limb with catch angle/length and knee. For TPKF, the limb, catch angle/length and interaction between the limb and catch angle/length were inserted. Following that, if the limb and the interaction of the limb with the other inputs were not significant, these inputs were excluded, and the regression was re-run again. Then, the hip or the knee position (input) was excluded if found insignificant, and the regression was re-run again until the final model was achieved (Appendix A). Only models where the catch angle/length was found to be a significant input were taken further and analysed. To determine the dominance effect of the RF catch angle/length on hip or knee joint, R^2 values and p-values of the linear regression models for the hip and knee joint were

compared at each point/period of gait. The dominance relationship was identified based on the highest R^2 value and significance of p-values of the regression and inputs. For linear regression with two inputs, the value of adjusted R^2 were considered while, for linear regression with one input the value of R^2 was taken. This method of regression is formally known as 'backward selection' with all variables being entered and then removed if non-significant.[237, 238]

3.3 Results

3.3.1. Sample size

The sample size recruited for this study was small. This is because this study is a retrospective study and the inclusion criteria for this study is very specific. Given the sample size in each sub-sample, all the results below may be under Powered and interpreted with caution. Never the less, these results may infer models which may be tested in future amongst larger samples.

3.3.2 Groups A & B (dynamic & true shortness)

3.3.2.1 Maximum hip extension in stance

Group A (catch angle/length<60°)

A dominance relationship was found to be between the catch angle/length and the knee joint (Tables 3-2 & 3-3). The limb and MHE were excluded as they were found to be non-significant inputs. A simple linear regression model of the KMHE with only catch angle/length as an input was found weak (R^2=24.5%) but significant (p=0.031) (Table 3-4). Average KMHE was found to increase by 0.411° with each degree increase in the catch angle/length. All limbs exhibited a significant difference in MHE from

normal (Table 3-1). Hip extension was only seen in five limbs with four limbs amongst them exhibiting increased KMHE. The remaining limbs did not achieve hip extension during gait (Table 3-5).

Group B (catch angle/length≥60°)

The limb in both models for the MHE and KMHE was found not significant. A simple linear regression model for the MHE and KME with only catch angle/length as an input was found not significant and weak (Tables 3-2 & 3-3). Eight limbs achieved hip extension during gait (Table 3-5). Amongst these eight limbs, five had increased KMHE. From the remaining 10 limbs, which did not achieve hip extension, eight showed increased KMHE, while decreased KMHE was observed in two limbs (Table 3-5).

3.3.2.2 Early swing

Group A (catch angle/length<60°)

A dominance relationship was found between the catch angle/length and hip joint (Tables 3-2 & 3-3). Catch angle/length was found to be a significant input in the model of RHFES, while it was not a significant input in the model of RKFES. In the model of RHFES, the limb was excluded because it was found to be a non-significant input while, RKFES was found to be a significant input. The overall regression for this model was found to be significant ($p<0.001$) and good ($R^2=79.9\%$). Average RKFES was found to increase by 0.338º with each degree increase in the catch angle/length if RKFES was kept constant (Table 3-4). All limbs showed a significant difference in RKFES in comparison to normal (Table 3-1). Reduction in RKFES accompanied with decrease in RHFES was observed in thirteen limbs (Table 3-6). Three limbs had

decreased RKFES with increased RHFES. The remaining three limbs showed increased RKFES and RHFES.

Group B (catch angle/length≥60°)

No effect of the RF catch angle/length was found on the knee or the hip joint as the catch angle/length was found to be a non-significant input in both models (Tables 3-2 & 3-3). A non-significant difference was observed in RHFES between the included limbs and normal, while a significant difference in RKFES was found (Table 3-1). Eight limbs showed a reduction in RKFES and RHFES, while six limbs had increased RKFES and RHFES (Table 3-6). The remaining three limbs had increased RHFES accompanied with decreased RKFES and one limb showed normal RHFES.

3.3.2.3 Peak knee flexion in swing

Group A (catch angle/length<60°)

A dominance relationship of the catch angle/length of the RF was found with the knee joint at PKF (Tables 3-2 & 3-3). The limb and HPKF were found to be non-significant inputs; hence, they were excluded. The simple linear model for the PKF with only catch angle/length as an input was found to be weak (R^2=25%) but significant (p=0.029) (Table 3-4). Average PKF was found to increase by 0.640° with each degree increase in catch angle/length. Sixteen limbs exhibited reduction in PKF, while three limbs showed increase in PKF (Table 3-7). All limbs showed reduction in HPKF (Table 3-7). Additionally, PKF and HPKF in all limbs were found to be significantly different from normal (Table 3-1).

Group B (catch angle/length≥60°)

The catch angle/length of the RF was found to be a non-significant input in both models of PKF and HPKF. This may indicate that no effect of catch angle/length is seen on the knee or hip joint at PKF (Tables 3-2 & 3-3). All the limbs demonstrated a significant increase from normal in HPKF (Table 3-1). Ten limbs exhibited reduced PKF, while six limbs showed increased PKF (Table 3-7). The remaining two limbs showed normal PKF.

3.3.2.4 Peak hip flexion in swing

Group A (catch angle/length<60°)

At PHF, a dominance relationship of the catch angle/length was found with the knee joint. The limb and PHF were excluded as they were not significant (Tables 3-2 & 3-3). A simple linear regression model for the KPHF with only the catch angle/length as an input resulted to be weak (R^2=33.9%) and significant (p=0.009) (Tables 3-2 & 3-3). Average KPHF was found to increase by 0.739º with each degree increase in catch angle/length (Table 3-4). All limbs exhibited a significant difference in PHF and KPHF from normal (Table 3-1). Seventeen limbs had increased PHF (Table 3-8). Amongst these 17 limbs, 15 limbs showed increased KPHF as well. Two limbs had decreased PHF accompanied with increased KPHF.

Group B (catch angle/length≥60°)

As above, a dominance relationship of the catch angle/length was found with the knee joint. The limb and PHF were excluded because they were found not significant (Tables 3-2 & 3-3). The simple linear regression model for the KPHF with only the

catch angle/length as an input was found to be weak (R^2=32.8%) but significant (p=0.013) (Tables 3-4). Average KPHF was found to decrease by 0.414º with each degree increase in catch angle/length. All limbs exhibited a significant difference in PHF and KPHF from normal (Table 3-1). Increased PHF and KPHF were observed in fifteen limbs (Table 3-8). Two limbs had decreased PHF accompanied with increased KPHF. The reaming one limb showed increased PHF with decreased KPHF.

3.3.2.5 Timing of the peak knee flexion in swing

Group A (catch angle/length<60°)

The limb was found to be a non-significant input. The simple model for the TPKF with only catch angle/length as an input was found to be weak (R^2=8.5%) and not significant (p=0.227) (Table 3-14). This may be explained because TPKF remains approximately constant with any change in catch angle/length (Figure 3-1). A significant difference from normal in TPKF was found for all the limbs (Table 3-1). Sixteen limbs showed delay in the TPKF (Table 3-10), while one limb exhibited advanced TPKF. The remaining two limbs had normal TPKF.

Group B (catch angle/length≥60°)

The limb was found to be a non-significant input. The simple model for the TPKF with only catch angle/length as an input was found to be weak (R^2=18.8%) and not significant (p=0.072) (Table 3-9). As above, Figure 3-2 illustrates that no linear relationship exists between TPKF and catch angle/length, i.e. as the catch angle/length increases, TPKF remains approximately constant. A significant difference from

normal in TPKF was found for all the limbs (Table 3-1). Seventeen limbs showed delay in TPKF, while the remaining limb exhibited normal TPKF (Table 3-10).

3.3.2 Group C (dynamic shortness)

3.3.2.1 Maximum hip extension in stance

Group C1 (catch angle<60°)

No effect of RF dynamic shortening was observed on the hip or the knee joint at MHE. The limb was excluded from the models of KMHE and MHE as it was found not significant (Tables 3 & 4). The simple linear regression models for KMHE and MHE with only catch angle as an input were found weak and not significant (Tables 3-11 & 3-12). All limbs exhibited a significant difference in MHE from normal (Table 3-1). Hip extension was only seen in three limbs, and the remaining limbs did not achieve hip extension during gait (Table 3-13). Eight limbs had increased KMHE, while five limbs showed a reduction in KMHE.

Group C2 (catch angle≥60°)

As above, no relationship was found between the catch angle and the knee or the hip joint at MHE. This is because the catch angle was found to be a non-significant input in both models (Tables 3-11 & 3-12). Simple linear regression models of MHE and KMHE with only catch angle as an input were found not significant and weak. Three limbs achieved hip extension during gait (Table 3-13). Five limbs had increased KMHE.

3.3.2.2 Early swing

Group C1 (catch angle<60°)

A dominance relationship was found between the catch angle and the hip joint in early swing. The limb was excluded because it was found to be a non-significant input, while the catch angle and RKFES were found to be significant inputs in the model of RHFES (Tables 3-11 & 3-12). This regression model was found to be significant ($p<0.001$) and good (R^2=90.2%) (Table 3-14). Average RHFES was found to increase by 0.341º with each degree increase in catch angle if RKFES was held constant (Table 3-14). All limbs showed a significant difference in RKFES in comparison to normal (Table 3-1). Reduction in RKFES was observed in ten limbs (Table 3-15). Three limbs had increased RKFES. Increased RHFES was seen in four limbs. Additionally, eight limbs exhibited a reduction in RHFES, while one limb had normal RHFES (Table 3-15).

Group C2 (catch angle≥60°)

A dominance relationship of the catch angle in this group was found to be with the knee joint in early swing. The model of RKFES was significant ($p=0.004$) and good (R^2=90.3%) (Tables 3-11 & 3-12). Additionally, the limb was a non-significant input; hence, it was excluded. In contrast, the catch angle and RHFES were found to be significant inputs in the model of RKFES. Average RKFES was found to decrease by 0.34º with each degree increase in catch angle if RHFES was held constant (Table 3-14). However, a non-significant difference was observed in RKFES and RHFES between the included limbs and normal (Table 3-1). Four limbs showed a reduction in RKFES and RHFES, while three limbs had increased RKFES and RHFES (Table 3-15).

3.3.2.3 Peak knee flexion is swing

Group C1 (catch angle<60°)

The catch angle was found to be a significant input in models of PKF and HPKF. The limb and HPKF were found not significant; hence, both inputs were excluded in the model of PKF (Tables 3-11 & 3-12). Additionally, the limb and PKF were found to be non-significant inputs in the model of HPKF. Both simple linear regression models for PKF and HPK with only the catch angle as an input were found significant; however, the regression for PKF was higher. This may suggest that a dominance relationship of the catch angle is observed with the knee joint at PKF. The regression was found acceptable (R^2=40.0%), only explaining 40% of the variation in PKF. Average PKF increased by 0.84º for each degree increase in catch angle (Appendix A) (Table 3-14). All the limbs demonstrated a significant difference in PKF from normal (Table 3-1). Twelve limbs exhibited reduced PKF, while one limb showed increased PKF (Table 3-16). Looking at the hip joint, all the limbs demonstrated a significant increase in hip flexion at PKF in swing (Tables 3-1 & 3-16).

Group C2 (catch angle≥60°)

The catch angle was found to be a non-significant input in the models of PKF and HPKF (Tables 3-11 & 3-12). This indicates that it does not affect the hip or the knee joint at PKF. The simple regression models for PKF and HPKF with only the catch angle as an input were found to be weak and not significant. Three limbs showed a reduction in PKF, while two limbs exhibited an increase in PKF (Table 3-16). The remaining two limbs had normal PKF. However, all limbs showed a significant increase in HPKF (Table 3-1).

3.3.2.4 Peak hip flexion in swing

Group C1 (catch angle<60°)

The catch angle was found to be a significant input in models of PHF and KPHF, which may indicate that the catch angle influences the hip and the knee joints at PHF (Tables 3-11 & 3-12). The limb was excluded in both models as it was a non-significant input. Both regression models resulted to be significant; however, the regression for the KPHF was found higher (R^2=68.1%). This may suggest that a dominance effect of catch angle is found on the knee joint at PHF. The catch angle and KPHF were found to be significant inputs in this model. Average KPHF was found to increase by 1.237º with each degree increase in the catch angle if PHF was held constant (Appendix A) (Table 3-14). All limbs exhibited a significant difference in PHF and KPHF from normal (Table 3-1). Eleven limbs had increased PHF. Amongst these 11 limbs, 10 limbs showed increased KPHF as well (Table 3-17). Two limbs had decreased PHF accompanied with increased KPHF.

Group C2 (catch angle≥60°)

The catch angle was found to be a non-significant input in the models of PHF and KPHF (Tables 3-11 & 3-12). This indicates that the catch angle does not affect the hip or the knee joint. Simple linear regression models for PHF and KPHF with only catch angle as an input were found weak and not significant. As above, all limbs exhibited a significant difference in PHF and KPHF from normal (Table 3-1). Six limbs exhibited increased PHF accompanied with increased KPHF (Table 3-17). The remaining one limb showed a decrease in PHF and increase in KPHF.

3.3.2.5 Timing of peak knee flexion in swing

Group C1 (catch angle<60°)

The limb was found to be a non-significant input. The simple regression model for the TPKF with only the catch angle as an input was found to be weak (R^2=5.5%) and not significant (p=0.44) (Table 3-9). Figure 3-3 illustrates that TPKF remains approximately constant with any change in catch angle. This indicates that no linear relationship exists between TPKF and catch angle. A significant difference from normal in the TPKF was found for all the limbs (Table 3-1). Ten limbs showed delay in the TPKF, while one limb exhibited early PKF in swing (Table 3-18). The remaining two limbs had normal TPKF.

Group C2 (catch angle≥60°)

The limb was excluded because it was found not significant. The simple regression model for the TPKF with only the catch angle as an input was found to be weak (R^2 =1.1%) and not significant (p=0.820) (Table 3-9). As seen in Figure 3-4, which illustrates the relationship between TPKF and catch angle, TPKF remains approximately constant with any change in catch angle. As above, this indicates that no linear relationship exists between TPKF and catch angle. All limbs displayed a significant delay in the TPKF (Tables 3-1 & 3-18).

3.3.3 Group D (true shortness)

3.3.3.1 Maximum hip extension in stance

Group D1 (length<60°)

The length was found to be a non-significant input in the models of MHE and KMHE. This suggests that no relationship was observed between the length and the knee or hip joint at MHE (Tables 3-19 & 3-20). Simple linear regression models for MHE and KMHE with only length as an input were found weak and not significant. All limbs exhibited a significant difference in MHE and KMHE from normal (Table 3-1). Hip extension was only seen in two limbs, and the remaining limbs did not achieve hip extension during gait (Table 3-21). Two limbs showed a reduction in KMHE, while four limbs exhibited increased KMHE.

Group D2 (length≥60°)

As above, the length was found to be a non-significant input in both models (Tables 3-19 & 3-20). Simple linear regression models of MHE and KMHE with only length as an input were found to be not significant and weak. All limbs exhibited a significant difference in MHE and KMHE from normal (Table 3-1). Five limbs achieved hip extension during gait and eight limbs had increased KMHE (Table 3-21).

3.3.3.2 Early swing

Group D1 (length<60°)

The limb and RHFES were excluded in the model of RKFES and the limb and RKFES were excluded in the model of the RHFES as those inputs were found not significant (Tables 3-19 & 3-20). Simple linear regression models for RHFES and RKFES with

only length as an input were found to be weak and not significant. This implies that no effect of length was found on the knee and the hip joint in early swing. All limbs showed a significant difference in RKFES in comparison to normal (Table 3-1). Reduction in RKFES was observed in all limbs (Table 3-22). Increased RHFES was seen in four limbs. The remaining two limbs showed normal and increased RHFES, respectively.

Group D2 (length≥60°)

The overall regression for models of RKFES and RHFES was found to be good and significant (Tables 3-19 & 3-20). However, the length was found to be a non-significant input in these models. Additionally, the limb was excluded because it was found to be a non-significant input (Tables 3-19 & 3-20). As above, this may show that there is no relationship between the length and the knee or the hip joint in early swing. Reduction in RKFES was observed in eight limbs, while three limbs had increased RKFES (Table 3-22). Increased RHFES was seen in six limbs. Four limbs exhibited a reduction in RHFES, and one limb showed normal RHFES (Table 3-22).

3.3.3.3 Peak knee flexion is swing

Group D1 (length<60°)

The limb was found to be not significant; hence, it was excluded from both models. Length and HPKF were found to be significant inputs in the regression model for PKF (Tables 3-19 & 3-20). Additionally, length and PKF were found to be significant inputs in the model of HPKF. Both regressions were significant; however, the regression for the model of HPKF was found higher (R^2=91.1%) (Tables 3-19 & 3-20). This suggest

that a dominance relationship is seen between the length and the hip joint at PKF. For each degree increase in length, average HPKF decreases by 0.999º when PKF is held constant (Table 3-23). The difference in PKF in the limbs from normal was found to be not significant (Table 3-1). All the limbs demonstrated a significant increase from normal in hip flexion at PKF in swing (Table 3-1). Four limbs exhibited reduced PKF, while two limbs showed increased PKF (Table 3-24).

Group D2 (length≥60°)

The limb and length were found to be non-significant inputs in the models for PKF and HPKF (Tables 3-19 & 3-20). This may suggest that no effect of the length is seen on the hip or the knee joint at PKF. The regression model for PKF was found to be weak (R^2=33.8%) and not significant (p=0.79). Although the model for HPKF was found to be weak (R^2=47.3%) but significant (p=0.031), as stated above, the length was found to be a non-significant input. Seven limbs exhibited reduction in PKF, while four limbs showed increase in PKF (Table 3-24). All limbs showed a significant difference in HPKF from normal (Table 3-1).

3.3.3.4 Peak hip flexion in swing

Group D1 (length<60°)

The limb was excluded from both models as it was not significant (Tables 3-19 & 3-20). The length was found to be a non-significant input in both models for the PHF and KPHF, which may show no influence of the length on the hip or knee joint at PHF. Both simple linear regression models for PHF and KPHF with only length as an input were weak and not significant. All limbs exhibited a significant difference in PHF and

KPHF from normal (Table 3-1). Five limbs showed increased PHF and KPHF (Table 3-25). The remaining one limb exhibited increased PHF with decreased KPHF.

Group D2 (length≥60°)

As above, the limb and the length were found to be non-significant inputs. This may indicate that no relationship exists between the length and the hip or the knee joint at PHF (Tables 3-19 & 3-20). The simple linear regression models for PHF and KPHF with only length as an input were found not significant and weak. However, all limbs exhibited a significant difference in PHF and KPHF from normal (Table 3-1). Nine limbs showed increased PHF and KPHF (Table 3-25). A further limb exhibited decreased PHF with increased KPHF, while the remaining one limb showed the opposite.

3.3.3.5 Timing of peak knee flexion

Group D1 (length<60°)

The limb was found to be a non-significant input. The simple linear regression model for the TPKF with only length as an input was found to be weak (R^2=36.3%) and not significant (p=0.206) (Table 3-9). A non-significant difference from normal in the TPKF was found for all the limbs (Table 3-1). All the limbs showed delay in the TPKF in swing (Table 3-26). As the length increased, the TPKF remained approximately constant. This may explain the inability to detect the linear relationship (Figure 3-5).

Group D2 (length≥60°)

As above, the limb was excluded because it was found to be a non-significant input.

The simple linear regression model for the TPKF with only length as an input was found to be weak (R^2=21.6%) and not significant (p=0.149) (Table 3-9). Ten limbs displayed a delay in the TPKF (Tables 3-1 & 3-26). As seen in Figure 3-6, which illustrates the relationship between TPKF and length, TPKF remains approximately constant with any change in length.

3.4 Discussion

This study was carried out to examine the relationship between catch angle/length measured using the Duncan-Ely test and the hip or the knee joint at MHE, in early swing, at PKF and at PHF. Additionally, this study investigated if a dominance relationship was found between the catch angle/length and the hip or knee joint at the selected gait points/periods. Overall, the regression models for the KMHE, PKF, KPHF and RHFES in Group A (catch angle/length<60°) and KPHF in Group B (catch angle/length≥60°) were found to be significant (Tables 3-2 & 3-3). This may suggest that dominance relationship of the catch angle/length was found with the knee joint at MHE, PKF and PHF in Group A (catch angle/length<60°) and at PHF in Group B (catch angle/length≥60°). In contrast, a dominance relationship was found between the catch angle/length and the hip joint in early swing in Group A (catch angle/length<60°) (Tables 3-2 & 3-3).

Further analysis was performed to distinguish the effect of dynamic shortness from true shortness on the hip or the knee joint at the selected gait points/periods. Additionally, this further analysis examined if a dominance relationship with the knee or the hip joint changes in the presence of dynamic or true shortness. In Group C1

(catch angle<60°), a dominance relationship of the catch angle remained with the hip joint in early swing and with the knee joint at PKF and PHF. However, the resulting regression values for the RHFES, PKF and KPHF were found higher in Group C1 (catch angle<60°) when compared with Group A (catch angle/length<60°) (Tables 3-11 & 3-12). In contrast, in Group D1 (length<60°) a dominance relationship was found with the hip joint at PKF. The overall regression for the HPKF was higher in Group D1 (length<60°) in comparison to the regression model of the PKF in Groups A (catch angle/length<60°) and C1 (catch angle<60°) (Tables 3-19 & 3-20). In Group C2 (catch angle≥60°), a dominance relationship was found with the knee joint in early swing, with the regression value being good and significant (Tables 3-11 & 3-12). In Group D2 (length≥60°), all the regression models were found to be weak and not significant (Tables 3-19 & 3-20). Additionally, the regression models for TPKF in all groups were found to be weak and not significant (Table 3-9).

3.4.1 Maximum hip extension in stance

3.4.1.1 Groups A & B (dynamic & true shortness)

A dominance relationship of the catch angle/length at MHE was observed with the knee joint in Group A (catch angle/length<60°), while no relationship between the catch angle/length and hip or knee joint at MHE was found in Group B (catch angle/length≥60°) (Tables 3-2 & 3-3). In Group A (catch angle/length<60°), the simple linear regression with only the catch angle/length as an input was found to be weak but significant (Table 3-2). A significant and weak R^2 value suggests that the data can have a significant trend even with high variability. Additionally, it shows that the catch angle/length (prediction/input) still provides information about KMHE

(response/output). However, a model with a weak R^2 value has less precise predictions and a wider prediction interval (the range likely to contain the output/response value of a single new observation). On application of the output equation, a catch angle/length≥38° will result in normal KPHF, which increases as the catch angle/length increases (Table 3-4). However, this result should be applied with caution as the overall regression is weak. When looking at the data, only thirteen limbs achieved hip extension during gait (Table 3-5). Three limbs achieved hip extension with reduced knee flexion. Considering the RF length, this would be expected; as the hip extension increases, the knee flexion will reduce.

3.4.1.2 Group C (dynamic shortness)

In Groups C1 (catch angle<60°) and C2 (catch angle≥60°), the simple linear regression models for MHE and KMHE with only catch angle as an input were weak and not significant. This suggested that no linear relationship was found between the catch angle and the knee or the hip joint at MHE (Tables 3-11 & 3-12). A non-significant input (catch angle) indicates that the changes in the predictor/input are not associated with the changes in the response/output (MHE & KMHE). When looking at the data, only six limbs achieved hip extension during gait (Table 3-13). Two limbs with catch angles of 45º and 60º achieved hip extension with reduced knee flexion. As above, considering the RF length, this would be expected. However, an interesting observation was found in four limbs which exhibited hip extensions of 9º, 7º, 7º and 17º but had increased knee flexion above normal value of 19º, 28º, 29º and 27º, respectively (Table 3-13).

3.4.1.3 Group D (true shortness)

No linear relationship was found between RF length and the hip or knee joint at MHE in both groups. This is because the simple regression models for MHE and KMHE with only length as an input were found to be weak and not significant (Tables 3-19 & 3-20). As stated above, a non-significant input (length) indicates that the changes in the predictor/input are not associated with the changes in the response/output (MHE & KMHE). Two limbs in Group D1 (length<60°) and 5 limbs in Group D2 (length≥60°) achieved hip extension during gait (Table 3-21). From these 7 limbs, 1 limb achieved hip extension with reduced knee flexion, while another limb had normal KMHE. Of the remaining 5 limbs, 4 limbs had below normal MHE and above normal KMHE. One limb exhibited above normal MHE and increased KMHE (Table 3-21). When three limbs with lengths of 70º are compared, one limb achieved hip extension (above normal), while the remaining two limbs did not achieve hip extension. The limb with the hip extension showed the least knee flexion which could be justified due to the RF length. However, the reduction in the knee flexion (5° below normal) was not very large compared to the increase in MHE (12° above normal) (Table 3-21).

Overall, a significant relationship was found between the catch angle/length and the knee joint at MHE in Group A (catch angle/length<60°). This shows that the catch angle/length<60° has a dominance relationship with the knee joint at MHE. However, the regression was found to be weak. It was not possible to detect whether the effect of dynamic and true shortness of the RF on the knee and hip joint at MHE was different. This may be explained by the low number of limbs in Groups C (dynamic shortness) and D (true shortness) and the variance presented. As discussed above,

variability is seen in limbs with equal catch angles/lengths. This variability may have affected the relationship, resulting in weak regression. Additionally, this is a complicated multiple regression model being used for estimation from a sample of heterogeneous limbs.

3.4.2 Early swing

3.4.2.1 Groups A & B (dynamic & true shortness)

The model of RHFES resulted in good and significant regression in Group A (catch angle/length<60°), which illustrates that the hip had a dominance relationship with the catch angle/length in Group A (catch angle/length<60°) in early swing (Tables 3-2 & 3-3). Additionally, the models for RHFES and RKFES in Group B (catch angle/length≥60°) were found to be good and significant. However, the catch angle/length was found to be a non-significant input in both models. A non-significant input indicates that changes in the catch angle/length are not associated with changes in RKFES or RHFES. This indicates that no effect of the catch angle/length is found on the hip or the knee joint in early swing in Group B (Tables 3-2 & 3-3). The catch angle/length was found to be directly related to RHFES in Group A (catch angle/length<60°) when RKFES was held constant (Table 3-4). On application of the equation, an approximate catch of 35-38º should result in normal RHFES (24°) when RKFES is normal (20º). Eleven limbs showed increased RKFES accompanied with increased RHFES above normal, which is an expected to accommodation for RF dynamic or true shortness (Table 3-6). In contrast, twenty-two limbs exhibited reduced RKFES and RHFES. Three limbs had a similar catch angle/length of 70º (Table 3-6). Of these, one limb had increased RKFES (35º) and RHFES (56º). The remaining two

limbs had decreased RKFES (16º and 8°). Amongst these two limbs, the first limb had normal RHFES (24º), while the second limb showed above normal RHFES (35°) (Table 3-6). It might have been expected that the limb with higher RKFES would have higher RHFES compared to the other limb, as increased hip flexion is required to achieve increased knee flexion, but the opposite was found (Table 3-6).

3.4.2.2 Group C (dynamic shortness)

The models for RKFES and RHFES resulted in good and significant regression in both groups (Tables 3-11 & 3-12). In early swing, the hip had a dominance relationship with the catch angle in Group C1 (catch angle<60°), while the knee showed a dominance relationship with the catch angle in Group C2 (catch angle≥60°) (Tables 3-11 & 3-12). This indicates the importance of considering both the hip and knee joint positions during the gait. The catch angle was found to be directly related to RHFES (output) in Group C1 (catch angle<60°), while it was found to be inversely related to RKFES (output) in Group C2 (catch angle≥60°) when the other input was held constant (Table 3-14). It should be noted that the catch angle is directly related to RHFES and inversely related to RKFES in both groups (Table 3-14). On application of the equation, an approximate catch of 25-28º should result in normal RHFES when RKFES is normal (20º) in Group C1 (catch angle<60°). In contrast, in Group C2 (catch angle≥60°) a catch angle of approximately 60-62º should result in normal RKFES when RHFES is normal (24º). Six limbs showed increased RKFES accompanied with increased RHFES above normal, which is expected to accommodate for RF dynamic shortness (Table 3-15). On the other hand, twelve limbs exhibited reduced RKFES and RHFES. Two limbs had a similar catch angle of 55º. One limb had increased RKFES

(30º) and RHFES (41º), while the second limb had decreased RKFES (15º) and normal RHFES (24º) (Table 3-15). This, again, can be explained by the fact that further hip flexion is required to allow further knee flexion. A possible explanation for the difference seen between these two limbs with similar catch angles is the difference found in the tone of the RF. The first limb exhibited increased tone (MAS=1) while no increase in muscle tone was found in the second limb (MAS=0).

3.4.2.3 Group D (true shortness)

The model for RKFES and RHFES resulted to be weak and not significant in Group D1 (length<60°). This may indicate that no relationship was found between the length and the hip or knee joint in early swing. In Group D2 (length≥60°), the overall regression for RHFES and RHFES models was good and significant (Tables 3-19 & 3-20). However, the length was found to be a non-significant input in both models. A non-significant input indicates that changes in length are not associated with changes in RKFES or RHFES. This indicates, as above, that no influence of the length is found on the hip or the knee joint in early swing. Five limbs showed increased RKFES accompanied with increased RHFES above normal, which is expected to account for RF shortness (Table 3-22). The remaining limbs exhibited reduced RKFES with reduced, increased or normal RHFES. Three limbs had a similar length of 90º and showed a decrease in RKFES (16°, 9° and 10°). Amongst these three limbs, one limb had decreased RHFES (16°), while the remaining two limbs had increased RKFES (26° and 29°) (Table 3-22).

A dominance relationship of the catch angle/length was found with the hip joint in early swing in Group A (catch angle/length<60°). In contrast, the catch angle/length was found to be a non-significant input in the models of RKFES and RHFES in Group B (catch angle/length≥60°). This may highlight that a catch angle/length≥60° does not affect early swing. The further analysis illustrated that dynamic shortness resulted in a relationship with the knee and hip joint in early swing, while true shortness showed no relationship. The major limiting factor when comparing between Groups C (dynamic shortness) and D (true shortness) is the difference in the number of the limbs included in each group. This may have affected the regression model and p-values. Group C1 (catch angle<60°) showed a dominance relationship of the catch angle with the hip joint, while Group D1 (length<60°) showed no relationship. In Group D2 (length≥60°), the length was found to be a non-significant input, demonstrating no relationship. In contrast, in Group C2 (catch angle≥60°) a dominance relationship was seen between the catch angle and knee joint. However, cautions must be applied when using these results specifically results of Groups D2 (length≥60°) and C2 (catch angle≥60°) due to the low number of limbs included. Additionally, the variance seen in RHFES and RKFES of the limbs with equal catch angles/lengths may explain the weak regression/no relationship found in some groups. Additionally, the heterogeneous nature of CP may have influenced the results as well.

3.4.3 Peak knee flexion in swing

3.4.3.1 Groups A & B (dynamic & true shortness)

A significant relationship was demonstrated between catch angle/length and PKF in Group A (catch angle/length<60°) (Tables 3-2 & 3-3). This indicates that a dominance

relationship is observed between the catch angle/length and the knee joint at PKF. However, the simple linear regression for PKF in Group A (catch angle/length<60°) was weak. As above, if the regression is found to be significant but weak it demonstrates that highly variable data can have a significant trend. Additionally, it shows that the catch angle/length (prediction/input) still provides information about PKF (response/output). It should be noted that a model with a low R^2 value has less precise predictions and a wider prediction interval (the range likely to contain the output/response value of a single new observation). Catch angle/length is directly proportional to PKF, i.e. as catch angle increases, PKF increases (Table 3-4). On application of the resulting equation, a catch angle≥55º should result in normal PKF. However, this result should be used with caution due to the weak R^2 value. In Group B (catch angle/length≥60°), no relationship between the catch angle/length and the hip or the knee joint was found at PKF (Tables 3-2 & 3-3). This may indicate that a catch angle/length≥60° has no effect/minimum effect on PKF or HPKF because the maximum PKF required during gait is approximately 60°. Twenty-six limbs exhibited reduced PKF. All the limbs in Group A (catch angle/length<60°) and sixteen limbs in Group B (catch angle/length≥60°) which exhibited normal, increased or decreased PKF had an increase in hip flexion measured at PKF (Table 3-7). This may indicate that further hip flexion is required to achieve further PKF, which may or may not reach normal levels. As illustrated in Table 3-7, for example, three limbs had a catch angle/length of 70º. Amongst these three limbs, one limb showed increased PKF (69°). The remaining limbs demonstrated decreased PKF of 55° and 34°, respectively (Table 3-7). The increase in HPKF was greater in the first limb (68°) followed by the second (62°), while the third limb showed normal HPKF. This is expected and can be

explained by fact that further hip flexion is required to achieve further knee flexion. Similar variability is also seen in the remaining limbs with equal catch angles/lengths. This variability may have affected the relationship, resulting in weak and/or not significant regression.

3.4.3.2 Group C (dynamic shortness)

A significant relationship was demonstrated between the catch angle and PKF and the catch angle and HPKF in Group C1 (catch angle<60°) (Tables 3-11 & 3-12). This indicates that both the hip and the knee joint exhibited a relationship with the catch angle at PKF. The R^2 value was higher for the model of PKF (Tables 3-11 & 3-12). However, it should be noted the overall regression was acceptable (R^2=40.0%), only explaining 40% of the variation in PKF. Catch angle is directly proportional to PKF, i.e. as the catch angle increases, PKF increases (Table 3-14). On application of the resulted equation, a catch angle≥55º should result in normal PKF. Although the overall regression was significant, the R^2 value was acceptable; hence, results should be used with care. In Group C2 (catch angle≥60°), no relationship between the catch angle and the hip or the knee joint at PKF was found (Tables 3-11 & 3-12). This may indicate that a catch angle≥60° has no effect/minimal effect on PKF or HPKF because the maximum PKF required during gait is approximately 60°. Overall, three limbs showed increased PKF (Table 3-16). Additionally, further 2 limbs had normal PKF, while the remaining limbs showed reduced PKF. All the limbs which exhibited normal, increased or decreased PKF had an increase in HPKF (Table 3-16). This may indicate that further hip flexion is required to achieve further PKF, which may or may not reach normal levels. As illustrated in Table 3-16, for example, two limbs had a catch angles

of 55º. When these two limbs were compared, the first limb showed increased PKF (76°) while the second limb demonstrated decreased PKF (53°). However, the increase in HPKF was greater in the second limb, which is opposite from what is expected. This might be explained due to the difference in the RF tone found in both limbs. The first limb had increased tone (MAS=1), while the second limb had no increase in tone (MAS=0). Variance was also observed in the remaining limbs with the same catch angles, which may have resulted in weak and/or non-significant regression.

3.4.3.3 Group D (true shortness)

In Group D1 (length<60°), the regression model for HPKF was found to be higher than the model of PKF (Tables 3-19 & 3-20). This may illustrate that a dominance relationship is found with the hip joint at PKF. The overall regression was good and significant. Additionally, the PKF was also found to be a significant input in the model of the HPKF, which may highlight the influence of the knee position on the hip position in the presence of RF shortness at PKF (Tables 3-19 & 3-20). The length was found to be inversely proportional to HPKF when PKF was held constant, i.e. as the length increased, HPKF decreased towards normal value (23°) (Table 3-23). On the application of the output equation, an approximately 20º increase in knee flexion is required to obtain normal HPKF (23º) when the length of the RF is 36º (Table 3-23). In Group D2 (length≥60°), no relationship between the length and PKF or HPKF was found as the length was found to be a non-significant input (Tables 3-19 & 3-20). Four limbs in Group D1 (length<60°) and 7 limbs in Group D2 (length≥60°) showed decreased PKF (Table 3-24). It should be noted that all the limbs in Group D1 (length<60°) and 9 limbs in Group D2 (length≥60°) had an increase in HPKF (Table

3-24). This may show that more hip flexion is mandatory to achieve further PKF to accommodate for RF shortness. Two limbs in Group D2 (length≥60°) showed a reduction of 26º and 12º of PKF with normal HPKF, which again relates to the RF length. As shown in Table 3-24, for example, three limbs had a length of 70º. When these three limbs were compared, one limb showed increased PKF, while the other two limbs demonstrated decreased PKF of 26º and 5º. The limb with the reduction of 26º in PKF showed normal HPKF, while the other two limbs exhibited increased HPKF (Table 3-24). This may be explained by the RF muscle length as further hip flexion is required to allow further knee flexion.

A dominance relationship of the catch angle/length was found with the knee joint at PKF in Group A (catch angle/length<60°) (Tables 3-2 & 3-3). However, the regression was weak but significant, which may suggest a significant trend in the data. In contrast, in Group B (catch angle/length≥60°) the catch angle/length was found to be a non-significant input in the models of PKF and HPKF (Tables 3-2 & 3-3). The further analysis demonstrated that the R^2 value increased to a higher level in Group C1 (catch angle<60°) when compared with Group A (catch angle/length<60°) (Table 3-11). However, a dominance relationship was still found to be between the catch angle and the knee joint at PKF (Tables 3-11 & 3-12). In contrast, in Group D1 (length<60°), a dominance relationship was found to be between the length and the hip at PKF (Tables 3-19 & 3-20). Additionally, the regression value was higher for the model of HPKF when compared with the regression model of the PKF in Groups A (catch angle/length<60°) and C1 (catch angle<60°). As above, the major limiting factors when comparing between Groups C (dynamic shortness) and D (true shortness) are the

difference in the number of limbs included in each group and the heterogeneous nature of CP. It should be noted that no relationship between the catch angle and/or length and hip joint or knee joint at PKF was found in Groups B (catch angle/length≥60°), C2 (catch angle≥60°) and D2 (length≥60°) (Tables 3-2 & 3-11 & 3-19). This may highlight that a catch angle and/or length≥60° does not affect or has a minimal effect on PKF.

3.4.4 Peak hip flexion in swing

3.4.4.1 Groups A & B (dynamic & true shortness)

A dominance relationship was found between the catch angle/length and the knee joint in both Groups. However, the overall regression was found to be weak but significant; hence, these results should be treated with caution (Tables 3-2 & 3-3). As explained above, a significant simple linear regression model indicates that the catch angle/length (predictor/input) provides information about KPHF (response/output) even if the values fall further from the estimated regression line. Then again, the precision of the prediction is low with a wide prediction interval. On the application of the output equation in Group A (catch angle/length<60°), a catch angle≥28° should result in normal KPHF (26°). The catch angle/length is directly related to KPHF, i.e. as the catch angle increases, KPHF increases (Table 3-4). In Group B (catch angle/length≥60°), the catch angle/length is inversely related to KPHF, i.e. as the catch/length increases, HPKF moves towards normal. On application of the equation a catch angle/length≥112° should result in normal KPHF (26°) (Table 3-4). Overall, thirty limbs showed increased PHF accompanied with increased KPHF (Table 3-8). This is expected as further knee flexion is required to achieve further hip flexion with

dynamic or true shortening. As seen in Table 3-8, for example, two limbs had a catch angle of 70º. When these two limbs were compared, both limbs showed an increase of PHF above normal (72° and 77°). However, the first limb had KPHF of 29° and the second 62°. The difference in KPHF is wider than the difference in PHF. The remaining limbs with equal catch angles/lengths illustrated similar variance. This variability may have affected the relationship, resulting in weak and/or not significant regression.

3.4.4.2 Group C (dynamic shortness)

The catch angle resulted to be a significant input in models of PHF and KPHF in Group C1 (catch angle<60°) (Tables 3-11 & 3-12). However, the overall regression was found to be higher for the model of KPHF. This indicates that a dominance relationship is seen between the catch angle and the knee joint at PHF. Additionally, PHF was found to be a significant input in this model (Tables 3-11 & 3-12). This may highlight the effect of the knee on the hip joint at PHF. On the application of the output equation, to achieve normal PHF (37°) with a catch angle of 28°, KPHF should increase by approximately 8° (Table 3-14). The catch angle is directly related to KPHF, i.e. as the catch angle increases, KPHF increases. In Group C2 (catch angle≥60°), no relationship was found between the catch angle and the hip or the knee joint at PHF (Tables 3-11 & 3-12). Overall, seventeen limbs showed increased PHF (Table 3-17). Additionally, further 2 had normal PHF. Amongst these 17 limbs, sixteen limbs also exhibited increased KPHF above normal, which is expected in the presence of RF dynamic shortening (Table 3-17). Variation is seen between limbs with the same catch angle, which may have affected the resulted linear regression. For example, when two limbs

with a catch angle of 48° were compared, one limb showed an increase of 23° of PHF above normal, while the other limb demonstrated an increase of 8° in PHF above normal. However, the increase in the knee joint flexion was equal in both limbs. This may be explained due to the difference found in the RF tone of both limbs, as the limb with the greater increase in PHF had no increase in tone. In contrast, the other limb showed increased tone (MAS=2). This indicates that the second limb is stiffer to move.

3.4.4.3 Group D (true shortness)

No relationship was found between the length and the hip or knee joint at PHF as the length was found to be a non-significant input in both models in both groups (Tables 3-19 & 3-20). Overall, fourteen limbs showed increased PHF and KPHF (Table 3-25). As above, this is expected, as further knee flexion is required to achieve further PHF. As seen in Table 3-25, two limbs had a length of 90º. When these two limbs were compared, one limb had PHF higher than the other limb by approximately 1°. However, the increase in KPHF in one limb was almost double the other limb. Similar variability is also seen in the remaining limbs with equal lengths, which may have influenced the resulted linear regression.

In Group A (catch angle/length<60°) and Group B (catch angle/length≥60°), a dominance relationship of the catch angle/length was found with the knee joint at PHF (Tables 3-2 & 3-3). However, the regression was weak but significant, which may suggest a significant trend in the data. The regression was higher in Group C1 (catch angle<60°) in comparison to Group A (catch angle/length<60°) and Group B (catch

angle/length≥60°) (Tables 3-11 & 3-12). No relationship was established between the length and the knee or the hip joint at PHF (Tables 3-19 & 3-20).

3.4.5 Timing of peak knee flexion in swing

3.4.5.1 Groups A & B (dynamic & true shortness)

Overall, no relationship was detected between the catch angle/length of the RF and TPKF (Table 3-9). A delay in TPKF was observed in 33 limbs. A further limb showed early TPKF, while the remaining 3 limbs exhibited normal TPKF (Table 3-10). TPKF ranged from 66 to 88% of gait of the included limbs. Additionally, the difference in TPKF in all limbs from normal was found to be significant (Table 3-1). This may indicate that although a linear relationship was not established, the effect of dynamic and true shortness of the RF can be observed on TPKF (Figures 3-1 & 3-2).

3.4.5.2 Group C (dynamic shortness)

The relationship between TPK and the catch angle in both groups was found to be weak and not significant (Tables 3-11 & 3-12). When looking at the data, 11 limbs in Group C1 (catch angle<60°) showed a delay in TPKF (Table 3-18). The remaining 2 limbs had normal and below normal TPKF. Two limbs with the same catch angle, 40º, showed variability in TPKF. The first limb had a 12% delay in TPKF while the second limb exhibited normal TPKF (Table 3-18). In Group C2 (catch angle≥60°), all limbs displayed a delay in TPKF. Additionally, in Group C2 (catch angle≥60°) it was noted that as the catch angle increases, TPKF stays around approximately the same values (0.8-0.88%) (Table 3-18). The inability to detect a relationship between the catch angle

and TPKF could be explained due to the difference in TPKF values between different catch angles being small and inconsistent (Figures 3-3 & 3-4).

3.4.5.3 Group D (true shortness)

As above, the relationship between TPKF and length in both groups was found to be weak and not significant (Tables 3-19 & 3-20) (Figures 3-5 & 3-6). All limbs in Group D1 (length<60°) and 10 limbs in Group D2 (length≥60°) showed a delay in TPKF (Table 3-26). The remaining one limb in Group D2 (length≥60°) had normal TPKF. Three limbs with the same length, 70º, showed variability in TPKF as follows: 0.78, 0.80 and 0.84 (Table 3-26).

It could be noted that no relationship is detected between the catch angle and/or length and TPKF in all groups. However, most limbs exhibited a delay in TPKF, and all limbs showed significant difference from normal (Table 3-1). This may suggest that although a linear relationship is not established, the effect of dynamic/true shortening of the RF is observed in TPKF. The difference in TPKF values between different catch angles/lengths was small and inconsistent, which may have influenced the ability to detect a linear relationship (Figures 3-1 to 3-6).

The current study showed that only thirteen limbs achieved hip extension during gait, raising the question regarding the hip joint position during the Duncan-Ely test. The position of the hip joint and knee joint during the currently used assessment procedure does not closely match the position of these joints during the period of gait when the RF affects gait. During the transition from the stance-to-swing, the hip moves from

approximately 10º extension to flexion as the knee flexes further. On a theoretical basis, a short RF (dynamic or true) should limit knee flexion if the hip is in extension. This raises the issue of whether the RF muscle length/spasticity should be measured while the hip is in extension and whether a catch measured in hip extension will have a different effect on gait when compared with a catch measured at hip neutral. This study illustrated that even in subjects where the hip does not achieve hip extension or hip neutral during the gait, a catch angle/length measured at hip neutral still affects the gait.

A study by Lee et al.[233] found that the sensitivity and specificity of the Duncan-Ely test measured at the slow and fast velocities to indicate the presence of abnormal PKF and TPKF is low and/or not significant amongst subjects with hemiplegic and diplegic CP. The current study found similar results for TPKF. However, a significant relationship between catch angle/length and the hip or knee joint at the PKF in Groups A (catch angle/length<60°), C1 (catch angle<60°) and D1 (length<60°) was observed in the current study. Another study by Marks et al.[32] reported low sensitivity and specificity of the Duncan-Ely test amongst subjects with hemiplegic, diplegic and quadriplegic CP in predicting delayed TPK. The results found by Marks et al.[32] agree with the findings of the current study.

In this study, thirty-seven limbs were involved. These limbs were divided into different groups; hence, the number of limbs in each group did not meet the required power calculations. Given that same variables have been used in different regression models and the small sample size in each sub-sample, caution is required in any interpretation

or generalisation of the results. Furthermore, the heterogenous nature of CP may have affected the results. Repeating this study with a larger sample size may lead to successful and meaningful models, and a relationship may be established. Additionally, this study only included subjects with diplegic CP. It would be useful to investigate the relationship amongst subjects with hemiplegic and quadriplegic CP.

This study investigated the relationship between the RF catch angle/length measured using the Duncan-Ely test with the hip or the knee joint at selected gait points/periods. The regression was significant/dominant for KME, PKF, KPHF and RHFES in Group A (catch angle/length<60°), while in Group B (catch angle/length≥60°) the model for KPHF was significant/dominant. In Group C1 (catch angle<60°), the regression models for RHFES, PKF and KPHF were significant/dominant. In contrast, the regression for RKFES was significant/dominant in Group C2 (catch angle≥60°). Meanwhile, in Group D1 (length<60°) the significant/dominant model was the model for HPKF, and all the regressions models were weak and not significant in Group D2 (length≥60°). Furthermore, the regression models for TPKF in all groups were weak and not significant. This study illustrates that a dominance relationship of the catch angle/length with the knee or the hip joint varies at different gait points/periods. This suggests that the RF can influence the knee or the hip joint. Additionally, this study shows that the effect of the dynamic shortness of the RF differs from true shortness of RF on the hip or the knee joint during gait.

It has been demonstrated in this chapter that a comprehensive assessment of bi-articular muscles enhances the understanding of the influence of the muscle

length/spasticity on the gait. Awareness of the presenting neurological and orthopaedic abnormalities will allow careful preparation of the AFO-FC prescription and any required adjunct therapy. Accurate measurement is essential. In the following chapter, the commonly used measurement tools for joint ROM measurements and bi-articular muscle length will be evaluated. The results of a literature review evaluating the reliability of different designs of goniometer will be presented. Furthermore, in this review different factors influencing the reliability of goniometers will be discussed. Following that, the feasibility of a 2D video analysis system (*PnO CMD*) for passive joint ROM measurement during physical assessment will be examined. The results of studies investigating the reliability and accuracy of the *PnO CMD* amongst healthy subjects will be demonstrated. Finally, the results of a reliability study of the *PnO CMD* amongst subjects with diplegic CP will be presented.

3.5 Figures

Figure 3-1: Timing of peak knee flexion in swing (TPKF) versus catch angle/length in Group A (catch angle/length<60°).

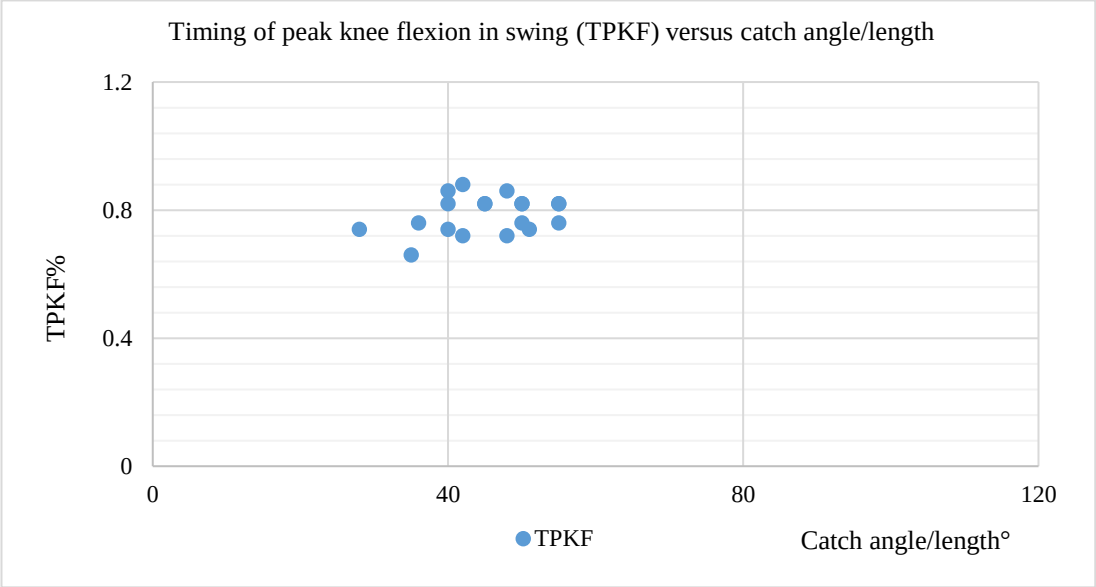


Figure 3-2: Timing of peak knee flexion in swing (TPKF) versus catch angle/length in Group B (catch angle/length≥60°).

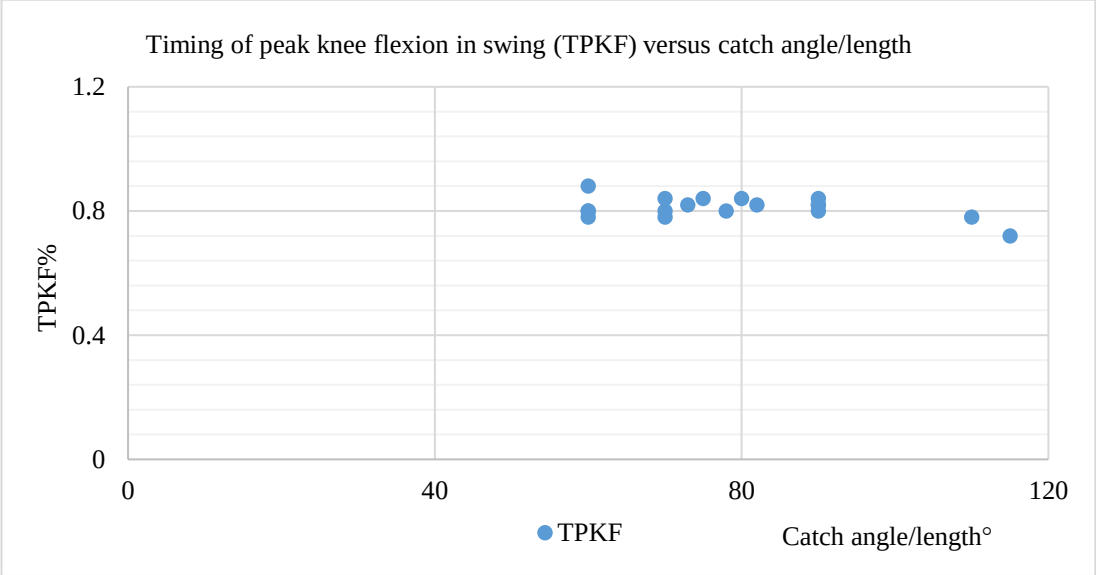


Figure 3-3: Timing of peak knee flexion in swing (TPKF) versus catch angle in Group C1 (catch angle<60°).

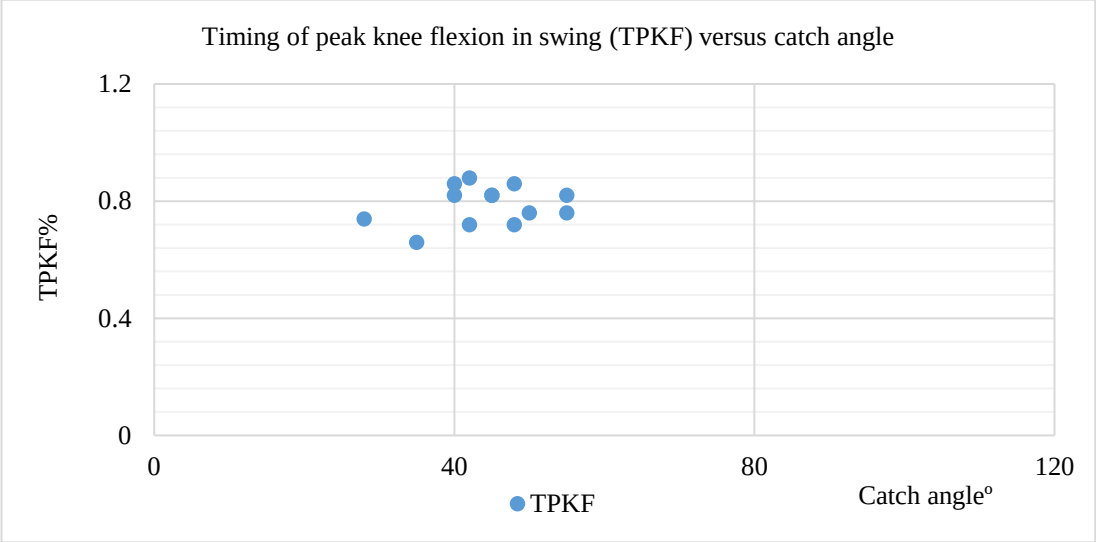


Figure 3-4: Timing of peak knee flexion in swing (TPKF) versus catch angle in Group C2 (catch angle≥60°).

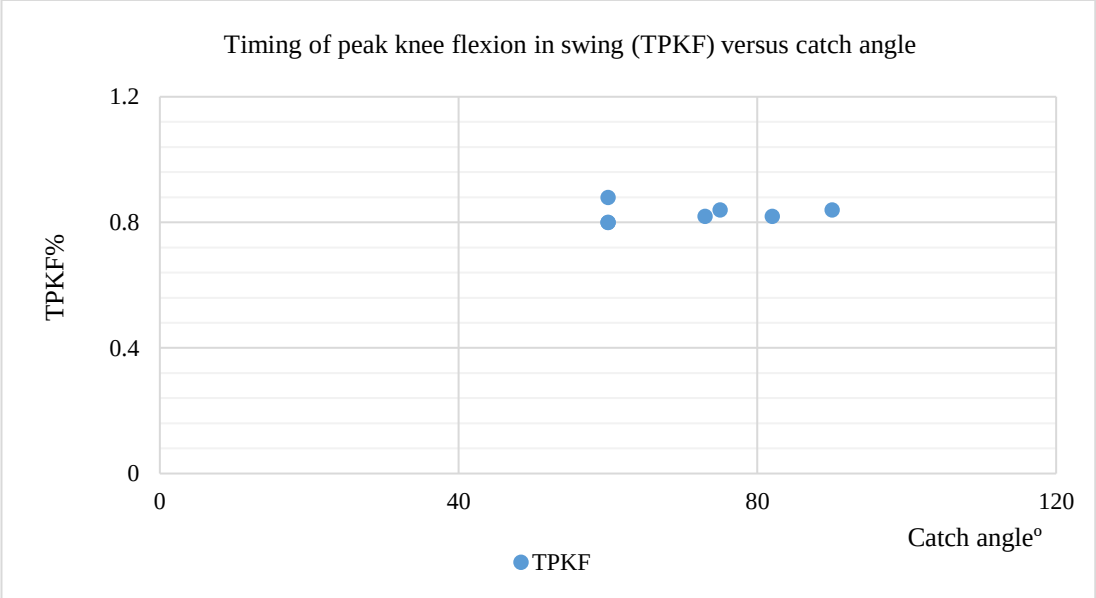


Figure 3-5: Timing of peak knee flexion in swing (TPKF) versus length in Group D1 (length<60°).

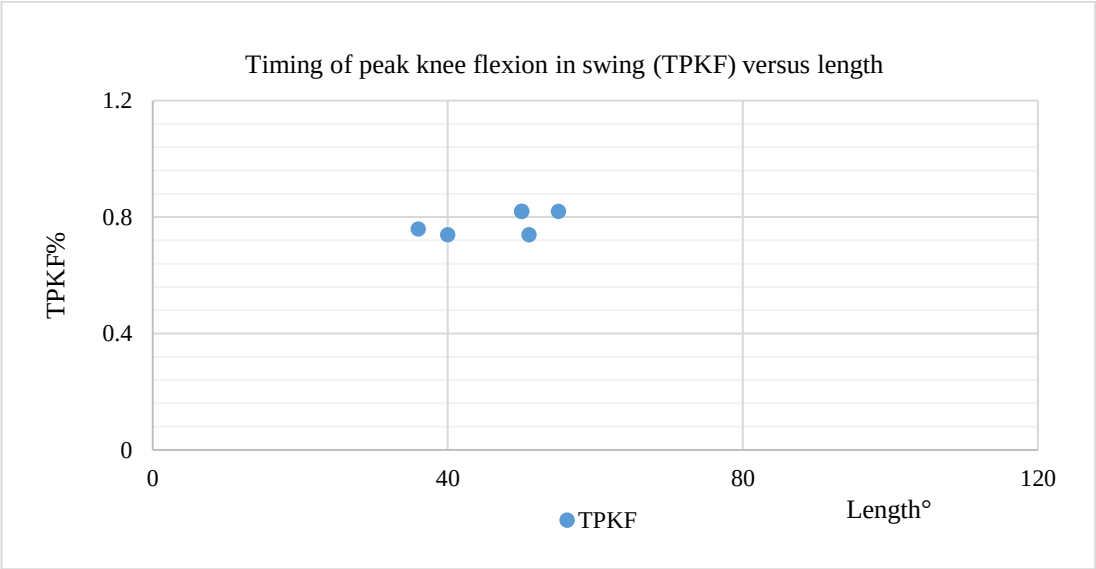
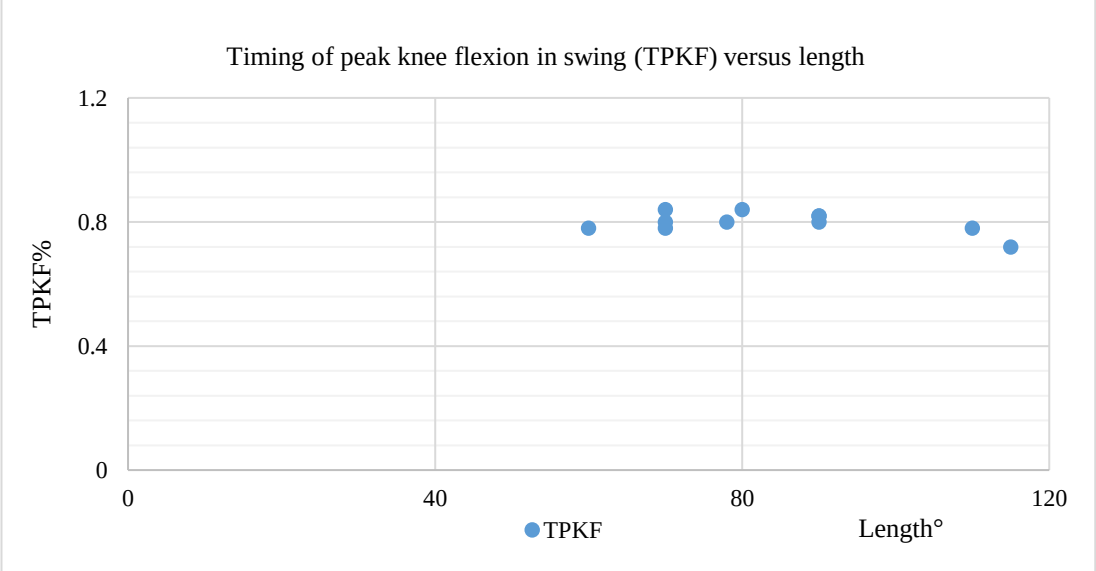


Figure 3-6: Timing of peak knee flexion in swing (TPKF) versus length in Group D2 (length≥60°).



3.6 Tables

Table 3-1: The mean and SD of gait variables measured in normal and in the groups.

Measures of gait	Normal Mean° (SD)	Groups A&B				Group C				Group D			
		Group A: catch angle/length<60		Group B: catch angle/length≥60		Group C1: catch angle<60°		Group C2: catch angle≥60°		Group D1: Length<60°		Group D2: Length≥60°	
		Mean° (SD)	P-value	Mean° (SD)	P-value	Mean° (SD)	P-value	Mean° (SD)	P-value	Mean° (SD)	P-value	Mean° (SD)	P-value
KMHE	14 (4.5)	17.1 (6.16)	0.02	25 (10.79)	<0.001	17 (7)	0.06	29 (13.4)	0.013	17 (4.5)	0.091	22 (8.4)	0.004
MHE	-8 (4.8)	6.35 (11.4)	<0.001	4.2 (12.1)	<0.001	5.6 (10.2)	<0.001	5.4 (13.6)	0.02	7.7 (14.6)	0.024	3.4 (11.7)	0.004
RKFES	20 (4)	14.11(9.2)	0.006	15.6 (9.4)	0.032	14 (10.8)	0.043	15.7 (10.6)	0.165	14 (4.8)	0.012	15.5 (9.1)	0.06
RHFES	24	21.84 (10.16)	0.183	25.3 (12.1)	0.325	21 (10.9)	0.2	22.5 (9.9)	0.358	23 (9.8)	0.463	27 (13.5)	0.22
PKF	60 (5.3)	53 (9.4)	0.002	56.5 (9.2)	0.063	50.5 (10)	0.002	57.8 (7.5)	0.24	59 (5.6)	0.272	56 (10.3)	0.108
HPKF	23 (5.5)	39.7 (8.2)	<0.001	37.7 (13.1)	<0.001	37 (6.4)	<0.001	36 (5.3)	<0.001	44.5 (10.3)	0.002	39 (16.5)	0.004
KPHF	26 (5.3)	38.76 (9.4)	<0.001	40.5 (11.7)	<0.001	39 (9.5)	<0.001	45 (6.1)	<0.001	37 (9.8)	0.019	38 (13.7)	0.009
PHF	37 (5.8)	48.97 (10.1)	<0.001	48.9 (12.9)	<0.001	46 (8.6)	0.001	45 (9.7)	0.038	55 (10.9)	0.004	51.5 (14.5)	0.004
TPKF	0.72 (0.03)	0.78 (0.05)	<0.001	0.81 (0.03)	<0.001	0.78 (0.06)	0.001	0.82 (0.02)	<0.001	0.78 (0.04)	0.006	0.79 (0.03)	<0.001

KMHE: knee at maximum hip extension, MHE: maximum hip extension, RKFES: range of knee flexion in early swing, RHFES: range of hip flexion in early swing, PKF: peak knee flexion in swing, HPKF: hip at peak knee flexion, KPHF: knee at peak hip flexion, PHF: peak hip flexion, TPKF: timing of the peak knee flexion, SD: standard deviation, P-values were obtained using the t-test (normality tested), P-values in red indicate that the value is statistically different from normal.

Table 3-2: Regression models between the catch angle/length and the knee or the hip joint at the investigated gait points/periods in Groups A & B (dynamic & true shortness).

Model output	Group A: catch angle/length<60°			Group B: catch/length≥60º		
	Model input/p-value	R^2	Model p-value	Model input/p-value	R^2	Model p-value
KMHE	Catch/length/0.031	24.5%	0.031	Catch/length/0.691	1%	0.691
MHE	Catch/length/0.134	12.7%	0.134	Catch/length/0.190	10.5%	0.190
RKFES	Catch/length/0.177 RHFES/<0.001	76.4%	<0.001	Catch/length/0.594 RHFES/<0.001	62%	<0.001
RHFES	Catch/length/0.038 RKFES/<0.001	79.9%	<0.001	Catch/length/0.970 RKFES/<0.001	61.3%	<0.001
PKF	Catch/length/0.029	25%	0.029	Catch/length/0.870 HPKF/0.013	29%	0.030
HPKF	Catch/length/0.385	4.5%	0.385	Catch/length/0.181 PKF/0.013	37.2%	0.012
KPHF	Catch/length/0.009	33.9%	0.009	Catch/length/0.013	32.8%	0.013
PHF	Catch/length/0.467	3.2%	0.467	Catch/length/0.321	6.1%	0.321

KMHE: knee at maximum hip extension, MHE: maximum hip extension, RKFES: range of knee flexion in early swing, RHFES: range of hip flexion in early swing, PKF: peak knee flexion in swing, HPKF: hip at peak knee flexion, KPHF: knee at peak hip flexion, PHF: peak hip flexion.

Table 3-3: The dominance relationship between the catch angle/length and knee or hip joint at the investigated gait points/periods in Groups A & B (dynamic & true shortness).

Catch angle/length°	Group A (catch angle/length<60°)		Group B (catch angle/length≥60º)	
	KMHE	MHE	KMHE	MHE
	Y	-	-	-
	RKFES	RHFES	RKFES	RHFES
	-	Y	-	-
	PKF	HPKF	PKF	HPKF
	Y	-	-	-
	KPHF	PHF	KPHF	PHF
	Y	-	Y	-

KMHE: knee at maximum hip extension, MHE: maximum hip extension, RKFES: range of knee flexion in early swing, RHFES: range of hip flexion in early swing, PKF: peak knee flexion in swing, HPKF: hip at peak knee flexion, KPHF: knee at peak hip flexion, PHF: peak hip flexion, Y: yes, indicating a relationship, (-): indicates no relationship.

Table 3-4: Significant linear regression models in Groups A & B (dynamic & true shortness).

Model output	Output equation	Model inputs	R^2	P-value	Input relationships
Group A (catch angle/length<60°)					
KMHE	KMHE=-1.393+ 0.411*catch angle/length	Catch angle/length	24.5%	0.031	Catch angle/length vs KMHE=direct relationship Ratio=1:0.411
RHFES	RHFES=-6.004+0.338*catch angle/length+0.896*RKFES	Catch angle/length and RKFES	79.9%	<0.001	-Catch angle/length vs RHFES=direct relationship Ratio=1:0.338 -RKFES vs RHFES=direct relationship Ration=1:0.896 -Catch angle vs RKFES=inverse relationship Ratio= 1:0.377
PKF	PKF=24.214+0.640*catch angle/length	Catch angle/length	25%	0.029	Catch angle/length vs PKF=direct relationship Ratio=1:0.640
KPHF	KPHF=5.522+0.739*catch angle/length	Catch angle/length	33.9%	0.009	Catch angle/length vs KPHF=direct relationship Ratio=1:0.739
Group B (catch angle/length≥60º)					
KPHF	KPHF=73.274-0.414*catch angle/length	Catch angle/length	6.1%	0.321	-Catch angle vs KPHF=inverse relationship Ratio=1:0.414

KMHE: knee at maximum hip extension, RKFES: range of knee flexion in early swing, RHFES: range of hip flexion in early swing, PKF: peak knee flexion in swing, KPHF: knee at peak hip flexion.

Table 3-5: Values of catch angle/length, MHE and KMHE in Groups A & B (dynamic & true shortness).

Catch angle/length°	MHE°	N (8°)-MHE°	KMHE°	N (14°)-KMHE°	Catch angle/length°	MHE°	N (8°)-MHE°	KMHE°	N (14°)-KMHE°
Group A (catch angle/length<60°)					Group B (catch angle/length≥60°)				
28	-11	19	13	1	60	2	6	11	3
35	-8	16	6	8	60	-32	40	46	-32
36	-24	32	11	3	60	-12	20	39	-25
40	-1	9	19	-5	60	-9	17	29	-15
40	-8	16	21	-7	70	20	-12	9	5
40	-4	12	19	-5	70	-27	35	13	1
42	-11	19	12	2	70	-5	13	20	-6
42	-19	27	25	-11	73	7	1	28	-14
45	3	5	6	8	75	-6	14	12	2
45	9	-1	19	-5	78	-12	20	28	-14
48	-8	16	13	1	80	2	6	20	-6
48	-7	15	18	-4	82	7	1	29	-15
50	-17	25	23	-9	90	-4	12	38	-24
50	-18	26	18	-4	90	-3	11	27	-13
50	-18	26	13	1	90	-11	19	30	-16
51	0.3	7.7	17	-3	90	2	6	36	-22
55	16	-8	18	-4	110	2	6	22	-8
55	17	-9	27	-13	115	3	5	14	0
55	-12	20	27	-13					

MHE: maximum hip extension, N-MHE: normal MHE-included limbs MHE, KMHE: knee at MHE, N-KMHE: normal KMHE-included limbs KMHE.

Table 3-6: Values of catch angle/length, RKFES and RHFES in Groups A & B (dynamic & true shortness).

Catch angle/length°	RKFES°	N (20°)-RKFES°	RHFES°	N (24°)-RHFES°	Catch angle/length°	RKFES°	N (20°)-RKFES°	RHFES°	N (24°)-RHFES°
Group A (catch angle/length<60°)					Group B (catch angle/length≥60°)				
28	6	14	12	12	60	33	-13	33	-9
35	3	17	4	20	60	9	11	10	14
36	16	4	24	0	60	13	7	18	6
40	14	6	21	3	60	15	5	21	3
40	18	2	23	1	70	8	12	35	-11
40	16	4	21	3	70	16	4	24	0
42	10	10	19	5	70	35	-15	56	-32
42	6	14	12	12	73	24	-4	33	-9
45	37	-17	37	-13	75	7	13	19	5
45	26	-6	31	-7	78	8	12	13	11
48	5	15	12	12	80	25	-5	43.8	-19.8
48	14	6	29	-5	82	21	-1	32	-8
50	11	9	19	5	90	3	17	13	11
50	3	17	13	11	90	16	4	29	-5
50	5	15	11	3	90	9	11	16	8
51	16	4	20	4	90	10	10	26	-2
55	18	2	41	-17	110	23	-3	25	-1
55	30	-10	41	-17	115	5	15	10	14
55	15	5	24	0					

RKFES: range of knee flexion in early swing, N-RKFES: normal RKFES-included limbs RKFES, RHFES: range of hip flexion in early swing, N-RHFES: normal RHFES-included limbs RHFES.

Table 3-7: Values of catch angle/length, PKF and HPKF in Groups A & B (dynamic & true shortness).

Catch angle/length°	PKF°	N (60°)-PKF°	HPKF°	N (23°)-HPKF°	Catch angle/length°	PKF°	N (60°)-PKF°	HPKF°	N (23°)-HPKF°
Group A (catch angle/length<60°)					Group B (catch angle/length≥60°)				
28	41	19	31	-8	60	58	2	35	-12
35	40	20	29	-6	60	67	-7	45	-22
36	54	6	62	-39	60	60	0	36	-13
40	68	-8	38	-15	60	65	-5	38	-15
40	58	2	37	-14	70	34	26	23	0
40	46	14	32	-9	70	55	5	62	-39
42	43	17	36	-13	70	69	-9	68	-45
42	43	17	40	-17	73	64	-4	39	-16
45	58	2	32	-9	75	45	15	35	-12
45	53	7	37	-14	78	54	5.7	29	-6
48	43	17	35	-12	80	67	-7	61	-38
48	48	12	50	-27	82	60	0	35	-12
50	56	4	45	-22	90	51	9	27	-4
50	56	4	49	-26	90	51	9	34	-11
50	53	7	48	-25	90	51	9	29	-6
51	62	-2	34	-11	90	53	7	29	-6
55	58	2	37	-14	110	67	-7	33	-10
55	53	7	47	-24	115	48	12	23	0
55	76	-16	35	-12					

PKF: peak knee flexion, N-PKF: normal PKF-included limbs PKF, HPKF: hip at PKF, N-HPKF: normal HPKF-included limbs HPKF.

Table 3-8: Values of catch angle/length, PHF and KPHF in Groups A & B (dynamic & true shortness).

Catch angle/length°	PHF°	N (37°)-PHF°	KPHF°	N (26°)-KPHF°	Catch angle/length°	PHF°	N (37°)-PHF°	KPHF°	N (26°)-KPHF°
Group A (catch angle/length<60°)					Group B (catch angle/length≥60°)				
28	44	-7	25	1	60	45	-8	46	-20
35	34	3	32	-6	60	66	-29	43	-17
36	74	-37	30	-4	60	40	-3	50	-24
40	48	-11	39	-13	60	50	-13	49	-23
40	41	-4	32	-6	70	72	-35	29	-3
40	50	-13	32	-6	70	77	-40	62	-36
42	46	-9	37	-11	70	33	4	29	-3
42	40	-3	41	-15	73	45	-8	55	-29
45	46	-9	42	-16	75	43	-6	36	-10
45	35	2	54	-28	78	41	-4	33	-7
48	45	-8	36	-10	80	69	-32	58	-32
48	60	-23	35	-9	82	41	-4	45	-19
50	59	-22	45	-19	90	35	2	41	-15
50	56	-19	36	-10	90	47	-10	35	-9
50	58	-21	43	-17	90	44	-7	40	-14
51	48	-11	25	2	90	42	-5	34	-8
55	43	-6	50	-24	110	53	-16	29	-3
55	42	-5	62	-36	115	39	-2	16	10
55	62	-25	41	-15					

PHF: peak hip flexion, N-PHF: normal PHF-included limbs PHF, KPHF: knee at PHF, N-KPHF: normal KPHF-included limbs KPHF.

Table 3-9: Linear regression models of the timing of peak knee flexion with catch angle/length in all groups.

Groups	R^2	Model p-value
A	8.5%	0.227
B	18.8%	0.072
C1	5.5%	0.44
C2	1.1%	0.820
D1	36.3%	0.206
D2	21.6%	0.149

Table 3-10: Values of catch angle and TPKF in Groups A & B (dynamic & true shortness).

Catch angle/length°	TPKF%	N (0.72%)- TPKF%	Catch angle/length°	TPKF%	N (0.72%)-TPKF%
Group A (catch angle/length<60°)			Group B (catch angle/length≥60°)		
28	0.74	-0.02	60	0.8	-0.08
35	0.66	0.06	60	0.8	-0.08
36	0.76	-0.04	60	0.88	-0.16
40	0.82	-0.1	60	0.78	-0.06
40	0.86	-0.14	70	0.78	-0.06
40	0.74	-0.02	70	0.84	-0.12
42	0.88	-0.16	70	0.8	-0.08
42	0.72	0	73	0.82	-0.1
45	0.82	-0.1	75	0.84	-0.12
45	0.82	-0.1	78	0.8	-0.08
48	0.86	-0.14	80	0.84	-0.12
48	0.72	0	82	0.82	-0.1
50	0.82	-0.1	90	0.84	-0.12
50	0.76	-0.04	90	0.8	-0.08
50	0.82	-0.1	90	0.82	-0.1
51	0.74	-0.02	90	0.82	-0.1
55	0.82	-0.1	110	0.78	-0.06
55	0.82	-0.1	115	0.72	0
55	0.76	-0.04			

TPKF: timing of peak knee flexion, N-TPKF: normal TPKF-included limbs TPKF.

Table 3-11: Regression models between the catch angle and the knee or the hip joint at the investigated gait points/periods in Group C (dynamic shortness).

Model output	Group C1: catch angle<60º			Group C2: catch angle≥60º		
	Model input/p-value	R^2	Model p-value	Model input/p-value	R^2	Model p-value
KMHE	Catch angle/0.062	28.1%	0.062	Catch angle/0.996	0%	0.996
MHE	Catch angle/0.380	7.1%	0.380	Catch angle/0.278	22.9%	0.278
RKFES	Catch angle/0.116 RHFES/<0.001	87.9%	<0.001	Catch angle/0.41 RHFES/0.002	90.3%	0.004
RHFES	Catch angle/0.035 RKFES/<0.001	90.2%	<0.001	Catch angle/0.057 RKFES/0.002	88.7%	0.006
PKF	Catch angle/0.020	40%	0.020	Catch angle/0.254	24.9%	0.254
HPKF	Catch angle/0.026	37.4%	0.026	Catch angle/0.085	47.8%	0.085
KPHF	Catch angle/<0.001 PHF/0.006	68.1%	0.001	Catch angle/0.549	7.6%	0.549
PHF	Catch angle/0.001 KPHF/0.006	60.5%	0.004	Catch angle/0.152	36.4%	0.152

KMHE: knee at maximum hip extension, MHE: maximum hip extension, RKFES: range of knee flexion in early swing, RHFES: range of hip flexion in early swing, PKF: peak knee flexion in swing, HPKF: hip at peak knee flexion, KPHF: knee at peak hip flexion, PHF: peak hip flexion.

Table 3-12: The dominance relationship between the catch angle and knee or hip joint at the investigated gait points/periods in Group C (dynamic shortness).

Catch angle°	Group C1 (catch angle<60º)		Group C2 (catch angle≥60º)	
	KMHE	MHE	KMHE	MHE
	-	-	-	-
	RKFES	RHFES	RKFES	RHFES
	-	Y	Y	-
	PKF	HPKF	PKF	HPKF
	Y	Y	-	-
	KPHF	PHF	KPHF	PHF
	Y	Y	-	-

KMHE: knee at maximum hip extension, MHE: maximum hip extension, RKFES: range of knee flexion in early swing, RHFES: range of hip flexion in early swing, PKF: peak knee flexion in swing, HPKF: hip at peak knee flexion, KPHF: knee at peak hip flexion, PHF: peak hip flexion, Y: yes, indicating a relationship, (-): indicates no relationship.

Table 3-13: Values of catch angle, MHE and KMHE in Group C (dynamic shortness).

Catch angle^{o}	MHE^{o}	N (8^{o})-MHE^{o}	KMHE^{o}	N (14^{o})-KMHE^{o}
Group C1 (catch angle<60^{o})				
28	-11	19	13	1
35	-8	16	6	8
40	-1	9	19	-5
40	-8	16	21	-7
42	-11	19	12	2
42	-19	27	25	-11
45	3	5	6	8
45	9	-1	19	-5
48	-8	16	13	1
48	-7	15	18	-4
50	-18	26	18	-4
55	17	-9	27	-13
55	-12	20	27	-13
Group C2 (catch angle≥60^{o})				
60	2	6	11	3
60	-32	40	46	-32
60	-12	20	39	-25
73	7	1	28	-14
75	-6	14	12	2
82	7	1	29	-15
90	-4	12	38	-24

MHE: maximum hip extension, N-MHE: normal MHE-included limbs MHE, KMHE: knee at MHE, N-KMHE: normal KMHE-included limbs KMHE.

Table 3-14: Significant linear regression models in Group C (dynamic shortness).

Model output	Output equation	Model inputs	R^2	P-value	Input relationships
Group C1 (catch angle<60º)					
RHFES	RHFES=-5.926+0.341*catch angle+0.853*RKFES	Catch angle and RKFES	90.2%	<0.001	Catch angle vs RHFES=direct relationship Ratio=1:0.341 -RKFES vs RHFES=direct relationship Ration=1:0.853 -Catch angle vs RKFES=inverse relationship Ratio= 1:0.399
PKF	PKF=13.8+0.84*catch angle	Catch angle	40%	0.020	Catch angle vs PKF=direct relationship Ratio=1:0.840
KPHF	KPHF=18.66+1.237*catch angle-0.733*PHF	Catch angle and PHF	68.1%	0.001	Catch angle vs KPHF=direct relationship Ratio=1:1.237 -PHF vs KPHF=inverse relationship Ration=1:0.733 -Catch angle vs PHF=direct relationship Ratio= 1:1.688
Group C2 (catch angle≥60º)					
RKFES	RKFES=18.027-0.336*catch angle+0.961*RHFES	Catch angle and RHFES	90.3%	0.004	Length vs RKFES=inverse relationship Ratio=1:0.336 -RHFES vs RKFES=direct relationship Ration=1:0.961 -Length vs RHFES=direct relationship Ratio= 1:0.35

RKFES: range of knee flexion in early swing, RHFES: range of hip flexion in early swing, PKF: peak knee flexion in swing, KPHF: knee at peak hip flexion, PHF: peak hip flexion.

Table 3-15: Values of catch angle, RKFES and RHFES in Group C (dynamic shortness).

Catch angle°	RKFES°	N (20°)-RKFES°	RHFES°	N (24°)-RHFES°
Group C1 (catch angle<60°)				
28	6	14	12	12
35	3	17	4	20
40	14	6	21	3
40	18	2	23	1
42	10	10	19	5
42	6	14	12	12
45	37	-17	37	-13
45	26	-6	31	-7
48	5	15	12	12
48	14	6	29	-5
50	3	17	13	11
55	30	-10	41	-17
55	15	5	24	0
Group C2 (catch angle≥60°)				
60	33	-13	33	-9
60	9	11	10	14
60	13	7	18	6
73	24	-4	33	-9
75	7	13	19	5
82	21	-1	32	-8
90	3	17	13	11

RKFES: range of knee flexion in early swing, N-RKFES: normal RKFES-included limbs RKFES, RHFES: range of hip flexion in early swing, N-RHFES: normal RHFES-included limbs RHFES.

Table 3-16: Values of catch angle, PKF and HPKF in Group C (dynamic shortness).

Catch angle°	PKF°	N (60°)-PKF°	HPKF°	N (23°)-HPKF°
Group C1 (catch angle<60°)				
28	41	19	31	-8
35	40	20	29	-6
40	58	2	37	-14
40	46	14	32	-9
42	43	17	36	-13
42	43	17	40	-17
45	58	2	32	-9
45	53	7	37	-14
48	43	17	35	-12
48	48	12	50	-27
50	56	4	45	-22
55	53	7	47	-24
55	76	-16	35	-12
Group C2 (catch angle≥60°)				
60	58	2	35	-12
60	67	-7	45	-22
60	60	0	36	-13
73	64	-4	39	-16
75	45	15	35	-12
82	60	0	35	-12
90	51	9	27	-4

PKF: peak knee flexion, N-PKF: normal PKF-included limbs PKF, HPKF: hip at PKF, N-HPKF: normal HPKF-included limbs HPKF.

Table 3-17: Values of catch angle, PHF and KPHF in Group C (dynamic shortness).

Catch angle°	PHF°	N (37°)-PHF°	KPHF°	N (26°)-KPHF°
Group C1 (catch angle<60°)				
28	44	-7	25	1
35	34	3	32	-6
40	48	-11	39	-13
40	41	-4	32	-6
42	46	-9	37	-11
42	40	-3	41	-15
45	46	-9	42	-16
45	35	2	54	-28
48	45	-8	36	-10
48	60	-23	35	-9
50	56	-19	36	-10
55	42	-5	62	-36
55	62	-25	41	-15
Group C2 (catch angle≥60°)				
60	45	-8	46	-20
60	66	-29	43	-17
60	40	-3	50	-24
73	45	-8	55	-29
75	43	-6	36	-10
82	41	-4	45	-19
90	35	2	41	-15

PHF: peak hip flexion, N-PHF: normal PHF-included limbs PHF, KPHF: knee at PHF, N-KPHF: normal KPHF-included limbs KPHF.

Table 3-18: Values of catch angle and TPKF in Group C (dynamic shortness).

Catch angle^{o}	TPKF%	N (0.72%)-TPKF%
Group C1 (catch angle<60^{o})		
28	0.74	-0.02
35	0.66	0.06
40	0.82	-0.1
40	0.86	-0.14
42	0.72	0
42	0.88	-0.16
45	0.82	-0.1
45	0.82	-0.1
48	0.86	-0.14
48	0.72	0
50	0.76	-0.04
55	0.82	-0.1
55	0.76	-0.04
Group C2 (catch angle≥60^{o})		
60	0.80	-0.08
60	0.80	-0.08
60	0.88	-0.16
73	0.82	-0.1
75	0.84	-0.12
82	0.82	-0.1
90	0.84	-0.12

TPKF: timing of peak knee flexion, N-TPKF: normal TPKF-included limbs TPKF.

Table 3-19: Regression models between the length and the knee or the hip joint at the investigated gait points/periods in Group D (true shortness).

Model output	Group D1: length<60º			Group D2: length≥60º		
	Model input/p-value	R^2	Model p-value	Model input/p-value	R^2	Model p-value
KMHE	Length/0.397	18.4%	0.397	Length/0.897	0.2%	0.897
MHE	Length/0.238	32.5%	0.238	Length/0.441	6.7%	0.441
RKFES	Length/0.655	5.5%	0.655	Length/0.569 RHFES/0.004	61.5%	0.009
RHFES	Length/0.689	4.4%	0.689	Length/0.261 RKFES/0.004	66%	0.005
PKF	Length/0.024 HPKF/0.010	87.2%	0.021	Length/0.383 HPKF/0.029	33.8%	0.079
HPKF	Length/0.014 PKF/0.010	91.1%	0.012	Length/0.115 PKF/0.029	47.3%	0.031
KPHF	Length/0.240	32.3%	0.240	Length/0.068	32.2%	0.068
PHF	Length/0.126	48.2%	0.126	Length/0.322	10.9%	0.322

KMHE: knee at maximum hip extension, MHE: maximum hip extension, RKFES: range of knee flexion in early swing, RHFES: range of hip flexion in early swing, PKF: peak knee flexion in swing, HPKF: hip at peak knee flexion, KPHF: knee at peak hip flexion, PHF: peak hip flexion.

Table 3-20: The dominance relationship between the length and knee or hip joint at the investigated gait points/periods in Group D (true shortness).

Length°	Group D1 (length<60º)		Group D2 (length≥60º)	
	KMHE	MHE	KMHE	MHE
	-	-	-	-
	RKFES	RHFES	RKFES	RHFES
	-	-	Y	-
	PKF	HPKF	PKF	HPKF
	Y	Y	-	-
	KPHF	PHF	KPHF	PHF
	-	-	-	-

KMHE: knee at maximum hip extension, MHE: maximum hip extension, RKFES: range of knee flexion in early swing, RHFES: range of hip flexion in early swing, PKF: peak knee flexion in swing, HPKF: hip at peak knee flexion, KPHF: knee at peak hip flexion, PHF: peak hip flexion, Y: yes, indicating a relationship, (-): indicates no relationship.

Table 3-21: Values of length, MHE and KMHE hip extension in Group D (true shortness).

Length°	MHE°	N (8°)-MHE°	KMHE°	N (14°)-KMHE°
Group D1 (length<60°)				
36	-24	32	11	3
40	-4	12	19	-5
50	-17	25	23	-9
50	-18	26	13	1
51	0.3	7.7	17	-3
55	16	-8	18	-4
Group D2 (length≥60°)				
60	-9	17	29	-15
70	20	-12	9	5
70	-27	35	13	1
70	-5	13	20	-6
78	-12	20	28	-14
80	2	6	20	-6
90	-3	11	27	-13
90	-11	19	30	-16
90	2	6	36	-22
110	2	6	22	-8
115	3	5	14	0

MHE: maximum hip extension, N-MHE: normal MHE-included limbs MHE, KMHE: knee at MHE, N-KMHE: normal KMHE-included limbs KMHE.

Table 3-22: Values of length, RKFES and RHFES in Group D (true shortness).

Length°	RKFES°	N (20°)-RKFES°	RHFES°	N (24°)-RHFES°
Group D1 (length<60°)				
36	16	4	24	0
40	16	4	21	3
50	11	9	19	5
50	5	15	11	3
51	16	4	20	4
55	18	2	41	-17
Group D2 (length≥60°)				
60	15	5	21	3
70	8	12	35	-11
70	16	4	24	0
70	35	-15	56	-32
78	8	12	13	11
80	25	-5	44	-20
90	16	4	29	-5
90	9	11	16	8
90	10	10	26	-2
110	23	-3	25	-1
115	5	15	10	14

RKFES: range of knee flexion in early swing, N-RKFES: normal RKFES-included limbs RKFES, RHFES: range of hip flexion in early swing, N-RHFES: normal RHFES-included limbs RHFES.

Table 3-23: Significant linear regression models in Group D (true shortness).

Model output	Output equation	Model inputs	R^2	P-value	Input relationships
Group D1 (length<60º)					
HPKF	HPKF=176.64-1.453*PKF-0.999*length	Length and PKF	91.1%	0.012	Length vs HPKF=inverse relationship Ratio=1:0.999 -PKF vs HPKF=inverse relationship Ratio=1:1.453 -Length vs PKF=inverse relationship Ratio= 1:0.688

PKF: peak knee flexion in swing, HPKF: hip at peak knee flexion.

Table 3-24: Values of length, PKF and HPKF in Group D (true shortness).

Length°	PKF°	N (60°)-PKF°	HPKF°	N (23°)-HPKF°
Group D1 (length<60°)				
36	54	6	62	-39
40	68	-8	38	-15
50	56	4	49	-26
50	53	7	48	-25
51	62	-2	34	-11
55	58	2	37	-14
Group D2 (length≥60°)				
60	65	-5	38	-15
70	34	26	23	0
70	55	5	62	-39
70	69	-9	68	-45
78	54	6	29	-6
80	67	-7	61	-38
90	51	9	34	-11
90	51	9	29	-6
90	53	7	29	-6
110	67	-7	33	-10
115	48	12	23	0

PKF: peak knee flexion, N-PKF: normal PKF-included limbs PKF,
HPKF: hip at PKF, N-HPKF: normal HPKF-included limbs HPKF.

Table 3-25: Values of length, PHF and KPHF in Group D (true shortness).

Length°	PHF°	N (37°)-PHF°	KPHF°	N (26°)-KPHF°
Group D1 (length<60°)				
36	74	-37	30	-4
40	50	-13	32	-6
50	58	-21	43	-17
50	59	-22	45	-19
51	48	-11	25	2
55	43	-6	50	-24
Group D2 (length≥60°)				
60	50	-13	49	-23
70	72	-35	29	-3
70	77	-40	62	-36
70	33	4	29	-3
78	41	-4	33	-7
80	69	-32	58	-32
90	44	-7	40	-14
90	42	-5	34	-8
90	47	-10	35	-9
110	53	-16	29	-3
115	39	-2	16	10

PHF: peak hip flexion, N-PHF: normal PHF-included limbs PHF, KPHF: knee at PHF, N-KPHF: normal KPHF-included limbs KPHF.

Table 3-26: Values of length and TPKF in Group D (true shortness).

Length°	TPKF%	N (0.72%)-TPKF%
Group D1 (length<60°)		
36	0.76	-0.04
40	0.74	-0.02
50	0.82	-0.10
50	0.82	-0.10
51	0.74	-0.02
55	0.82	-0.10
Group D2 (length≥60°)		
60	0.78	-0.06
70	0.80	-0.08
70	0.78	-0.06
70	0.84	-0.12
78	0.80	-0.08
80	0.84	-0.10
90	0.82	-0.10
90	0.82	-0.10
90	0.80	-0.08
110	0.78	-0.06
115	0.72	0

TPKF: timing of peak knee flexion, N-TPKF: normal TPKF-included limbs TPKF.

Chapter 4 Clinical tools for joint range of motion measurement

'Intratester reliability of universal goniometer'

(An entry to the University of Strathclyde 'Images of Research 2013' competition (top 50))

'Measure for Measure'

(An entry to the University of Strathclyde 'Images of Research 2016' competition (top 50))

4.1 Factors influencing the reliability of different designs of goniometer in measurement of lower limb range of motion: a literature review

4.1.1 Introduction

A variety of measurement tools are currently used for joint ROM measurement, ranging from simple visual estimation to advanced 3D video analysis systems.[239-241] The most practical, and most frequently used, clinical tool is the goniometer, and several different designs have been developed over the years. The simplest and most inexpensive type is the UG, with more advanced designs including electrical (EG) and gravity-dependent goniometers (inclinometers). The UG is a 180º or 360º protractor with a single axis joining two arms. One arm is movable around the axis while the other arm is stationary.[242] UGs are available in different sizes to suit the joint being measured. The inclinometer contains a pointer and a fluid level, which is affected by gravity to measure joint ROM. It resembles the carpenter's level meter.[243] The EG has two arms connected by a potentiometer. Changes in joint position cause changes in the resistance of the potentiometer. The resulting change in voltage is used to measure the amount of joint ROM.[243] For measurements obtained using the UG, EG and inclinometer to be clinically useful, results must be accurate and reliable.

Intratester reliability is important as the same clinician may take the same measurement on different occasions to document change. In the clinical setting, however, more than one tester may be involved in the measurement. As this has potential for further error, intertester reliability must also be considered clinically relevant.[244, 245]

Several factors may affect the reliability of the measurements obtained using the UG, EG and inclinometer. Reliability might vary between different joints as each joint has different characteristics. Arguably this may make it easier or harder to obtain reliable measurements depending on the joint measured.[246-251]

The type of movement measured may also affect reliability, and reliability of active movements may differ when compared to passive movements, as the force applied by the therapist to move the joint varies. This may cause different angles to be obtained each time measurements are taken.[251] In addition, following a standard instruction procedure and prior training may affect the reliability as this minimises the error associated with different procedures.

Different pathologies may influence the UG, EG and inclinometer reliability. Upper motor neuron disorders may cause variation in muscle tone. Hypertonicity and spasticity may influence the ability to accurately define end range of joint motion, which in turn may affect the reliability of the measurements. Additionally, the presence of bony deformations may cause difficulties in clearly identifying bony landmarks, which may compromise the reliability of measurements.

It is essential to understand how reliability may be affected when such variables are introduced since incorrect interpretation of measurements obtained may lead to inappropriate treatment. The aim of this part of the thesis is, therefore, to investigate the intratester and intertester reliability of the UG, EG and inclinometer and to examine how different factors influence measurement reliability.

4.1.2 Methods

An electronic and manual literature search was conducted to investigate the intratester and intertester reliability of the UG, EG and inclinometer for measurement of lower limb joints. A variety of search terms were used to search different medical and engineering databases such as Medline, EMBASE, NHS Scotland e-library, Science Direct, PubMed and Google Scholar. Search terms included reliability and/or universal goniometer, universal goniometer, goniome* measurement reliability, intertester reliability of universal goniometer or UG, intratester reliability of universal goniometer or UG, reliability and/or electrical goniometer, electrical goniometer, goniome* measurement reliability, intertester reliability of electrical goniometer or EG, intratester reliability of electrical goniometer or EG, reliability and/or inclinometer, inclinometer, inclionme* measurement reliability, intertester reliability of inclinometer or gravity dependent goniometer, and intratester reliability of inclinometer or gravity dependent goniometer. Identified secondary references from the articles were found and related books were also reviewed.

The review investigated the reliability of the UG, EG and inclinometer in measuring joint ROM of the lower limb. Inclusion criteria were as follows: studies that evaluated intratester reliability and/or intertester reliability of the UG, EG or inclinometer, studies that included subjects with pathology and/or healthy subjects, studies that measured hip, knee and/or ankle joint, and studies that used ICC to calculate reliability. Exclusion criteria included: studies that only investigated measurement of the upper limb and studies that did not use ICC to calculate the reliability. References from 1980 to July 2015 were included to ensure the number of the studies was manageable.

Papers sourced were graded according to the Scottish Intercollegiate Guideline Network (SIGN),[252] and thematic tables of evidence were constructed for each design of goniometer and the pathology of the subjects tested.

Different statistical methods were used in order to measure the reliability, such as ICC, Pearson product moment correlation coefficient, analysis of variance reliability, and coefficient of variation and generalisability theory. However, this review concentrated on papers that used ICC values to calculate reliability. This method was considered the most appropriate method for reliability measurement as data is centred and scaled using a pooled mean and standard deviation. Furthermore, as the correlation line between the values is drawn at a 45º angle, this was considered to reflect the most accurate reliability value.[253, 254] Most studies did not report the reason for the chosen method of analysis, although one paper stated that ICC best reflected errors associated with measurements.[255] It has been suggested that the Pearson product moment correlation coefficient may produce high reliability values even when large inconsistency between paired scores is found.[256] This statistical method may overestimate reliability as each variable is centred and scaled by its own mean and standard deviation. Additionally, the correlation line is drawn at its best position without specifying location.[253]

4.1.3 Results

4.1.3.1 Search results

The initial search yielded seventy-one studies, of which twenty-nine papers matched the inclusion criteria and were fully reviewed.[246-249, 251, 255, 257-274] All the studies were case series (SIGN grade 3).

4.1.3.2 Statistical analysis

Intraclass correlation coefficient (ICC or reliability) values were rated as weak (ICC=0-0.60), good (ICC=0.60-0.80), or excellent (ICC>0.80).[254, 257, 258, 275, 276] Six studies reported on intratester reliability only.[246, 248, 263, 273, 277, 278] Six studies reported on intertester reliability only.[247, 249, 262, 267, 271, 279] Seventeen studies reported on both intratester and intertester reliability (Tables 4-1 to 4-3).[251, 255, 257-261, 264-266, 268-270, 272, 274, 280, 281]

4.1.3.3 Motion measured and measurement procedure:

Twenty-one studies examined passive motion[246-249, 251, 255, 257-260, 263, 265-271, 274, 279, 280] and eight studies active motion.[261, 262, 264, 272, 273, 277, 278, 281] Four studies did not give testers standard instructions to follow or prior training.[255, 257, 258, 261]

4.1.3.4 Participants

Thirteen studies included healthy subjects,[249, 262, 263, 267-270, 272-274, 277, 278, 281] while twelve studies included subjects with various pathologies including diabetes,[259] neurological conditions,[246, 247, 251, 264-266] orthopaedic conditions[261, 279] and neurological and orthopaedic conditions.[255, 258] One study stated only that the participants were

nursing home residents.[260] Two studies included both subjects with neurological conditions and healthy subjects,[248, 271] and one study included both subjects with orthopaedic conditions and healthy subjects.[280] Sample sizes varied widely (range: 6-150).

4.1.3.5 Goniometer design and comparison

Twenty-one studies investigated the reliability of the UG only.[246-249, 251, 255, 257-264, 266, 268-271, 273, 274] Three studies investigated the reliability of inclinometer only,[272, 279, 281] and none of the studies investigated the reliability of the EG only. One study compared the reliability of the EG and UG,[277] and two studies compared the reliability of the UG and inclinometer.[267, 278] A further study compared the reliability of the UG and inclinometer but did not state any numerical ICC values for the UG.[280] Another study investigated the reliability of the UG and inclinometer but did not compare them.[265]

4.1.3.6 Reliability of universal goniometer

Hip joint

Healthy subjects

Active motion

One study concluded that intertester reliability for measuring internal and external rotation was excellent (ICC=0.90-0.94) (Table 4-1).[262] Another study found excellent intratester reliability (ICC=0.80-0.96) for measurement of hip adduction (ICC=0.80), abduction (ICC=0.86), lateral rotation (0.80), medial rotation (ICC=0.92), extension (ICC=0.83) and flexion (ICC=0.95) ROM.[277]

Passive motion

One study reported good to excellent intratester reliability for measurement of hip extension (ICC=0.70-0.96).[263] Two studies reported weak to excellent intratester reliability for measurement of hip extension (ICC=0.09-0.92).[248, 268] Two studies found weak to good intertester reliability for measurement of hip extension (ICC=0.10-0.65).[268, 271] In contrast, another study found excellent intertester reliability for measurement of hip extension (ICC=0.92).[267] A single study reported on intratester reliability for hip flexion and found weak to excellent reliability (ICC=0.52-0.99) (Table 4-1).[248] Another study stated that the measurements obtained for hip rotation were reliable, but no numerical ICC values were provided.

Subjects with pathology

Active motion

No study was found investigating measurement of active hip motion.

Passive motion

Three studies reported good to excellent intratester and intertester reliability (ICC=0.61-0.981) for measurements of hip extension, flexion, abduction and lateral rotation amongst subjects with neurological conditions.[246, 251, 266] Two studies also found excellent intratester reliability for measurement of hip abduction (ICC=0.82-0.95) and hip extension (ICC=0.98) but weak intertester reliability for measurement of hip extension (ICC=0.24) and hip abduction (ICC=0.37-0.47) amongst the same subject group.[260, 265] Another study reported weak intertester reliability for measurement of hip extension (ICC=0.19-0.50).[271] One study found inconsistent

results and significant variation in intratester reliability within one session and between sessions for measurement of hip extension (ICC=0.17-0.91) and flexion (ICC=0.55-0.80) (Table 4-1).[248] Another study stated that the measurements obtained for hip rotation amongst subjects with orthopaedic conditions was reliable, but no numerical ICC values were provided.[280]

Knee joint

Healthy subjects

Active motion

One study was found which concluded that intratester reliability was excellent for knee flexion (ICC=0.91) and knee extension (ICC=0.80).[277]

Passive motion

Two studies reported excellent intratester reliability for measurement of knee flexion (ICC=0.96-0.99) and knee extension (ICC=0.83-0.97).[257, 269] One study reported good intratester reliability for measurement of knee flexion (ICC=0.65-0.72),[270] while another study found weak to excellent intratester reliability for knee extension measurement (ICC=0.34-0.99).[248] Other studies found good to excellent intertester reliability during measurement of flexion (ICC=0.88-0.99) and extension (ICC=0.64-0.71) (Table 4-2).[249, 257, 269] On the other hand, three studies found weak to good intertester reliability for measurement of knee extension (ICC=0.21-0.68) and flexion (ICC=0.44-0.59).[269-271]

Subjects with pathology

Active motion

A single study was found which reported excellent intratester and intertester reliability of flexion and extension amongst subjects with orthopaedic conditions (ICC=0.89-0.99) (Table 4-2).[264]

Passive motion

Intratester reliability was found to be excellent in four studies investigating measurements of knee flexion (ICC=0.99) and extension (ICC=0.81-0.98) amongst subjects with neurological and orthopaedic conditions.[246, 251, 255, 260] However, one study reported weak to excellent intratester reliability for measurement of knee extension (ICC=0.57-0.92) amongst subjects with neurological conditions.[248] Two studies reported weak intertester reliability for measurement of knee extension (ICC=0.26[271] and 0.58[251]) amongst subjects with neurological conditions. On the other hand, three studies reported good to excellent intertester reliability for measurement of knee extension (ICC=0.78-0.96) (Table 4-2) amongst subjects with neurological and orthopaedic conditions.[247, 255, 260] One study reported excellent intertester reliability (ICC=0.90) for measuring knee flexion amongst subjects with neurological and orthopaedic conditions.[255]

Ankle joint

Healthy subjects

Active motion

Two studies found that measurement of ankle dorsiflexion produced excellent intratester reliability (ICC=0.85-0.96).[273, 278] Another study concluded excellent intratester reliability for ankle dorsiflexion (ICC=0.92) and plantarflexion (ICC=0.96).[277]

Passive motion

Two studies were found which reported good to excellent intratester and intertester reliability for ankle dorsiflexion (ICC=0.63-0.99) (Table 4-3).[248, 274]

Subjects with pathology

Active motion

A single study reported weak to excellent intratester reliability (ICC=0.47-0.93) and weak intertester reliability (ICC=0.25-0.28) for the measurement of plantarflexion and dorsiflexion amongst subjects with orthopaedic conditions (Table 4-3).[261]

Passive motion

A single study examined measurement of dorsiflexion and plantarflexion amongst subjects with diabetes and found excellent intratester reliability (ICC=0.89-0.96), while the intertester reliability varied between good to excellent (0.74-0.89).[259] Five studies reported excellent intratester reliability (ICC=0.81-0.99) during measurement of dorsiflexion and plantarflexion amongst subjects with neurological and orthopaedic

conditions.[246, 248, 251, 258, 266] By contrast, two other studies reported weak to good intertester reliability (ICC=0.12-0.73) during measurement of dorsiflexion and plantarflexion amongst the same subject group,[251, 258] excluding two studies where excellent intertester reliability was found for measurements of plantarflexion and dorsiflexion (ICC=0.87-0.88).[247, 266]

4.1.3.7 Reliability of electrical goniometer

Hip joint

Healthy subjects

Active motion

One study was found which concluded good to excellent intratester reliability (ICC=0.72-0.86) for measurement of hip adduction (ICC=0.77), abduction (ICC=0.79), lateral rotation (ICC=0.86), medial rotation (0.86), extension (ICC=0.72) and flexion (ICC=0.89) ROM.[277]

Passive motion

No study was found investigating measurement of passive hip motion.

Subjects with pathology

No study was found investigating measurements of active or passive hip motion.

Knee joint

Healthy subjects

Active motion

One study was found which concluded that intratester reliability was excellent for knee flexion (ICC=0.91) and knee extension (ICC=0.80) (4-2).[277]

Passive motion

No study was found investigating measurement of passive knee motion.

Subjects with pathology

No study was found investigating measurements of active or passive knee motion.

Ankle joint

Healthy subjects

Active motion

One study was found which concluded good to excellent intratester reliability for ankle dorsiflexion (ICC=0.80) and plantarflexion (ICC=0.93).[277]

Passive motion

No study was found investigating measurement of passive ankle motion.

Subjects with pathology

No study was found investigating measurements of active or passive ankle motion.

4.1.3.8 Reliability of inclinometer

Hip joint

Healthy subjects

Active motion

No study was found investigating measurement of active hip motion.

Passive motion

One study found excellent intertester reliability for measurement of hip extension (ICC=0.91).[267] Another study found excellent intratester and intertester reliability (ICC=0.96-0.99) for measurement of hip rotation.[280]

Subjects with pathology

Active motion

No study was found investigating measurement of passive hip motion.

Passive motion

Excellent intratester and intertester reliability (ICC=0.95-0.97) was found during measurement of hip rotation amongst subjects with low back dysfunction.[280] On the other hand, weak intertester reliability (ICC=0.43-0.48) was found for measurement of hip rotation amongst subjects with symptoms of osteoarthritis.[279] A further study found excellent intratester and intertester (ICC=0.85-0.97) reliability for measurement of hip abduction amongst subjects with neurological conditions.[265]

Knee joint

No study was found investigating measurements of active or passive knee motion amongst healthy subjects or subjects with pathology.

Ankle joint

Healthy subjects

Active motion

Two studies reported that measurement of ankle dorsiflexion produced excellent intratester (ICC=0.85-0.98) and intertester reliability (ICC=0.97).[278, 281] Another study found good to excellent intratester reliability (ICC=0.77-0.91) and excellent intertester reliability for measurement of ankle dorsiflexion.[272]

Passive motion

No study was found investigating measurement of passive ankle motion.

Subjects with pathology

No study was found investigating measurements of passive or active ankle motion.

4.1.4 Discussion

Several different designs of goniometers have been developed over the years including the UG, EG and gravity-dependent goniometer (inclinometers).[267, 282, 283] The UG is the most frequently used tool in the clinical environment. However, the disadvantage of using the UG is the requirement of having to use both hands to move the joint while simultaneously aligning the UG with bony landmarks, which may compromise

reliability.[258, 284] Inclinometers are less frequently used compared with UGs as they tend to be more expensive. The inclinometer measures the angle of the segment against vertical.[21] Conversely, inclinometers are easier to use compared to UGs as only one hand is required.[30] EGs convert angular motion to an electrical signal, which is then displayed on the device as ROM measurement. The EG requires careful calibration prior to attaching it to the subject and is time consuming. Straps are used to attach the EG to the subject with cables, which may restrict movement. New technologies are recently emerging for joint ROM measurements, such as 2D and 3D video analysis systems.[285, 286]

This review included 29 studies investigating intratester and intertester reliability of the UG, EG and inclinometer in measuring active or passive ROM of lower limb joints amongst subjects with pathology or healthy subjects.

In general, reliability of the goniometers varied across different pathologies, proving to be most reliable amongst healthy subjects. A number of studies stated that the presence of spasticity is a major cause of error[265, 275, 287, 288] and concluded that care should be taken when using the measurements obtained using the UG or inclinometer for assisting in clinical judgment.[275, 287] However, Kilgour et al.[248] compared measurement reliability of the UG amongst healthy subjects to those with spastic diplegia and found equal reliability.[248] This study concluded that a major cause of error was in defining the end range of the joint ROM rather than the presence of spasticity. Furthermore, Lee et al.[271] also compared measurement reliability of the UG amongst healthy subjects to those with CP and found higher reliability amongst subjects with

CP. Elveru et al.[258] reported a higher intertester reliability for the UG for ankle plantarflexion ROM in subjects with general orthopaedic conditions in comparison with subjects with neurological conditions. Another study stated that equal reliability is achieved for hip rotation measurements using the UG and inclinometer amongst subjects with low back pain and healthy subjects.[280] However, numerical ICC values were only stated for inclinometer.

Some studies included more than one form of pathology and grouped results without reporting on each pathology individually.[255, 257, 258, 260, 264] Three studies included more than one form of CP and did not report on each group separately, hence reducing the ability to interpret results.[247, 265 271]

Watkins et al.[255] investigated intratester and intertester reliability of the UG for knee joint ROM amongst subjects with knee joint problems and reported excellent reliability (Table 4-2). A posterior analysis was performed in the study to determine the effect of different pathologies on the reliability. Overall, pathology did not influence the intratester and intertester reliability. However, intertester reliability for knee extension was found to be weak amongst below-knee amputees, which may be explained due to short distal limb segment causing difficulties in aligning the UG.

Passive ROM is the motion mostly measured in clinical environments, and only 8 studies included in this review reported on active motion, while twenty-one studies reported on passive motion. Two studies reported excellent intratester and intertester reliability of the UG for measuring hip and knee joint active ROM (Tables 4-1 & 4-

2).[262, 264] Another study reported weak to excellent intratester reliability and weak intertester reliability for the UG for measuring active ankle ROM (Table 4-3).[261] Thomas and Rome[273] concluded excellent intratester reliability for the UG for measurement of active ankle dorsiflexion. Clapper and Wolf[277] concluded good to excellent intratester reliability for measurements of active ROM of hip, knee and ankle using the UG and EG. Furthermore, three studies reported good to excellent intratester and intertester for measurement of active ankle dorsiflexing using the inclinometer.[272, 278, 281] No study was found reporting on the reliability of the EG for measurement of passive motion. However, one study suggested that the low intertester reliability could be explained due in part to the difference in the force applied by therapists during assessments of passive motion, causing different angles to be obtained during each testing session.[251]

It was noted that reliability varied across the joints measured due to the different joint characteristics and ease of identifying bony landmarks. Overall, reliability varied from weak to excellent across the hip, knee and ankle joint. The knee joint was found to be reliable to measure, and this is supported with the high ICC values found (Table 4-2).[249, 255, 257, 264] Measurement of knee flexion appears to be more reliable than measurement of knee extension ROM (Table 4-2).[255, 257, 264] Similar results were found for measurement of hip joint ROM, as some studies reported excellent intratester and intertester reliability for the measurement of hip extension, abduction and external rotation (Table 4-1).[246, 251, 260, 265-267] This suggests that the length of lever arms may have more effect. Aligning the goniometer to follow the long bones in the thigh, calf and mid-lateral trunk may assist knee and hip joint ROM measurements, making this

more reliable. Furthermore, it has been reported that even complex motions can be measured reliably when strict standard position is applied.[289, 290]

Most studies provided the testers with a standard measurement procedure and prior training in order to minimise associated error. However, Rothstein et al.[257] deliberately did not standardise the measurement procedure using the UG (measuring technique and subject's position) to mimic the clinical setting and stated that "measurement technique will often vary between the therapists, partially because of their training and preferences and partially because of adaptation, such as positioning which are necessary with different patients."(p 1611) The study reported high intratester reliability during measurement of passive knee flexion and extension ROM and high intertester for knee flexion ROM but lower intertester reliability for knee extension ROM amongst subjects with pathology (Table 4-2).[257] To examine the effect of the different patient positions used in the study, a posterior analysis of the results was carried out, and an increase in intertester reliability for knee extension was reported from (ICC=0.20-0.69) to (ICC=0.74-0.84) when the same position was used. It was suggested that using different patient positions while measuring causes variability due to the bi-articular muscles (hamstrings) affecting the knee extension.[257] The hamstring muscles cross both hip and knee joint limiting knee extension ROM when the hip is flexed; hence, variation in position of the hip joint during measuring knee joint extension can cause differences in the measurements obtained. Subject position varied across a number of studies. The positions used to measure hip joint ROM were the Thomas test, modified Thomas test, prone hip extension test, supine position with knee maintained in different degrees of flexion, prone position and seated position. Kilgour

et al.[248] reported higher intratester ICC values for the Staheli test (ICC=0.78-0.91) in comparison to the Thomas test (ICC=0.17-0.66) using the UG in subjects with CP. Furthermore, intratester ICC values for the prone hip extension test (ICC=0.80-0.92) were found to be higher than intratester ICC values for the Thomas test (ICC=0.09-0.91) amongst healthy subjects in the previous study. A further study found weak intertester reliability of the UG when using the Thomas test (ICC=0.58).[247] In contrast, Lee et al.[271] reported higher ICC values for the Thomas test (ICC=0.20-0.50) in comparison to the prone hip extension test (ICC=0.10-0.19) amongst subjects with CP and healthy subjects using the UG. Calpis et al.[267] reported excellent intertester reliability for hip extension measurement using the modified Thomas test with the UG and inclinometer amongst healthy subjects. Van Dillen et al.[263] compared 4 positions using the UG for measuring hip extension, which included femur maintained in 0° of abduction with knee maintained in 80° of flexion, femur maintained in 0° of abduction with knee fully extended, femur fully abducted with knee maintained in 80° of flexion and femur in full abduction and knee fully extended. In this study, the higher intratester reliability was achieved for the position where the femur was fully abducted and knee fully extended (ICC=0.96). Simoneau et al.[262] measured hip external and internal rotation using the UG in prone and seated positions and concluded higher ICC values were achieved for internal rotation (ICC=0.94) and external rotation (ICC=0.93) in the prone position. The study recommended documentation of the position of the hip to allow repeated reliable measurements. It has been reported that proper aligning of the UG when measuring using the Thomas test or prone hip extension test can be difficult as one hand is used to ensure the lumbar spine is flat, and the other hand is used to align the UG while maintaining the position of both arms of the UG.[247] Ellison et al.[280]

reported that no significant difference was found in measuring hip rotation in prone and seated position using the UG and inclinometer amongst healthy subjects and subjects with pathologies. However, no ICC values were provided for comparison. Additionally, it was stated that measuring in the prone position is better as patient could be more easily stabilised. For knee ROM measurements, the following positions were used: popliteal angle and supine position with hip extended. Kilgour et al.[248] compared measuring knee extension with the hip in neutral and in 90° of flexion using the UG and found higher ICC values with the hip in a neutral position in subjects with CP and in healthy subjects. On the other hand, Cadenhead et al.[246] found equal reliability for the UG when measuring knee extension while maintaining the hip in neutral or 90° of flexion. For ankle joint ROM measurements, the following positions were used: supine position with knee extended, supine position with knee flexed, prone position and weight-bearing lung position. A study found equal intratester reliability when measuring ankle dorsiflexion with knee extended and knee flexed using the UG.[248] Diamond et al.[259] and Jonson and Gross[274] measured ankle dorsiflexion in prone position using the UG and reported good to excellent intratester and intertester reliability. Three studies found good to excellent intratester and intertester reliability for the UG and inclinometer using the weight-bearing lunge measure of ankle dorsiflexion amongst healthy subjects.[272, 278, 281] Thomas and Rome investigated the effect of 3 different subject positions (prone, supine and sitting) on the reliability of the UG for measuring active ankle dorsiflexion; however, only one ICC value was reported with no further information about the position.

Another source of error may be due to discrepancies in identification of bony landmarks and goniometric alignment between testers. Peeler and Anderson[270] carried out a pilot test with 3 testers and found that differences were reported between testers when identifying the lateral epicondyle of the femur (used to align the axis of the goniometer), especially in subjects with pathological changes at the knee.[257] In addition, they reported that difficulties were found in maintaining the position of the axis of the UG when trying to align the two arms. Differences in identification of bony landmarks can increase the error with any design of goniometer, but the error associated with identification of bony landmarks can be reduced by using inclinometers rather than UGs, as fewer bony landmarks require identification.[261, 265]Additionally, Watkins et al.[255] followed non-standard measuring procedure using the UG and reported excellent intratester and intertester reliability for measurements of knee flexion and extension in subjects with knee pathologies. A posterior analysis of the results showed that non-standardisation of the measurement procedure contributed slightly to measurement error but still suggested that standard procedures be applied to minimise this error (when the same position was used, ICC for flexion increased by 0.02 and for extension by 0.01).[255] A further study found weak intertester reliability when measuring active ankle ROM using the UG amongst subjects with pathology when the position was not standardised, suggesting that a standard protocol should be established and followed.[261] The weak intertester reliability reported in this study may be explained due to the variation in the UG alignment using bony landmarks. In measuring the ankle joint, the fixed arm is aligned over the long axis of fibula; however, the moveable arm could be aligned with the heel, fifth metatarsal or plantar surface of the foot causing variation in measurements amongst testers.[261] In

addition, the variation found in the study by Youdas et al.[261] may be explained due to the effect of the gastrocnemius muscle, which crosses the knee and ankle joint, limiting the available ankle dorsiflexion ROM when the knee is extended. Different knee joint positions used in the previous study may have caused different ankle dorsiflexion ROM to be recorded, leading to variation in results (wide range of intratester and intertester ICC reported). In contrast, another study stated that lack of standardisation was not a significant factor for difference amongst the testers (intertester) during measurements of passive ankle plantarflexion ROM using the UG.[258] However, the opposite was reported for ankle dorsiflexion ROM with the UG (ICC increased by 0.09 when using different positions and decreased by 0.10 when using the same position) but still rated as weak intertester reliability (Table 4-3).[258] Furthermore, it was stated that involving two testers in the measurement procedure may increase the reliability of the measurements amongst subjects with CP as one tester stabilises the limb and the second tester takes the measurements.[247]

Generally, it was found from the studies included in this review that intratester reliability was higher than intertester reliability (Tables 4-1 to 4-3).[251, 257-261, 265, 269, 271] One study suggested that averaging two measurements each session increases the reliability of the measurements obtained,[261] agreeing with the findings of Low.[291] However, a study by Rothstein et al.[257] found that no greater reliability is obtained when the mean of measurements is used, suggesting that reliability can be achieved by taking a single measurement in clinical settings. In addition, a posterior analysis of the results by Elveru et al.[258] suggested that no increase in reliability was achieved when using the mean of two measurements, agreeing with the finding of Boone et al.[292] The

results of Kilgour et al.[248] showed no increase in reliability when averaging two measurements but stated that "taking two duplicate measures in clinical practice could help therapists to identify measurements within on session that might need to be repeated".[(p 399)]

The ability to make a direct comparison between the studies was compromised due to the differences in methodology adopted in the studies, such as the level of experience of the testers, number of sessions and time between the sessions.

Most studies included testers who were physical therapists with experience levels ranging from 1[257, 265] to 30[248] years. McWhirk and Glanzman[247] included two therapists with different levels of experience (1 and >10 years' experience) to investigate the intertester reliability of the UG in subjects with CP. They found good to excellent reliability for all the motions measured, excluding hip extension ROM (ICC=0.58) (Tables 4-1 to 4-3). Elveru et al.[258] performed a posterior analysis of the results to investigate the effect of experience on reliability of the UG and reported an increase in intratester ICC from 0.90 to 0.91 for ankle dorsiflexion and from 0.86 to 0.92 for ankle plantarflexion, and an increase in intertester ICC from 0.50 to 0.54 for ankle dorsiflexion and a decrease from 0.72 to 0.70 for ankle plantarflexion when more experienced testers took the measurements. Although an increase of ICC was reported, this increase did not affect the overall rating of the ICC values. Bennell et al.[281] reported excellent intratester reliability for both testers (second year physiotherapist student and a qualified physiotherapist with 9 years' experience) for ankle dorsiflexion measurements using inclinometer. Another study concluded that a novice tester (fourth

year exercise science students) with no previous experience can obtain reliable ankle dorsiflexion measurements using the UG and inclinometer.[278] Another study reported lower intratester ICC values for fourth year undergraduate podiatry students in comparison to experienced podiatrists (3 to 20 years' experience) for measurements of active ankle dorsiflexion using the inclinometer.[272] However, limited information about the tester experience was provided. Remaining studies did not provide additional evidence to show the effect of the testers' experience on the reliability obtained.

The number of testing sessions and the period between each session varied across the studies. Most studies used a test-retest design to calculate intratester reliability of measurements taken by different testers on the same day.[246, 247, 249, 255, 257-264, 267, 271] Kilgour et al.[248] investigated intratester reliability of the UG within and between sessions (one week apart) for passive ROM of hip, knee and ankle joint amongst subjects with CP and healthy subjects. In this study, all intra-sessional ICC values were found to be higher than inter-sessional ICC values (Tables 4-1 to 4-3). Wakefield et al.[268] also reported weak intratester inter-sessional (between sessions) ICC values for the UG for measurement of hip extension amongst healthy subjects (Table 4-2). In contrast, Mutlu et al.[266] and Herrero et al.[265] reported high intratester intersessional ICC values for all the motions measured using the UG and inclinometer amongst subjects with CP (Tables 4-1 & 4-3). Pandya et al.[251] reported excellent intratester intersessional ICC values for all measurements obtained using the UG amongst subjects with Duchenne muscular dystrophy, and Peeler and Anderson[270] and Thomas and Rome[273] reported similar results amongst healthy subjects using the UG (Tables 4-1 to 4-3). Good intratester intersessional ICC value was reported in another study

for measurement on ankle dorsiflexion using the UG.[274] Bennell at al.[281] and Munteanu et al.[272] also reported good to excellent intratester intersessional ICC values for ankle dorsiflexion using the inclinometer.

It is important to consider measurement reliability in the clinical context. It is reported that an error of ±5° in measurement may be clinically acceptable.[211] Hence, clinicians should be cautions when interpreting results of reliability studies and must select studies appropriate to pathology. Although Mutlu et al.[266] reported good to excellent intertester reliability of the UG, a variation of 0-28º was found in intertester measurements. In addition, another study reported a variation of 15-20º in the measurements of the UG between sessions.[248] The clinical effect of such findings must be considered, especially when using measurements to determine treatment effect.

A limited number of studies were found which compared between the UG, EG and inclinometer. Konor et al.[278] reported higher intratester ICC values for inclinometer (ICC=0.96-0.97) in comparison to the UG (ICC=0.85-0.96) for ankle dorsiflexion measurements. On the other hand, Clapis et al.[267] reported a lower intertester ICC value for inclinometer (ICC=0.89) in comparison to the UG (ICC=0.92) for measurement of hip extension amongst healthy subjects.[267] However, the differences between the ICC values reported for the UG and inclinometer by Konor et al.[278] and Clapis et al.[267] are not large (ICC values reported for both devices are excellent). Clapis et al.[267] proposed that the low ICC values obtained using the inclinometer may be due to the limited amount of training. Additionally, it was found that inclinometer and the UG can be used interchangeably for measurement of hip extension amongst healthy subjects.[267]

Ellison et al.[280] concluded that inclinometers are easier to use and more reliable compared to UGs during measurement of passive hip rotation amongst healthy subjects and subjects with low back pain.[280] However, ICC values for the UG amongst healthy subjects and subjects with low back pain and ICC values for inclinometer amongst healthy subjects were not reported. A study investigated the reliability of active hip, knee and ankle ROM measurements using the UG and EG amongst healthy subjects.[277] ICC values for the UG were found to be higher in comparison to the ICC values for the EG, except for hip external rotation. No study was found which compared reliability of inclinometer with EG or reliability of the UG with EG and inclinometer.

This review aimed to investigate the intratester and intertester reliability of the UG, EG and inclinometer and to examine how different factors can influence the reliability (Table 4-4). Twenty-nine studies were included which investigated the reliability of measuring hip, knee and ankle joint ROM. This literature review highlights variation in the methodology employed, which reduced the ability to compare studies directly as the number of testers, experience level, number of sessions, time between the sessions and subject position varied across the studies. Most studies indicated that the reliability was best when used to measure ROM in healthy subjects and that reliability may be reduced in the presence of different pathologies. Passive ROM is the motion mostly measured in the clinical environment, hence a larger number of studies have examined the reliability of measurement of passive motion rather than active motion. The limited number of studies measuring active motion compromised the ability to make comparisons with measurement of passive ROM. It was stated that the low

intertester reliability could be explained due in part to the difference in the force applied by therapists during assessments of passive motion, causing different angles to be obtained during each session. Generally, it was found that intratester reliability was higher than intertester reliability. Reliability varied from weak to excellent across the hip, knee and ankle joints due to the different joint characteristics and ease of identifying bony landmarks. It has been reported that even complex motions can be measured reliably when a strict standard position is applied. Standardisation of the measurement procedure and prior training were found to increase measurement reliability, and one study suggested that involvement of more than one tester in the measurement procedure may have a beneficial effect on reliability. A limited number of studies were found which compared the different designs of the goniometers, hence affecting the conclusions. No study was found which compared the 3 designs of goniometers. Further research is required to investigate and compare the reliability of the UG, EG and inclinometer and the possibility of using protocols and technology to increase reliability when measuring joint ROM.

4.2 *PnO Clinical Movement Data*

The previous literature review examining the reliability of the UG, EG and inclinometer demonstrated considerable variation in results for joint ROM measurements. Inconsistent agreement was observed between different studies, which restricted the ability to make a direct comparison due to the lack of similarity in the methodology implemented in those studies. The number of testers, experience level, number of sessions, time between the sessions, and subject position varied across the studies found. The review also highlighted the gap in current research about the reliability of the UG and the requirement for a more reliable measuring tool. Currently, 3D motion analysis systems such as Vicon have been shown to produce accurate and reliable measurement.[293] Additionally, these systems have been employed in several studies as a benchmark to assess the feasibility of other potential systems/devices.[271, 293-296] 3D gait analysis systems are often financially and technically inaccessible in clinical environments especially where space is a challenge. A certain level of expertise is required to operate these systems. Additionally, the outcomes produced by these systems may be hard for clinicians to understand.[293] Hence, the use of alternative systems such as the 2D video analysis systems might be clinically useful.

In this thesis, the *PnO CMD* (Previously known as *Siliconcoach*^{TM}) was chosen as an example of 2D analysis systems (*Prosthetic and Orthotic Data Solutions*, Florida, USA).[297] 2D analysis systems such as the *PnO CMD* was considered because these systems are applicable to capture motion and analyse gait. Additionally, the use of the *PnO CMD* in clinical practice may aid in documentation of the fitting and alignment process and reduction of trial and error. It can also promote increased communication

between allied health professionals. The *PnO CMD* system is affordable, practical and easy to use in clinical environments. A single video camera is required to capture sagittal plane ROM. Following that, the video can be uploaded into the *PnO CMD* software for joint ROM measurement. Hence, the use of this software can be applicable in clinical environments where space could be a challenge. The option to play back the video frame by frame aids in identifying the joint end ROM. Additionally, the web-based application of *PnO CMD* enhances the use of this system and reduces the requirements for hardware, as this system can be used for example on tablets.

The reliability of the measurement of ankle, knee and hip angles at initial contact, midstance and terminal stance phases of gait using *PnO CMD* has previously been investigated.[298] This study also investigated whether the use of predefined anatomical markers improved measurement reliability, with the authors concluding that using predefined anatomical markers increased reliability. The intratester reliability varied across gait and was best reported in terminal stance (ICC=1.00).[298] Additionally, this study concluded a high intertester ICC value (ICC=0.86) for measurement of knee angle at initial contact. A further study[286] aimed to determine the reliability of *PnO CMD* in assessing dynamic and static ROM of the knee joint. Acceptable ICC values (ICC>0.6) for dynamic and static motion were found.[286] Additionally, a study[299] investigated the intratester and intertester reliability of measuring resisted isometric knee flexion during single leg squats using *PnO CMD*. Intertester reliability was found to be acceptable for all three testers (ICC>0.6).[299] It was concluded that intratester reliability was found to be better than intertester reliability.

The overall aim of the following studies is to test the feasibility of 2D analysis system (*PnO CMD*) in clinical practice for passive joint ROM measurement during physical assessment. The findings of these studies may help clinicians to use advanced technologies such as 2D analysis systems in clinics during physical assessment. Additionally, it may help to introduce a system which is cost effective and more practical in comparison to Vicon. Initially, the reliability of *PnO CMD* will be investigated amongst healthy subjects in comparison to the UG in measurement of passive joint ROM. Additionally, measurements taken using a standard developed measurement procedure with/without markers will be compared to establish the most reliable measurement procedure. Following that, the accuracy of *PnO CMD* will be researched in comparison to Vicon (benchmark) amongst healthy subjects. It is mandatory to examine the reliability and accuracy of *PnO CMD* amongst healthy subjects, as to the best of the author's knowledge there is no published work investigating this. In addition, this step is carried out to establish the most reliable and accurate measurement procedure that can be taken forward and tested amongst subjects with diplegic CP.

4.2.1 The reliability of *PnO Clinical Movement Data* and universal goniometer in the measurement of hip, knee and ankle motion amongst healthy subjects

4.2.1.1 Aim

This pilot study aims to investigate the intratester and intertester reliability of *PnO CMD* compared to the UG in measuring passive ROM of the lower limb joints during physical assessment amongst healthy subjects. Additionally, this study aims to establish the effect of markers on measurement reliability. This will help to determine the most reliable measurement procedure.

4.2.1.2 Methods

Participants and testers

After gaining appropriate ethical approval, recruitment posters for testers and participants were displayed within the Department of Biomedical Engineering of the University of Strathclyde (DEC.BioMed.2013.21). Individuals who showed interest were asked to contact the research team and were provided with additional information and the participant information sheet. A period of three days was given to each individual to make a decision on their participation. Following that, informed consent was obtained from the participants and testers at the introductory session. Inclusion criteria were as follows: Participants were adult (age>18) and did not suffer from any musculoskeletal or neurological conditions, or from any condition resulting in any lower limb sensory deficit. Participants were excluded if they were unable to attend

the scheduled measuring and recording sessions or if there was a change in physical status or injury during the trial period.

Testers were required to be a qualified allied health professional or a final year prosthetics and orthotics student with current experience in measurement of joint ROM. Final year prosthetics and orthotics students use the UG throughout the course and have sufficient training in utilising it. Additionally, only students who have completed their clinical placement were included. Testers were excluded if they were unable to attend the scheduled measuring sessions or were unable to complete the video analysis within the given time frame or had no experience in measuring joint ROM.

Study design

Sagittal plane ROM of the hip, knee and ankle joint of the dominant limb during physical assessment was measured with both tools, with and without markers. Markers were applied on the following bony landmarks: shoulder, greater trochanter, lateral femoral condyle of the knee, lateral malleolus and fifth metatarsal head. Markers consisted of bright coloured adhesive *Velcro*™ cut into circular shapes (25 mm) and placed by the same researcher to reduce the variability.

Participants

An introductory session was arranged where explanation about the trial was given to the participants. For practical reasons, the participants were divided into two groups. Each group attended two half-day sessions: one with markers and one without markers,

with approximately a one-week gap between the sessions. Participants were provided with *Lycra*™ suits to wear for all the sessions to limit any movement/obstruction which may be caused by loose clothing. Additionally, each participant was given a time slot to attend for video recording within a separate video recording session. In the video recording session, which lasted for approximately 30 minutes, the camera (Sony HD video camera with 3.1 mega pixels) was positioned using a tripod perpendicular to the bed where the participant was lying down at an appropriate distance to capture the image of the participant from the shoulder to toe. The researcher moved each joint individually into maximum flexion and extension while a video of the motion was captured. Additionally, care was taken to ensure that the joint measured was perpendicular to the video camera to minimise the parallax type error. The same procedure was repeated using markers.

Testers

An introductory session was arranged where a PowerPoint presentation explaining the measuring method with *PnO CMD* and the UG was given by the researcher in order to standardise the measuring methods. A standard measurement procedure was developed and implemented for this study. Additional information was provided in a measuring instruction manual (Appendix B). For the purpose of this study, each tester was asked to record the following measurements on the dominant limb using both devices: maximum hip flexion, maximum hip extension, maximum knee flexion, maximum knee extension, maximum ankle plantarflexion and maximum ankle dorsiflexion. The testers were asked to attend four half-day sessions: two marker sessions and two no marker sessions with approximately a one-week gap between each

session. In each session, each tester measured the hip, knee and ankle joint ROM of four participants (Figure 4-1).

Intratester reliability of PnO Clinical Movement Data

Each participant's joint ROM of the dominant limb was video recorded with and without markers. Following that, each tester was instructed to evaluate each video three times in a time frame of three weeks with approximately a one-week gap between each evaluation (Figure 4-1). Each session lasted for approximately two hours. The order of the joints (with and without markers) was randomised between each evaluation. The testers were guided to pause the video at the stage when the researcher holds the joint at the end of the range for 3 seconds and to take the measurement at that position. The order of evaluation of the joints and video sequences was randomised. Special assessment sheets were used to record the ROM measurements using the identification codes provided to each tester and participant for blind analysis.

Intratester reliability of the universal goniometer

Prior to each session, the inclusion and exclusion criteria were checked for each participant to ensure that the participant's status had not changed. Each tester measured the ROM of each participant (hip, knee and ankle) 3 times in each session. Each session lasted for approximately 3 hours. The order of measuring participants was randomised and the order of each session (marker/no marker) was also randomised (Figure 4-1). Each tester had approximately 10 minutes to measure maximum sagittal plane motion of each participant's hip, knee and ankle joints. Special assessment sheets were used

to record the ROM measurements using the identification codes provided for each tester and participant for blind analysis.

Intertester PnO Clinical Movement Data and the universal goniometer

The mean of the three repetitions for each device was calculated with and without markers and compared for each joint between the two devices. It should be noted that intertester values were only able to be calculated if intratester values across all the testers were above the satisfactory level.

Statistical analysis

To achieve power of 80% at the 5% level of significance, 3 testers (final year prosthetics and orthotics students) and 8 participants (healthy subjects) were included in the study. ICC model (2, 1) was used after initial summary statistics was produced. This ICC reliability tool was used to assess and compare the reliability of the *PnO CMD* and the UG along with Bland & Altman plots and an appropriate paired test assessing the significance of actual differences. ICC values above 0.60 were considered to be satisfactory for research purposes.[300]

4.2.1.3 Results

Participants and testers

In this study, a total of eight healthy subjects were recruited. Additionally, three final year prosthetics and orthotics students were recruited as testers.

Intratester reliability

With markers

PnO Clinical Movement Data

The lowest ICC values found were for ankle dorsiflexion for all the testers (ICC=0.63, 0.65 & 0.78). However, the values were above satisfactory limits (ICC>0.6). The highest ICC values for all the testers were found for knee flexion measurements (ICC=0.98, 0.98 & 0.99) (Table 4-5). All ICC values were above satisfactory limits (Figures 4-2 to 4-4). All Bland & Altman plots illustrated small dispersion and equal distribution of the points above and below zero confirming the high ICC results found.

Universal goniometer

Some ICC values were found to be below the satisfactory limits (ICC<0.06) including hip flexion for tester 1 (ICC=0.46), hip extension for tester 1 & 2 (ICC=0.39 & 0.50), knee flexion for tester 1 & 2 (ICC=0.56 & 0.43), knee extension for tester 1 & 3 (ICC=0.54 & 0.41) and ankle dorsiflexion for tester 2 & 3 (ICC=0.40 & 0.56). The lowest ICC value was found for hip extension for one tester (ICC=0.39), while the highest ICC value was found for ankle plantarflexion and dorsiflexion for one tester (ICC=0.94) (Table 4-5) (Figures 4-2 to 4-4). Additionally, the Bland & Altman plots validated the ICC results achieved.

Without markers

PnO Clinical Movement Data

ICC values for all the joints measured were found to vary from 0.24 to 0.98. ICC values for ankle dorsiflexion for all the testers were found to be lower in comparison to the

other motions measured and below the satisfactory limits (ICC=0.24, 0.32 & 0.58) (ICC<0.6). Additionally, the ICC value for one tester for hip extension was found to be lower than the satisfactory limits (ICC=0.53). The highest ICC values for all the testers were found for hip flexion measurements as indicated in Table 4-5 (ICC=0.92, 0.93 & 0.98). All the Bland & Altman plots verified the ICC results found.

Universal goniometer

ICC values across all the joints measured were found to vary from 0.39 to 0.93. The lowest ICC value was found for ankle dorsiflexion for one tester (ICC=0.34), while the highest was found for ankle plantarflexion for another tester (ICC=0.93) (Table 4-5) (Figures 4-2 to 4-4). Furthermore, the Bland & Altman plots illustrated widespread scattering of the points confirming the low ICC results achieved.

Intertester reliability

With markers

PnO Clinical Movement Data

ICC values for all the joints measured ranged from 0.94 to 0.99 and were above the satisfactory level (ICC>0.60) (Table 4-5) (Figure 4-5). Additionally, all Bland & Altman plots showed random scattered points equally distributed above and below zero; hence, validating the high ICC values obtained.

Universal goniometer

Only ICC value for ankle plantarflexion could be calculated, and this was found to be below the satisfactory level (ICC=0.39) (Table 4-5) (Figure 4-5). The Bland & Altman plot illustrated large dispersion confirming the low ICC result found.

Without markers

PnO Clinical Movement Data

ICC values for all the joints measured except ankle dorsiflexion and hip extension were found to range from 0.91 to 0.97. The ICC value for hip extension for one of the testers was below the satisfactory level (ICC=0.53), which prevented the calculation for intertester reliability. All the ICC values across all the testers for ankle dorsiflexion were below the satisfactory levels (ICC<0.6) (ICC=0.24, 0.32 & 0.58); hence, intertester reliability was not concluded for this joint (Table 4-5) (Figure 4-5). All Bland & Altman plots confirmed the high ICC values achieved. The plots illustrated small dispersion and equal distribution of the points above and below zero.

Universal goniometer

Only intratester ICC values for ankle plantarflexion across all the testers were above the satisfactory level, but the resulting intertester reliability was lower than the satisfactory level (ICC=0.47) (Table 4-5) (Figure 4-5). This low ICC value was confirmed by the Bland & Altman plot.

4.2.1.4 Discussion

All intratester and intertester ICC values obtained using *PnO CMD* with markers for all testers were found to be above the satisfactory limit (ICC>0.60) with small variations in values, which demonstrates the reliability of this tool using markers and the standard measurement procedure developed (Table 4-5) (Appendix B). Most intratester and intertester ICC values for *PnO CMD* with no markers were found to be above the satisfactory limit (ICC>0.60) for all the testers, except for one tester's value for hip extension measurements (ICC=0.53) and ankle dorsiflexion measurements for all testers (ICC=0.24, 0.32 & 0.58) (Table 4-5) (Figures 4-2 to 4-5). Furthermore, it was observed in this study that all intratester ICC values for the UG (with/without markers) across all joints ranged considerably and, in some cases, were below the satisfactory limits (ICC<0.60) (Table 4-5). Due to the variation found in the ICC values for the UG, the effect of the markers on reliability of the UG was inconsistent (Table 4-5). Additionally, only two values of intertester reliability for the UG could be calculated including ankle plantarflexion with/without markers (ICC=0.39 & 0.47) and, and these were found to be lower than the intratester reliability and below the satisfactory limits (ICC<0.60). This demonstrates the unreliability of using this tool in comparison to *PnO CMD* (Table 4-5).

Intratester ICC values for ankle plantarflexion were the only values found to be above satisfactory limits (ICC>0.6) for both tools across all the testers. The resulting intertester reliability for ankle plantarflexion for *PnO CMD* with/without markers were found to be higher than intratester reliability (Table 4-5). On the other hand, the intertester reliability for ankle plantarflexion for the UG with/without markers were

found to be lower than intratester reliability (Table 4-5). *PnO CMD* (with/without marker) ICC values for ankle dorsiflexion for all testers were found to be lower in comparison to the other motions measured (Table 4-5).

To the best of the author's knowledge, only ICC values of the UG can be compared to the results previously found in the literature, as no study was found investigating the reliability of *PnO CMD* for passive motion of hip, knee and ankle joint. One study, Kilgour et al.,[248] was found, which investigated the intratester reliability for the measurement of hip, knee and ankle motion using the UG with markers amongst healthy subjects. ICC values reported for hip extension, knee extension and ankle dorsiflexion using a similar testing position were all above the satisfactory limits, which does not agree with the findings of this study (Table 4-5). Peters et al.[269] used a similar testing position as in the current study to investigate intratester reliability of the UG without markers for measurements of knee joint motion amongst healthy subjects. Again, reported ICC values were higher than the values reported in this study. The difference in experience of the testers between the study by Peters et al.[269], the study by Kilgour et al.[248] and the current study may explain the variance found in the ICC values reported. Qualified clinicians were included in the study by Kilgour et al.[248] and Peters et al.[269], while final year students were included in this study.

Rothstein et al.[257] and Youdas et al.[261] reported an increase in ICC values for the UG when subject's position was standardised. As recommended, subject's position and a standard measurement procedure was applied in this study with both tools (Appendix B). Another source of error stated in the literature is the variance found between the

clinicians in the identification of bony landmarks.[270] Markers proved to be useful with *PnO CMD* as the increased reliability. However, the effect of markers on the UG reliability was not clear. It was noted that the ICC values for the UG (with/without markers) ranged from weak to excellent with no pattern observed, making it hard to draw any conclusions. A possible explanation is the unequal manual force applied while measuring. Testers were instructed in each UG session to move each joint to maximum ROM, which may have resulted in different force applied between the testers and sessions. On the other hand, this variance was not present with *PnO CMD* as all the joints were moved by the same researcher and the same video captured on a single occasion was used for all the evaluation.

This study represents an initial step in using advanced technology (*PnO CMD)* in clinical practice to measure passive joint ROM. High reliability has been illustrated for sagittal plane passive ROM of hip, knee and ankle joints, and this increases with the use of markers. Markers can be created from cheap available material, such as adhesive *Velcro*^{TM}, and used effectively. Additionally, the used measurement procedure can be easily applied. However, the intratester reliability was calculated before the intertester reliability and the data used to calculate the intratester reliability was averaged and used for intertester reliability. Theoretically, since the data has been re-used, Bonferroni correction factor may be considered appropriate. The significance of each intertester ICC values was <0.001 (with the exclusion of one value as shown in Table 4-5); hence, even with the Bonferroni correction factor the ICC values would still be significant. Furthermore, these results can only be generalised amongst healthy subjects, and further research is recommended amongst subjects with different

pathologies such as cerebral palsy. The presence of spasticity and rotational deformity increases the challenge in obtaining reliable measurements.[240, 287, 288] Additionally, as this study only involved final year prosthetics and orthotics students, it would be beneficial to investigate if experience or professional background have an effect on reliability of *PnO CMD* and the UG as this has not been reported sufficiently in the literature.[33] Test-retest design (intra-sessional intratester) was applied in this study where all the repeated measurements were taken in the same session. Kilgour et al.[248] and Wakefield et al.[268] reported lower inter-sessional intratester reliability in comparison to intra-sessional intratester reliability for the UG of hip, knee and ankle joint. Thus, investigating the intersessional intratester for *PnO CMD* is recommended as clinicians measure on different occasions. This study only investigated the reliability of both tools for a single testing position used for hip, knee and ankle motion. Further investigation of reliability of *PnO CMD* is required using different test positions for joint ROM.

In conclusion, *PnO CMD* was found to be more reliable than the UG in measuring passive sagittal ROM of the lower limb joints' motion amongst healthy subjects. In addition, it was found that using markers increased the intratester and intertester reliability of *PnO CMD*. The present work opens possibilities for using new technology in joint ROM measurements to achieve more reliable measurements.

4.2.1.5 Error associated with the marker displacement using the *PnO Clinical Movement Data*

Introduction

In the previous study, it was established that the use of markers increased the measurement reliability of the *PnO CMD*. However, the use of markers can lead to measurement error due to the inaccuracy and variance in marker placement. Errors in marker placement can occur for several reasons. The first and most obvious reason is the inability to palpate the anatomical landmarks and place the marker precisely on these landmarks. Most of the anatomical landmarks are covered by skin, fat tissue, tendons and muscles which compromises the ability to accurately palpate these landmarks. Another reason for marker placement error is the difference found between examiners in identifying the anatomical landmarks. The process of palpating anatomical landmarks is subjective to some degree as the examiner is left with the decision of identifying the exact location of the marker placement, especially on flat surfaces. Intertester variability and between-sessions variability in marker placement have been identified as major sources of error in several studies.[301-304] This sub-study aimed to establish the measurement error caused by the marker displacement using the *PnO CMD*.

Methods

For this research, a model demonstrating the knee joint was constructed (Figure 4-6). Published body segment parameters (Nordic)[305-307] with the average United Kingdom male height (177 cm)[308] were used to calculate the length of the femur and tibia. Each

marker (A, B and C) was displaced in four directions from the original position by 1 cm (Figure 4-6). The arm representing the femur was moved through the range while a video of the motion was captured using and HD video camera (Sony HD video camera with 3.1 mega pixels). Following that, the video was uploaded into the *PnO CMD* system. Then, the video was paused when points ABC formed the following six angles: 180°, 170°, 160°, 150°, 110° and 90°. A total of 125 combinations were measured for each angle. The researcher completed all the measurements.

To establish the error associated with the marker displacement using the *PnO CMD*, the *PnO CMD* measurements were compared to trigonometry (benchmark) measurements for each angle. The standard error of measurement (SEM) was calculated for the *PnO CMD* using the ICC value (ICC=0.99) previously obtained from the research investigating the reliability of the *PnO CMD* (Section 4.2.1). Additionally, Bland & Altman plots were constructed for each angle to assess the degree of agreement between the methods.[309] Since trigonometry is obtained using an equation, the ICC can be considered to be equal to 1; hence, the SEM=0.

Results & discussion

Table 4-6 illustrates the mean and SD for each angle measured using the *PnO CMD* and trigonometry. The SEM for the *PnO CMD* ranged from 0.196° to 0.294°. It could be noted that as the angle increased, the SEM increased, exhibiting an approximately linear relationship (Figure 4-7).

Bland & Altman plots for 90° and 110° illustrated random scattered points equally distributed above and below zero (Figures 4-8 & 4-9). This indicated that no consistent bias of one approach versus the other existed. As the angle increased, the plots moved towards a linear relationship (Figures 4-10 to 4-13). Bland & Altman plots for 170° and 180° showed that there is clearly a linear tendency in the *PnO CMD*-trigonometry score. The differences between the methods moved from negative to positive as the size of angle created by the pattern of markers increased (Figures 4-12 & 4-13). This indicated that for small angles the trigonometry measurement of the angle is larger than the measurement obtained using the *PnO CMD*, but for bigger angles the opposite is found. In all the plots, most of the points were within the +/-2° (Figures 4-8 to 4-13).

This sub-study was carried out to measure the possible error caused by marker displacement using the *PnO CMD*. It has been reported that an error of ±5° in measurement may be clinically acceptable.[211] The measurement error caused by marker displacement using the *PnO CMD*, which ranged from 0.196° to 0.294°, is very small and is clinically acceptable (Table 4-6). One of the limitations of this sub-study is that the model used was a simple mechanical model. The use of a simple mechanical model eliminates the possible causes of variation in marker displacement commonly presented amongst subjects. These possible reasons may include skin movement, non-standard measurement procedure, intertester difference, between-sessions difference, presence of bony deformities and subject's performance. Another limitation of this sub-study is that the sample used is not independent. However, this sample may be possibly considered as been measured on different subjects. This is because

displacement of markers can be caused by the presence of long bone torsions. Furthermore, it should be noted that the SEM calculated for the *PnO CMD* represents a combination of the error caused by the marker displacement and the error associated with the use of the *PnO CMD*.

Overall, the measurement error caused by marker displacement using *the PnO CMD* is small and clinically acceptable especially when compared to current measurement standards such as the UG. The SEM increased as the angle increased, indicating that the error is higher with the larger angles.

4.2.2 Accuracy of *PnO Clinical Movement Data* to measure dynamic hip, knee and ankle joint range of motion during gait amongst healthy subjects

4.2.2.1 Aim

This research aims to investigate the accuracy of *PnO CMD* in comparison to Vicon (benchmark) to measure dynamic motion of the lower limb joints amongst healthy subjects during the gait.

4.2.2.2 Methods

Participants

Subjects were recruited from the Department of Biomedical Engineering of the University of Strathclyde. Ethical approval for this study was granted from the Biomedical Engineering Departmental Ethics Committee, University of Strathclyde (DEC.BioMed.2015.60). Inclusion criteria were as follows: adult participants aged over 18 with body mass index≤28 who did not suffer from any musculoskeletal or neurological conditions or from any condition resulting in any lower limb sensory deficit. Participants were excluded if they were unable to attend the scheduled session or if there was a change in their physical status or injury during the trial period.

Study design

Lab setup

The start and the stop/finish points were marked on the floor using coloured tape to form a walking path. The force platforms were located midway between the two points. Force platforms were used to accurately identify initial contact and toe-off

using the Vicon system. One video camera (Sony HD video camera with 3.1 mega pixels) was used to capture the sagittal plane view of the dominant limb using the *PnO CMD* system. The camera was positioned on a tripod perpendicular to the midpoint of the force platforms. The height of the video camera was adjusted to capture the participant from shoulder to toe. Prior to each measurement session, both systems were calibrated following the appropriate guidelines. Both systems were adjusted to capture 50 frames per second.

Gait capturing session

Participants were asked to wear close fitting black leggings and a close fitting black top for the capturing session. At the beginning of each session, a brief introduction was given to the participant and consent was obtained. The following measurements were recorded: weight, height, knee width, ankle width and leg length. These measurements are mandatory for static calibration of the Vicon system. Following that, two sets of markers were applied using hypoallregenic double-sided tape to mark identified bony landmarks. The first set of markers were the Vicon system markers. These markers are spherical shaped markers coated with a highly reflective material which can be tracked by the infrared light emitting cameras of the Vicon system. Vicon markers were applied as indicated by the Plug-In-Gait guidelines on the following bony landmarks of both sides: anterior superior iliac spine, posterior superior iliac spine, lateral femoral epicondyle, lateral malleolus, thigh, shank, second metatarsal head and calcaneus. The second set of markers were applied for *PnO CMD*. These markers were constructed from bright coloured adhesive *Velcro*™ cut into round-shaped markers (25 mm). Markers used for *PnO CMD* were applied on the following bony landmarks of the

dominant side: pelvis, greater trochanter, lateral femoral epicondyle, lateral malleolus and fifth metatarsal head.

After participant preparation, static calibration of the Vicon system was carried out. The participant was asked to stand in the anatomical position, and appropriate system guidelines were followed. Then, the participant was asked to walk from the start point to the stop/finish point (points marked on the floor) at their own speed 3 times while both systems captured gait at the same time.

Pelvis marker

The position of this marker was determined following a primary test. Several locations were tested to determine the position that would yield the most accurate results. The location of the pelvis marker was located as follows: the anterior superior iliac spine (point A) and posterior superior iliac spine were first identified (point B). The midpoint of the projection of the line between A and B on the skin in sagittal plane was then marked (point C). Following that, the midway between points A and C was marked and the pelvis marker was applied (point D).

The effect of changing the location of the marker (from the shoulder to pelvis) on the reliability of the hip ROM measurement using *PnO CMD* was investigated. The same methodology stated in Section 4.2.1 for reliability of *PnO CMD* with markers was followed. All intratester and intertester values were high, as stated in Table 4-7. However, the intratester reliability was calculated before the intertester reliability and the data used to calculate the intratester reliability was averaged and used for intertester

reliability. Theoretically, since the data has been re-used, Bonferroni correction factor may be considered appropriate. The significance of each intertester ICC values was <0.001 (Table 4-7); hence, even with the Bonferroni correction factor the ICC values would still be significant.

Analysis

To ensure a clear identification of the initial contact, gait with clean foot strike on the force platforms and a clear sagittal plane camera view was analysed. This method enhances a more repeatable and accurate analysis. If more than one gait cycle with clean foot strike was available, gait analysed was chosen randomly. The researcher measured the dominant limb, hip, knee and ankle joints ROM using both systems at 5 predetermined points of gait. These points were selected because they can be repeatedly identified using both systems. The points used were: point 1: the instant the heel of the dominant foot touches the ground, point 2: prior to toe-off of the other foot, point 3: the swinging foot (other foot) is in line with stance foot (dominant foot), point 4: the hip of the dominant side is at maximum extension/other foot initial contact, point 5: prior to toe-off of the dominant foot.

Statistical analysis

To achieve 80% power, 12 healthy subjects were involved in this study. The hip, knee and ankle sagittal plane angles were compared for accuracy using the ICC model (2, 1) after initial summary statistics was produced. This ICC reliability tool was used to assess the accuracy of the *PnO CMD* system against the Vicon system along with Bland & Altman plots and an appropriate paired test assessing the significance of

actual differences. ICC values above 0.60 were considered to be satisfactory for research purposes.[300]

4.2.2.3 Results

All ICC values were high (ICC>0.60), ranging from 0.90 to 0.98 (Table 4-8). In addition, all results were highly significant ($p<0.05$) with narrow to moderate CI range (Table 4-8). The lowest ICC value was found for the hip joint at point 2 (ICC=0.90) and for ankle joint at point 1 of gait (ICC=0.90), whereas the highest ICC value was found for the knee joint at points 2 & 4 of gait (ICC=0.98) (Table 4-8). All Bland & Altman plots illustrated small dispersion, confirming the high ICC results achieved. A total of 15 Bland & Altman plots were plotted. Five plots showed that the measurements taken using *PnO CMD* were higher than the measurements taken using Vicon, while six plots showed the opposite. Four plots illustrated equal distribution of the values above and below zero.

4.2.2.4 Discussion

All ICC values were found to be high and above the satisfactory level (ICC>0.6). This demonstrated the accuracy of *PnO CMD* in measurements of hip, knee and ankle joint sagittal plane ROM at 5 predefined points of gait (Table 4-8). The highest ICC value (ICC=0.98) with the narrowest CI range (CI=0.94-0.99) was found for the knee joint at point 4 of gait (Table 4-8). The lowest ICC value (ICC=0.90) with the widest CI range (CI=0.69-0.97) was reported for the hip joint at point 2 of gait. Overall, the ICC values for the knee joint were the highest with the narrowest CI range (Table 4-8). Similarly, the intratester and intertester ICC values for the knee joint were found to be

the highest amongst the joints measured (Table 4-8). However, Evans[310] compared the accuracy of *PnO CMD* with Vicon in measurement of hip, knee and ankle ROM during a rugby union place kick. The study reported that the least difference between the two systems was found for the hip joint, which does not agree with the findings for the current study.

The results of the current study can only be generalised amongst healthy subjects with BMI≤28, and further research is recommended amongst subjects with pathology and subjects with higher BMI. As discussed previously, presence of spasticity and rotational deformity may affect the accuracy of the measurements taken. The 2D nature of *PnO CMD* does not consider any rotational movement. In addition, further research investigating the accuracy during physical assessment and different activities such as ascending and descending stairs may be beneficial. Due to the high reliability of *PnO CMD* in measurements of joint ROM with markers, only one gait with clean foot strike per participant was chosen randomly and analysed. Four studies found no increase in the reliability when mean of repeated measurements was calculated.[248, 257, 258, 292] However, two previous studies reported an increase in reliability when mean of repeated measurements was taken.[261, 291] Repeating this study and averaging measurements taken from more than one gait might narrow the CI range achieved.

For the Vicon system, force platforms data was used to identify initial contact and toe-off. On the other hand, these points were visually estimated using the *PnO CMD* system, which may have caused differences between the measurements achieved. Another source of difference might be due to the variation in the measurement method

between the two systems. The Vicon system measures the angles from the centre of the joint, while the angles are measured from the bony landmarks using *PnO CMD*. Presence of sunlight or other bright light is mandatory to capture a clear picture quality with *PnO CMD*. This was compromised in this study due to the absence of sunlight in gait lab, which affected the ability to clearly identify the centre of the markers.

The current work establishes the accuracy of *PnO CMD* for joint ROM measurements amongst healthy subjects in comparison to the Vicon system. Hip, knee and ankle joint sagittal plane ROM was measured at 5 predefined points of gait using both systems. Joint ROM measurements with a higher degree of accuracy can be achieved using *PnO CMD*. This work helps in introducing this practical and accurate advanced system for use in the clinical practice for joint ROM during physical assessment.

The above two studies were carried out to investigate the reliability and the accuracy of *PnO CMD* amongst healthy subjects. The initial study compared *PnO CMD* and the UG in measuring passive joint ROM. Additionally, the study compared two measurement procedures, measuring with markers and without markers. *PnO CMD* with markers was found to be the most reliable measurement tool and procedure. The second study examined the accuracy of *PnO CMD* with markers in comparison to Vicon (benchmark). This study confirmed that the best accuracy is achieved using *PnO CMD* with markers. The above two studies established that *PnO CMD* with markers is a reliable and a clinically accurate measurement tool and procedure; hence, this was taken forward and tested amongst subjects with diplegic CP.

4.2.3 The reliability of *PnO Clinical Movement Data* in the measurement of hip, knee and ankle motion amongst subjects with diplegic cerebral palsy

4.2.3.1 Aim

This pilot study aims to investigate the intratester and intertester reliability of *PnO CMD* with markers in measuring passive ROM of the lower limb joints during physical assessment amongst subjects with diplegic cerebral palsy.

4.2.3.2 Methods

Participants and testers

Appropriate ethical approvals were obtained from the West of Scotland Research Ethics Service, Clinical Research & Development Office of Greater Glasgow & Clyde Health Board and hospital managements. Recruitment posters and flyers were displayed in the common/waiting areas of NHS hospitals and schools. Additionally, clinicians informed and distributed flyers to participants who showed interest with contact information to contact the research team. Posters for recruiting testers were displayed in the department where participants worked and in the Department of Biomedical Engineering of the University of Strathclyde. Participant information sheets were sent out to participants and testers who contacted the research team showing interest. Participants and testers were given one week after receiving the participant information sheet to make their decision. This time allowed participants to familiarise themselves with all the given information and ask for any help required from family or friends to fully understand the steps included. Following that, the testers

were asked to come in for an introductory session which lasted for approximately 1 hour. In this session, a PowerPoint presentation explaining the measuring method with *PnO CMD* was given to standardise the measuring methods. In addition, each participant was invited to come in for an initial screening session, which lasted for approximately one hour. In the screening session, a physical assessment was carried out to determine if the participant was suitable to take part in the study (see inclusion criteria below). The first session was carried out following the initial screening session in order to minimise the number of sessions for each participant. Informed consent was obtained from the participants and testers at the initial session.

Inclusion criteria were as follows: Participants were subjects with diplegic CP who were aged≥7 years of age and did not have any lower limb sensory deficit. Participants were excluded if they were unable to attend the scheduled recording sessions or reported any change in physical status or injury during the trial period.

Testers were required to be qualified allied health professionals with current experience in measurement of joint ROM. Testers were excluded if they were unable to complete the video analysis within the given time frame or had no experience in measuring joint ROM.

Study design

Sagittal plane ROM of the hip, knee and ankle joints of the dominant limb during physical assessment was measured with markers. Markers were applied on the following bony landmarks: pelvis (as explained above in Section 4.2.2.2), greater

trochanter, lateral femoral condyle of the knee, lateral malleolus and fifth metatarsal head. Markers were placed by the same researcher to reduce the variability and consisted of bright coloured adhesive *Velcro*™ cut into circular shapes (25 mm). Testers were provided with instructions for the measurement procedure. Furthermore, a training session for the use of *PnO CMD* was arranged.

Participants

Each participant was given a time slot to attend a single video recording session. A brief introduction was given prior to the session where explanation about the trial was provided to the participants. The video recording session lasted for approximately 1 hour. Participants were provided with *Lycra*™ suits to wear for the session to limit any movement/obstruction which may be caused by loose clothing. The video camera (Sony HD video camera with 3.1 mega pixels) was positioned using a tripod perpendicular to the bed where the participant was lying down at an appropriate distance to capture the image of the participant from the shoulder to toe. The researcher moved each joint individually into maximum flexion and extension while a video of the motion was captured. Additionally, care was taken to ensure that the joint measured was perpendicular to the video camera to minimise the parallax type error.

Testers

An introductory session was arranged where a PowerPoint presentation explaining the measuring method with *PnO CMD* and the UG was given by the researcher in order to standardise the measuring methods. The same developed measurement procedure used in section (4.2.1) was applied (Appendix B). Additional information was

provided in a measuring instruction manual. For this study, each tester was asked to record the following measurements on the dominant limb using both devices: maximum hip flexion, maximum hip extension, maximum knee flexion, maximum knee extension, maximum ankle plantarflexion and maximum ankle dorsiflexion.

Intratester reliability

Each participant's joint ROM of the dominant limb was video recorded. Following that, each tester was instructed to evaluate each video three times in a time frame of three weeks with approximately a one-week gap between each evaluation. The testers were guided to pause the video at the stage when the researcher holds the joint at the end of the range for 3 seconds and to take the measurement at that position. The order of evaluation of the joints and video sequences was randomised.

Intertester reliability

The mean of the three repetitions was calculated for each joint. It should be noted that intertester values could only be calculated if intratester values across all the testers were above the satisfactory level.

Statistical analysis

ICC models were used to calculate the reliability of *PnO CMD*. To achieve power of 80% at the 5% level of significance, 3 testers (prosthetist or orthotist) and 8 participants (subjects with cerebral palsy) were included in this study. ICC model (2, 1) was used after initial summary statistics was produced. This ICC reliability tool was applied to assess and compare the reliability of the *PnO CMD* and the UG.

Additionally, Bland & Altman plots and an appropriate paired test was used to assess the significance of actual differences. ICC values above 0.60 were considered to be satisfactory for research purposes.[300]

4.2.3.3 Results

Intratester reliability

ICC values for all the joints measured were found to range from 0.91 to 1.00. All ICC values were found to be highly significant ($p<0.05$). ICC values for knee extension for all the testers were lower than other motions measured (ICC=0.94, 0.91 & 0.94). Additionally, the highest ICC values for all the testers were found to be for knee flexion measurements (ICC=1.00, 0.99 & 1.00), as indicated in Table 4-9. Additionally, all Bland & Altman showed random scattered points equally distributed above and below zero; hence, validating the high ICC values obtained.

Intertester reliability

ICC values across all the measured joints were found to be above the satisfactory level (ICC>0.6). The lowest ICC value was for knee extension (ICC=0.84), while the highest ICC value was recorded for knee flexion (ICC=1.00). CI range for knee extension was found to be wide (CI=0.47-0.98), which was confirmed by the Bland & Altman plot. The rest of the CI range was found to be narrow (Table 4-9).

4.2.3.5 Discussion

This study illustrated high reliability of *PnO CMD* for sagittal plane joint ROM measurements amongst subjects with diplegic CP. All intratester and intertester ICC

values were found to be above the satisfactory level and highly significant. The lowest intratester and intertester ICC values were shown for measurement of knee extension (ICC=0.94, 0.91 & 0.94), while the highest ICC values were reported for knee flexion (ICC=1.00, 0.99 & 1.00). The findings of the current study partially agree with the findings of the study amongst healthy subjects (Section 4.2.1). Intratester and intertester ICC values for hip flexion, hip extension, and knee flexion turned out to be approximately equal between healthy subjects and subjects with diplegic CP (Table 4-5 & 4-9). Kilgour et al.[248] also reported equal reliability of the UG amongst healthy subjects and subjects with CP.[248] The highest intratester and intertester ICC values amongst healthy subjects and subjects with diplegic CP were reported for knee flexion measurements (Table 4-5 & 4-9). The lowest intratester and intertester ICC values reported amongst healthy subjects were found to be of ankle dorsiflexion. Meanwhile, ICC values reported for ankle dorsiflexion amongst subjects with diplegic CP were found to be very high with very narrow CI range. Additionally, intratester and intertester ICC values for ankle plantarflexion were also found to be higher amongst subjects with diplegic CP compared with healthy subjects (Table 4-5 & 4-9). This agrees with the findings of a study by Lee et al.[271], where higher reliability for the UG was found amongst subjects with CP when compared with healthy subjects.[271] However, intratester and intertester reliability for knee extension was found to be lower amongst subjects with diplegic CP when compared to healthy subjects.

The current study established a more reliable measurement tool and measurement procedure for passive joint ROM measurement amongst subjects with diplegic CP when compared with the UG. All intratester and intertester ICC values reported in the

literature for the UG for measurement of hip, knee and ankle ROM amongst subjects with diplegic CP were found to be lower than the ICC values found in the current study using *PnO CMD*.[246, 247, 266, 271] Lee et al.[271] reported an intertester ICC value of 0.26 for knee extension for the UG amongst subjects with diplegic CP. The intertester ICC value for knee extension increased with the use of *PnO CMD* to 0.84. Intratester and intertester ICC values reported in the current study for hip flexion and extension (ICC=0.99-1.00) were higher than the reported ICC values by Mutlu et al.[266] (ICC=0.60-0.86) for the UG amongst subjects with diplegic CP. It has been reported that involving two testers in the measurement procedure may increase the reliability of the UG amongst subjects with CP, as one tester stabilises the limb and the second tester takes the measurements.[247] In busy clinical environment, this may not be feasible. The use of *PnO CMD* increases the reliability of the joint ROM measuring and allows for a more practical assessment procedure. Furthermore, all the assessment procedures can be carried out by a single clinician and captured using a single video. Following the video recording, the assessment video can be analysed at a suitable time. Further research aiming to investigate *PnO CMD* for measurement of bi-articular muscle length assessment and spastic catch may be beneficial. The ability to play back the video frame by frame enables the clinician to clearly identify the catch point and measure it reliably and accurately. This system enables the clinics to customise and fill a report for each patient. Having the recorded video alongside the report aids in clearly identifying the effect of the treatment. The markers required to be applied for *PnO CMD* are fewer and cheaper in comparison to the markers required for the Vicon system. Furthermore, this system is easy to learn and use. All the testers involved in the studies had no experience using *PnO CMD* but could measure reliably following a

training session which lasted approximately one hour. However, these testers were experienced clinicians, which may have affected that results. A learning curve was observed as the third *PnO CMD* measuring session was shorter in both reliability studies. Additionally, it would have been beneficial to measure using the UG and compare the results amongst these subjects. It should be noted that the intratester reliability was calculated before the intertester reliability and the data used to calculate the intratester reliability was averaged and used for intertester reliability. Theoretically, since the data has been re-used, Bonferroni correction factor may be considered appropriate. The significance of each intertester ICC values was <0.001 (Table 4-9); hence, even with the Bonferroni correction factor the ICC values would still be significant. Furthermore, the testers measured the join ROM using the same video which lead to eliminate/reduce the challenge faced due to presence of spasticity and defining the end range of joint.

This study concluded high reliability of *PnO CMD* in sagittal plane joint ROM measurements amongst subjects with diplegic CP. High intratester and intertester values were reported for all the joints measured.

In this chapter, the reliability of the different designs of goniometer was evaluated. Additionally, the feasibility of the *PnO CMD* for passive joint ROM measurements was examined. The results indicate that reliable and accurate results can be obtained by the *PnO CMD*. The final chapter provides an overview and the implications of the results of this thesis. This is followed by a discussion of the limitations of this thesis

and possible future avenues of research. Finally, the overall conclusions of the thesis will be presented.

4.3 Figures

Figure 4-1: Flow chart explaining the measurement method using the *PnO CMD* and universal goniometer (UG).

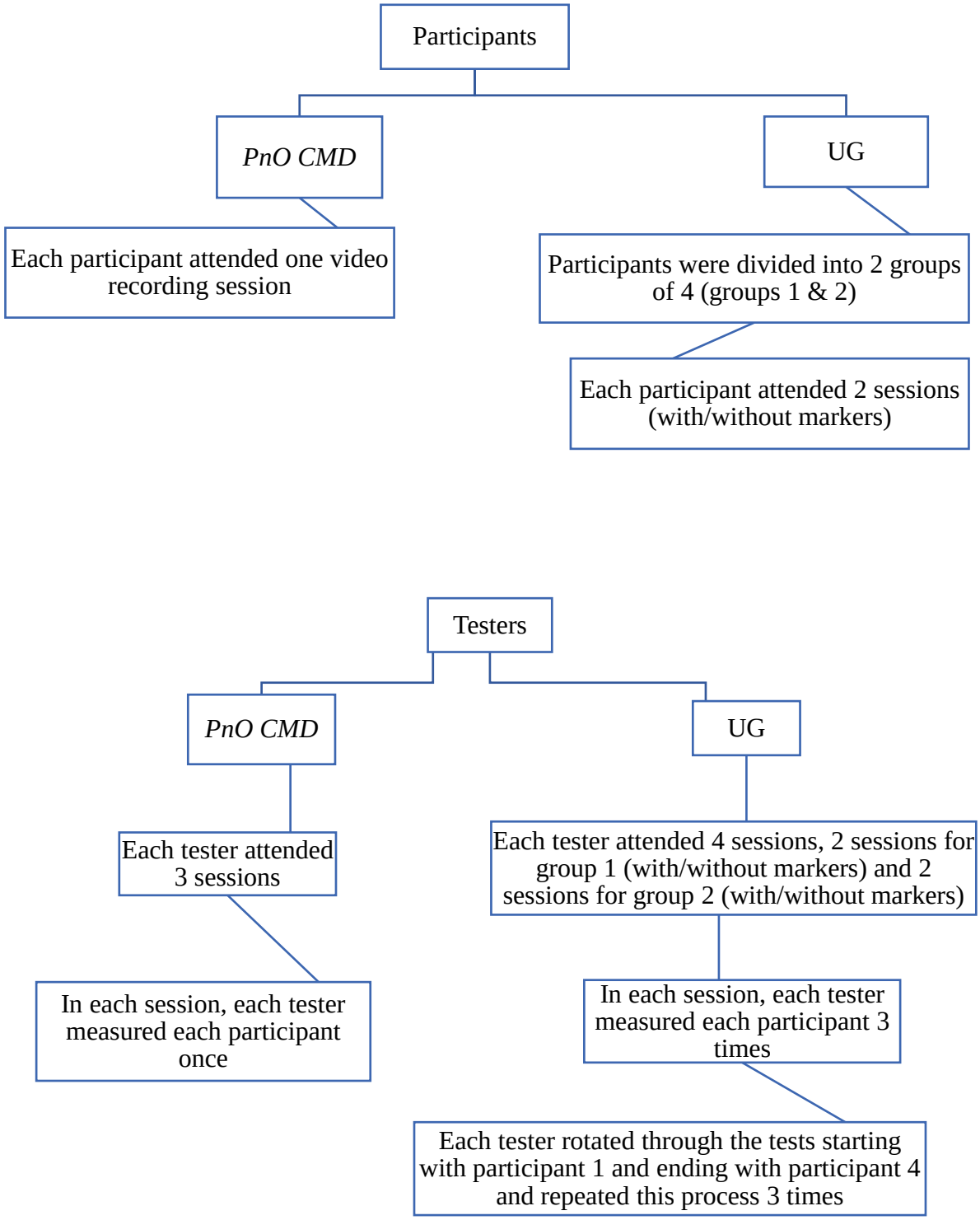


Figure 4-2: Intratester ICC values for tester 1 for all the motion measured using both tools with/without markers.

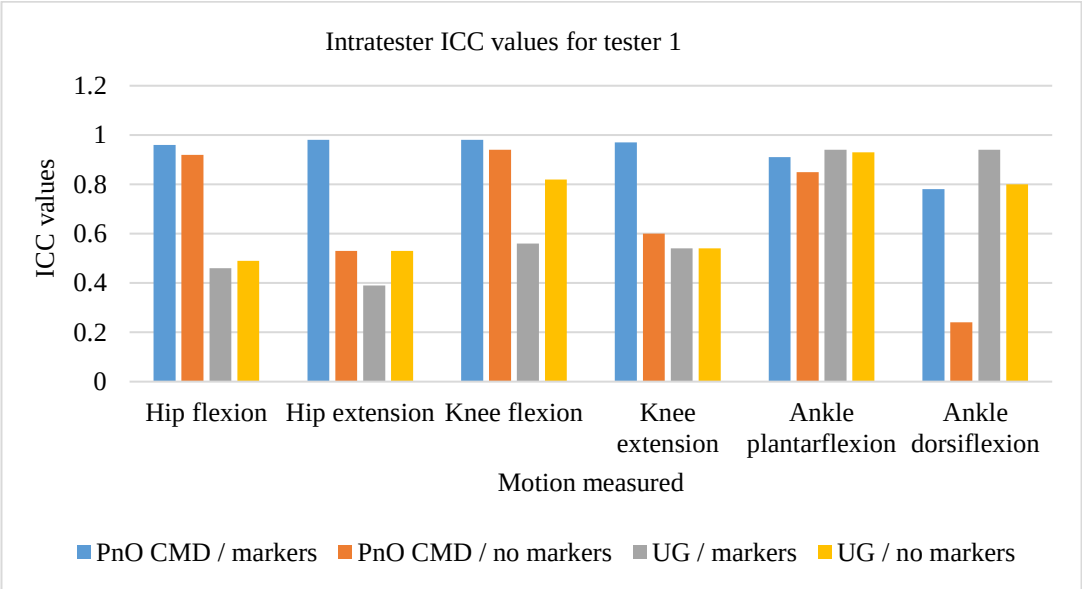


Figure 4-3: Intratester ICC values for tester 2 for all the motion measured using both tools with/without markers.

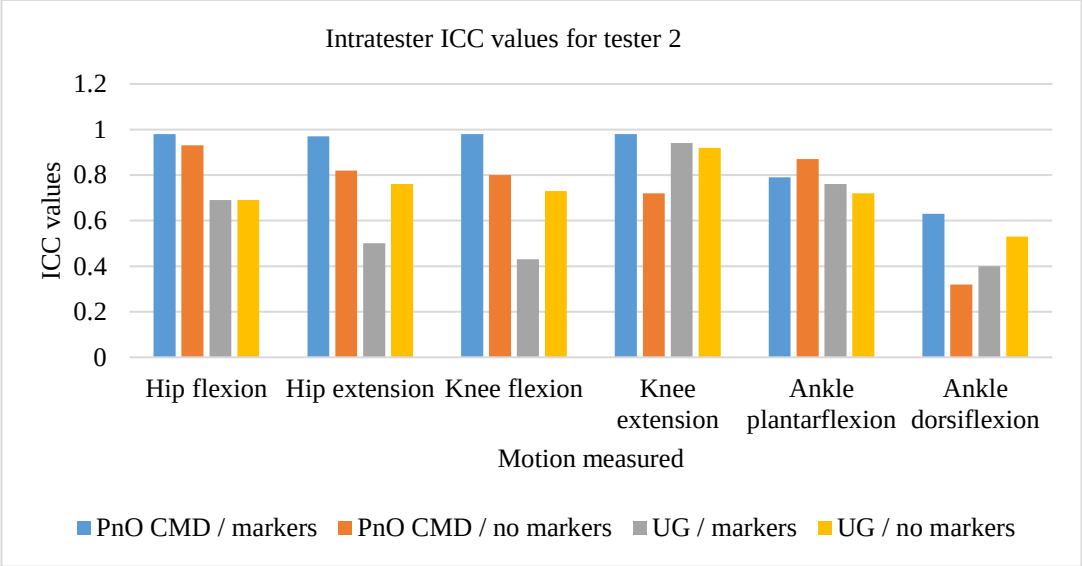


Figure 4-4: Intratester ICC values for tester 3 for all the motion measured using both tools with/without markers.

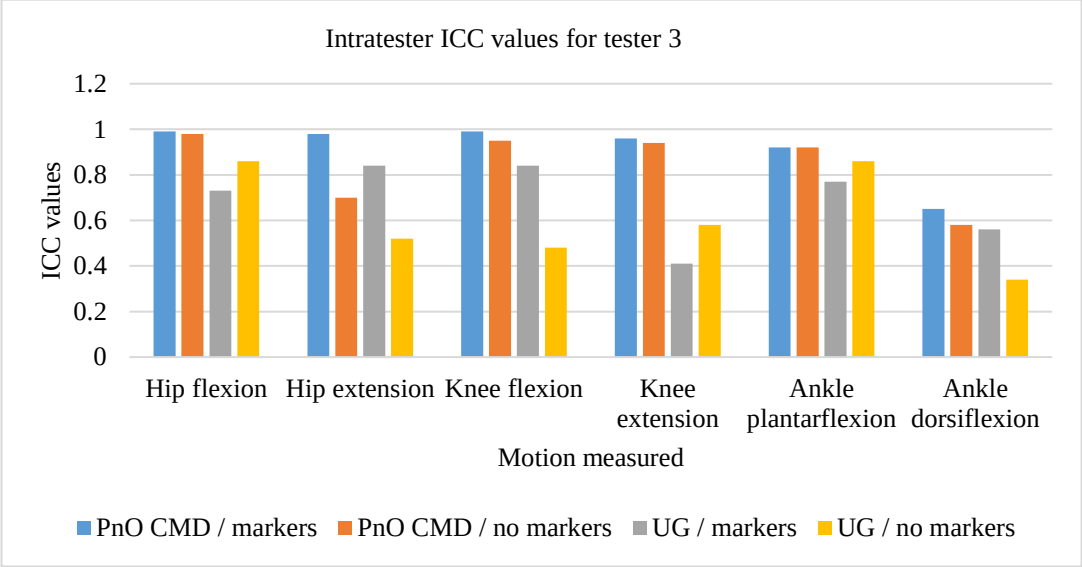


Figure 4-5: Intertester ICC values for all testers for all the motion measured using both tools with/without markers.

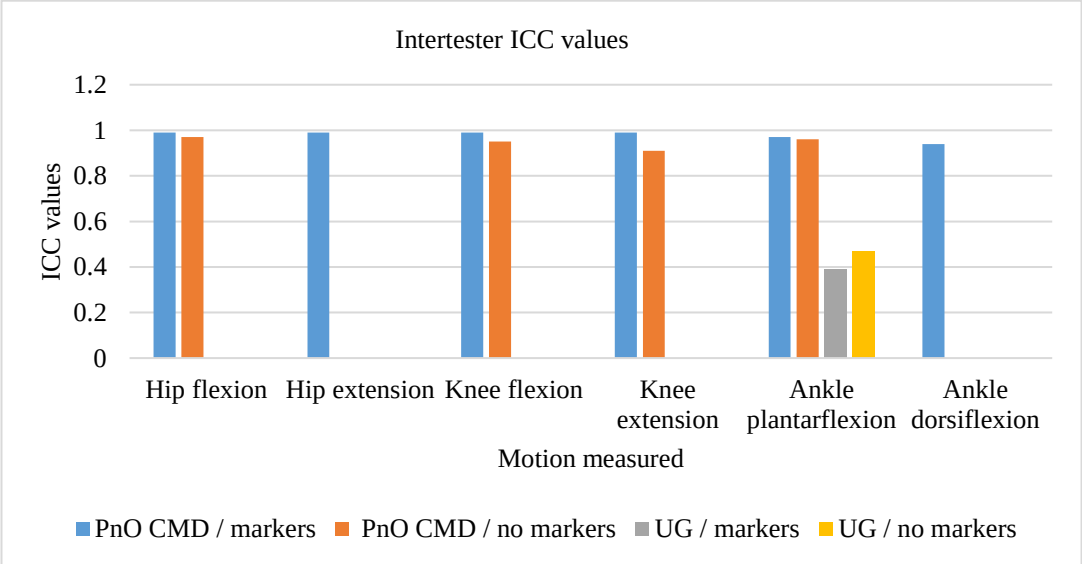


Figure 4-6: The module representing the knee joint. Each marker was displaced by 1 cm in four directions.

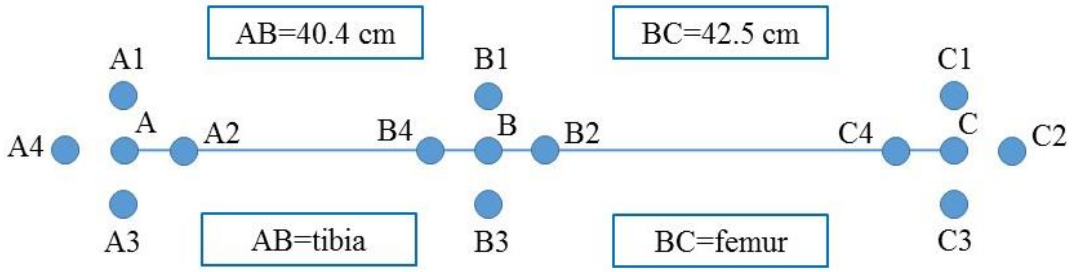


Figure 4-7: The relationship between the standard error of measurements (SEM) of the *PnO CMD* and the ABC angle

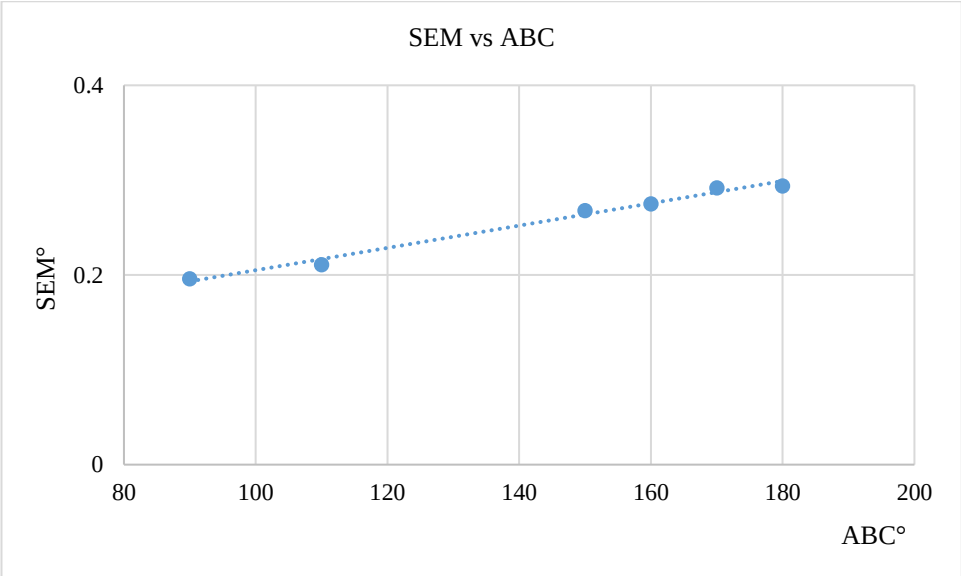


Figure 4-8: Bland & Altman plot for ABC=90°. Y-axis=*PnO CMD*-trigonometry. X-axis=mean value of *PnO CMD* and trigonometry

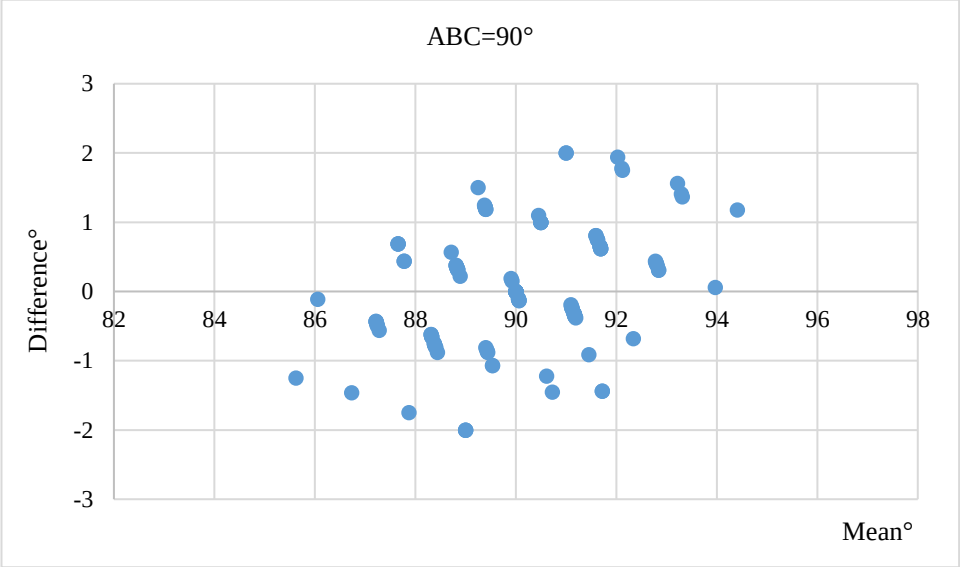


Figure 4-9: Bland & Altman plot for ABC=110°. Y-axis=*PnO CMD*-trigonometry. X-axis=mean value of *PnO CMD* and trigonometry

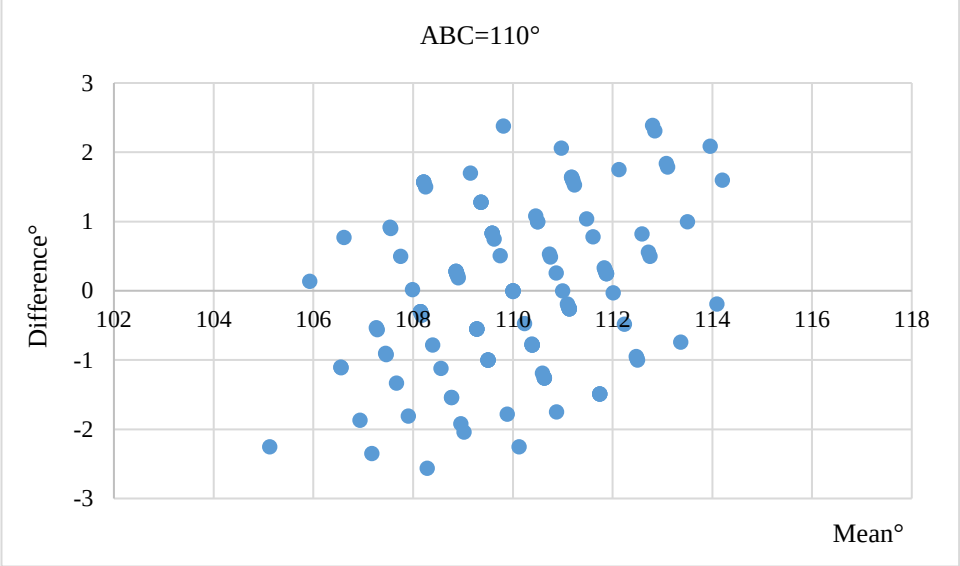


Figure 4-10: Bland & Altman plot for ABC=150°. Y-axis=*PnO CMD*-trigonometry. X-axis=mean value of *PnO CMD* and trigonometry

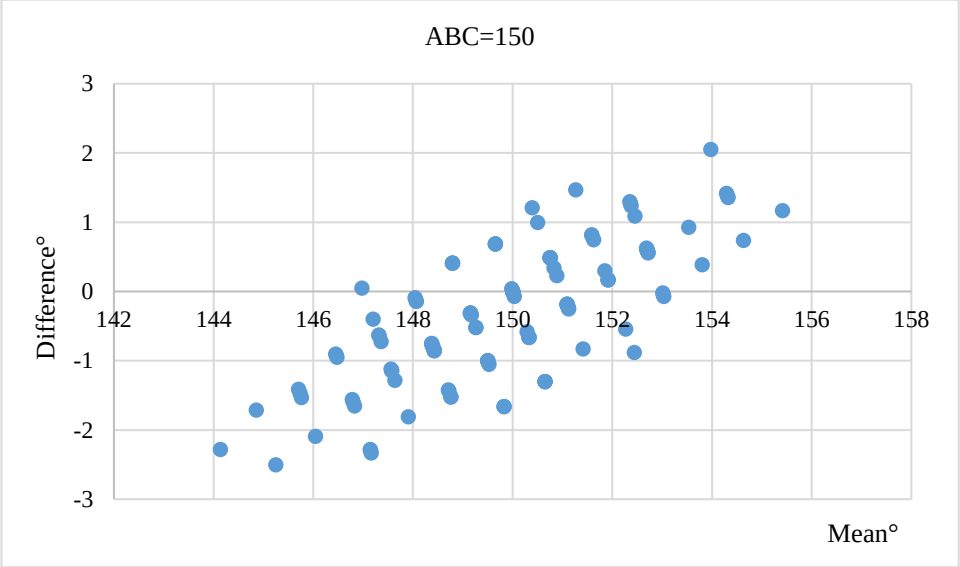


Figure 4-11: Bland & Altman plot for ABC=160°. Y-axis=*PnO CMD*-trigonometry. X-axis=mean value of *PnO CMD* and trigonometry

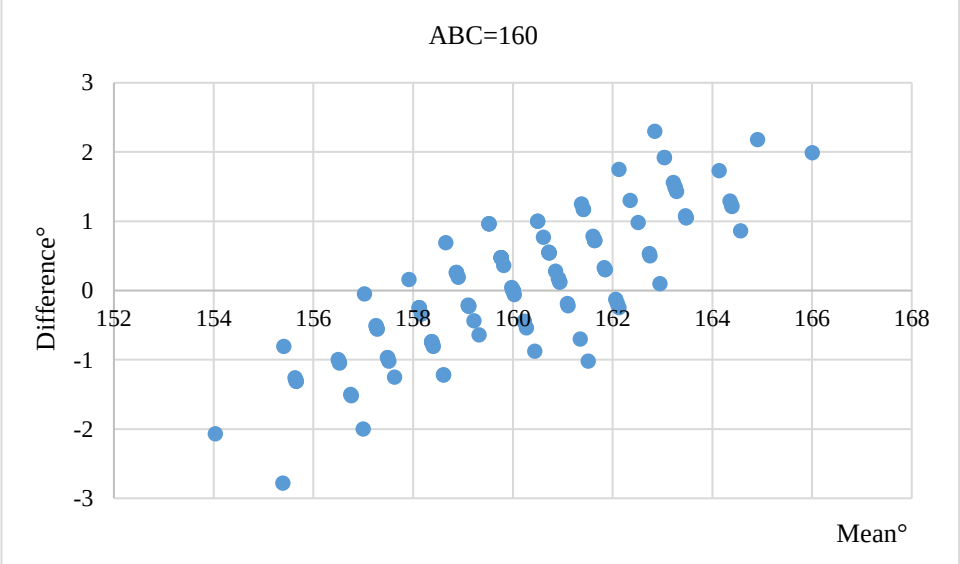


Figure 4-12: Bland & Altman plot for ABC=170°. Y-axis=*PnO CMD*-trigonometry. X-axis=mean value of *PnO CMD* and trigonometry

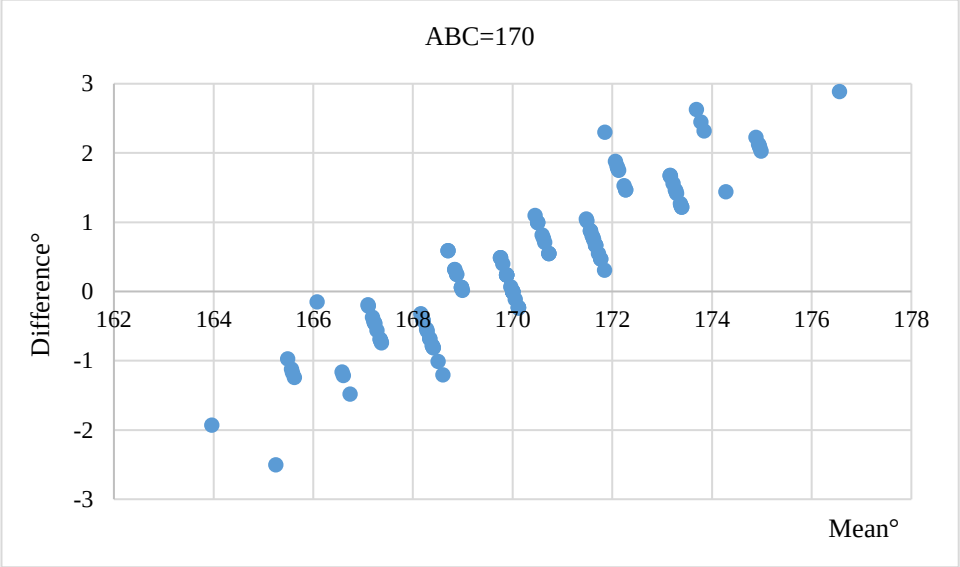
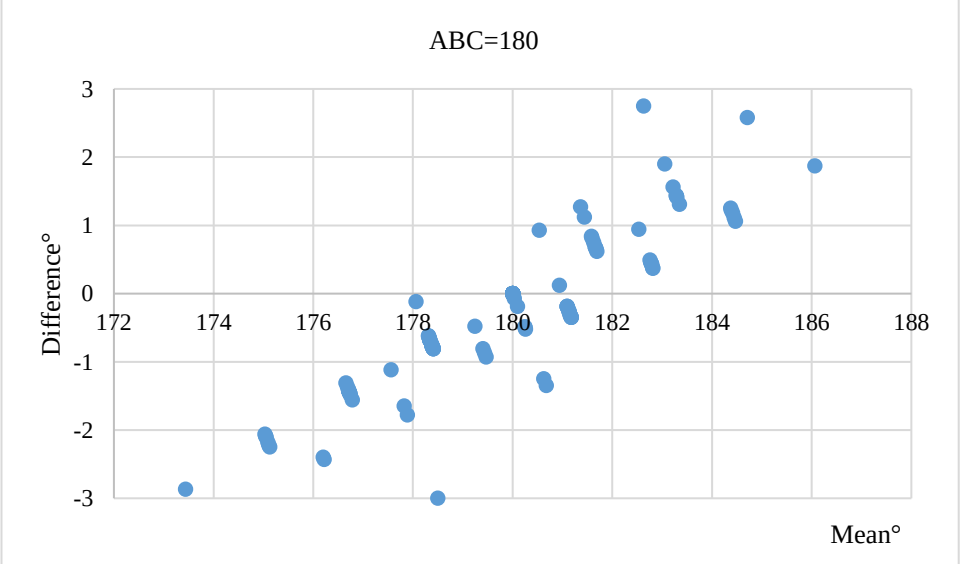


Figure 4-13: Bland & Altman plot for ABC=180°. Y-axis=*PnO CMD*-trigonometry. X-axis=mean value of *PnO CMD* and trigonometry



4.4 Tables

Table 4-1: Summary of the papers which studied the reliability of the UG, EG and inclinometer for measuring hip joint amongst healthy subjects and subjects with pathology.

Source	Goniometer design	Reliability / condition	Movement	Results (ICC)
Healthy subjects				
Active motion				
Clapper and Wolf (1988)[277]	UG & EG	Intra	Flexion, extension, abduction, adduction, external rotation, internal rotation	-UG (flexion=0.95 (E), extension=0.83 (E), abduction=0.86 (E), adduction=0.80 (E), external rotation=0.80 (G), internal rotation=0.92 (E))/ EG (flexion=0.89 (E), extension=0.72 (G), abduction=0.79 (G), adduction=0.77 (G), external rotation=0.86 (G), internal rotation=0.86 (G))
Simoneau et al. (1998)[262]	UG	Inter	External, internal rotation	-Internal rotation=0.82-0.97 (G/E), external rotation=0.76-0.98 (E)
Passive motion				
Ellison, Rose and Sahrmann (1990)[280]	UG & Inclinometer	Intra+inter	Internal rotation, external rotation	Inclinometer, intra (internal rotation=0.98-0.99 (E), external rotation=0.96-0.97 (E)), inter (internal rotation=0.98-0.99 (E), external rotation=0.96 (E))/ UG, no values reported
Van Dillen et al. (2000)[263]	UG	Intra	Extension	0.70-0.96 (G/E)
Kilgour, McNair and Stott (2007)[248]	UG	Intra	Flexion, extension	Extension=0.09-0.92 (W/E), flexion=0.52-0.99 (W/E)
Clapis, S Davis and R Davis (2008)[267]	UG & inclinometer	Inter	Extension	UG=0.92 (E)/ Inclinometer=0.89 (E)
Lee et al. (2011)[271]	UG	Inter	Extension	0.10-0.27 (W)
Wakefield et al. (2015)[268]	UG	Intra+inter	Extension	Intra=0.51-0.54 (W)/ Inter=0.30-0.65 (W/G)
Subjects with pathology				
Passive motion				
Pandya et al. (1985)[251]	UG	Intra+inter/Duchene muscular dystrophy	Extension	Intra=0.85 (E)/ Inter=0.74 (G)
Ellison, Rose and Sahrmann (1990)[280]	UG & Inclinometer	Intra+inter/orthopaedic conditions	Internal rotation, external rotation	Inclinometer, intra (internal rotation=0.96-0.97 (E), external rotation=0.95 (E)), inter (internal rotation=0.96-0.97 (E), external rotation=0.95-0.96 (E))/UG, no values reported

Source	Goniometer design	Reliability / condition	Movement	Results
Mollinger, and Steffen (1993)[260]	UG	Intra+inter/nursing home residents	Extension	Intra=0.98 (E)/Inter=0.24 (W)
Croft et al. (1996)	Inclinometer	Inter/orthopaedic conditions	Flexion, internal rotation, external rotation	Flexion=0.87 (E), internal rotation=0.48 (W), external rotation=0.43 (W)
Cadenhead, McEwen and Thompson (2002)[246]	UG	Inter/neurological conditions	Extension, abduction, external rotation	Extension=0.94-0.98 (E), abduction=0.96 (E), external rotation=0.78-0.86 (G/E)
McWhirk and Glanzman (2006)[247]	UG	Inter/neurological conditions	Extension, abduction	Extension=0.58 (W), abduction=0.90 (E)
Kilgour, McNair and Stott (2007)[248]	UG	Intra/neurological conditions	Flexion, extension	Extension=0.17-0.91(W/E), flexion=0.62-0.98 (G/E)
Mutlu, Livanelioglu and Gunnel (2007)[266]	UG	Intra+inter/neurological conditions	Flexion, extension, abduction, external rotation	Intra (extension=0.73- 0.99 (G/E), abduction=0.48-0.70 (W/G), external rotation=0.80-0.84 (G/E), flexion=0.60-0.86 (G/E))/Inter (extension=0.92-0.95 (E), abduction=0.61-0.77 (G), external rotation=0.91-0.92 (E), flexion=0.77-0.83 (G/E))
Herrero et al. (2011)[265]	UG & inclinometer	Intra+inter/neurological conditions	Abduction	UG, intra=0.82-0.95 (E), inter=0.37-0.47 (W)/Inclinometer, intra=0.85-0.97 (E), inter=0.96-0.97 (E)
Lee et al. (2011)[271]	UG	Inter/neurological conditions	Extension	0.19-0.50 (W)

E: excellent reliability (ICC>0.80), G: good reliability (ICC=0.60-0.80) and W: weak reliability (ICC=0.00-0.60), ICC: intraclass correlation coefficient, UG: universal goniometer, EG: electrical goniometer.

Table 4-2: Summary of the papers which studied the reliability of the UG, EG and inclinometer for measuring knee joint amongst healthy subjects and subjects with pathology.

Source	Goniometer design	Reliability / condition	Movement	Results (ICC)
Healthy subjects				
Active motion				
Clapper and Wolf (1988)[277]	UG & EG	Intra	Flexion, extension	UG (flexion=0.95 (E), extension=0.85 (E))/EG (flexion=0.91 (E), extension=0.80 (G))
Passive motion				
Rothstein, Miller, Roettger (1983)[257]	UG	Intra+inter	Flexion, extension	Intratester (extension=0.91-0.97 (E), flexion=0.97-0.99 (E))/Intertester (flexion=0.91-0.99 (E), extension=0.64-0.71 (G))
Gogia et al. (1987)[249]	UG	Inter	Flexion	0.99 (E)
Kilgour, McNair and Stott (2007)[248]	UG	Intra	Extension	0.34-0.99 (W/E)
Peeler and Anderson (2008)[270]	UG	Intra+inter	Flexion	Intratester=0.65-0.72 (G)/Intertester=0.44-0.59 (W)
Lee et al. (2011)[271]	UG	Inter	Extension	0.20 (W)
Peters et al. (2011)[269]	UG	Intra+inter	Flexion, extension	Intratester (flexion=0.96-0.98 (E), extension=0.83-0.87 (E))/Intertester (flexion=0.88 (E), extension=0.21(W))
Subjects with pathology				
Active motion				
Brosseau et al. (2001)[264]	UG	Intra+inter/neurological conditions	Flexion, extension	-Intratester (flexion=0.99 (E), extension=0.97-0.98 (E))/Intertester (flexion=0.97-0.98 (E), extension=0.89-0.92 (E))
Passive motion				
Pandya et al. (1985)[251]	UG	Intra+inter/Duchene muscular dystrophy	Extension	Intratester=0.93 (E)/Intertester=0.58 (W)
Watkins et al. (1991)[255]	UG	Intra+inter/neurological and orthopaedic conditions	Flexion, extension	Intratester (flexion=0.99 (E), extension=0.98 (E))/Intertester (extension=0.86 (E), flexion=0.90 (E))
Mollinger, and Steffen (1993)[260]	UG	Intra+inter/nursing home residents	Extension	Intratester=0.99 (E)/Intertester=0.96 (E)
Cadenhead, McEwen and Thompson (2002)[246]	UG	Intra/neurological conditions	Extension	0.81-0.98 (E)
McWhirk and Glanzman (2006)[247]	UG	Inter/neurological conditions	Extension	0.78-0.92 (G/E)
Kilgour, McNair and Stott (2007)[248]	UG	Intra/neurological conditions	Extension	0.57-0.99 (W/E)
Lee et al. (2011)[271]	UG	Inter / neurological conditions	Extension	0.26 (W)

E: excellent reliability (ICC>0.80), G: good reliability (ICC=0.60-0.80) and W: weak reliability (ICC=0.00-0.60), ICC: intraclass correlation coefficient, UG: universal goniometer, EG: electrical goniometer

Table 4-3: Summary of the papers which studied the reliability of the UG, EG and inclinometer for measuring ankle joint amongst healthy subjects and subjects with pathology.

Source	Goniometer design	Reliability	Movement	Results (ICC)
Healthy subjects				
Active motion				
Clapper and Wolf (1988)[277]	UG & EG	Intra	Dorsiflexion, plantarflexion	-UG (dorsiflexion=0.92 (E), plantarflexion=0.96 (E))/EG (dorsiflexion=0.80 (G), plantarflexion=0.93 (E))
Bennell et al. (1998)[281]	Inclinometer	Intra+inter	Dorsiflexion	Intra=0.98 (E)/Inter=0.97 (E)
Thoms and Rome (1997)[273]	UG	Intra	Dorsiflexion	0.93 (E)
Munteanu et al. (2009)[272]	Inclinometer	Intra+inter	Dorsiflexion	Intra=0.77-0.91 (G/E)/Inter=0.95 (E)
Konor et al. (2012)[278]	UG & inclinometer	Intra	Dorsiflexion	UG=0.85-0.96 (E)/Inclinometer=0.96-0.97 (E)
Passive motion				
Jonson and Gross (1997)[274]	UG	Intra+inter	Dorsiflexion	Intra=0.74 (G)/Inter=0.65 (G)
Kilgour, McNair and Stott (2007)[248]	UG	Intra	Dorsiflexion	0.63-0.99 (G/E)
Subjects with pathology				
Active motion				
Youdas, Bogard, Suman (1993)[261]	UG	Intra+inter/orthopaedic conditions	Dorsiflexion, plantarflexion	Intratester (dorsiflexion=0.64-0.92 (G/E), plantarflexion=0.47-0.96 (W/E))/Intertester (dorsiflexion=0.28 (W), plantarflexion=0.25 (W))
Passive motion				
Pandya et al. (1985)[251]	UG	Intra+inter/Duchene muscular dystrophy	Dorsiflexion	Intratester=0.90 (E)/Intertester=0.73 (G)
Elveru, Rothstein and Lamb (1988)[258]	UG	Intra+inter/neurological and orthopaedic conditions.	Dorsiflexion, plantarflexion	Intratester (dorsiflexion=0.90 (E), plantarflexion=0.86 (E))/Intertester (dorsiflexion=0.50 (W), plantarflexion=0.72 (G))
Diamond et al. (1989)[259]	UG	Intra+inter/diabetes	Dorsiflexion	Intratester=0.89-0.96 (E)/Intertester=0.74-0.87 (G/E)
Cadenhead, McEwen and Thompson (2002)[246]	UG	Intra/neurological conditions	Dorsiflexion	0.81-0.98 (E)
McWhirk and Glanzman (2006)[247]	UG	Inter/neurological conditions	Dorsiflexion	0.87 (E)
Kilgour, McNair and Stott (2007)[248]	UG	Intra/neurological conditions	Dorsiflexion	0.63-0.99 (G/E)
Mutlu, Livanelioglu and Gunnel (2007)[266]	UG	Intra+inter/neurological conditions	Dorsiflexion	Intratester=0.81-0.90 (E)/Intertester=0.88 (E)

E: excellent reliability (ICC>0.80), G: good reliability (ICC=0.60-0.80) and W: weak reliability (ICC=0.00-0.60), ICC: intraclass correlation coefficient, UG: universal goniometer, EG: electrical goniometer.

Table 4-4: Summary of the key findings of the literature review investigating the reliability of different designs of goniometer.

Factors affecting reliability	Key findings
Search results & overall impression	-Twenty-nine studies were included -Variation in the methodology applied reduced the ability of direct comparison -Number of testers, experience level, number of sessions, time between the sessions and subject position varied between the studies
Reliability (intratester & intertester)	-Intratester reliability was found to be higher than intertester reliability -Low intertester reliability could be a result of difference in the force applied by therapists during assessment
Motion measured	-Most studies examined reliability of passive motion -Limited number of studies investigated active motion, which affected the comparison between active and passive motion
Measurement procedure	-Standardisation of procedure and prior training increased reliability -Involvement of more than one tester can increase the reliability
Subjects	-Reliability was best amongst healthy subjects -Reliability may be reduced in the presence of different pathologies
Goniometer design	-Limited number of studies compared between the different designs of the goniometers -No study compared between the 3 designs of goniometers
Joint measured	-Reliability ranged from weak to excellent across the hip, knee and ankle joints -Different joint characteristics and ease of identifying bony landmarks affected reliability -Strict standard position can increase the reliability of complex motions

Table 4-5: Intratester and intertester ICC values, p-values and 95% CI across all the testers for both tools with/without markers

Motion/ Tester	Tester 1				Tester 2				Tester 3				Tester 1, 2 & 3			
Reliability	Intratester												Intertester			
	ICC	95% CI		P-value	ICC	95% CI		P-value	ICC	95% CI		P-value	ICC	95% CI		P-value
		Lower bound	Upper bound			Lower bound	Upper bound			Lower Bound	Upper bound			Lower bound	Upper bound	
PnO CMD/markers																
Hip Flexion	0.96	0.88	0.99	<0.001	0.98	0.94	0.99	<0.001	0.99	0.97	0.99	<0.001	0.99	0.98	0.99	<0.001
Hip Extension	0.98	0.94	0.99	<0.001	0.97	0.90	0.99	<0.001	0.98	0.94	0.99	<0.001	0.99	0.97	0.99	<0.001
Knee Flexion	0.98	0.96	0.99	<0.001	0.98	0.96	0.99	<0.001	0.99	0.98	0.99	<0.001	0.99	0.99	1.00	<0.001
Knee extension	0.97	0.91	0.99	<0.001	0.98	0.93	0.99	<0.001	0.96	0.89	0.99	<0.001	0.99	0.98	0.99	<0.001
Ankle plantarflexion	0.91	0.76	0.98	<0.001	0.79	0.47	0.94	<0.001	0.92	0.77	0.98	<0.001	0.97	0.91	0.99	<0.001
Ankle dorsiflexion	0.78	0.461	0.94	<0.001	0.63	0.29	0.91	<0.001	0.65	0.26	0.90	<0.001	0.94	0.82	0.98	<0.001
PnO CMD/no markers																
Hip Flexion	0.92	0.76	0.98	<0.001	0.93	0.80	0.98	<0.001	0.98	0.93	0.99	<0.001	0.97	0.91	0.99	<0.001
Hip Extension	0.53*	0.13	0.86	<0.001	0.82	0.53	0.95	<0.001	0.70	0.33	0.92	<0.001	-	0.46	0.97	<0.001
Knee Flexion	0.94	0.82	0.98	<0.001	0.80	0.23	0.95	<0.001	0.95	0.86	0.99	<0.001	0.95	0.44	0.99	<0.001
Knee extension	0.60	0.15	0.89	<0.001	0.72	0.37	0.92	<0.001	0.94	0.7	0.98	<0.001	0.91	0.35	0.98	<0.001
Ankle plantarflexion	0.85	0.60	0.96	<0.001	0.87	0.60	0.97	<0.001	0.92	0.77	0.98	<0.001	0.96	0.88	0.99	<0.001
Ankle dorsiflexion	0.24*	-0.19	0.73	0.15	0.32*	-0.10	0.77	0.07	0.58*	0.18	0.88	<0.001	-	-	-	-

Motion/ Tester	Tester 1				Tester 2				Tester 3				Tester 1, 2 & 3			
Reliability	Intratester												Intertester			
	ICC	95% CI		P-value	ICC	95% CI		P-value	ICC	95% CI		P-value	ICC	95% CI		P-value
		Lower bound	Upper bound			Lower bound	Upper bound			Lower Bound	Upper bound			Lower bound	Upper bound	
UG/markers																
Hip Flexion	0.46*	0.071	0.83	0.01	0.69	0.27	0.92	<0.001	0.73	0.37	0.93	<0.001	-	-	-	-
Hip Extension	0.39*	-0.05	0.81	0.045	0.50*	0.05	0.85	0.01	0.84	0.56	0.96	<0.001	-	-	-	-
Knee Flexion	0.56*	0.15	0.87	0.005	0.43*	0.02	0.82	0.02	0.84	0.59	0.96	<0.001	-	-	-	-
Knee extension	0.54*	0.08	0.87	0.01	0.94	0.82	0.98	<0.001	0.41*	-0.04	0.81	0.04	-	-	-	-
Ankle plantarflexion	0.94	0.84	0.98	<0.001	0.76	0.44	0.94	<0.001	0.77	0.45	0.94	<0.001	0.39*	-0.07	0.80	<0.001
Ankle dorsiflexion	0.94	0.82	0.98	<0.001	0.40*	-0.05	0.81	0.04	0.56*	0.16	0.87	<0.001	-	-	-	-
UG/no markers																
Hip Flexion	0.49*	0.04	0.85	0.01	0.69	0.31	0.92	<0.001	0.86	0.60	0.97	<0.001	-	-	-	-
Hip Extension	0.53*	0.08	0.86	0.01	0.76	0.41	0.94	<0.001	0.52*	0.08	0.86	0.01	-	-	-	-
Knee Flexion	0.82	0.54	0.95	<0.001	0.73	0.36	0.93	<0.001	0.48*	0.04	0.84	0.01	-	-	-	-
Knee extension	0.54*	0.11	0.86	<0.001	0.92	0.76	0.98	<0.001	0.58*	0.19	0.88	<0.001	-	-	-	-
Ankle plantarflexion	0.93	0.79	0.98	<0.001	0.72	0.36	0.92	<0.001	0.86	0.64	0.96	<0.001	0.47*	-0.22	0.86	0.07
Ankle dorsiflexion	0.80	0.51	0.95	<0.001	0.53*	0.09	0.86	0.01	0.34*	-0.11	0.79	0.075	-	-	-	-

ICC: intraclass correlation coefficient, CI: confidence interval, ICC values marked with an asterisk* are below satisfactory level (ICC<0.60). P-values in red font are significant values at 0.05 level.

Table 4-6: Mean, SD of both methods and SEM of *PnO CMD*.

ABC°	Mean (SD)°		SEM°
	PnO CMD	Trigonometry	*PnO CMD*
90	90.06 (1.96)	90 (1.64)	0.196
110	110 (2.11)	110.01(1.77)	0.211
150	149.68(2.68)	150 (1.96)	0.268
160	160.11(2.75)	160.01(1.99)	0.275
170	170.39(2.92)	169.99(1.99)	0.292
180	179.71(2.94)	179.99(1.98)	0.294

SD: standard deviation, SEM: standard error of measurement

Table 4-7: Intratester and intertester ICC values, p-values and 95% CI across all the testers.

Motion/ Tester	Tester 1				Tester 2				Tester 3				Tester 1, 2 & 3			
Reliability	Intratester												Intertester			
	ICC	95% CI		P-value	ICC	95% CI		P-value	ICC	95% CI		P-value	ICC	95% CI		P-value
		Lower bound	Upper bound			Lower Bound	Upper bound			Lower bound	Upper bound			Lower bound	Upper bound	
Hip flexion	0.99	0.99	1.00	<0.001	0.99	0.97	0.99	<0.001	0.99	0.98	0.99	<0.001	1.00	0.99	1.00	<0.001
Hip extension	0.99	0.98	0.99	<0.001	0.97	0.90	0.99	<0.001	0.99	0.96	0.99	<0.001	0.99	0.98	0.99	<0.001

ICC: intraclass correlation coefficient, CI: confidence interval, ICC values marked with an asterisk* are below satisfactory level (ICC<0.60). P-values in red font are significant values at 0.05 level.

Table 4-8: ICC values, p-values and 95% CI for each point of gait.

Points of gait	Hip				Knee				Ankle			
	ICC	95% CI		P-value	ICC	95% CI		P-value	ICC	95% CI		P-value
		Lower bound	Upper bound			Lower bound	Upper bound			Lower bound	Upper bound	
Point 1	0.94	0.75	0.98	<0.001	0.96	0.78	0.99	<0.001	0.90	0.70	0.97	<0.001
Point 2	0.90	0.69	0.97	<0.001	0.98	0.91	0.99	<0.001	0.92	0.75	0.97	<0.001
Point 3	0.96	0.79	0.99	<0.001	0.93	0.71	0.98	<0.001	0.94	0.83	0.98	<0.001
Point 4	0.93	0.80	0.98	<0.001	0.98	0.94	0.99	<0.001	0.93	0.79	0.98	<0.001
Point 5	0.95	0.86	0.98	<0.001	0.95	0.84	0.98	<0.001	0.93	0.78	0.98	<0.001

ICC: intraclass correlation coefficient, CI: confidence interval, ICC values marked with an asterisk* are below satisfactory level (ICC<0.60). P-values in red font are significant values.

Table 4-9: Intratester and intertester ICC values, p-values and 95% CI across all the testers.

Motion/ Tester	Tester 1				Tester 2				Tester 3				Tester 1, 2 & 3			
Reliability	Intratester												Intertester			
	ICC	95% CI		P-value	ICC	95% CI		P-value	ICC	95% CI		P-value	ICC	95% CI		P-value
		Lower bound	Upper bound			Lower bound	Upper bound			Lower bound	Upper bound			Lower bound	Upper bound	
Hip Flexion	0.99	0.97	0.99	<0.001	0.97	0.88	0.99	<0.001	0.99	0.97	0.99	<0.001	0.98	0.94	0.99	<0.001
Hip Extension	0.97	0.88	0.99	<0.001	0.97	0.89	0.99	<0.001	0.98	0.92	0.99	<0.001	0.98	0.93	0.99	<0.001
Knee Flexion	1.00	0.99	1.00	<0.001	0.99	0.99	1.00	<0.001	1.00	0.99	1.00	<0.001	1.00	0.99	1.00	<0.001
Knee Extension	0.94	0.76	0.99	<0.001	0.91	0.67	0.99	<0.001	0.94	0.78	0.99	<0.001	0.84	0.47	0.98	0.001
Ankle plantarflexion	0.95	0.82	0.99	<0.001	0.93	0.74	0.99	<0.001	0.96	0.86	0.99	<0.001	0.98	0.92	0.99	<0.001
Ankle dorsiflexion	0.99	0.97	1.00	<0.001	0.99	0.96	0.99	<0.001	0.99	0.96	0.99	<0.001	0.99	0.97	0.99	<0.001

ICC: intraclass correlation coefficient, CI: confidence interval, ICC values marked with an asterisk* are below satisfactory level (ICC<0.60). P-values in red font are significant values at 0.05 level.

Chapter 5 Discussion

5.1 Summary and implications of the findings

The overarching aim of this thesis was to investigate the effect of optimisation of the temporal midstance shank kinematics with the use of an AFO- FC on several variables including shank kinematics, thigh kinematics, magnitude of FZ2 and GRF alignment in relation to hip and knee joint in temporal midstance to late stance. The understanding of the implication of, and the tools used to conduct assessment procedures and measurement processes were tested in the assessment of bi-articular muscles (specifically rectus femoris), to ascertain if results were predictive of presenting gait deviations and hence provide information to optimise treatment.

The initial study of this thesis examined the effect of optimisation of the temporal midstance shank kinematics with the use of an AFO-FC, on thigh kinematics in temporal midstance and on thigh and shank kinematics at FZ2 and at the maximum TVA. Furthermore, this study investigated the effect of optimisation of the temporal midstance shank kinematics on FZ2 and on the alignment of the GRF in relation to the hip and knee joints in temporal midstance, at FZ2 and at the maximum TVA. The results of this study suggested that a positive influence on the shank kinematics (at FZ2 and at the maximum TVA) and thigh kinematics (in temporal midstance, at FZ2 and at the maximum TVA) was observed with the use of an AFO-FC (Table 2-4). Additionally, 9 participants exhibited a positive increase in magnitude of FZ2 and reduction in the degree of Ben Lomonding (Table 2-9). Amongst these 9 participants, FZ2 increased to greater than or equal to BW in 5 participants (Table 2-9).

Some effects of optimisation of temporal midstance shank kinematics were observed on improving the GRF alignment in relation to the hip and knee joints at the 3 selected points (Table 2-6). The above findings may indicate that optimising the temporal midstance shank kinematics not only affects the temporal midstance but terminal stance as well. Each of these points has an important impact on the gait. A study by Ridgewell[178] found similar results: optimising the shank kinematics affects both phases. The study reported improvements in knee extension and knee moments that occur in terminal stance.

Optimising temporal midstance shank kinematics by optimising the AFO-FC characteristics and fine-tuning is thought to normalise gait. However, in the wider body of literature, the definition of normal gait is unclear.[178] Additionally, the systemic effect of an AFO-FC across a number of variables is ambiguous. Further research is required

The study described in Chapter Two aimed to investigate if examined gait variables optimised simultaneously using an AFO-FC. For this, the participants were divided into two groups according to their barefoot SVA. Group One included participants with SVAs less inclined than normal and normal (n=7), and Group Two included participants with SVAs more inclined than normal (n=9). Following that, the temporal midstance SVA was correlated with the investigated elements barefoot and with an AFO-FC. The correlation values differed between both groups. This indicates that the initial barefoot SVA alignment affects the results obtained with the use of an AFO-FC. The correlation values in Group One indicated that the alignment of the GRF at

the knee joint in temporal midstance and at the hip joint at FZ2 may optimise simultaneously with the use of an AFO-FC (Table 2-7). In contrast, the correlation values in Group Two showed that FZ2, thigh inclination and GRF alignment at knee joint at the maximum TVA may optimise simultaneously with the use of an AFO-FC (Table 2-8). However, caution is required in any interpretation or generalisation of these results because of the small sample size in each group. However, these results may help provide evidence in future studies designed with appropriate sample size/power.

Furthermore, agreement across the variables with positive influence, including shank kinematics, thigh kinematics and FZ2, was found in 2 participants (Table 2-10). Two participants also had agreement of 5 out of 6 variables. The most positive effect of optimising the temporal midstance shank kinematics was seen on the thigh kinematics at maximum TVA. In contrast, shank kinematics at maximum TVA was affected negatively with the use of an AFO-FC (Table 2-10).

In 9 participants, the SVA measured while standing differed from the SVA measured in temporal midstance (Table 2-12). This suggests that the AFOs were insufficiently stiff, which allowed further shank inclination (above 10-12° inclination) during temporal midstance (Table 2-12). The further shank inclination observed in 9 participants may have caused some of the negative results found amongst these participants, e.g. GRF alignment in relation to the hip and the knee joints (Tables 2-5 & 2-6). Further possible causes which may have influenced/compromised the results obtained include FW characteristics, un-accommodated gastrocnemius length, bone

rotations and muscle tone/spasticity/shortness. This study highlighted the effect of optimising the temporal midstance shank kinematics with the use of an AFO-FC on gait and the importance of paying attention to this matter. Additionally, this study emphasised the importance of choosing the appropriate AFO-FC characteristics, including stiffness, followed by tuning to achieve the optimum temporal midstance SVA alignment. However, as discussed in Chapter Two, the results of this study should be applied with an understanding of their clinical implications.

As discussed in Chapter One, detailed clinical assessment is a mandatory step in the preparation of the optimum intervention plan and forming the prescription of an AFO-FC. Comprehensive physical assessment identifies the orthopaedic and neurological challenges presented. One of the common challenges found amongst subjects with diplegic CP is shortness and/or spasticity of the multi-articular and bi-articular muscles, specifically the hamstrings, RF and gastrocnemius. Appropriate assessment of these muscles enables optimum AFO-FC prescription and any required adjacent therapy. In this thesis, the RF muscle was chosen as an exemplar to emphasise the importance of bi-articular muscle assessment.

Initially, a literature review about the RF was completed. This review highlighted that the exact timing of activation of the RF during gait and the effect of the RF dysfunction on gait were found to be inconsistent. As discussed in Chapter Three, early studies recommended that diminished or delayed peak knee flexion in swing associated with excessive RF activity is an indication for RF transfer surgery.[24, 25] This is because excessive activity in the RF during swing causes larger than normal internal knee

extension moments, thus limiting the knee flexion. This abnormal gait is known as SKG. However, the outcome of the RF transfer surgery was not always found to be successful; hence, research was carried out to provide further insight into this. A study by Goldberg et al.[27] found that most subjects with SKG exhibited normal or below normal internal knee extension moments during swing phase. This finding opposes the link commonly made between altered RF muscle activity and decreased peak knee flexion in swing. Additionally, further research showed that other factors, such as abnormal activity of the RF during double support and preswing, decrease knee flexion velocity at toe-off. Reduced knee flexion velocity at toe-off can lead to reduction of the peak knee flexion in swing.[28, 30, 226, 311]

The literature review related to the function of RF was found inconclusive. This uncertainty found in the literature led to examining the literature related to the method used to examine the RF. The Duncan-Ely test is commonly used to measure the length and spasticity of the RF. In total, four studies were found.[32, 233, 235, 312] Amongst these four studies, two studies examined the reliability of the Duncan-Ely test. One study reported good intertester reliability amongst subjects with CP.[233] On the other hand, the second study reported lower intertester reliability and weak to excellent intratester reliability amongst healthy subjects.[235] A further study found that the Duncan-Ely test had a good positive predictive value, indicating that subjects who test positively would actually have SKG.[232] However, another study reported that the sensitivity, specificity and the negative predictive values were found to be low for the Duncan-Ely test.[32] This limited evidence found in the literature about the Duncan-Ely test affected the ability to draw clear conclusions. The RF is a bi-articular muscle which crosses the hip and

the knee joints. Spasticity in the RF muscle can affect the knee and/or the hip joint. However, none of the studies investigating the sensitivity, specificity and predictive values of the Duncan-Ely test considered the position of the hip joint when gait variables, such as peak knee flexion, were measured.[32, 232, 233, 313] Several studies did examine the hip moments across gait cycle and the hip flexion velocity at toe-off in subjects with SKG.[27, 224] However, these studies did not measure the hip and the knee joints at the same instant.[27, 28, 30, 226]

The study in Chapter Three examined the relationship between the catch angle/length of the RF measured using the Duncan-Ely test and the hip or the knee joint at selected points/periods of gait. These points/periods included maximum hip extension in stance, early swing (from toe-off to peak knee flexion in swing), peak knee flexion in swing and peak hip flexion in swing. Furthermore, this study evaluated if a dominance relationship between the catch angle/length of the RF was found with the hip or the knee joint. The relationship between the catch angle/length of the RF measured using the Duncan-Ely test and the timing of peak knee flexion in swing was also examined. Finally, the study investigated if the effect of dynamic shortness of the RF on gait differed from the effect of true shortness of the RF, and if the dominant relationship was found with the hip or the knee joint. To best of the author's knowledge, this is the first study which considers the hip joint position and the effect of RF length/spasticity on the hip joint.

Thirty-seven limbs were included and were divided into main groups per the catch angle/length: Group A: limbs with catch angle/length<60° (n=19) and Group B: limbs

with catch angle/length≥60° (n=18). To differentiate between the effect of true and dynamic shortness, all limbs were divided based on the presence of dynamic or true shortening: Group C: limbs with dynamic shortness (n=20) and Group D: limbs with true shortness (n=17). Within Group C, limbs were further divided: Group C1: limbs with catch angle<60º (n=13) and Group C2: limbs with catch angle≥60º (n=7). Within Group B, limbs were further divided: Group D1: limbs with length<60º (n=6) and Group D2: limbs with length≥60º (n=11). Significant regression models were analysed and considered.

Overall, the regression models between the catch angle/length and the knee joint at maximum hip extension, peak knee flexion and peak hip flexion in Group A (catch angle/length<60°) were found to be weak but significant (Tables 3-2 & 3-3). This may suggest that a dominance relationship of the catch angle/length was found with the knee joint at these points. In contrast, a dominance relationship was found between the catch angle/length and the hip joint in early swing in Group A (catch angle/length<60°). In Group B, the regression model between the catch angle/length and the knee at peak hip flexion was found significant, indicating that the knee joints exhibited a dominance relationship with the catch angle/length at this point (Tables 3-2 & 3-3).

Further analysis showed that the effect of dynamic shortness on gait differed from the effect of true shortness. In Group C1 (catch angle<60°), a dominance relationship of the catch angle was found to be with the hip joint in early swing and with the knee joint at peak knee flexion and peak hip flexion (Tables 3-11 & 3-12). In contrast, in

Group D1 (length<60°) a dominance relationship was found with the hip joint at peak knee flexion (Tables 3-19 & 3-20). The overall regression at peak knee flexion was higher in Group D1 (length<60°) in comparison with C1 (catch angle<60°). In Group C2 (catch angle≥60°), a dominance relationship was found with the knee joint in early swing, with the regression value being good and significant. In Group D2 (length≥60°), all the regression models were found to be weak and not significant (Tables 3-19 & 3-20).

The linear regression models for the timing of peak knee flexion in all groups were found to be weak and not significant (Table 3-9). Although these models were weak and not significant, the effect of dynamic/true shortness of the RF was seen on the timing of peak knee flexion in swing. Thirty-three limbs showed delay in timing of peak knee flexion in swing (Table 3-10). In general, the sample size in each sub-sample and the variance observed between similar catch angles/lengths may have affected the regression models. Some of this variance found between similar catch angles can be explained due to the difference observed in the tone of the RF (difference in MAS score). The difference observed in tone leads to varying degree of resistance to motion. For example, MAS score=1 means that slight increase is observed in muscle tone which is demonstrated by a catch and release or by minimal resistance at the end of ROM while, MAS=+1 indicates a slight increase in muscle tone accompanied with a catch and followed by minimal resistance throughout the remaining ROM.[49, 50] When comparing two limbs with similar catch angle and different tone, the limb with minimal tone (MAS=1) will be less resistance to motion following the catch in comparison to a limb with increased tone (MAS=+1).

Since advanced gait analysis systems are not always available in the clinical environment, this study enables clinicians to understand how a catch angle/length measured using the Duncan-Ely test will affect gait. Additionally, it may be possible that linear regression output equations and the ratios between the input and output variables may be useful clinically to estimate the effect on gait (Tables 3-4, 3-14 & 3-23). However, as before these results must be applied with caution due to the low sample number and regression values. Furthermore, the research methodology applied in Chapter Three can be adopted to examine the effect of other bi-articular and multi-articular muscles on gait, e.g. hamstrings and gastrocnemius.

Overall, despite the fact that not all regression models were significant, data of this study suggests that in most included limbs dynamic/true shortness of the RF can cause reduction in maximum hip extension in stance, increased knee flexion at maximum hip extension in stance, reduced range of knee flexion in early swing, reduced range of hip flexion in early swing, reduced peak knee flexion in swing, increased hip at peak knee flexion, increased peak hip flexion, increased knee at peak hip flexion and delay in timing of peak knee flexion in swing. Some of these results agree with the findings in the literature about the effect of the RF on gait, i.e. reduced range of knee flexion in early swing, reduced peak knee flexion in swing and delayed timing of peak knee flexion in swing.[24, 25] However, the data included in this study suggests that some other gait variables, including the maximum hip extension in stance and peak hip flexion in swing, should be considered when examining the effect of the RF on gait. Additionally, both the hip and the knee joints should be examined as a dominance effect may vary during different points/periods.

In Chapter Three, the benefits/importance of understanding the impact of the RF on gait has been demonstrated. This will enable clinicians to determine the most appropriate treatment plan. The catch angles/lengths obtained using the Duncan-Ely test were analysed and examined to understand the impact of these angles on gait. The findings of this study led to the next essential element to examine, which was the reliability of the most common measurement tools used to measure the muscle length and joint ROM.

Different designs of goniometers are commonly used to measure joint ROM and bi-articular muscle length and spasticity. Dynamic and true shortness of the muscles can be measured by measuring the available joint ROM. A literature review examining the reliability of different designs of goniometers was compiled. This review aimed to investigate the intratester and intertester reliability of the UG, EG and inclinometer to measure hip, knee and ankle joint ROM. In addition, the review aimed to examine how different factors influence measurement reliability. These factors included motion measured, joint measured, measurement procedure, subjects, goniometer design and number of testers. Understanding how these factors affect the reliability is essential and helpful in clinical settings. The literature review provided an overview of how the reliability changed with these factors (Table 4-4).

Twenty-nine studies were included in the review. Of these 29 studies, only five studies were found which investigated the reliability of the UG amongst subjects with CP.[247, 248, 265, 266, 271] One of these studies also examined the reliability of the inclinometer.[265] Amongst these studies, three studies included more than one type of CP and did not

report on each group separately, hence reducing the ability to understand how results change with different types of CP.[247, 265, 271] The other two studies specifically investigated the reliability of the UG amongst subjects with diplegic CP. Kilgour et al[248] examined the intratester reliability of the UG in the measurement of hip flexion, hip extension, knee extension and ankle dorsiflexion amongst subjects with diplegic CP. The reliability of the hip flexion and ankle dorsiflexion measurements ranged from good to excellent, while the reliability of the knee and hip extension measurements ranged from weak to excellent. Additionally, this study compared measurement reliability of the UG amongst healthy subjects to those with spastic diplegic CP and found equal intratester reliability in both groups.[248] Furthermore, Mutlu et al.[266] examined the intratester and intertester reliability of the UG for the measurement of hip flexion, hip extension, hip abduction, hip external rotation and ankle dorsiflexion. With the exception of hip abduction, intratester and intertester reliability for the motions measured were found to range from good to excellent reliability. The intratester reliability for the hip abduction measurements ranged from weak to good, while the intertester reliability for hip abduction measurement was found to be good. However, this study found a variation of 0-28º in intertester measurements obtained.[266] It has been stated that spasticity can increase measurement error.[265, 275, 287, 288] In contrast, another study concluded that a major cause of error was in defining the end range of the joint ROM rather than the presence of spasticity.[248] The evidence found from the literature about the reliability of the UG and inclinometer amongst subjects with CP is inconclusive. This indicates that clinicians must be cautious when interpreting the ICC values of the UG and inclinometer for assisting in clinical judgment amongst subjects with CP.[275, 287] Further research is required to understand

the reliability of different designs of the goniometer amongst subjects with diplegic CP and how different factors affect the reliability.

Overall, the ability to make direct comparisons between the studies was compromised because the number of testers, experience level, the number of sessions, the time between the sessions and subject position varied across the studies. Most studies indicated that the reliability was higher amongst healthy subjects in comparison to the reliability amongst subjects with pathology. A limited number of studies examined the reliability of measuring active motion. This decreased the ability to compare between the measurement of active and passive ROM. Generally, it was found that intratester reliability was higher than intertester reliability. Furthermore, the reliability varied from weak to excellent across the hip, knee and ankle joints due to the different joint characteristics and ease of identifying bony landmarks. Standardisation of the measurement procedure and prior training were found to increase measurement reliability. Additionally, a limited number of studies were found which compared between the different designs of the goniometers, which affected the ability to draw conclusions. No study was found which compared all 3 designs of goniometers. This review illustrated the gap in the current research about the UG, EG and inclinometer and the need to introduce a more reliable measuring tool.

The series of studies in Chapter Four aimed to test the feasibility of using a 2D video analysis system (*PnO CMD*) for passive joint ROM measurement during physical assessment. Initially, a study investigating the reliability of the *PnO CMD* in comparison to the UG amongst healthy subjects was performed. In this study, sagittal

plane passive ROM of the hip, knee and ankle joints was measured using both tools with/without markers. Additionally, a standardised measurement procedure was developed for this study (Appendix B). Intratester and intertester ICC values were calculated for both tools with/without markers. The highest ICC values were found for the *PnO CMD* with markers, which illustrated the reliability of this system and the measurement procedure used (Table 4-5). Additionally, a sub-study was done which aimed to establish the measurement error caused by marker displacement (1 cm) using the *PnO CMD*. The measurement error caused by marker displacement using the *PnO CMD* was found to be small and clinically acceptable. The SEM increased as the angle increased, indicating that the error is higher with the larger angles (Table 4-6). Following that, the accuracy of the *PnO CMD* with markers was tested in comparison to Vicon (benchmark) amongst healthy subjects. Hip, knee and ankle sagittal plane dynamic ROM was measured at 5 predefined points of gait. High ICC values were obtained using this system (Table 4-8). Finally, the reliability of the *PnO CMD* for measurement of joint ROM amongst subjects with diplegic CP was investigated. Hip, knee and ankle joint sagittal plane passive joint ROM with markers were measured using the same measurement procedure developed. Intratester and intertester ICC values were found to be high and above the satisfactory level (Table 4-9). Reliable passive joint ROM measurements can be obtained using this system amongst subjects with diplegic CP. These studies may help introduce the use of advanced technology in clinical practice. Further research can be performed to test the reliability and accuracy of the *PnO CMD* for bi-articular muscle length and spasticity measurements.

This research demonstrated that using 2D analysis systems in clinical settings has several potential benefits. The *PnO CMD* was found to be more reliable and accurate tool than the UG for all the joints measured. Additionally, from the Bland & Altman plots, it was noted that deviations from the mean were all within ±2° for high ICC values (<0.60). ICC values above 0.60 is also considered to be satisfactory for research purposes.[300] This indicates that an acceptable level of reproducibility can be achieved using the *PnO CMD*. The markers used can be constructed from cheap and affordable materials such as *Velcro*™, and the developed measurement procedure can be followed easily. The use of this system is applicable in the clinical environment as only one video camera is required to capture the sagittal plane ROM. Furthermore, the testers could measure reliably using this system following a training session which lasted for approximately one hour. This indicates that this system is easy to use and practical. The option to play back the video frame by frame aids in identifying the joint end ROM. Furthermore, this system allows for construction of customised reports where photos of the subjects and the joints measured can be added. This enables the clinicians to share the joint ROM measurements, the point at which the measurements were taken and the measurement procedure followed. The use of this system enhances communication between the allied health professionals.

5.2 Limitations of the research

The limitations of the individual studies were discussed in detail within each chapter. The below limitations are the main limitations of the whole thesis.

The sample included in this thesis will most likely not represent the wider diplegic CP population due to the heterogeneous nature of diplegic CP. Additionally, in this thesis the barefoot gait patterns were not considered due to the small sample, limiting the generalisability of the results. Variations in muscle spasticity/tone/length, surgical intervention, bone deformities, joint ROM and contractures were found between the participants, increasing the dissimilarities. Additionally, within some of the experiments, data was re-used; hence, appropriate comments regarding multiple testing have been added where appropriate within Chapters 2,3 and 4.

All the participants included in the study outlined in Chapter Two used solid AFOs which were prescribed by their local health centres. Information related to the ankle angle and the SVA were recorded. However, information related to the AFO-FC characteristics, such as material, trimlines and length of the sole plate, was not available. This limits the ability to understand the results in depth. Variability would have been expected between the AFOs because they were supplied by different local health services. However, this makes the results of the study more realistic, applicable and representative of real clinical environments.

The testers included in the study investigating the reliability of the *PnO CMD* and the UG amongst healthy subjects were final year prosthetic and orthotics students, while the testers included in the study investigating the reliability of the *PnO CMD* amongst subjects with diplegic CP were qualified and experienced clinicians. This may have influenced the results of both studies. Although students receive sufficient training to use the UG during their course and additional training was given for the use of the

PnO CMD, their experience is considered limited. In contrast, the qualified testers are experienced testers who are familiar with the measurement procedure, and this may have altered the ICC values.

5.3 Recommendations for future research

The research discussed in Chapter Two indicated positive results of optimisation of the temporal midstance shank kinematics using an AFO-FC on shank kinematics, thigh kinematics, FZ2 and GRF alignments in relation to the hip and knee joints in temporal midstance to late stance. However, these results were not clinically significant. This may be due to the AFO-FC allowing further shank inclination. Repeating this study with the optimum AFO-FC prescription may produce more positive results and the effect may be more apparent on the gait.

In this thesis, the barefoot patterns were not considered; rather, the barefoot temporal midstance SVA was considered. Participants were divided into two groups per their barefoot shank inclination. The study found that the correlation results between the temporal midstance SVA and the investigated variables in barefoot and with an AFO-FC differed between the two groups. Repeating this study and dividing the participants according to their barefoot gait patterns may produce different results.

The Duncan-Ely test is used to measure the rectus femoris length where the hip joint is positioned in anatomical neutral position and the knee joint is flexed to measure the available ROM. Looking at the phase of gait where the length of the RF has its greatest effect, the hip joint is not positioned in anatomical neutral position. Rather, the hip

joint moves from approximately 10º extension to flexion as the knee flexes further. On a theoretical basis, a short RF should limit the knee flexion if the hip is in extension. Although most of the limbs included did not achieve hip extension during gait, a significant relationship was seen between the catch angle/length measured in neutral position and the knee or hip at several points/periods presented. However, some relationships were found not significant, raising the following questions: will changing the hip position during the Duncan-Ely test effect the relationship between the measured catch angle/length and the hip and knee joints at different points/periods of gait? Will a catch angle/length measured in hip extension have a different effect on gait when compared with a catch angle/length measured at hip neutral? Further research is required to be able to understand the relationship between the catch angle/length measured and the hip or knee during gait.

The series of studies in Chapter Four demonstrated that reliable and accurate results of passive joint ROM can be achieved using the *PnO CMD*. The length of the bi-articular muscle can be obtained by measuring the available joint ROM. As discussed previously, the use of a 2D analysis system is beneficial. Further research is required to establish the reliability of the *PnO CMD* in the measurement of bi-articular muscle length/spasticity using a specialised test, such as the Duncan-Ely test.

This thesis focused on subjects with diplegic CP. As discussed in Chapter One, this form of CP is challenging due to the variation found between both limbs. Additionally, only subjects with diplegic CP were included in order to draw clear conclusions and

minimise the number of variables. Similar protocols used in this thesis could be applied on other populations to achieve similar aims.

5.4 Overall conclusions

The overarching aim of this thesis was to investigate the effect of optimisation of the temporal midstance shank kinematics with the use of an AFO- FC on several variables including shank kinematics, thigh kinematics, magnitude of FZ2 and GRF alignment in relation to hip and knee joint in temporal midstance to late stance. The understanding of the implication of, and the tools used to conduct assessment procedures and measurement processes were tested in the assessment of bi-articular muscles (specifically rectus femoris), to ascertain if results were predictive of presenting gait deviations and hence provide information to optimise treatment.

A series of studies were completed to achieve the overall aim. These studies included a study investigating the influence of optimisation of the temporal midstance shank kinematics with the use of an AFO-FC on several gait variables in temporal midstance to late stance. This study demonstrated that optimising shank kinematics in temporal midstance using an AFO-FC has a positive influence on optimising thigh kinematics in temporal midstance and on thigh and shank kinematics when FZ2 and maximum TVA occur. Additionally, this study found that optimising the shank kinematics with the use of AFO-FC may increase FZ2 and lead to reduction in the degree of Ben Lomonding. The results of this study indicated some effect of optimising the temporal midstance shank kinematics on improving the GRF alignment in relation to the hip and knee joints. A further analysis showed that several variables of gait may optimise

systemically when the temporal midstance shank kinematics is optimised . This systematic optimisation was found to differ based on the barefoot SVA. The correlation values in limbs with barefoot SVA less inclined than normal and normal suggested that the alignment of the GRF at the knee joint in temporal midstance and at the hip joint at FZ2 may optimise simultaneously with the temporal midstance shank kinematics is optimised. In contrast, the correlation values in limbs with the SVA more inclined than normal implied that FZ2, thigh inclination and GRF alignment at knee joint at the maximum TVA may optimise simultaneously when the temporal midstance shank kinematics is optimised.

This was followed by evaluation of the currently used assessment method and measurement tool for measurement of length and spasticity of a commonly affected bi-articular muscle, the RF. A study which aimed to investigate the relationship between the RF catch angle/length measured using the Duncan-Ely test with the hip or the knee joint at selected gait points/periods was completed. In limbs with catch angle/length<60°, the regression models for knee at maximum hip extension, peak knee flexion, knee at peak hip flexion and RHFES were significant/dominant. In contrast in limbs with catch angle/length≥60°, only the model for knee at peak hip flexion was found significant/dominant. Additionally, the results show that the effect of the dynamic shortness of the RF differs from true shortness of the RF on the hip or the knee joint. In limbs catch angle<60°), the regression models for range of hip flexion in early swing, peak knee flexion and knee at peak hip flexion were significant/dominant. In contrast, the regression for range of knee flexion in early swing was significant/dominant in limbs with catch angle≥60°. Meanwhile, in limbs

with length<60°, the only significant/dominant model was the model for hip at peak knee flexion, while and all the regression models were weak and not significant in limbs with length≥60°. Furthermore, the regression models for timing of peak knee flexion in all groups were weak and not significant. This study illustrated that a dominance relationship of the catch angle/length with the knee or the hip joint varies at different gait points/periods. This suggests that the RF can influence the knee or the hip joint. This study demonstrated the clinical implications of RF length/spasticity assessment on predicting the effect observed on the hip or the knee joint during gait. This understanding of the presented challenges will enhance improving the intervention plan.

After evaluating the commonly used measurement method for RF muscle length and spasticity, the commonly used measurement tool, the goniometer, was examined. A literature review investigating the reliability of different designs of goniometer, including the EG, UG and inclinometer, was completed. Additionally, this review examined how different factors can influence the reliability of the UG, EG and inclinometer. This review highlighted the variation found in the methodology employed between the studies. This variation reduced the ability to compare studies directly, as the number of testers, experience level, number of sessions, time between the sessions and subject position varied across the studies. Several factors were found to influence the reliability, which included motion measured, joint measured, measurement procedure, subjects, goniometer design and type of reliability. The literature review provided an overview of how the reliability changes with these factors. Additionally, this review demonstrated that the evidence for the reliability of

these tools is largely inconclusive, especially amongst subjects with diplegic CP. The presence of spasticity and defining the end range of the joint ROM have been identified as major causes of measurement error using the goniometer. Additionally, it has been stated that care should be taken when using the measurements obtained with the UG for assisting in clinical judgment. Stabilising the limb while measuring the muscle length, ROM and spasticity using the recommended test such as the Duncan-Ely is challenging. To minimise the measurement error and to be able to stabilise the limb, it has been recommended to involve two testers in the measurement procedure. This will allow one tester to stabilise the limb and the second tester to record the measurements. However, in busy clinical environments this is not feasible. The findings of this review illustrated that further research is required to investigate and compare the reliability of the UG, EG and inclinometer and the possibility of using protocols and technology to increase reliability when measuring joint ROM, especially amongst subjects with CP.

The final investigation of this thesis tested the feasibility of using a 2D analysis system for passive joint ROM measurements. Initially, a study investigating the reliability of the *PnO CMD* in comparison to the UG amongst healthy subjects was performed. In this study, sagittal plane passive ROM of the hip, knee and ankle joints was measured using both tools with/without markers. Intratester and intertester ICC values were calculated for both tools with/without markers. The highest ICC values were found for the *PnO CMD* with markers, which illustrated the reliability of this system. Following that, the accuracy of the *PnO CMD* with markers was tested in comparison to Vicon (benchmark) amongst healthy subjects. Hip, knee and ankle sagittal plane dynamic ROM were measured at 5 predefined points of gait. High ICC values with a moderate

to narrow CI range were found for all the points measured. Finally, the reliability of the *PnO CMD* for measurement of joint ROM amongst subjects with diplegic CP was investigated. Hip, knee, and ankle joint sagittal plane passive joint ROM with markers were measured. Intratester and intertester ICC values were found to be high and above the satisfactory level. The series of studies which investigated the *PnO CMD* reliability and accuracy concluded that this analysis system is accurate and reliable for passive joint ROM measurements during physical assessment.

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Appendices

Appendix A

Below is an example of the steps followed to run the linear regression as explained in Section 3.2.3.

Peak knee flexion (PKF)

Group C1: catch angle<60°

1- The knee (PKF) is set as an input, while the hip (HPKF) is set as an output.

Regression

Variables Entered/Removed[a]			
Model	Variables Entered	Variables Removed	Method
1	Limb*catch angle, Catch angle, PKF, Limb, Limb*PKF[b]	.	Enter

a. Dependent Variable: HPKF
b. All requested variables entered.

Model Summary				
Model	R	R Square	Adjusted R Square	Std. Error of the Estimate
1	.716[a]	.512	.164	5.89594

a. Predictors: (Constant), Limb*catch angle, Catch angle, PKF, Limb, Limb*PKF

ANOVA[a]						
Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	255.742	5	51.148	1.471	.309[b]
	Residual	243.335	7	34.762		
	Total	499.077	12			

a. Dependent Variable: HPKF
b. Predictors: (Constant), Limb*catch angle, Catch angle, PKF, Limb, Limb*PKF

Coefficients[a]								
Model		Unstandardised Coefficients		Standardised Coefficients	t	Sig.	95.0% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	17.036	17.539		.971	.364	-24.437	58.508
	PKF	-.178	.425	-.276	-.418	.688	-1.183	.827
	Limb	6.039	24.803	.486	.243	.815	-52.611	64.688
	Catch angle	.649	.385	.762	1.685	.136	-.262	1.559
	Limb*PKF	-.190	.512	-.874	-.371	.722	-1.401	1.021
	Limb*catch angle	.130	.601	.502	.217	.835	-1.292	1.553

a. Dependent Variable: HPKF

Limb is not significant; hence, it can be excluded along with the multiplied terms.

Regression

Entering only PKF and catch angle

Variables Entered/Removed[a]			
Model	Variables Entered	Variables Removed	Method
1	Catch angle, PKF[b]	.	Enter

a. Dependent Variable: HPKF
b. All requested variables entered.

Model Summary				
Model	R	R Square	Adjusted R Square	Std. Error of the Estimate
1	.688[a]	.473	.368	5.12874

a. Predictors: (Constant), Catch angle, PKF

ANOVA[a]						
Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	236.038	2	118.019	4.487	.041[b]
	Residual	263.039	10	26.304		
	Total	499.077	12			

a. Dependent Variable: HPKF
b. Predictors: (Constant), Catch angle, PKF

Coefficients[a]								
Model		Unstandardised Coefficients		Standardised Coefficients	t	Sig.	95.0% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	18.053	9.119		1.980	.076	-2.265	38.372
	PKF	-.262	.191	-.406	-1.370	.201	-.687	.164
	Catch angle	.739	.252	.868	2.930	.015	.177	1.301

a. Dependent Variable: HPKF

Only catch angle is significant; hence, PKF can be excluded.

Regression

Entering catch angle only

Variables Entered/Removed[a]			
Model	Variables Entered	Variables Removed	Method
1	Catch angle[b]	.	Enter

a. Dependent Variable: HPKF
b. All requested variables entered.

Model Summary				
Model	R	R Square	Adjusted R Square	Std. Error of the Estimate
1	.612[a]	.374	.317	5.32918

a. Predictors: (Constant), Catch angle

ANOVA[a]						
Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	186.675	1	186.675	6.573	.026[b]
	Residual	312.402	11	28.400		
	Total	499.077	12			

a. Dependent Variable: HPKF
b. Predictors: (Constant), Catch angle

Coefficients[a]								
Model		Unstandardised Coefficients		Standardised Coefficients	t	Sig.	95.0% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	14.441	9.070		1.592	.140	-5.523	34.405
	Catch angle	.521	.203	.612	2.564	.026	.074	.967

a. Dependent Variable: HPKF

Overall, the regression is significant, but the R^2 is weak (37.4%).

2- The hip (HPKF) is set as an input, while the knee (PKF) is set as an output

Regression

Variables Entered/Removed[a]			
Model	Variables Entered	Variables Removed	Method
1	Limb*HPKF, Catch angle, HPKF, Limb, limb*catch angle	.	Enter

a. Dependent Variable: PKF

b. All requested variables entered.

Model Summary				
Model	R	R Square	Adjusted R Square	Std. Error of the Estimate
1	.809[a]	.655	.408	7.69716

a. Predictors: (Constant), Limb*HPKF, HPKF, Catch angle, Limb, Limb*catch angle

ANOVA[a]						
Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	786.353	5	157.271	2.655	.118[b]
	Residual	414.724	7	59.246		
	Total	1201.077	12			

a. Dependent Variable: PKF

b. Predictors: (Constant), Limb*HPKF, HPKF, Catch angle, Limb, Limb*catch angle

Coefficients[a]								
Model		Unstandardised Coefficients		Standardised Coefficients	t	Sig.	95.0% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	28.379	19.111		1.485	.181	-16.810	73.568
	Catch angle	.599	.529	.453	1.132	.295	-.652	1.849
	Limb	5.392	32.638	.280	.165	.873	-71.785	82.569
	HPKF	-.188	.570	-.121	-.329	.752	-1.535	1.160
	Limb*Catch angle	.912	.768	2.263	1.187	.274	-.904	2.727
	Limb*HPKF	-1.069	.895	-2.218	-1.195	.271	-3.185	1.047

a. Dependent Variable: PKF

Limb is not significant hence it can be excluded along with the multiplied terms.

Regression

Entering only HPKF and catch angle

Variables Entered/Removed[a]			
Model	Variables Entered	Variables Removed	Method
1	HPKF, Catch angle[b]	.	Enter

a. Dependent Variable: PKF
b. All requested variables entered.

Model Summary				
Model	R	R Square	Adjusted R Square	Std. Error of the Estimate
1	.703[a]	.495	.394	7.78891

a. Predictors: (Constant), HPKF, Catch angle

ANOVA[a]						
Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	594.406	2	297.203	4.899	.033[b]
	Residual	606.671	10	60.667		
	Total	1201.077	12			

a. Dependent Variable: PKF
b. Predictors: (Constant), HPKF, Catch angle

Coefficients[a]								
Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95.0% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	23.601	15.528		1.520	.160	-10.998	58.200
	Catch angle	1.156	.394	.880	2.931	.015	.277	2.035
	HPKF	-.649	.520	-.375	-1.248	.241	-1.808	.510

a. Dependent Variable: PKF

Only catch angle is significant; hence, HPKF can be excluded.

Regression

Entering catch angle only

Variables Entered/Removed[a]			
Model	Variables Entered	Variables Removed	Method
1	Catch angle	.	Enter

a. Dependent Variable: PKF
b. All requested variables entered.

Model Summary				
Model	R	R Square	Adjusted R Square	Std. Error of the Estimate
1	.633[a]	.400	.346	8.09331

a. Predictors: (Constant), Catch angle

ANOVA[a]						
Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	480.558	1	480.558	7.337	.02
	Residual	720.519	11	65.502		
	Total	1201.077	12			

a. Dependent Variable: PKF
b. Predictors: (Constant), Catch angle

Coefficients[a]								
Model		Unstandardised Coefficients		Standardised Coefficients	t	Sig.	95.0% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	13.803	13.775		1.002	.338	-16.516	44.121
	Catch angle	.835	.308	.633	2.709	.020	.157	1.514

a. Dependent Variable: PKF

Overall, the regression is significant, but the R^2 is acceptable (40%). The model of the PKF is taken forward as the R^2 is larger in comparison to the model of the HPKF.

The output equation is:

Average PKF=13.803 + 0.835*catch angle

Rounding it up=13.8+0.84*catch angle

Appendix B

Universal goniometer instruction manual for testers

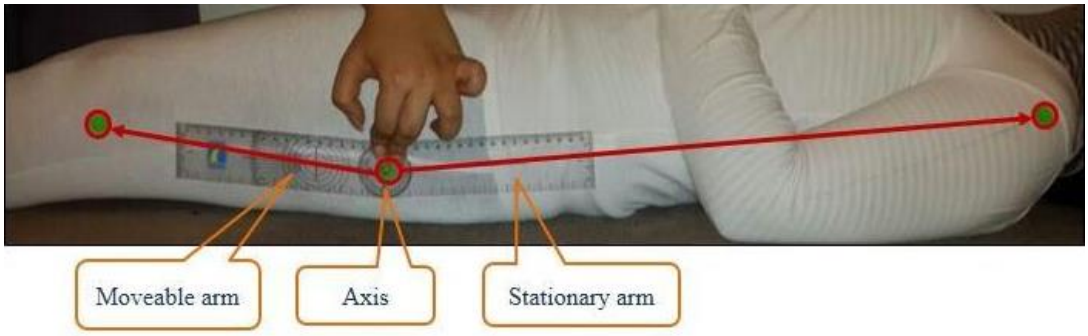
Measuring with markers on bony landmarks

1-UG alignment for joint ROM measurement:

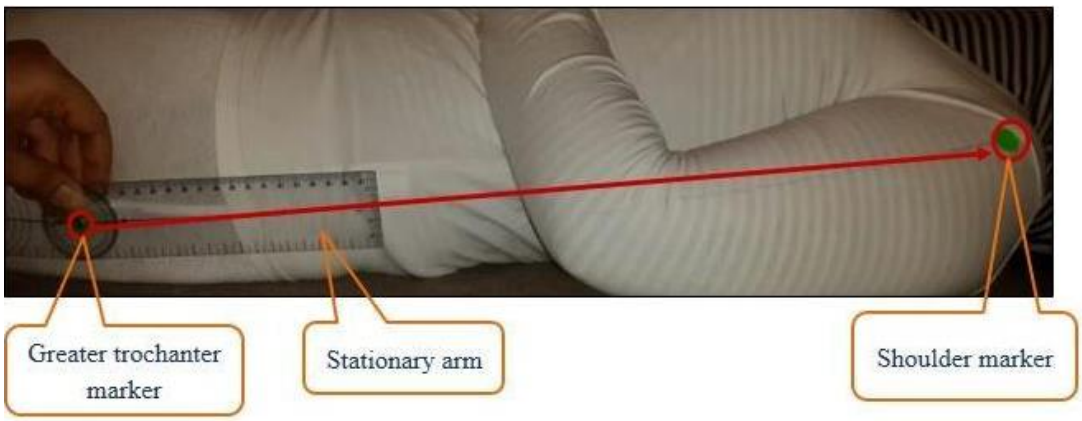
This is a method of measuring with UG which may be different to your clinical practice, therefore please read the demonstration section below carefully

A-Hip (use this alignment when measuring flexion and extension):

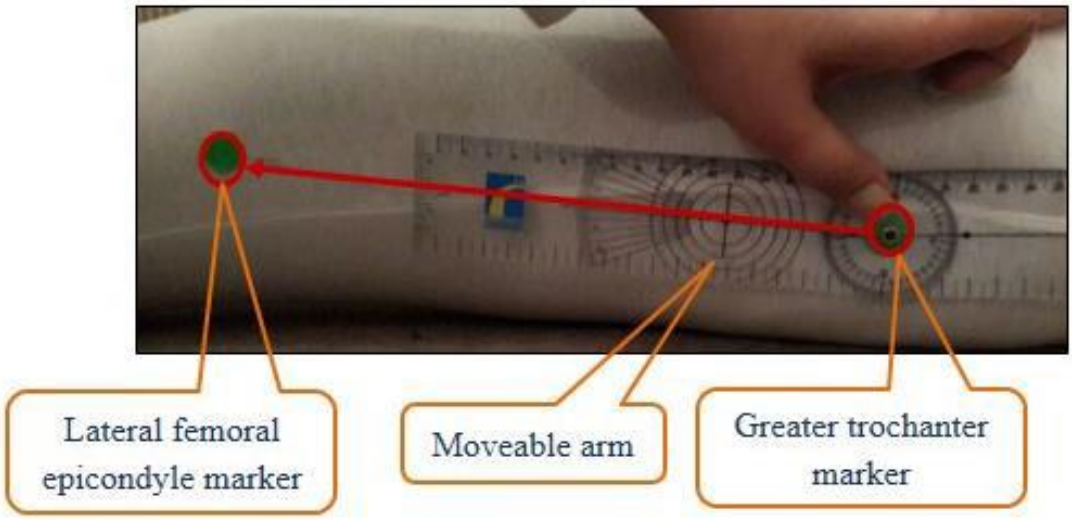
Axis: on the marker on the greater trochanter of the femur as shown below



Stationary arm: pointing towards the marker on the shoulder as shown below

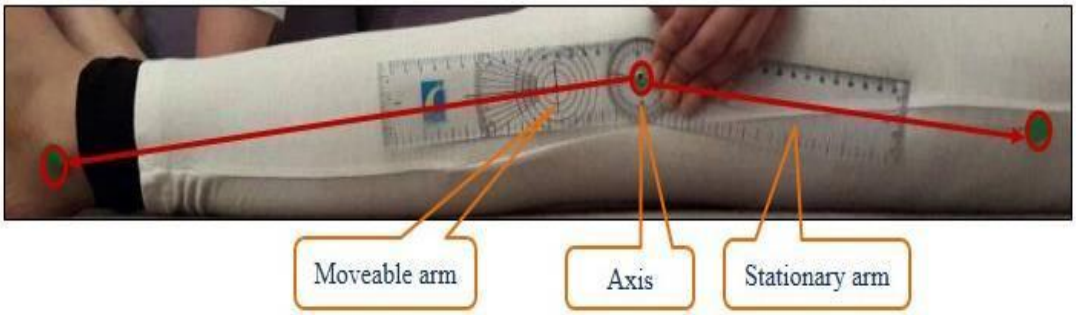


Moveable arm: pointing towards the marker on the lateral femoral epicondyle as shown below

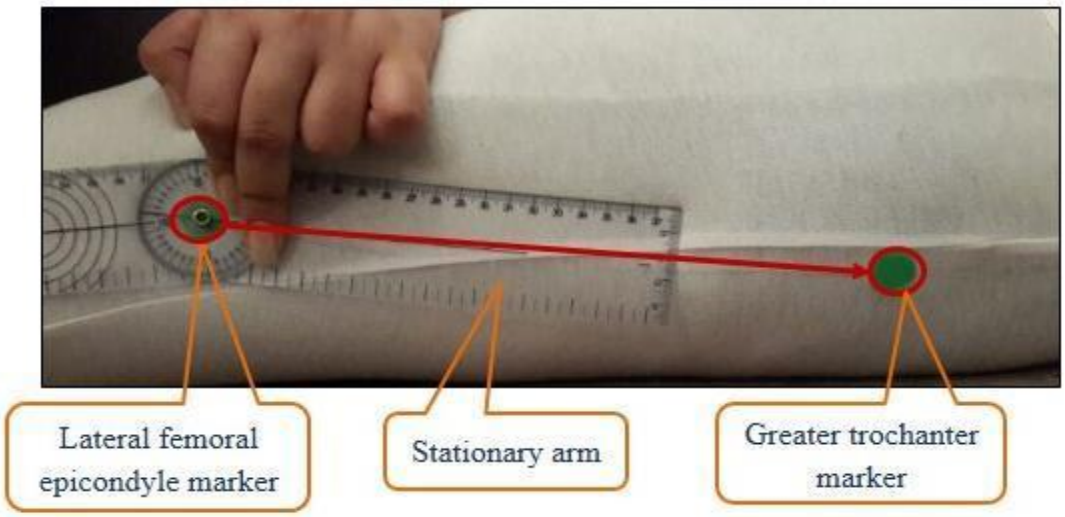


B- Knee (use this alignment when measuring flexion and extension):

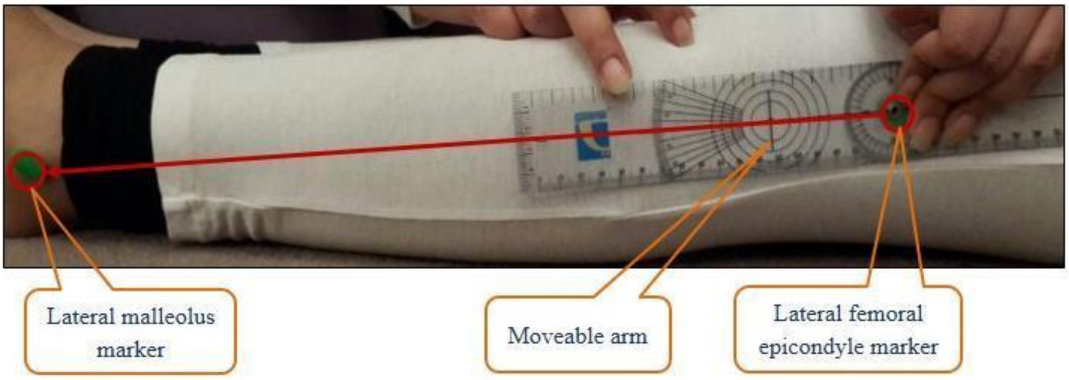
Axis: on the marker on the lateral epicondyle of the femur as required



Stationary arm: pointing towards the marker on the greater trochanter as shown below

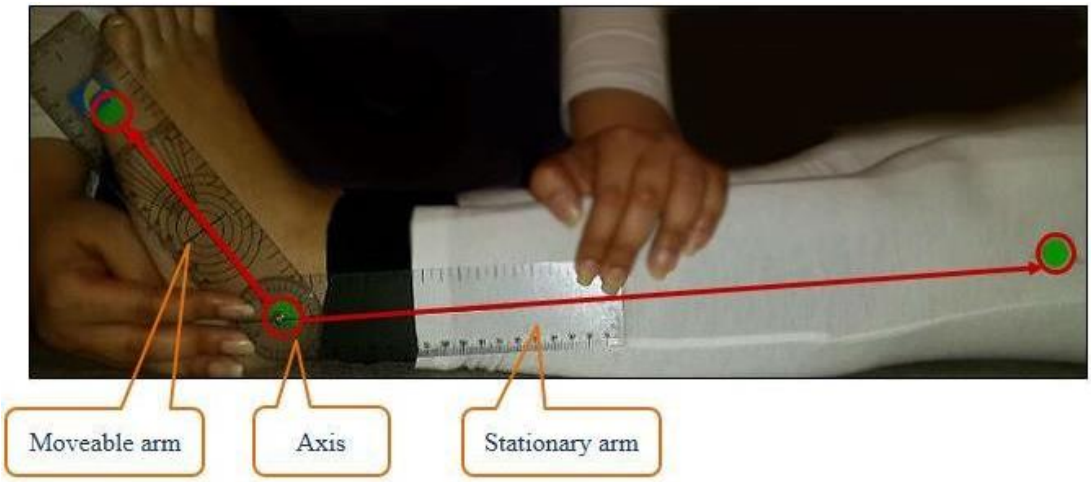


Moveable arm: pointing towards the marker on the lateral malleolus as shown below

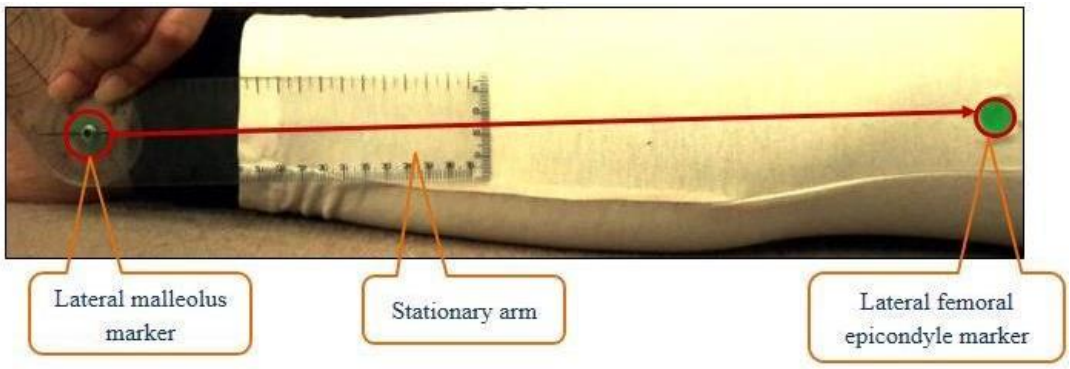


C- Ankle (use this alignment when measuring plantarflexion and dorsiflexion):

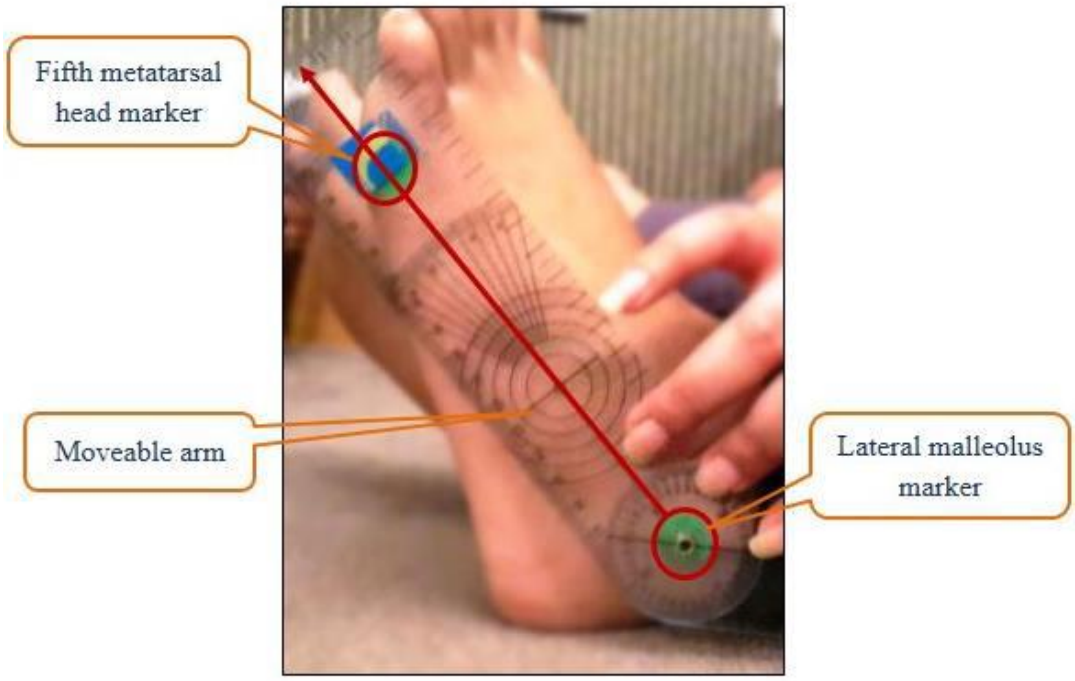
Axis: on the marker on the lateral malleolus as shown below



Stationary arm: pointing towards the marker on the lateral femoral epicondyle as shown below



Moveable arm: pointing towards the marker on the fifth metatarsal head



2- Measuring procedure:

A- Subject in supine position, measure:

Hip joint flexion ROM:

- Start position: subject in supine position with the hip (0° abduction, adduction, and rotation) and knee in neutral position (0°)
- Stabilisation: Stabilise the pelvis by placing one hand on the ipsilateral pelvis (the side being measured), for additional support keep the contralateral limb flat in neutral position
- Motion: lift the thigh and flex the hip allowing the knee to flex as well while maintaining the hip in neutral position. Apply pressure until the maximum hip flexion is achieved
- End feel: soft or firm. The hip should not be flexed beyond the point where it causes posterior tilting of the pelvis
- Align the goniometer as shown above and take the measurement
- Record the measurement in the appropriate space in the UG measurements recording sheet provided

Knee joint flexion ROM:

- Start position: subject in supine position with the hip (0° abduction, adduction, and rotation) and knee in neutral position (0°)
- Stabilisation: Stabilise the femur with one hand to prevent rotation, abduction and adduction of the hip
- Motion: lift the thigh to flex the hip to approximately 90° and stabilise the thigh to prevent further motion while moving the knee joint into flexion.
- End feel: soft or firm, the knee should not be flexed beyond the point where it causes further hip flexion
- Align the goniometer as shown above and take the measurement
- Record the measurement in the appropriate space in the UG measurements recording sheet provided

Knee joint extension ROM:

- Start position: subject in supine position with the hip (0° abduction, adduction, and rotation) and knee in neutral position (0°). A towel is placed under the ankle to allow maximum extension
- Stabilisation: None
- Motion: Apply pressure above the knee to extend
- End feel: Firm
- Align the goniometer as shown above and take the measurement
- Record the measurement in the appropriate space in the UG measurements recording sheet provided

Ankle joint plantarflexion ROM:

- Stabilisation: grasp the posterior aspect of the calcaneus (heel) and hold the joint in neutral position maintaining the knee in full extension with the foot perpendicular to the lower leg
- Motion: with the other hand hold the forefoot at the level of the metatarsal heads and apply pressure to push the ankle into plantarflexion, maintain the subtalar joint in neutral position
- End feel: Firm or hard
- Align the goniometer as shown above and take the measurement
- Record the measurement in the appropriate space in the UG measurements recording sheet provided

Ankle joint dorsiflexion:

- Stabilisation: grasp the posterior aspect of the calcaneus (heel) and hold the joint in neutral position maintaining the knee in full extension with the foot perpendicular to the lower leg
- Motion: with the other hand hold the forefoot at the level of the metatarsal heads and apply pressure to push the ankle into dorsiflexion, maintain the subtalar joint in neutral position
- End feel: firm or hard
- Align the goniometer as shown above and take the measurement
- Record the measurement in the appropriate space in the UG measurements recording sheet provided

B- Move the subject to prone position and measure:

Hip joint extension ROM:

- Start position: with hip (0° abduction, adduction, and rotation) and knee in neutral position (0°)
- Stabilisation: Stabilise the pelvis by placing one hand on the ipsilateral pelvis (the side being measured), for additional support keep the contralateral limb flat in neutral position
- Motion: lift the thigh and extend the hip while maintaining the knee in an extended position and the hip in neutral position. Apply pressure until the maximum hip extension is achieved
- End feel: Firm. The hip should not be extended beyond the point where it causes anterior tilt of the pelvis or extension of the lumbar spine
- Align the goniometer as shown above and take the measurement
- Record the measurement in the appropriate space in the UG measurements recording sheet provided

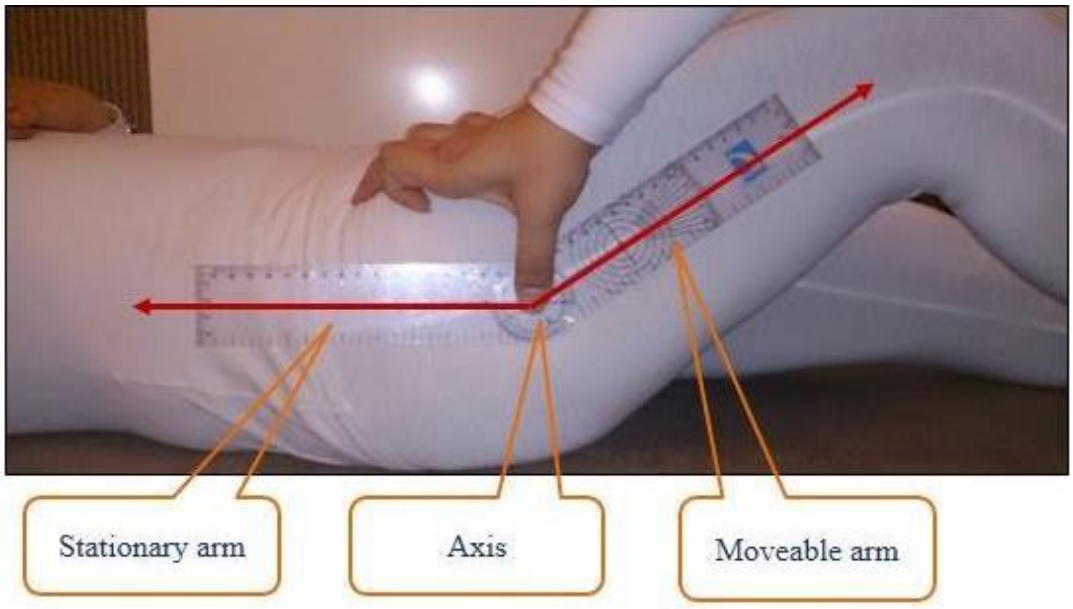
Measuring with no markers

1-UG alignment for joint ROM measurement

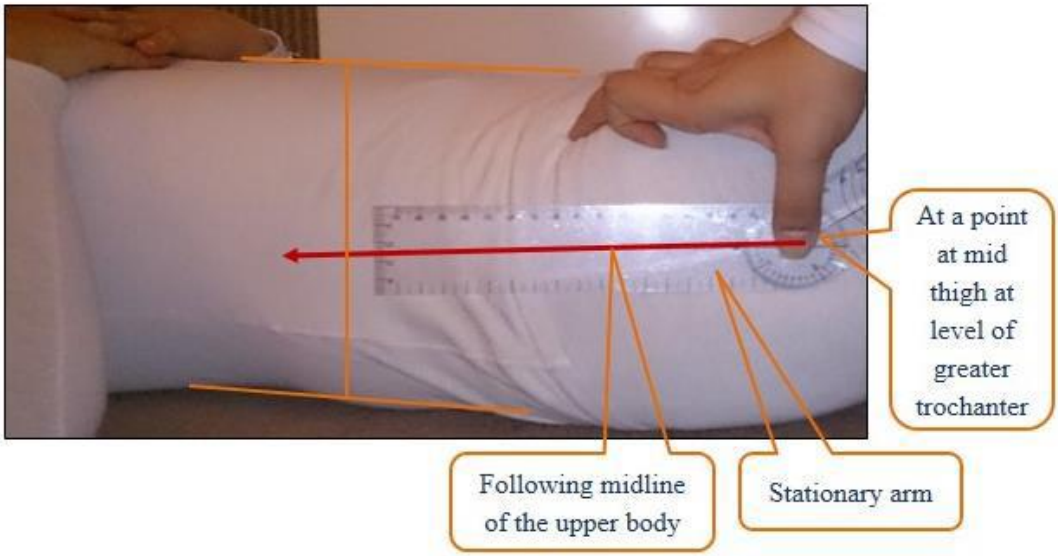
This is a method of measuring with UG which may be different to your clinical practice, therefore please read the demonstration section below carefully

A- Hip (use this alignment when measuring flexion and extension):

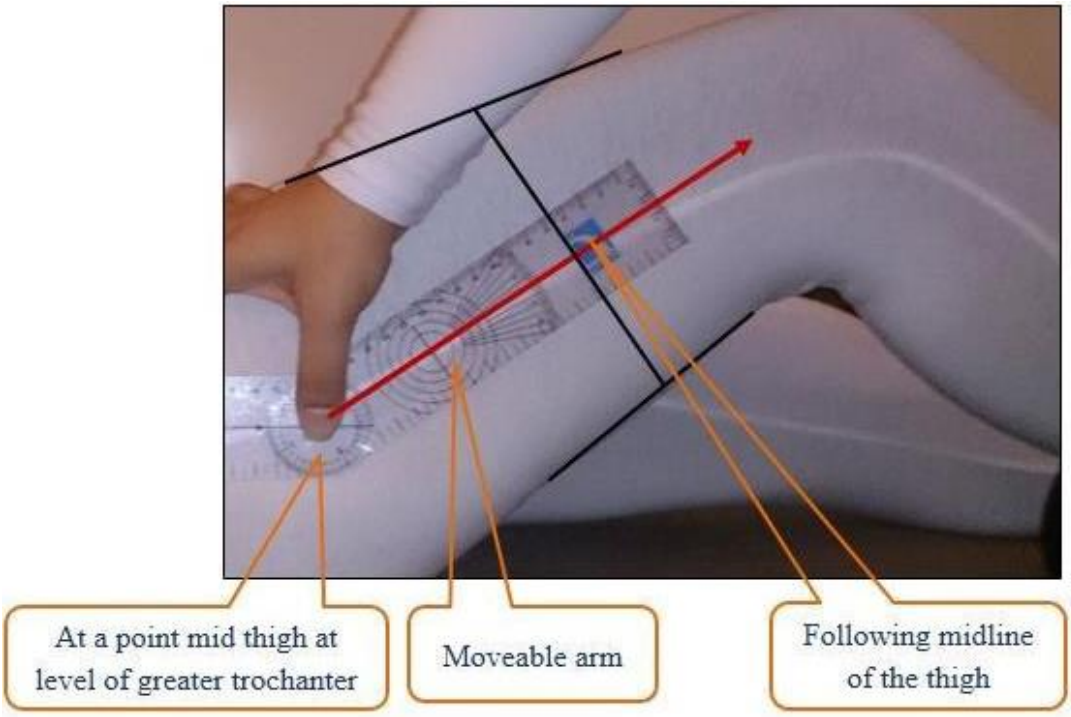
Axis: at a point at the mid-thigh at the level of greater trochanter as shown below



Stationary arm: following midline of the upper body as shown below

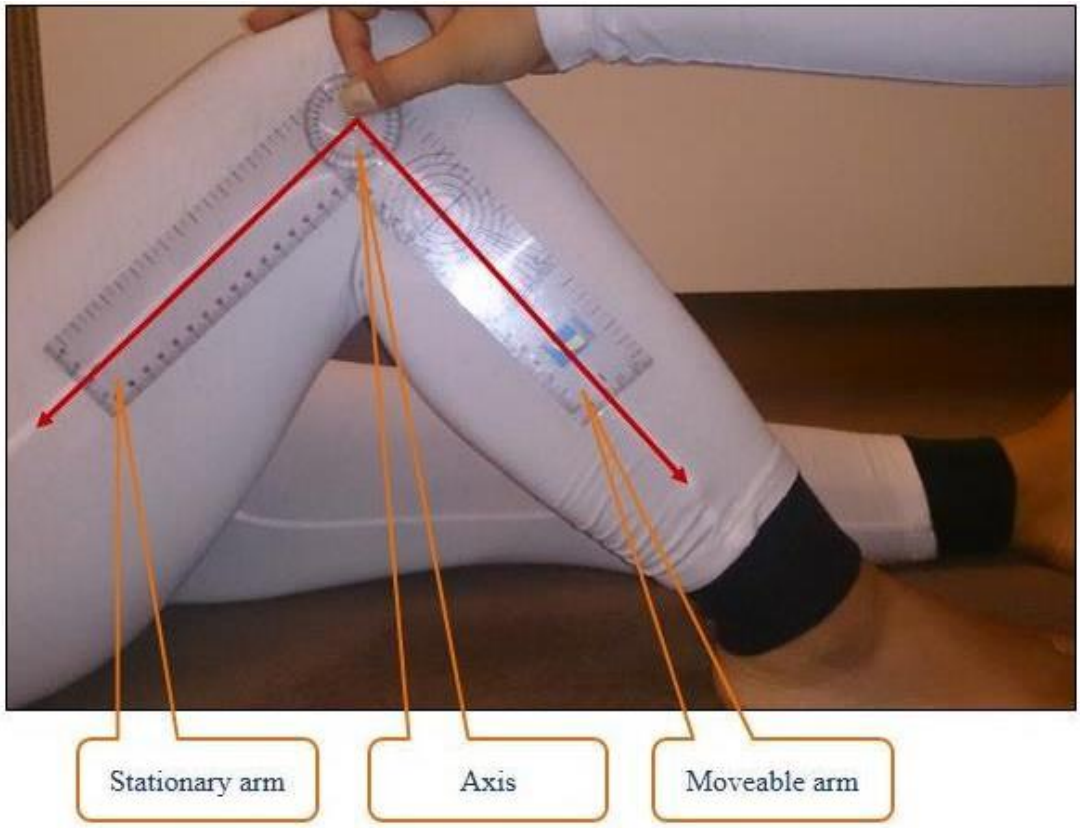


Moveable arm: following midline of thigh as shown below

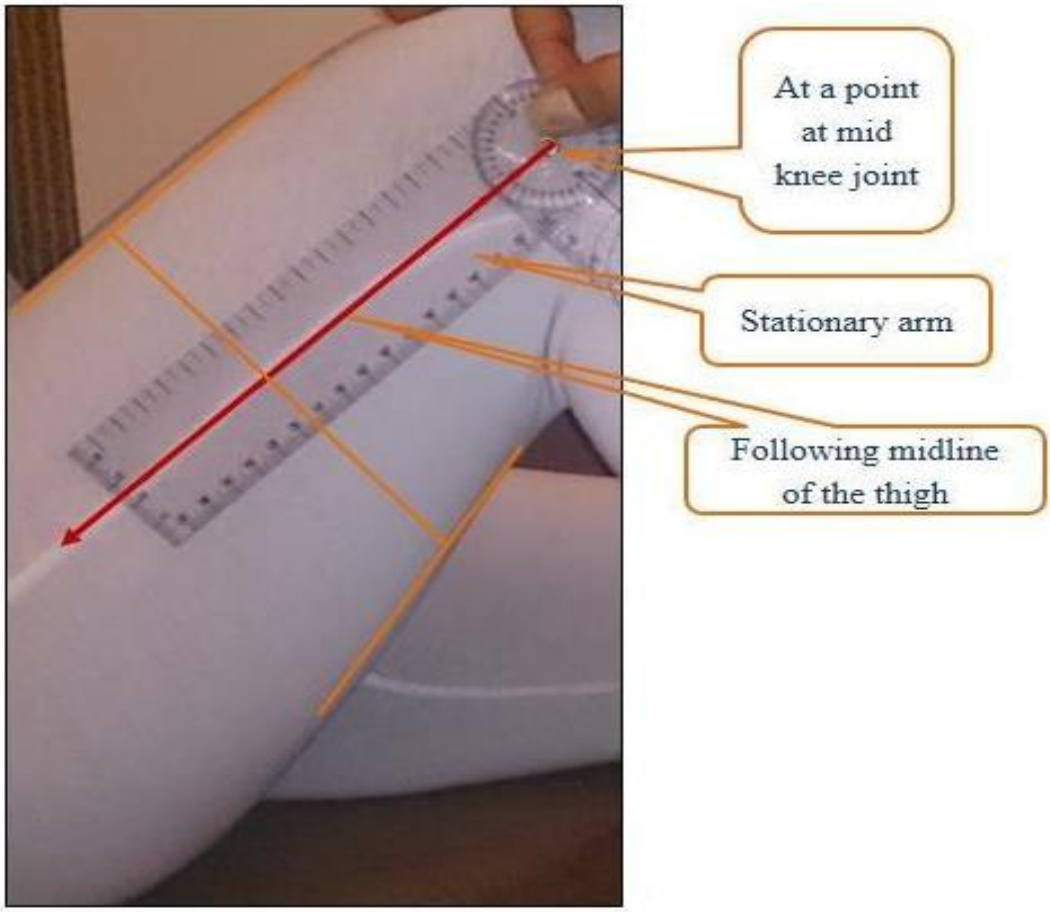


B- Knee (use this alignment when measuring flexion and extension):

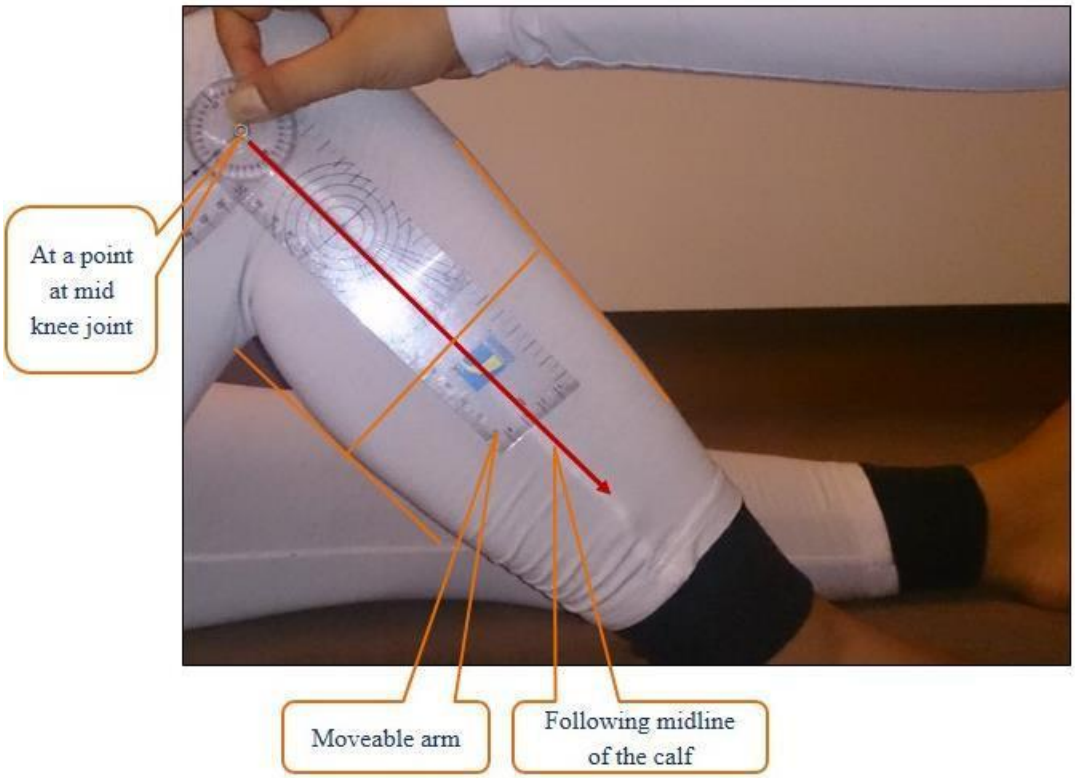
Axis: at a point at the mid knee joint as shown below



Stationary arm: following midline of the thigh as shown below

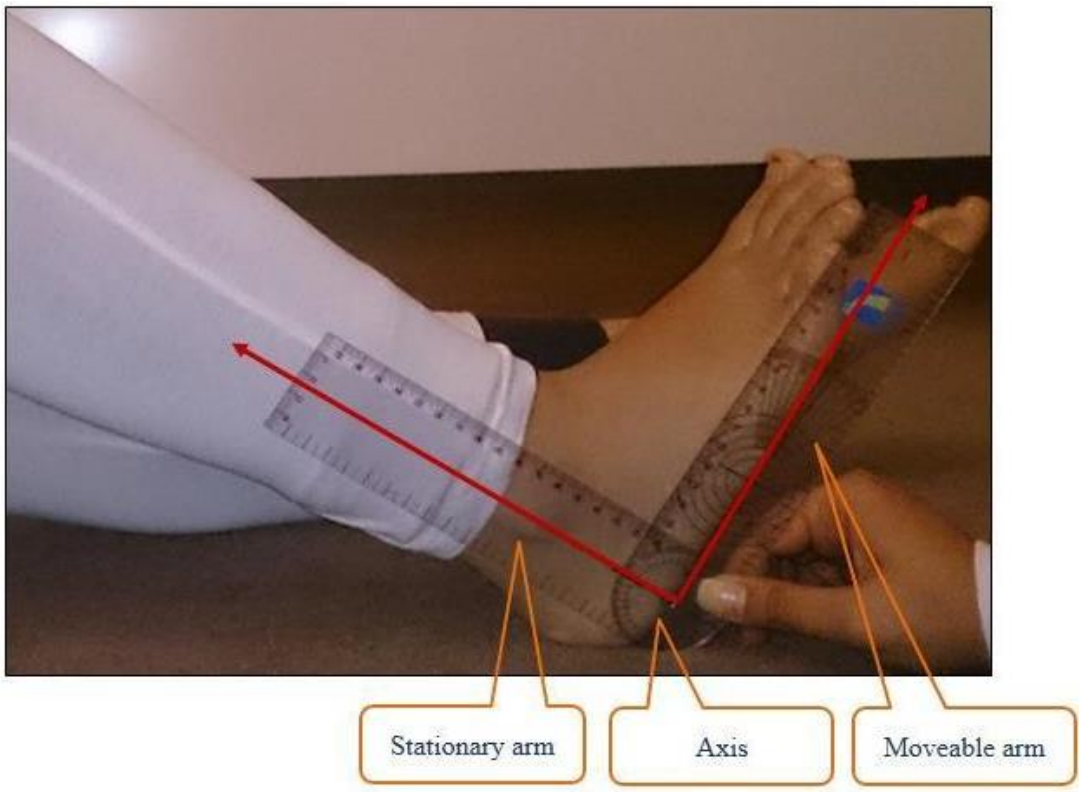


Moveable arm: following midline of the calf as shown below

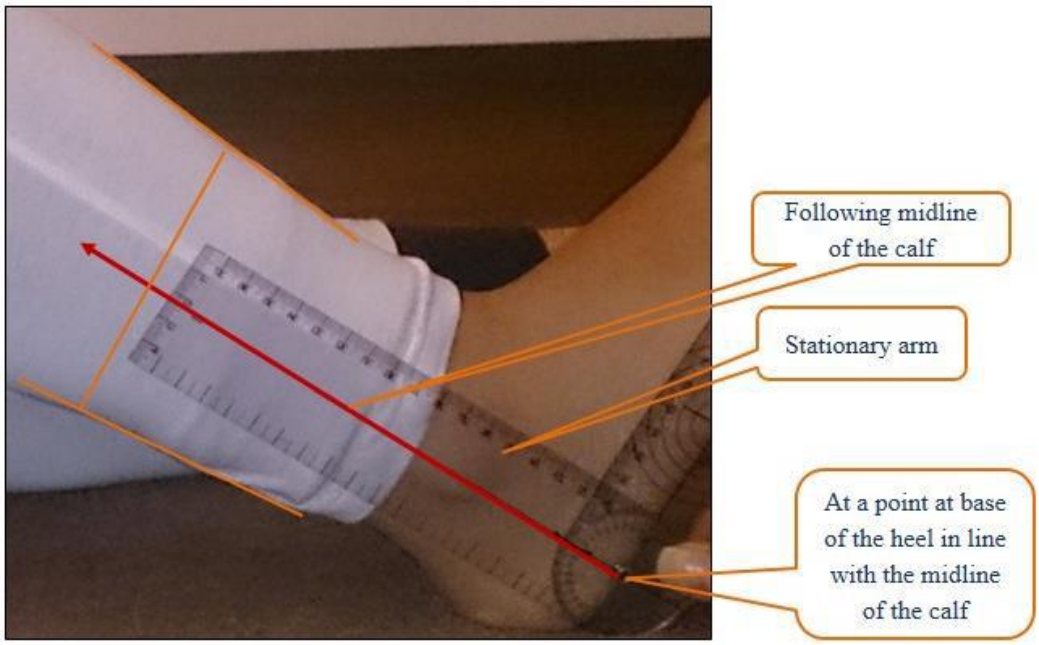


C- Ankle (use this alignment when measuring plantarflexion and dorsiflexion):

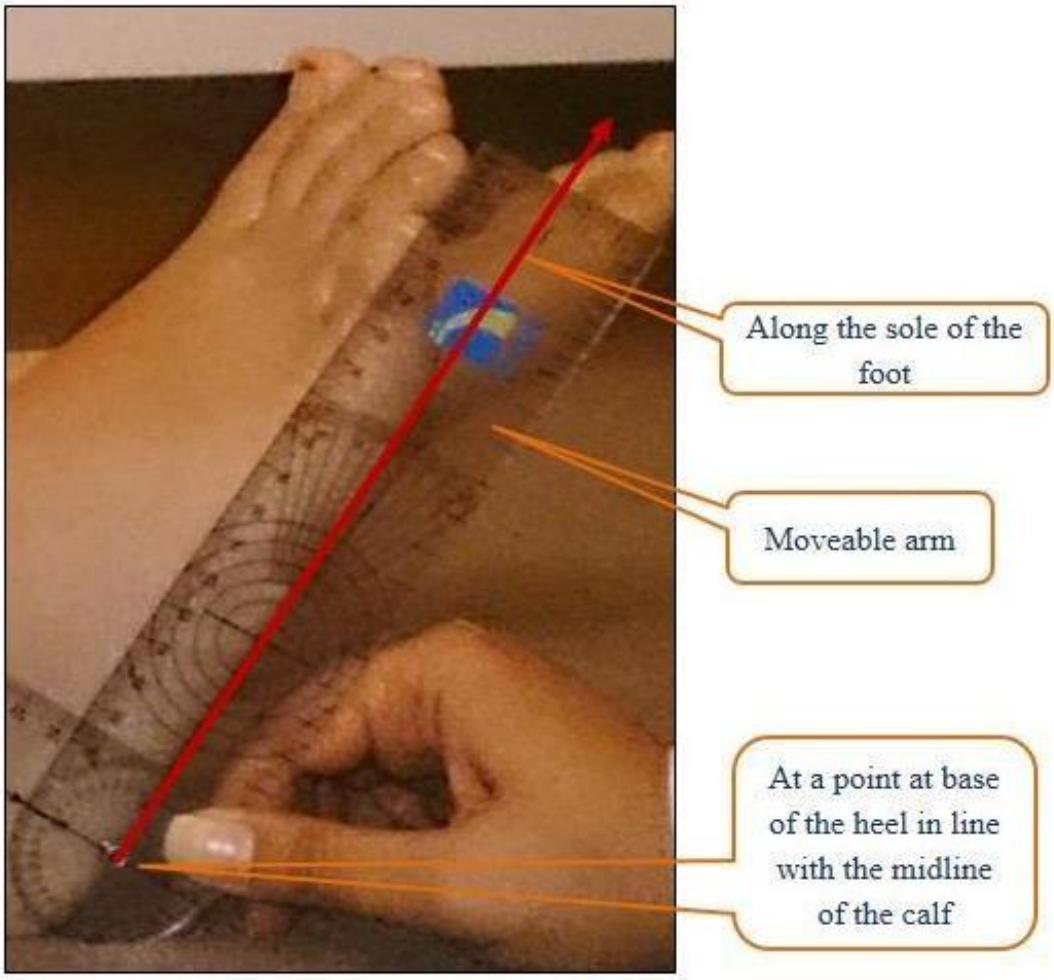
Axis: at a point at the base of the heel in line with midline of the calf as shown below



Stationary arm: following the midline of the calf as shown below



Moveable arm: along the sole of the foot as shown below



2- Measuring procedure:

A- Subject in supine position, measure:

Hip joint flexion ROM:

- Start position: subject in supine position with the hip (0° abduction, adduction, and rotation) and knee in neutral position (0°)
- Stabilisation: Stabilise the pelvis by placing one hand on the ipsilateral pelvis (the side being measured), for additional support keep the contralateral limb flat in neutral position
- Motion: lift the thigh and flex the hip allowing the knee to flex as well while maintaining the hip in neutral position. Apply pressure until the maximum hip flexion is achieved
- End feel: soft or firm. The hip should not be flexed beyond the point where it causes posterior tilting of the pelvis
- Align the goniometer as shown above and take the measurement
- Record the measurement in the appropriate space in the UG measurements recording sheet provided

Knee joint flexion ROM:

- Start position: subject in supine position with the hip (0° abduction, adduction, and rotation) and knee in neutral position (0°)
- Stabilisation: Stabilise the femur with one hand to prevent rotation, abduction and adduction of the hip
- Motion: lift the thigh to flex the hip to approximately 90° and stabilise the thigh to prevent further motion while moving the knee joint into flexion.

- End feel: soft or firm, the knee should not be flexed beyond the point where it causes further hip flexion
- Align the goniometer as shown above and take the measurement
- Record the measurement in the appropriate space in the UG measurements recording sheet provided

Knee joint extension ROM:

- Start position: subject in supine position with the hip (0° abduction, adduction, and rotation) and knee in neutral position (0°). A towel is placed under the ankle to allow maximum extension
- Stabilisation: None
- Motion: Apply pressure above the knee to extend
- End feel: Firm
- Align the goniometer as shown above and take the measurement
- Record the measurement in the appropriate space in the UG measurements recording sheet provided

Ankle joint plantarflexion ROM:

- Stabilisation: grasp the posterior aspect of the calcaneus (heel) and hold the joint in neutral position maintaining the knee in full extension with the foot perpendicular to the lower leg

- Motion: with the other hand hold the forefoot at the level of the metatarsal heads and apply pressure to push the ankle into plantarflexion, maintain the subtalar joint in neutral position
- End feel: Firm or hard
- Align the goniometer as shown above and take the measurement
- Record the measurement in the appropriate space in the UG measurements recording sheet provided

Ankle joint dorsiflexion:

- Stabilisation: grasp the posterior aspect of the calcaneus (heel) and hold the joint in neutral position maintaining the knee in full extension with the foot perpendicular to the lower leg
- Motion: with the other hand hold the forefoot at the level of the metatarsal heads and apply pressure to push the ankle into dorsiflexion, maintain the subtalar joint in neutral position
- End feel: firm or hard
- Align the goniometer as shown above and take the measurement
- Record the measurement in the appropriate space in the UG measurements recording sheet provided

B- Move the subject to prone position and measure:

Hip joint extension ROM:

- Start position: with hip (0° abduction, adduction, and rotation) and knee in neutral position (0°)
- Stabilisation: Stabilise the pelvis by placing one hand on the ipsilateral pelvis (the side being measured), for additional support keep the contralateral limb flat in neutral position
- Motion: lift the thigh and extend the hip while maintaining the knee in an extended position and the hip in neutral position. Apply pressure until the maximum hip extension is achieved
- End feel: Firm. The hip should not be extended beyond the point where it causes anterior tilt of the pelvis or extension of the lumbar spine
- Align the goniometer as shown above and take the measurement
- Record the measurement in the appropriate space in the UG measurements recording sheet provided

PnO Clinical Movement Data instruction manual for testers

A- Measurements required each session:

- 16 hip joints maximum flexion and 16 hip joints maximum extension measurements
- 16 knee joints maximum flexion and 16 knee joints maximum extension measurements
- 16 ankle joints maximum plantarflexion and 16 ankle joints maximum dorsiflexion measurements

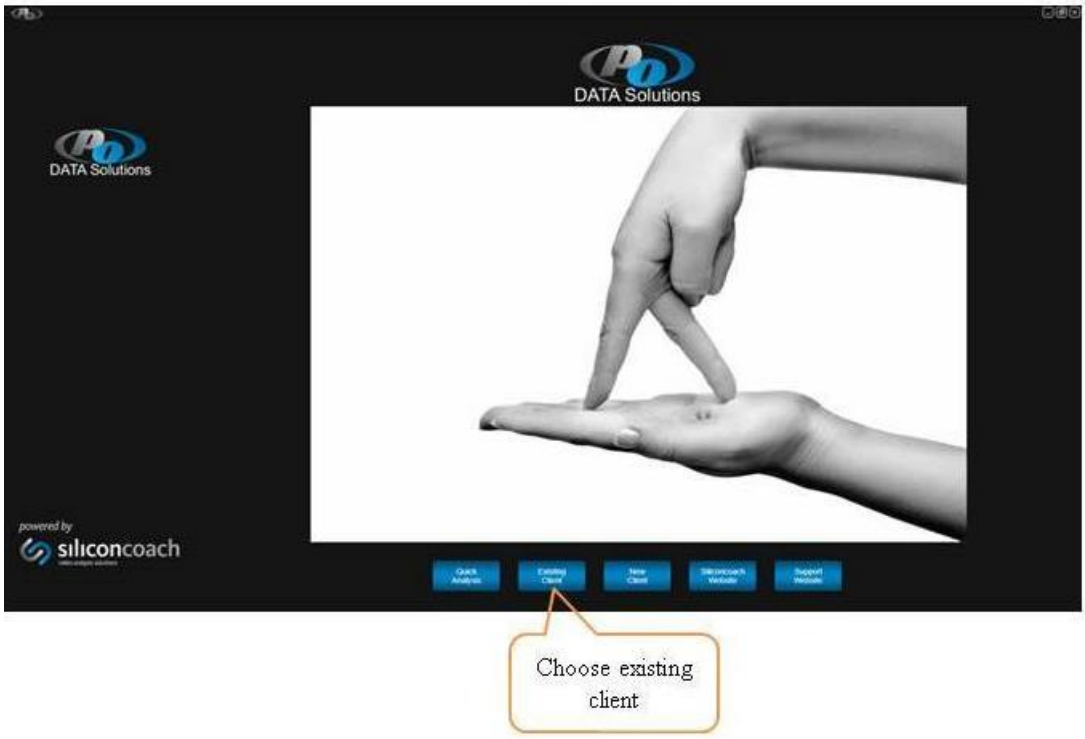
B- Turning on the computer:

- Open the steel doors on the bottom of the trolley
- Turn on the system unit by pressing the button on the top right (look under beneath the door opening)
- Turn on the monitor
- Press enter

C- Opening the software:

- On the desktop double click on P + O Clinical Movement Data Software.
- Choose Existing Client on the welcome screen as shown below

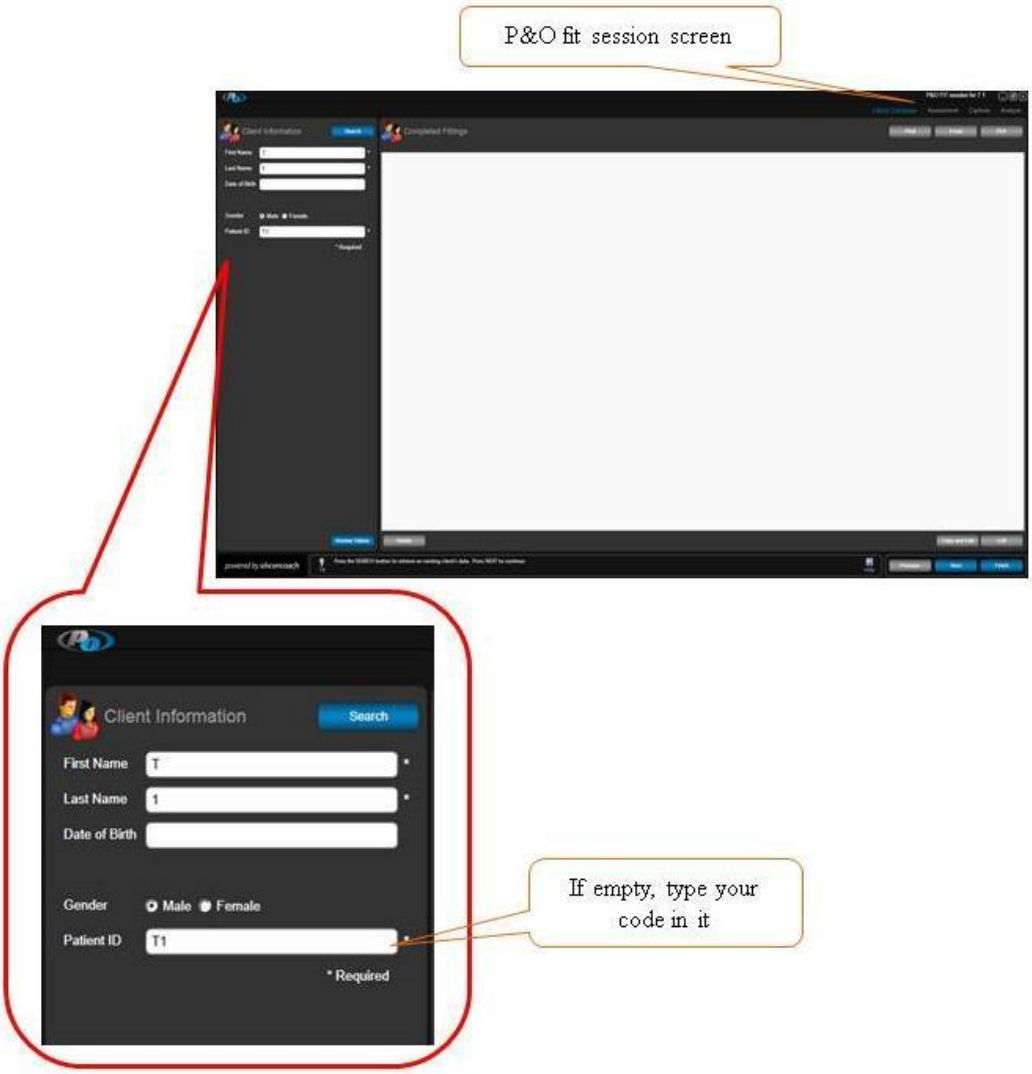
Welcome screen



3. In the search box, type the following as shown in the image below

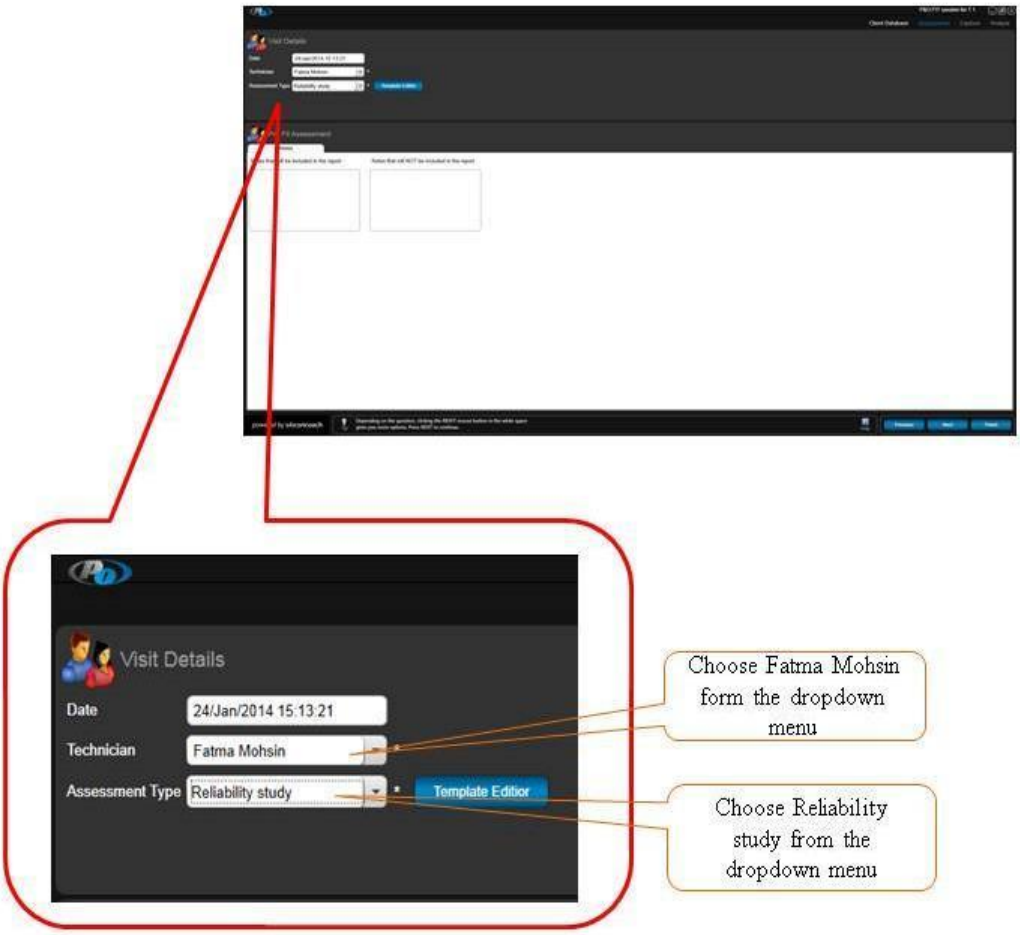
4. Click ok

- This will take you to another page called P&O fit session (written in the extreme top right)
- On the left side of client information, look for Patient ID
- Patient ID is your code; if the space is empty type your code in it as shown below



8. Click next

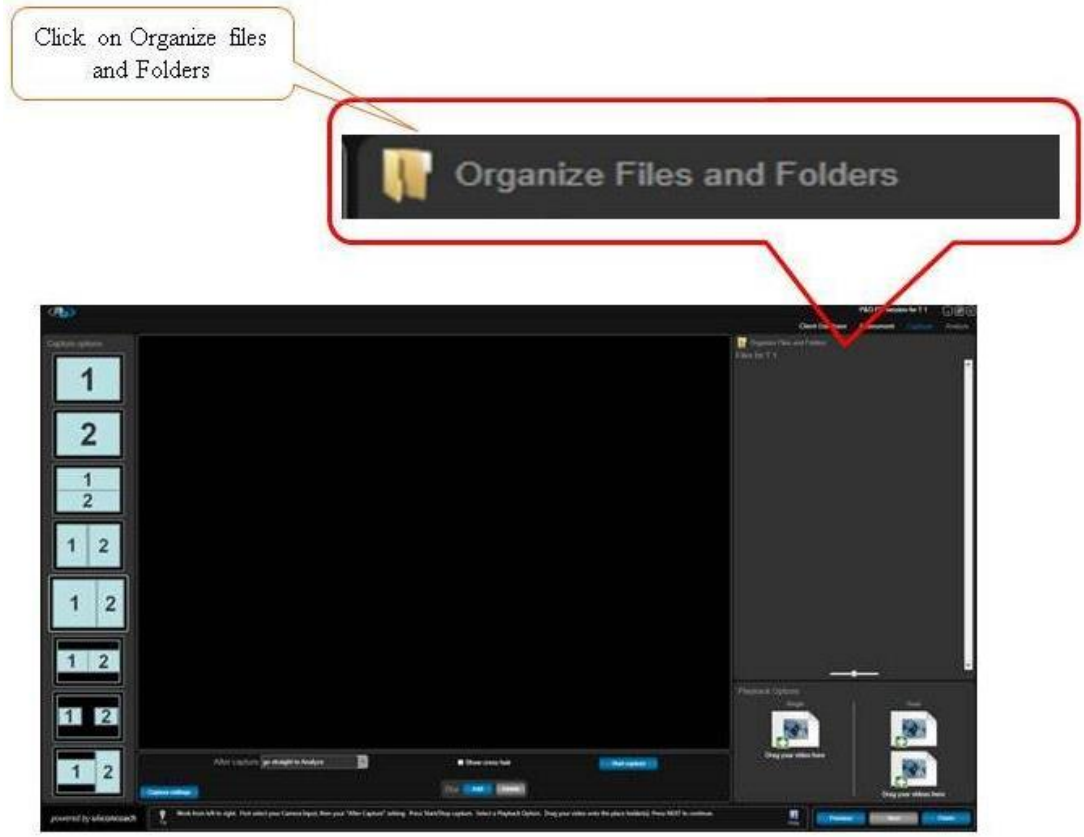
- From Technician drop down menu choose – Fatma Mohsin
- From Assessment Type drop down menu choose – Reliability study as shown below



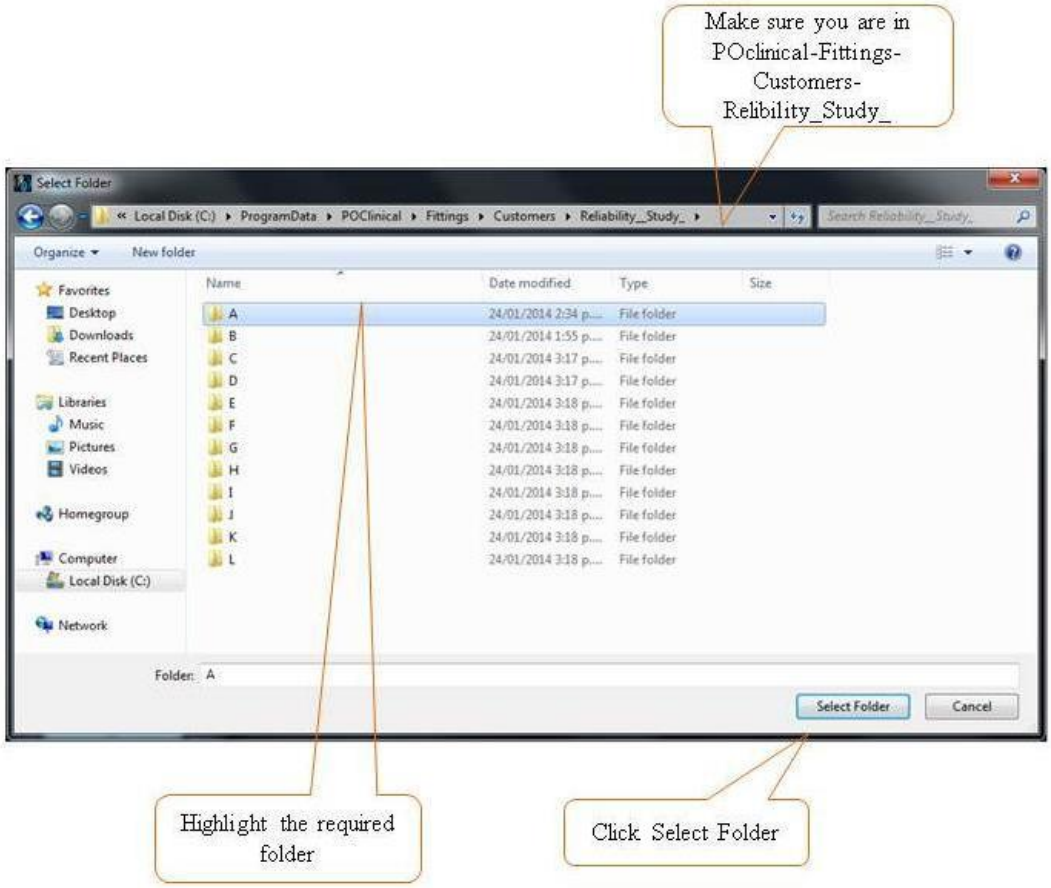
11. Click next

D- Opening the videos and analysing:

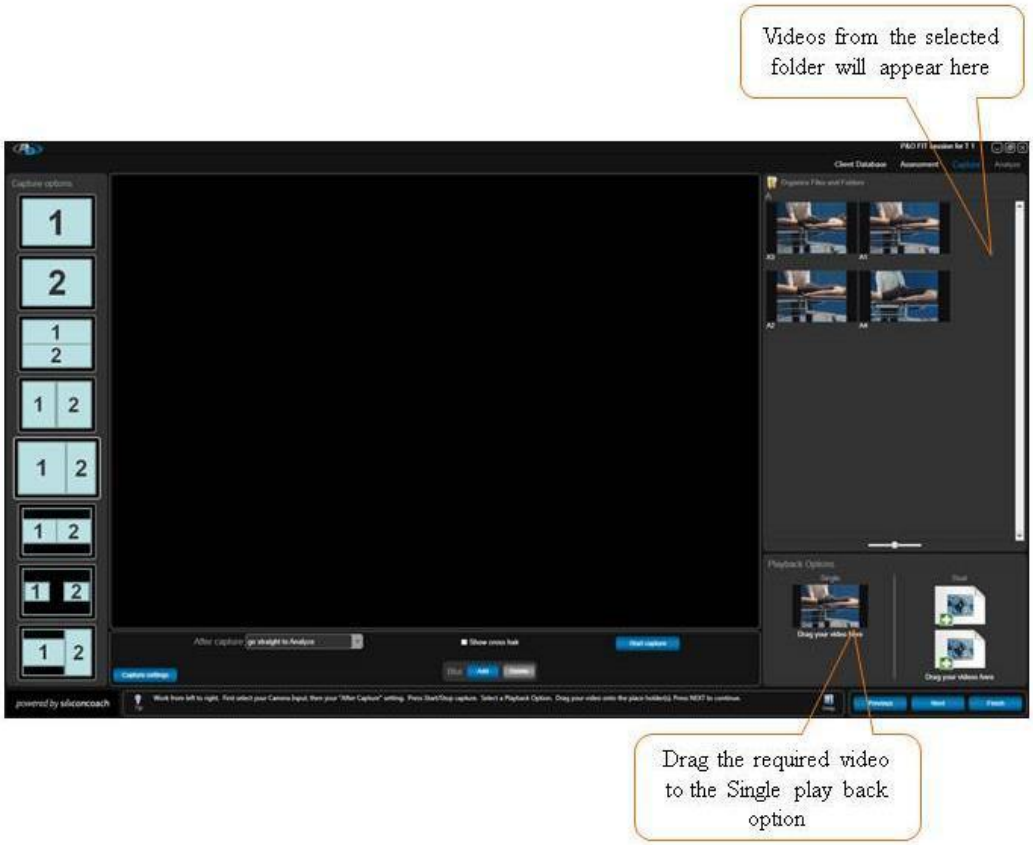
- Click on the icon of Organize Files and Folders on the top right hand side as shown below



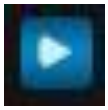
- Choose Select any folder
- Make sure you are at the folder POClinical –Fittings-Customers
- Search for Reliability_study_
- Double click on the folder -
- Highlight the required folder A (you will be asked later to repeat this step and choose the following folder which is folder B and work on alphabetical order)
- Click Select Folder as shown below



- Videos will appear on right hand side (as you face the screen)
- Drag video A1 down to Single play back option which is in the bottom right of the screen (you will be asked later to repeat this step and choose the following video which is video A2 and work on numerical order)
- The video will appear in the single play back box if dropped correctly as shown below
- Take a note of the number of the video you choose (file number-video number)



12. Click next

- Analyze screen will appear
- To play the clip use this button 
- When the maximum joint range of motion is achieved, the tester on the screen will hold the position for couple of seconds, pause the video and start analysing
- Make sure that you have a clear view of the joint measured and the joint is held in a stationary position
- Please move to section F of the document and read thoroughly to understand how to operate different tools
- Take the angle measurement following the steps mentioned in section H
- When you are done analysing the video, click previous to choose the next video
- If the videos are not there, click on the icon organize files and folders on the top right hand side as shown above
- Choose Select any folder
- Search for Relibility_study_
- Double click on the folder
- Highlight folder A
- Click Select folder as shown above
- Videos will appear on right hand side
- Select the next video (A2)
- Drag the video down to Single play back option which is in the bottom right of the screen (remember to work on numerical order)
- The video will appear in the single play back box if dropped correctly as shown below

- Take a note of the number of the video you choose (file number-video number)
- Click next
- Repeat the process from section D step 13 page 9 for videos A2, A3 and A4
- Each report allows you to capture 4 images (4 videos)
- Each folder contains 4 videos
- Once you are finished with folder A (4 videos) press the finish button and choose save this report and return to this Clients area and return to section C step 5 page 4 and start again with folder B and so on (remember to work on alphabetical order)

E- Finishing up and saving all the reports

- Make sure you save the report when finishing all the measurements by clicking finish and choose save this report and return to the welcome screen
- On the welcome screen click the exit button on the top right
- Click ok
- Shut down the computer and close the monitor screen

F- Tools:

On the analyse screen there are number of tools below the video, you will only need to use the following:

Measure angle tool 

To use the measure angle tool:

- Chose the tool from the tool bar
- Click the mouse button on the starting point and release, then the centre point of the angle and release then the end point and release.

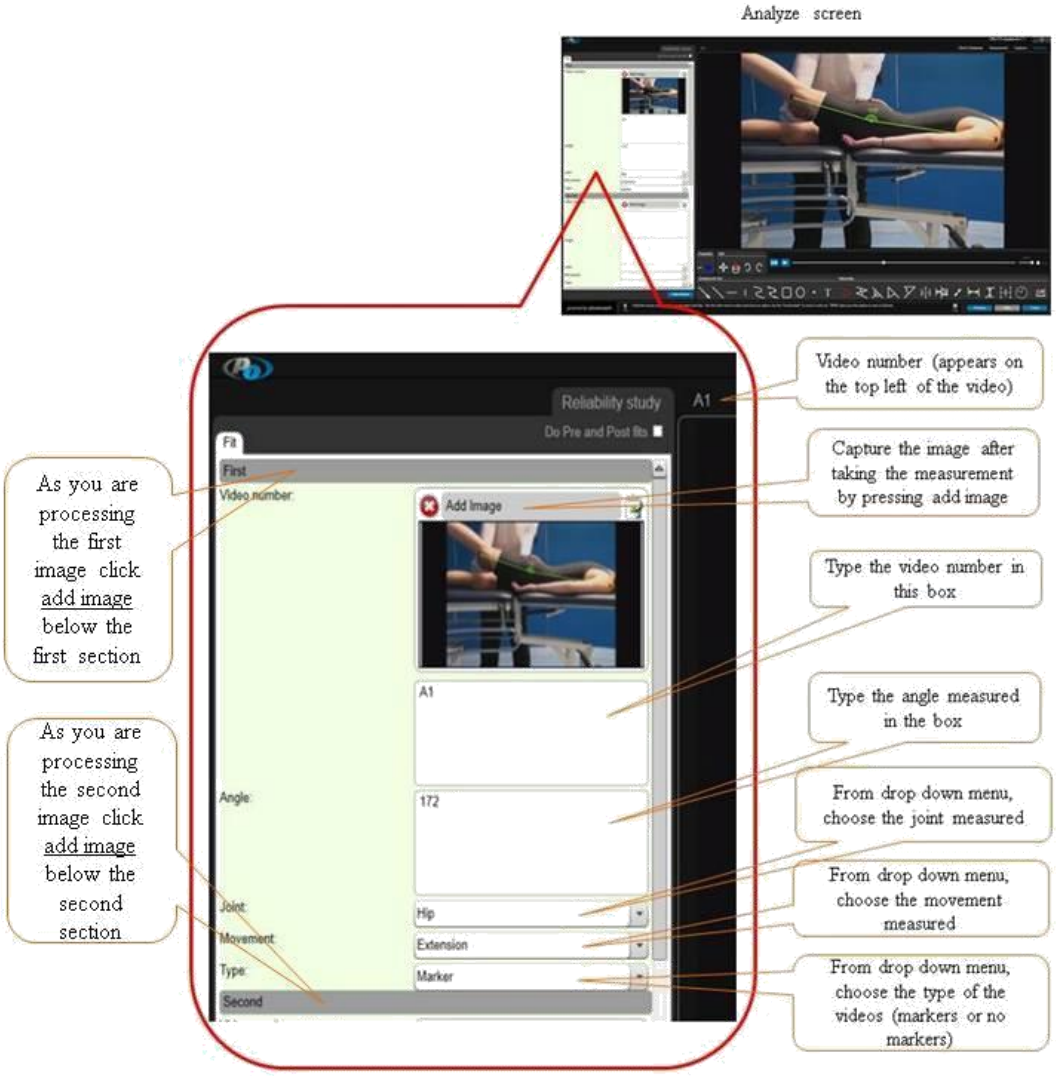
Delete all  and undo 

- Use these tools to delete any measurements taken
- The Undo tool clears one step at time and the Delete all tool deletes all the steps
- Delete all the measurements taken after capturing the still image and prior to starting the next stage

G- Creating the report:

After measuring each angle following the steps mentioned below in section H and when you are satisfied with the angle measured please:

- Capture the image by pressing add image on the report on the left-hand side after taking the measurement (if you are processing first image click add image in the first section and if you are processing the second image click add image in the second section and so on as shown below)
- **Please make sure to not overwrite any image**
- Manually type the video number (appears on the top left of the video) and angle measured
- From the drop-down menu choose the joint, movement and type as shown below
- Please fill the Report for *PnO CMD* videos provided as you analyse each video



- Delete all the measurements taken by using the delete all icon  after capturing the still image and prior to starting the next stage
- Return to section D step 19 page 9

H- Measurements

- For this study measure the ROM following the instructions below
- If you are looking at the hip videos, follow the Hip joint angle section below
- If you are looking at the knee videos, follow the Knee joint angle section below
- If you are looking at the knee videos, follow the Ankle joint angle section below

Hip joint angle:

If markers appear in the video, follow section H1 below and if no markers appear in the video follow section H2 below

H1

- Using the measure angle tool  click on the marker on the shoulder (start point), then on the marker on the greater trochanter (centre point), then on the marker on the lateral femoral epicondyle (end point)
- Make sure that you capture the image and add the required information as stated above in section G

H2

- Using the measure angle tool  click on a point at the shoulder (start point), follow the midline of the body and then click on a point on the mid-thigh at the level of greater trochanter (centre point), follow the mid-line of thigh and click on a point at the mid knee (end point)
- Make sure that you capture the image and add the required information as stated above in section G

Knee joint angle:

If markers appear in the video, follow section K1 below and if no markers appear in the video follow section K2 below

K1

- Using the measure angle tool  start at the marker on the greater trochanter (start point), then click on the marker on the lateral femoral epicondyle (centre point), then on the marker at the lateral malleolus (end point)

- Make sure that you capture the image and add the required information as stated above in section G

K2

- Using the measure angle tool  click on a point at the mid-thigh at the level of greater trochanter (start point), follow the midline of the thigh and click on a point at the mid knee (centre point), follow the midline of the calf and click on a point above the lateral malleolus (end point)
- Make sure that you capture the image and add the required information as stated above section G

Ankle joint angle:

If markers appear in the video, follow section A1 below and if no markers appear in the video follow section A2 below

A1

- Using the measure angle tool  start at the marker on the lateral femoral epicondyle (start point), then the marker on the later malleolus (centre point), then the marker on the fifth metatarsal head (end point)
- Make sure that you capture the image and add the required information as stated above in section G

A2

- Using the measure angle tool  start at a point at the mid-calf on the level of the fibula head (start point) following the mid-calf click at the point on the base of the heel (center point), then following the sole of the foot click at a point on the sole of the foot at the level of the fifth metatarsal head (end point)
- Make sure that you capture the image and add the required information as stated above in section G

Editing report (if required):

1. On the desktop click on P + O Clinical Movement Data Software.
2. Choose Existing Client
3. In the search box type: Last name: 1-3 (your code number)
4. In the result box highlight the required name (your code) as shown above
5. Click ok
6. This will take you to another page called P&O fit session (written in the extreme top right)
7. On the left side of client information, look for Patient ID
8. Patient ID is your code; if the space is empty type your code in it as shown above
9. The most recent report will appear on the screen.
10. On the top left of the report, several tabs will appear with different dates of the report saved
11. Click on the required report
12. Select Edit on the lower right hand side
13. Click next
14. Click on the icon of organize files and folders on the top right hand side.
15. Choose Select any folder
16. Search for Reliability_study_
17. Double click on the folder
18. Highlight the folder you want to edit (A-X)
19. Click select folder as shown above
20. Videos will appear on right hand side
21. Drag the required video to edit down to Single play back option which is in the

bottom right of the screen

22. The video will appear in the single play back box if dropped correctly as shown above
23. Click next
24. Analyze screen will appear.
25. Make sure you save the report when finishing all the measurements by clicking finish and choose save this report and return to the welcome screen
26. On the welcome screen click the exit button on the top right
27. Click ok
28. Shut down the computer and close the monitor screen

Publications and presentation

Publications

1- Mohsin F, McGarry A and Bowers R. Factors Influencing the Reliability of the Universal Goniometer in Measurement of Lower-Limb Range of Motion: A Literature Review. Journal of Prosthetics and Orthotics. 2015;27(4):140-8.

2 - Mohsin F, McGarry A and Bowers R (in press). The reliability of a video analysis system (*PnO Clinical Movement Data*^{TM}) and the universal goniometer in the measurement of hip, knee and ankle sagittal plane motion amongst healthy subjects Journal of Prosthetics and Orthotics.

Conference presentations

1- Mohsin F, McGarry A, Bowers R. Factors influencing the reliability of different designs of goniometer: a literature review. International Society for Prosthetics and Orthotics World Congress 2013, Hyderabad, India. 2013.

2- Mohsin F, McGarry A, Bowers R. The reliability of a video analysis system (*Siliconcoach*^{TM}) and the universal goniometer in the measurement of hip, knee and ankle motion. International Society for Prosthetics and Orthotics World Congress 2015, Lyon, France. 2015.

Posters

1- Mohsin F, McGarry A, Bowers R. A literature review on the reliability of different designs of goniometer. British Association of Prosthetists and Orthotists 2013, Glasgow, United Kingdom. 2013. Poster presentation runner up prize.

2 -Mohsin F, McGarry A, Bowers R. Literature review of the reliability of different designs of goniometers. American Orthotic and Prosthetic Association National Assembly 2014, Las Vegas, United States. 2014.

3- Mohsin F, McGarry A, Bowers R. The reliability of a video analysis system (*Siliconcoach*^{TM}) and the universal goniometer in the measurement of lower limb joints. American Academy of Orthotists and Prosthetists 41st Annual Meeting and Scientific Symposium 2015, New Orleans, United States. 2015.