APPLICATIONS OF GAIT ANALYSIS IN NORMAL AND PATHOLOGICAL GAIT

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DECLARATION

All the work described herein was undertaken at the Anderson Gait Analysis Laboratory Edinburgh where I have worked since its inception in 1988. This work could not have been undertaken without the help and collaboration of my co-authors.

The idea for the longitudinal studies in normal children was mine as were the ideas for the projects on the visual assessment of gait, the classification of hemiplegia and clinical decision making. I shared the ideas for the studies on rockers and ankle foot orthoses with Mr Michael Hullin FRCS and the hamstring study with Dr Marietta van der Linden. All the patients in the clinical studies were under my care and I also reviewed their gait and clinical data. I was one of the assessors of the patients in the projects on the visual assessment of gait and for the classification of hemiplegia.

I made a significant contribution to the writing of all eight manuscripts before publication. I am the senior author of all eight publications.

None of the work contained in these eight papers has been submitted for a higher degree in any other University.

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Gait analysis is a team activity and it is a pleasure to acknowledge the outstanding contributions from colleagues who have worked or who are still working at the Anderson Gait Analysis Laboratory. The James and Grace Anderson Trust have been generous benefactors to the laboratory over the years and much of what has been published from the laboratory could not have been achieved without this support.

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To our patients, children and their parents who have attended the laboratory.

CONTENTS

ABSTRACT

This thesis is a compilation of eight original papers in the field of gait analysis that have been published in peer-reviewed journals.

The first two studies show that age is not a factor in the characterisation of kinematics and kinetics of gait in children but rather normalised speed is the key factor. It was also found that ground reaction forces were mature by the age of five years and the sagittal joint kinematics, moments and powers were mature by the age of seven. These two papers have produced unique reference data for a population of children aged between five and twelve years old. They have also highlighted a technique of normalisation of gait data to dimensionless units, a technique which is appropriate for children who grow in height and weight.

The third paper describes how a validated method for assessing gait visually in cerebral palsy was developed. This visual gait score is capable of demonstrating change over time, thus providing clinicians with a simple tool to follow a patient's progress. The fourth paper provides a classification of hemiplegic cerebral palsy based on sagittal plane kinetics. This has extended previous classifications based only on kinematics.

The fifth and sixth papers consider practical applications of gait analysis. The fifth paper concerns rocker design for patients who wear a below knee plaster cast and the sixth paper, ankle foot orthosis function in myelomeningocele. Both papers illustrate how gait analysis can help in the understanding of rocker and ankle foot orthosis function and how it can improve their design.

The seventh paper shows how gait analysis can alter treatment planning for patients who have cerebral palsy. This paper extends previous knowledge by considering a more relevant clinical scenario in which an orthopaedic surgeon makes a clinical decision and subsequently modifies it in the light of gait data.

The eighth paper illustrates for the first time how converting gait data into dimensionless units does avoid the influence of changes in height and weight on the outcome of hamstring surgery in children. This normalisation technique will be of future value when comparing outcomes in populations with differing height and weight.

INTRODUCTION

This thesis is a compilation of eight papers in the field of gait analysis that have been published in peer-reviewed journals. Its aim is to show how gait analysis, originally a scientific curiosity but now available commercially, can be used to study gait in normal and pathological states.

Knowledge of normal gait is a prerequisite for understanding pathological gait. The two prospective studies of maturation of gait in children have shown for the first time that age is not a factor in the characterisation of kinematics and kinetics of gait in children but rather normalised speed is the key factor. These two prospective studies also confirm the findings from previous cross-sectional studies that ground reaction forces are mature by the age of five years and that sagittal joint kinematics, moments and powers are mature by the age of seven. These two studies have produced unique reference data for a population of children aged between five and twelve years old. They have also highlighted a technique of normalisation of gait data to dimensionless units. This technique is appropriate for children who grow in height and weight and has not been previously applied to studies of maturation of gait in normal children.

Having established these concepts for normal data in children, two studies show how information from instrumented gait analysis can be used to assess gait visually in cerebral palsy and to produce a sagittal plane kinetic classification of hemiplegia. The third study describes how a validated method was developed to assess gait visually in cerebral palsy. The purpose of this study was to use knowledge from three-dimensional analyses to produce a gait score that does not require special equipment other than a video camera, and is sensitive to change. This score is capable of demonstrating change over time, thus providing clinicians with a simple tool with

which to assess gait and to follow a patient's subsequent progress. At the time of writing the Edinburgh Visual Gait Score is the first new score to have been validated specifically for cerebral palsy, and it may be of value to those who do not have access to instrumented gait analysis. The fourth study continues with the theme of gait evaluation but additionally provides a classification of hemiplegic cerebral palsy. Previous studies have solely considered sagittal plane kinematics, and this study is the first based on sagittal plane kinetics. The use of kinetics for classification is more precise than kinematics alone. However none of the existing classifications are able to distinguish a primary problem in gait from a compensatory mechanism.

Having developed theoretical tools for the visual evaluation of gait and for the classification of hemiplegia, consideration is then given to ways in which gait analysis may have practical benefits for patients. Two papers show how gait analysis can be used to optimise gait by improving orthotic and rocker design for patients with myelomeningocele and for those who have to wear a plaster cast. The fifth paper describes rocker design for patients who have to wear a below knee plaster cast and the sixth paper, ankle-foot orthosis function in low-level myelomeningocele. Both papers have provided new biomechanical concepts for rockers and orthoses and show how gait analysis can be used to optimise rocker and orthotic design for the individual patient. Both papers also illustrate the practical benefits of gait analysis thus confirming that it is not merely a research tool. These papers consider some non-operative benefits of gait analysis.

The final two papers concern some surgical aspects of gait analysis in cerebral palsy. The seventh paper shows how gait analysis can affect surgical decision-making in cerebral palsy. It extends existing knowledge by considering a more relevant clinical scenario in which the clinical decision of an orthopaedic surgeon might be

modified in the light of subsequent gait data and shows clearly how surgical decision-making may be altered by gait analysis. It does not, however, establish that surgical outcomes are enhanced by gait analysis, a topic which remains a controversial issue and is only likely to be answered in the future by randomised trials. Such trials may be impractical for patients who have access to gait analysis. The final paper discusses the outcome of hamstring surgery in cerebral palsy as assessed by gait analysis. It also illustrates the importance of the use of normalisation techniques in the assessment of post-operative results to exclude the effects of a child's growth. The use of these techniques has been described more fully in the first two papers. Without normalisation it could be possible to attribute, erroneously, functional results from surgery that had occurred due to the patients' growth during the postoperative period. The use of this particular normalisation technique on surgical outcome in children is illustrated for the first time in this paper. This normalisation technique enabling comparisons between populations of differing height and weight is likely to be of future benefit.

The combination of these eight papers shows that gait analysis is an excellent research, evaluative and audit tool that benefits patients who have complex motor disorders.

HISTORY OF GAIT ANALYSIS

Although descriptions of gait and its disorders have been noted in historic times it was not until Borelli (1608-1679) that a detailed description of the mechanical function in locomotion in man was made. *De motu animalium* was published in two parts posthumously in 1680 and 1681. Borelli was Professor of Mathematics in Naples, and Queen Christian of Sweden, who resided in Rome, supported the costs for the posthumous publication of his work. Borelli was a mechanist who considered that Nature always acts using the simplest and most economic means. He recognised that muscles acting with short lever arms, balance the body-weight which usually acts with a much longer lever arm so that the joints must transmit forces at least equal and generally several times greater than the actual weight of the body or that part which they sustain (Maquet 1989). Borelli determined the centre of gravity of the body more that two centuries before Braune and Fischer (1892) by having a subject lie on a board balanced on the cutting edge of a wooden trihedral prism (Macquet 1992). He calculated the forces involved in various daily activities, including walking, recognising that when walking, subjects describe a wavy path which moves successively from right to left (Maquet 1989). Borelli thus anticipated the observations of Saunders, Inman, and Eberhart (1953) by about 300 years. He also described both the forward displacement of the centre of gravity beyond the supporting area, and the manner in which the forward swinging of the limbs saves the body from losing balance. These constituted a fundamental concept of propulsion and restraint in gait (Steindler 1953).

In Germany the Weber brothers (1836) worked extensively on the mechanics of walking and on measurements of the kinematics of gait, but their studies were limited to the movements, rather than to the forces involved. Steindler (1953) has

described their work as an 'observational' period of studies on gait. The Weber brothers also investigated the muscle effort involved in propulsion and restraint. Their calculations from their cadaver work were extremely accurate compared with those observed in real life. However they did consider, erroneously, that leg motion in swing was a pure pendulum effect and not dependent on muscle action.

The next period, forty years later, can be considered as a period of 'visual recording' (Steindler 1953) where photography and kymography enabled a visual record of gait and movement to be made. Carlet (1872; quoted in Macquet 1992) recorded the temporal parameters of gait using the kymograph, a mechanical device attached to the human trunk to record movements on drums covered in soot. He introduced an experimental shoe that had an air chamber between two soles. The chamber was connected by a tube to a drum to which was attached a lever that registered the duration and pressure of the foot. The subject carried the portable apparatus and the chamber was compressed as the subject loaded it.

Marey (1883) in France was the first to use photography to analyse body movements during locomotion. He used a method of multiple exposure photography in which several exposures were made on a single plate or frame. His subject was clad in black with white stripes indicating the position of various body parts. Recordings were made by rotating a slotted disc in front of the camera lens as the subject walked across the room. The resulting pictures were the very first "stick diagrams." A similar technique was still in use in the 1960's, (Murray, Drought and Kory 1964), but used precisely timed flashes of stroboscopic light to illuminate the reflective markers worn by subjects. Marey's photographs revealed kinematic curves similar to those obtained by Braune and Fischer 20 years later (Steindler 1953).

Muybridge (1887), in California, also working with photography, recorded the

successive phases of movements in gait. He used as many as 24 cameras in series, which were used in combination to demonstrate the sequence of events in slow motion. However, all his analyses considered only the movements themselves and did not take into account the forces involved. In the late 19th century the first motion picture cameras recorded patterns of locomotion in both humans and animals. In 1877 Muybridge used photographs to demonstrate that when a horse gallops there is a point during which all its hooves are off the ground. In 1887 he published *Animal locomotion.* In the following year the number of cameras was increased and he recorded motion in humans when he photographed members of the Olympic Club of San Francisco doing various exercises. Pathological gaits were photographed in 1885.

In Germany, Braune and Fischer (1895) used a similar method to that of Marey, but substituted Geisler tubes that were fluorescent strip lights, for the white stripes. Between six and eight hours was necessary to prepare the subject for analysis. Braune and Fischer also determined the motion of body parts calculating their centre of gravity and inertia by experimenting with a frozen cadaver and cadaver parts taken from a suicide. They oscillated various segments from the bodies as pendulums to determine the moments of inertia. By relating the photographs taken simultaneously by four cameras to a three co-ordinate system and by combining the body segments, mass and acceleration data, they were able to estimate the forces involved in locomotion. Braune and Fischer used photographic images combined with calculations to provide graphic representations of human gait. They plotted the centre of gravity and produced 31 phases of gait in which they calculated fluctuations of floor pressure for each of the 31 phases. This was the first study into the kinetics of gait. Their combination of a visual record and kinetics anticipated the introduction of commercial systems in the latter half of the 20th century. This work was subsequently

integrated into a monumental account and mathematical analysis of human gait, *Der Gang des Menschen* (1895). Braune and Fischer produced the first rational scientific investigation of gait and their work remains a classic. They showed that the Weber brothers were incorrect in their assertion that swinging of the unloaded leg in gait was not a simple pendulum movement since muscle forces are required for acceleration and deceleration (Maquet 1992).

In the early 20th century more attention was being paid to muscle action and its effect on joint movement. Von Baeyer (1933; quoted in Steindler 1953) reported on muscle combinations acting on the lower extremities that were an analysis of muscle action as they affected the proximal end of a lever arm. He formulated the concept of closed kinetic chains where peripheral resistance exceeds central resistance as seen in the mechanical concepts of propulsion and restraint considered by Borelli about 300 years earlier. Scherb (1936; quoted in Steindler 1953) investigated normal and pathological gait. In his work on kinetic diagnostic analysis of the disturbance of gait he investigated the sequence of muscle function in the lower extremities by palpation of muscle groups as a subject walked on a treadmill. He was able to confirm his clinical findings later with the advent of electromyography.

Advances in foot pressure measurement were made by Schwartz and Heath (1947) who refined Carlet's (1872) pneumatic shoe sole apparatus. In a series of publications they refined further the concept of foot pressure measurements. In 1947 they described the use of piezoelectric discs used to record foot pressure during gait. They used a photographic method to record the foot pressures via light beams projected from a galvanometer on to moving photographic paper.

Bresler and Frankl (1950) described forces and moments in the leg during level walking and studied four healthy males and used a force plate. They commented

on the tedium of the calculations and to calculate the quantities for one stride approximately 14,000 numerical calculations were performed, 72 curves plotted and 24 curves subjected to graphic differentiation. Initially this took 500 man-hours. They presented the mechanism of normal level walking in terms of displacements and the force systems in the joints of the lower extremities. The data were obtained from simultaneous recording of the positions of the leg in space and from floor reactions during normal level walking. They calculated the mass moments of inertia and related the forces and moments to a system of horizontal and vertical orthogonal axes.

Major advances in gait analysis were also made by Saunders, Inman and Eberhart (1953) who used a force plate, electromyography and an interrupted light technique similar to Marey's (1882). They also used high-speed motion photography but found this to be of qualitative value only. Pins were inserted into bones to measure the transverse rotations of limb segments. The subject's response was not noted (Sutherland 2002). Their studies showed how body segments move through all three planes of movement to smooth the path of motion and to minimise the excursion of the centre of gravity thus reducing energy requirements. Their determinants of gait were features of the movement pattern that minimised the centre of gravity excursions. They considered the centre of gravity to lie in front of the second sacral vertebra. During normal walking they described a smooth sinusoidal vertical curve in the plane of progression and at right angles to the direction of progression, thereby producing a horizontal curve. Their determinants of gait were as follows: pelvic rotation, pelvic tilt, knee flexion in stance, foot and knee mechanisms and lateral displacement of the pelvis. More recently Gage (1991) extended the concept of determinants to five attributes of normal gait: stability in stance, adequate foot

clearance, foot pre-positioning in terminal swing, adequate step length and energy conservation.

Measurement of gait movements from motion picture film was being used in 1972 (Sutherland and Hagy) but these techniques were very time consuming and impractical for routine use. The introduction of powerful computers revolutionised the length of time required to calculate data. In the latter part of the 20th century a number of systems were developed capable of automated and semi-automated computer aided motion analysis. The first systems to become commercially available required the operator to identify the location of each marker used for each frame manually which was still a very time consuming exercise. Since then the problems of automatic marker identification have been at the forefront of computer aided motion analysis developments. In 1974 Selspot (Innovision Systems), which allowed automatic tracking of active light emitting diode markers, became commercially available. Subsequently Watsmart and Optotrak (Northern Digital Inc.) used a similar technique. Vicon (Oxford Metrics), a television camera based system, became commercially available since 1982. A number of systems based on television camera technology have been developed since then. These include the Elite system (Motion Analysis Corporation), Coda (Charnwood Dynamics) and the MacReflex (Qualysis).

THE GAIT CYCLE

Walking is a complex activity involving movement in three planes - sagittal, coronal and transverse, and at five different anatomical levels - trunk, pelvis, hip, knee and ankle. The gait cycle is a continuum of flow of movement that does not have a readily definable end or beginning but conventionally can be considered to begin at the point when a foot makes contact with the ground. Winter's (1987) division of the gait cycle used a functional classification of weight acceptance, midstance, push off, lift off and reach. Sutherland et al (1988) substituted three periods of stance: initial double support, single limb stance, and second double support. Perry (1992) was responsible for the now widely accepted division of the gait cycle into five stance phase periods - initial contact, loading response, midstance, terminal stance, and preswing and three swing phase periods - initial, mid and terminal swing.

1. Periods of the gait cycle

Each gait cycle is divided into two periods of stance and swing. The stance phase, when the foot is contact with the ground, can be subdivided further into two periods of double support or stance and one period of single support when only one foot is in contact with the ground.

The timing of the cycle is normally 60% for stance and 40% for swing when the limb under consideration is off the ground. The cycle begins at initial contact (IC) and lasts for about 10% of the cycle. It is followed by single support that lasts for about 40% of the cycle and then stance ends with a second period of double support lasting for about 10% of the cycle. Walking speed alters the percentages of the stance period.

The gait cycle may also be described in terms of stride and step length. A stride is the equivalent of a gait cycle and is the interval between two sequential initial contacts of the same limb. Step length is the interval between the initial contact by the same foot. Two steps equal one stride.

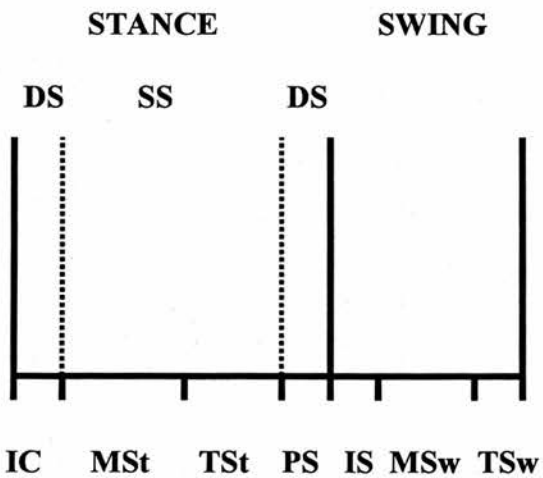

Figure 1 Summary of the gait cycle. DS Double Support, SS Single Support, IC Initial Contact, MSt Mid stance, TSt Terminal stance, PS Pre-swing, IS Initial Swing, MSw Mid swing, TSw Terminal Swing.

2. Phases of the gait cycle

Eight phases identify the functional significance of each part of the gait cycle. Perry (1992) suggested that the gait cycle also contains three tasks, those of weight acceptance, single limb support and limb advancement. Weight acceptance comprises initial contact (IC) and the loading response (LR). The goal at this point in the cycle is for the body's weight to be transferred onto a limb that has just stopped swinging. Initial contact lasts for about 2% of the cycle. This is followed by the loading response that may occupy about 10% of the cycle and continues until the opposite foot lifts off in swing. This is then followed by single limb support until the opposite foot make contact again with the ground. The next phase is mid stance, which

comprises the first half of single support, followed by terminal stance, which comprises the second half of single support. Mid stance may occupy 10-30% of the cycle while terminal stance occupies 30-50%. The final task of limb advancement comprises pre-swing, initial swing, mid swing and terminal swing. Pre-swing comprises about 50-60% of the cycle, initial swing 60-73%, mid swing 73-87% and terminal swing 87-100%. These divisions are based on timing.

Perry (1974) also described progression, or limb advancement, in stance as rocker sequences. The first rocker occurs from heel strike to foot flat. The second rocker is from foot flat to heel lift occurring as the tibia progresses over the stationary foot. The third rocker is the rotation of the foot over the metatarsal heads occurring as the ankle plantarflexes.

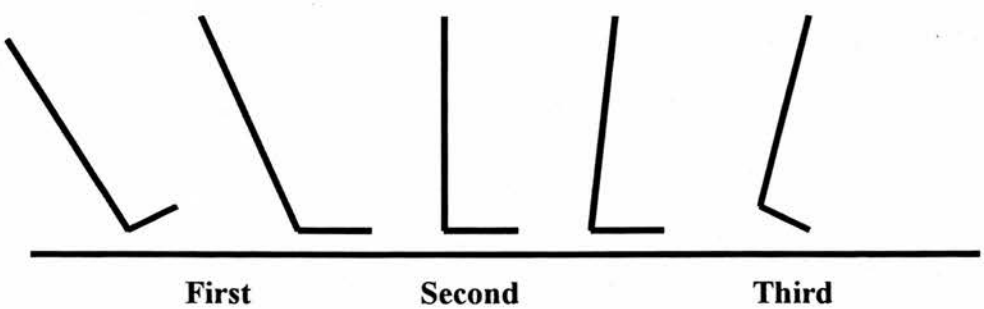

Figure 2 Sagittal stick diagrams of the shank and foot showing the rocker sequences.

3. Locomotor functions

Saunders, Inman and Eberhart (1953) described six determinants of normal gait: pelvic rotation, pelvic tilt, knee flexion in stance, foot and knee mechanisms and lateral displacement of the pelvis. These factors minimize the excursion of the body's center of gravity (CoG) in gait. Pelvic rotation, pelvic tilt, knee flexion and the foot

and knee mechanism reduce vertical excursion of the CoG. The tibio-femoral angle and hip adduction approximate the feet and reduce lateral displacement of the body. Such mechanisms produce an almost symmetrical displacement of the CoG in the horizontal and vertical planes.

Ralston (1958) suggested that individuals use a self-selected walking speed that is the most energy efficient for them. He studied 12 male and seven female untrained but healthy subjects and measured minute volumes and oxygen concentrations while the subjects walked around a track at four different speeds. He found that during level walking their energy expenditure was a linear function of the square of speed. Winter (1987) showed that when walking speed increases the angular velocity of the lower limb joints increased at the same proportions and the patterns of power produced in these joints maintained their usual timings.

Perry (1992) ascribed four functions in gait: propulsion, stance stability, shock absorption and energy conservation. Propulsion requires the input of kinetic energy for the lower limb to achieve maximum length followed by the conversion of kinetic energy to potential energy as the upper body progresses forwards and the lower limb shortens again before push off. Stability of the limbs is required for efficient propulsion. Transfer of energy from a swinging limb to a stance limb is a potentially unbalanced situation in which the upper body passes downwards towards the floor. System efficiency is the balance between work accomplished and energy expended. Intensity of muscular effort can be considered to be an individual's capability and the energy expended their endurance (Perry 1992). Gage (1991) extended Perry's concept to describe five attributes for normal gait: stance phase stability, swing phase clearance, adequate foot pre-positioning, adequate step length and energy efficiency.

4. Temporal/distance measurements of the gait cycle

Velocity is defined as the self selected speed of motion in the desired direction of travel and is measured in metres per second. *Cadence* is defined as the number of times a heel strike occurs per minute and equals the number steps per minute. *Step length* is defined as the distance between heel strike and opposite heel strike of the foot in the direction of travel, measured in metres. *Stride length* is defined as the distance between consecutive heel strikes of the same foot in the direction of travel, measured in metres. *Single support/stance* is the percentage of the gait cycle during which only one foot is in contact with the ground. *Double support/stance* is the percentage of the gait cycle when both feet are in contact with the ground. There are two double support phases in each gait cycle. Single and double support can be measured using timing or as a percentage of the gait cycle (*vide supra*).

5. Kinematics

Kinematics describe motion at a joint, angular displacement and accelerations. These may be described by considering angles, defined in three dimensions, between specific points e.g. a joint centre or limb segments. This motion can be measured in several ways. Electrogoniometers, which measure angles, can be applied directly to a subject's limb. They can be biaxial or multiaxial (Chao 1980); the former have the disadvantage of measuring movement in two planes only and the latter are bulky.

Assessment of movement can be obtained from cine film or video recordings by digitising the movement of markers attached to a subject. The data collected is two-dimensional and subject to inaccuracies if rotational abnormalities exist in the limbs, a feature often seen in neurological conditions.

Optoelectronic tracking systems rely on camera detection of emitted or reflected light. Reflected infra red light is commonly used in commercially available clinical gait analysis systems. Markers attached to a subject reflect infrared light that is then detected by cameras in the laboratory and the signals computed to generate a graphical display or stick diagram of joint motion.

6. Kinetics

These describe the external ground reaction forces acting on the limb, joint moments and powers. The external forces on the foot can be resolved into three orthogonal axes at right angles to each other, which are defined at the origin of the ground reaction force on the foot at any given point in the cycle. The three forces are: a vertical force acting upwards, a horizontal shear force passing in an antero-posterior direction and horizontal shear force acting in a medio-lateral direction. By convention Fz represents the vertical force, Fy the medio-lateral force and Fx the antero-lateral force. The vertical component is characterised by two peaks (Fz1 and Fz2) and one trough. The fore/aft shear force is characterised as a braking force initially directed posteriorly and subsequently passing anteriorly during the acceleration phase of gait. The medio-lateral shear force initially acts laterally and is short lived and subsequently acts medially on the foot. These forces are measured in Newtons.

A moment (Newton metres, Nm) can be defined as a turning force acting on a joint. Its magnitude can be calculated as the product of the force and its lever arm from the joint under consideration. The lever arm is the shortest perpendicular distance between its line of action and the joint. The external moment generated is measured in Newton/metres and will induce an internal moment that is usually generated by muscle activity. All the moments described here are external.

Power is a measure of work (Nm/sec) and in gait analysis power is defined as the product of moment and joint angular velocity. Power may be generated by muscles contracting concentrically and absorbed during eccentric muscle contraction.

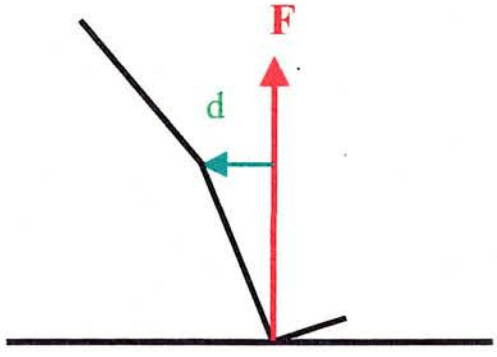

Figure 3 Stick diagram showing a lower limb in the sagittal plane at initial contact. The ground reaction force (F) is shown in red and the distance (d) from the knee joint centre in green. The moment (M) at the knee is calculated from the equation M=Fxd.

Winter (1987) criticised this method of joint moment calculation as an oversimplification, since it does not take into consideration segment masses, accelerations or segment weight and may not take into account passive stabilisation of joints by soft tissue restraints. However, Wells (1981) showed that the M=Fxd method is a sufficiently accurate approximation and this method is used routinely in clinical studies. Wells compared two methods of joint moment calculation - the above method and from link segment analysis. The latter method also contains approximations. Correlations were very good at the ankle, good at the knee and at mid stance at the hip. He concluded that the projection of the ground reaction force was a good predictor of joint moments but that care should be taken when considering walking at faster speeds and hip joint values at initial contact and foot off. Another

possible source of error when the M=Fxd method is used are rotational deformities and these should be taken into consideration during data interpretation (Mann 1998).

Mikosz et al (1978) evaluated the importance of limb segment inertia on joint loads during gait. They used the Selspot system and a PDP 11/40 computer and concluded that the inertial contributions of the limb segments to joint loads during stance were of secondary importance than obtaining accurate position data to compute the quasi-static forces during stance. They felt that in many cases the increased computational effort to obtain limb segment inertial loads was not always warranted.

Kadaba et al (1989) studied within-day and between-day repeatability for kinematic, kinetic and electromyography (EMG) analyses in 40 subjects. They found excellent intrasubject repeatability for kinematics in the sagittal plane within and between test days. The transverse and coronal plane kinematics repeatability was good within the test day, but poor between test days. The vertical and fore/aft forces were more repeatable than the mediolateral shear forces while sagittal moments were more repeatable than transverse and coronal moments. EMG was generally repeatable within and between visits. They concluded that it would be reasonable to base decision-making on a single gait evaluation. Nevertheless investigators must be aware of the shortcomings of data collection, marker placement and computational methods when considering kinematic and kinetic data derived from gait analysis.

7. Integration of kinematic and kinetic events in the normal gait cycle

Initial contact

The heel makes contact with the floor while the ankle is in neutral dorsiflexion, the knee is extended and the hip is flexed to approximately 30°. A rapid vertical ground reaction force (GRF) accompanies this as the body falls by about 1cm.

This force passes in front of the ankle, knee and hip. The gluteus maximus, hamstrings, and pretibial muscles are active at this point.

Loading response

The goal is to prevent excessive knee flexion, excessive ankle plantarflexion and stabilise the hip. Shock absorption is provided by the quadriceps, limiting knee flexion. The first rocker occurs as the heel rotates into plantarflexion to ensure contact of the remainder of the foot with the ground. The knee flexes to about 18°, the ankle to about 10° of plantar flexion and the subtalar joint goes into valgus. The GRF passes behind the ankle and knee but in front of the hip.

In the coronal plane there is an adductor moment at the hip activating the hip abductors to resist downward motion of the unloaded opposite hemi pelvis. The GRF is also adductor at the knee and is resisted by the ilio-tibial band. The GRF at the ankle is valgus and is resisted both by tibialis anterior and posterior calf muscle action. In the transverse plane there is an internal rotation of the hemi pelvis (anterior pelvic rotation) and an internal rotation of the talus.

Mid stance

The goals are to allow forward progression of the tibia over the foot (second rocker), the elongation of the lower limb by the quadriceps extending the knee and stabilisation of the hip by the abductors.

In the sagittal plane, limb stability is conferred by the action of the gastrocnemius and soleus. The GRF becomes anterior to the ankle and knee and posterior to the hip. Contralateral toe off shifts body weight to the stance limb and the hip reduces its flexed position from 30° to 10°. These critical events all occur in the sagittal plane.

Mid stance may be defined anatomically as the point of maximum leg length or maximum hip height (Hullin and Robb 1991), or the point at which there is reversal of the fore/aft shear forces on the foot.

Terminal stance

The goals are to allow a controlled forward fall of the upper body over the stationary foot and to stabilise the foot when this occurs. The third rocker provides stability and progression as the ankle dorsiflexes to 10° and the heel rises as knee extension is completed. The upper body drops, and a second peak of the GRF is produced. There is a large ankle dorsiflexor moment requiring a strong contraction from both the gastrocnemius and soleus to stabilise the tibia at the ankle. The knee then unlocks and allows knee flexion to begin.

Pre-swing

The goal is for the knee to flex, assisting with clearance of the limb. As the knee flexes through 40°, the ankle plantarflexes by 20° and the GRF progresses to the metatarsal heads and as third rocker is completed. The limb unloads, push off begins, the knee flexes and the thigh advances. The latter is assisted by the flexor action of the adductor longus muscle. In the coronal plane the vector passes lateral to the hip producing an abductor moment resisted by adductor longus activity.

Initial swing

The goal is foot clearance that is dependent on knee flexion and hip flexion that propels the thigh forward. The knee flexion progresses to 60°, hip flexion advances the thigh by 20° and the pretibial muscles dorsiflex the foot to assist with clearance.

Mid swing

Continued action of the pretibial muscles ensures foot clearance. Hip flexor activity is minimal and knee extension is a passive event. The hamstrings begin to contract at the end of mid stance slowing down the swinging shank. Mid swing can be defined anatomically as hip high point (Hullin 1990).

Terminal swing

The goals are: to decelerate hip flexion and prevent knee hyperextension and for the ankle to dorsiflex to a neutral position to prepare for initial contact. The hamstrings contract to control the hip and knee and the quadriceps become active to complete knee extension.

METHODS USED IN THE STUDIES

1. Laboratory locations

Data collection took place at the Anderson Gait Analysis Laboratory in the Princess Margaret Rose Orthopaedic Hospital, Edinburgh. In the earlier years (1988-1995) this laboratory was situated within the Physiotherapy Department of the Hospital and the walkway was 5.5m long. At this location, data was collected from a Vicon Motion analysis PDP 11/73 system, three Vicon cameras (Oxford Metrics, Oxford, England), and a Kistler 9281B eight-channel force plate (Kistler Instruments AG, Winterthur, Switzerland). The PDP 11/73 computer used the RSX 11M operating system (Digital Equipment Corporation, Maynard, Massachusetts, USA) (Hullin 1990).

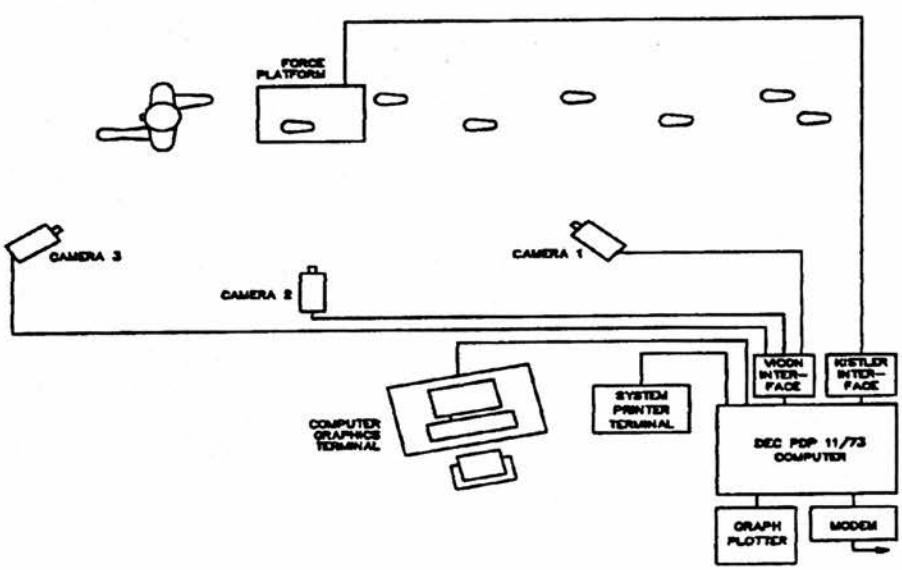

Figure 4 Laboratory set-up 1988-1995 (Hullin 1990)

In 1995 the gait laboratory was rehoused in a room dedicated for this purpose where the walkway was 7.5m long. Equipment was upgraded to a Vicon VX system

and five Vicon cameras (Oxford Metrics, Oxford, England) though the Kistler force plate continued to be used. Data acquired were analysed using the Vicon Clinical Manager software system (Oxford Metrics, Oxford, England) (Mann 1998).

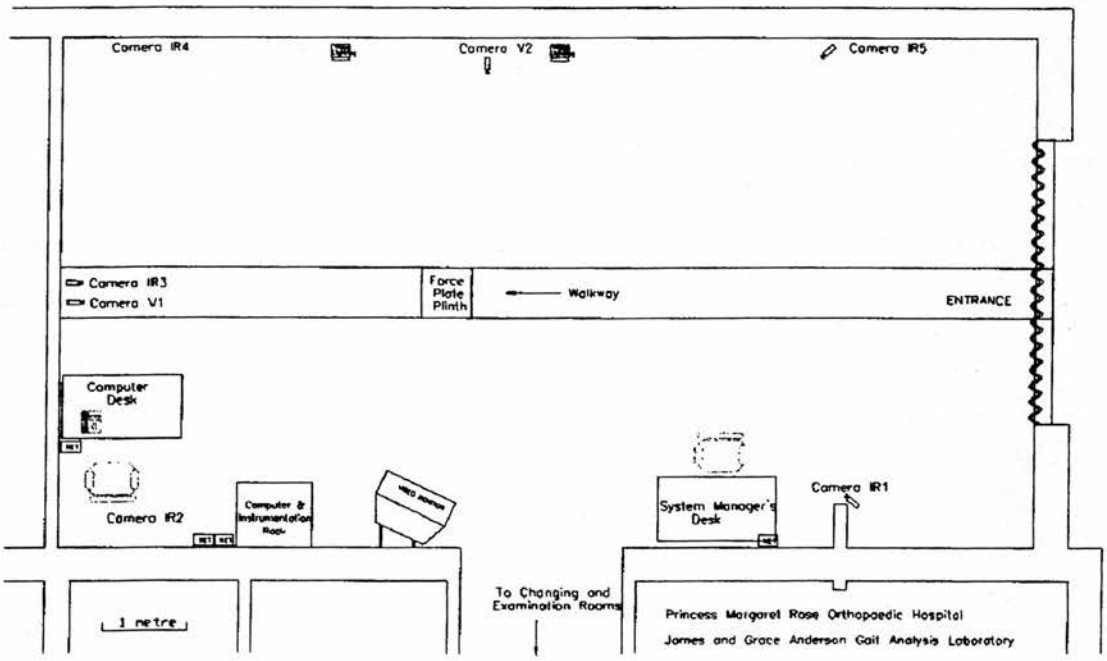

Figure 5 Laboratory set up 1993-2002 (Mann 1998)

In 2002 following the closure of the Princess Margaret Rose Orthopaedic Hospital, the laboratory was relocated once more to a dedicated room at the Eastern General Hospital, Edinburgh and the VX was replaced by a Vicon 512 system (Oxford Metrics) in 2000.

2. Markers

The markers applied to the subject were made from 'Scotchlite' (3M Corporation), a retro-reflective material with an adhesive backing. The markers were originally 15mm diameter but they were replaced by 25mm diameter markers mounted onto a flat disc to enable a more secure attachment to the subject. Rotation indicators designed in the Anderson Analysis Laboratory (Hillman et al 1998) were positioned on the patient to improve visual assessment of lower limb rotational alignment with the pelvis.

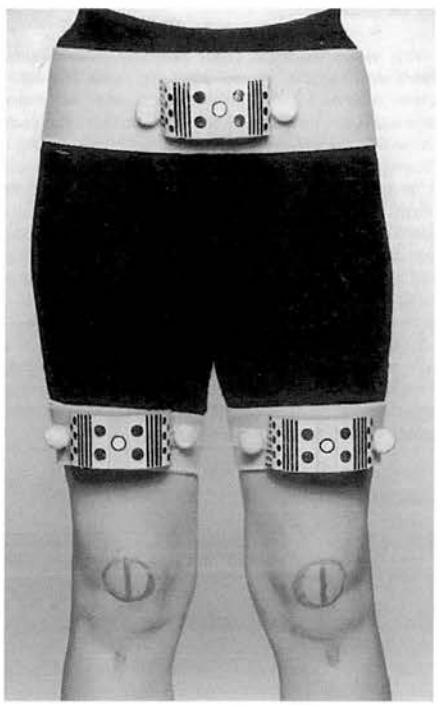

Figure 6 Rotation indicators (Hillman et al 1998)
(Reproduced with permission)

3. Cameras

The Vicon cameras have a ring of light emitting diodes around the lens. The cameras use infrared light so that data collection may take place in normal lighting conditions using artificial light, though external sources of infrared light must be excluded. The cameras operated on the European Standard Video Signal (CCIR), a

625 line interlaced picture. Each field of view was scanned twice per complete picture so that each marker could be detected 50 times per second. The cameras were synchronised using a charged couple device the signal digitised and the computer then calculated the marker positions.

4. Force Plate

All the studies used a Kistler 9281B eight-channel force plate (Kistler Instruments AG, Winterthur, Switzerland) to measure forces applied to the foot. Piezoelectric crystals mounted within the plate produce an electrical voltage proportional to the compressing forces. The force plate was mounted flush with the floor and the plate could be turned through 90° to accommodate differing foot sizes and step lengths.

5. Computers

Computers synchronised both the output of infrared flashes from the cameras and the input from the force plate. Vicon software calculated the three-dimensional trajectories of the markers 50 times per second and assigned them coordinates around three axes (XYZ) relative to a defined origin. The X and Y-axes were parallel to the edges of the force plate and the Z-axis was vertical. Sagittal, coronal and transverse axes, representing anatomical planes, were defined relative to the direction of progression of the subject. Since subjects do not always walk along the X and Y-axes as defined by Vicon such axes are useful in clinical practice. Data smoothing was embedded within the Vicon software.

6. Calibrating image space

The space in which images are to be captured must be calibrated to allow calculations with respect to a frame of reference. Fixed points within this space must be identified (Whittle 1982) and may be obtained from a calibration frame, points suspended from a ceiling or using a dynamic test. During calibration the Vicon system detects the calibration object with one camera and calculates where that camera has to be to have that view. This is then repeated for the other cameras. The Vicon system then reviews the line from each of the cameras to one specific marker and calculates the point of intersection. It then revises its concept of position and modifies the camera positions to achieve the best fit to the known position of the marker. The average distance between the measured and the actual value of the markers can be reported as the 'norm of residuals'. In practice a residual value of less than 5mm is deemed acceptable as the system is unlikely to produce a larger error than this when calculating a marker position. The residual for the PDP 11/73 was less than 5mm and for the VX less than 3mm (Mann 1998). In the early years a calibration frame constructed from steel tubing with aluminium angle joints measuring $0.75m^3$ was used. This was fitted with conical feet that could be placed accurately into studs set in the floor. Retro-reflective markers were attached to known points on the struts of the cube for calibration (Hullin 1990). When the VX was in use, four rods suspended from the ceiling were used for calibration. Dynamic calibration for the Vicon 512 requires the use of an 'L' shaped frame to which four markers are attached to give the absolute origin of the force plate. A 2m wand with two markers placed 50cm apart was then used for calibration. The norm of the residuals is currently 1mm.

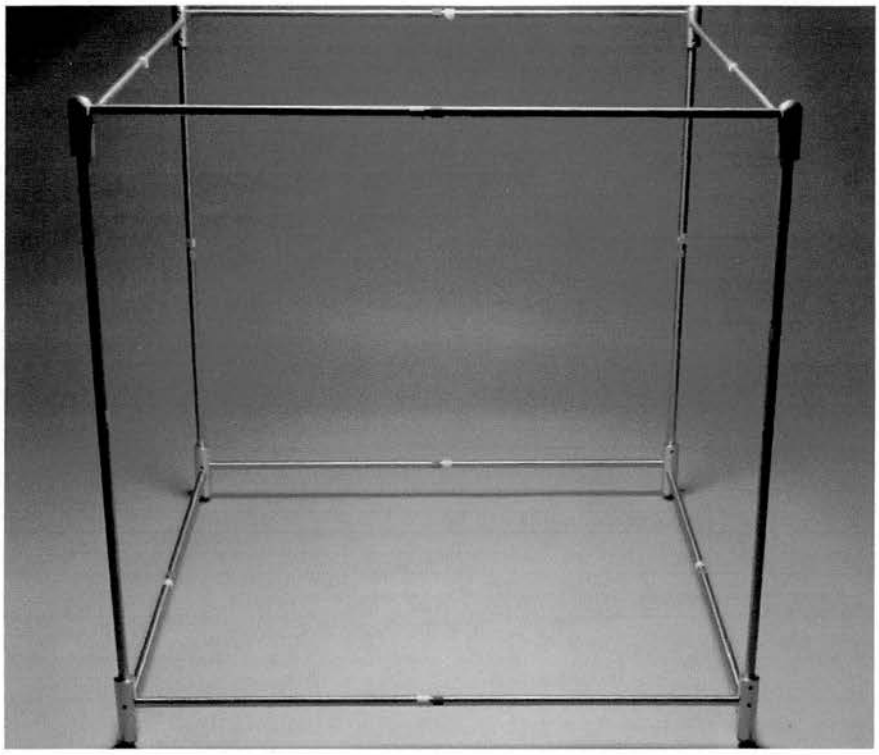

Figure 7 The calibration cube

7. Marker placement

Passive markers consisting of retro reflective material were attached to defined anatomical landmarks on the subjects. Two types of error can occur with marker placement. Relative errors occur as a result of relative movement between two or more markers that define a rigid segment. Absolute errors result from the marker moving in relation to the bony landmark it represents. These errors are caused by the movement of the soft tissue upon which the markers are placed. Although markers are generally placed on the skin overlying a bony landmark, this is not always acceptable to a subject. Hazlewood et al (1997) from the Anderson Gait Analysis Laboratory compared the placement of markers on skin and Lycra over the anterior superior iliac spine and found that there was significantly more marker movement when placed directly on skin rather than Lycra. Internal consistency checks of the force plate, computer hardware and marker accuracy are carried out in the Anderson Gait

Analysis Laboratory at six monthly intervals. This laboratory has an absolute accuracy of 3mm for maker placement (Anderson Gait Analysis Gait Laboratory internal data) and, for example, a coefficient of repeatability (Bland and Altman 1986) of no greater than 6.3° for dynamic hip rotation (Kerr et al 2003). These routine checks are sufficient to identify any inaccuracies that might significantly affect clinical decision-making.

The following landmarks based on Davis et al (1991) were used for marker placement:

a) Before 1996 the *sacral* marker was placed on a 10cm wand whose base was located between the dimples of Venus at the mid point between the posterior aspect of the two posterior superior iliac spines. The marker was aligned such that the wand would be in line with the anterior superior iliac spines in the sagittal plane. When five cameras became available the wand was replaced by a marker placed on the sacrum, midway between the posterior superior iliac spines.

b) Before 1996 the most prominent parts of the *anterior superior iliac spines* and *greater trochanters* were used. Subsequently the trochanteric marker was discarded in favour of a thigh wand and the anterior superior spine marker was retained.

c) After 1996 the *thigh* marker was placed on a wand and located on the mid thighs in a straight line from the greater trochanters to the knee marker. The wand enabled more accurate calculation of joint centres of the ankle, knee and hip.

d) The centre of rotation of the *knees* in the sagittal plane

e) Before 1996 a *shank* marker was placed directly on the shank in the midline on a line between the knee and ankle. This was later replaced by a *tibial* wand.

f) The centre of rotation of the *ankles*

g) Before 1996 the *foot* marker was placed on the base of the fifth metatarsal and subsequently replaced by a marker placed between the second and third metatarsal heads on the dorsal aspect of the foot for the *toe* markers.

h) Before 1996 the *heel* marker was placed on the posterior aspect of the lateral side of the calcaneum in the same horizontal plane as the toe marker. After 1996 it was placed on the midline of the posterior aspect of the heel.

Since 1999 a knee alignment device has allowed the automatic positioning of the tibial and thigh wands. Any limb for which the range of knee ab/adduction exceeded 10° would be considered potentially unreliable for hip rotation data, as errors in sagittal knee data appearing erroneously in the coronal plane would result in inaccurate hip rotation data (Ramakrishnan and Kadaba 1991). These errors may be introduced from marker placement and/or skin movement.

Figure 8 Marker placements; the force plate can be seen behind the child's feet.

8. Definitions of joint co-ordinate systems and joint angles

The Vicon Clinical Manager software was used to establish the joint co-ordinate systems using the above marker protocol. The joint (Cardan) angles were also calculated from the software.

In the sagittal plane, *pelvic tilt* is an antero-posterior movement, and is the angle between the laboratory sagittal axis and the projection of the pelvic sagittal axis. Posterior tilt indicates an upward movement while anterior tilt a downward one. *Pelvic obliquity* is the angle between the laboratory transverse axis and projection of the pelvic transverse axis. This motion may be up or down in the coronal plane. *Pelvic rotation* is the angle between the projection of the laboratory sagittal axis and the pelvic sagittal axis. This motion may be internal or external pelvic rotation.

Hip flexion and *extension* is motion occurring between the thigh sagittal axis and a line connecting the anterior and posterior superior spines. *Hip ab/adduction* is motion occurring between the projection of the pelvic transverse axis and the thigh frontal axis. *Hip rotation* is motion occurring between the projection of the pelvic sagittal axis and the thigh sagittal axis and may be internal or external.

Knee flexion/extension is motion occurring between the thigh sagittal axis and the shank sagittal axis. *Knee varus/valgus* is motion occurring between the thigh frontal axis and the shank frontal axis. Valgus indicates a deviation away from and varus a deviation towards the anatomical mid line. *Knee rotation* is motion occurring between the thigh sagittal axis and the shank sagittal axis and may be internal or external.

Since the marker system does not allow a more detailed analysis of separate motions occurring in the foot and ankle these are considered together. *Ankle dorsi/plantar flexion* is motion occurring between the shank sagittal axis and

projection of the foot. *Foot rotation* is motion occurring between the projection of the shank sagittal axis and the foot in the transverse plane. The *foot progression angle* is motion occurring between the laboratory sagittal axis and the laboratory transverse axis and may be internal or external.

9. Electromyography

Electromyographic activity of muscles is often measured as part of a gait analysis. For larger muscle groups the information may be obtained from surface electrodes and fine wire electrodes placed within smaller muscles (Sutherland 2001). This technique was not used in the studies reported here and therefore is not considered further.

10. Energy consumption

Direct or indirect measurement of energy or oxygen consumption is often used as part of a gait assessment but this technique was also not used for any of the studies reported here and is not considered further.

11. Data acquisition

A full explanation of the procedure was given and written consent obtained from the subject or parent for the examination, video and photographic records. The subject or parent completed the self-report Gillette Functional Assessment Questionnaire (Novacheck, Stout, and Tervo 2000) where appropriate. Each subject was allowed to acclimatise him/herself to the environment that was also child friendly. The subjects wore Lycra shorts and a T-shirt or a bathing costume during gait analysis and physical measurements.

The subject's height, weight and limb lengths were measured. The bony contours of the patellae and heels were marked using a pen and subjects wore rotation indicators (Hillman et al 1998) to assist the visual assessment of gait.

The subjects then performed several test walks to familiarise themselves with the laboratory environment. They were usually asked to walk at their normal, comfortable speed. The starting point on the walkway was altered until a clean heel strike was obtained for the limb under consideration. Normally a total of three right and three left strikes on the force plate were obtained. In the studies investigating the effects of speed on gait parameters, subjects were asked to walk more slowly or quickly at a speed of their choice. A metronome was not used to control cadence or step length as this was shown to affect normal gait (Sekiya et al 1996, Zijlstra et al 1985). The subjects' gait was also recorded on videotape. An example of kinematic and kinetic output from a patient who has spastic diplegia is illustrated in Figure 9a-c.

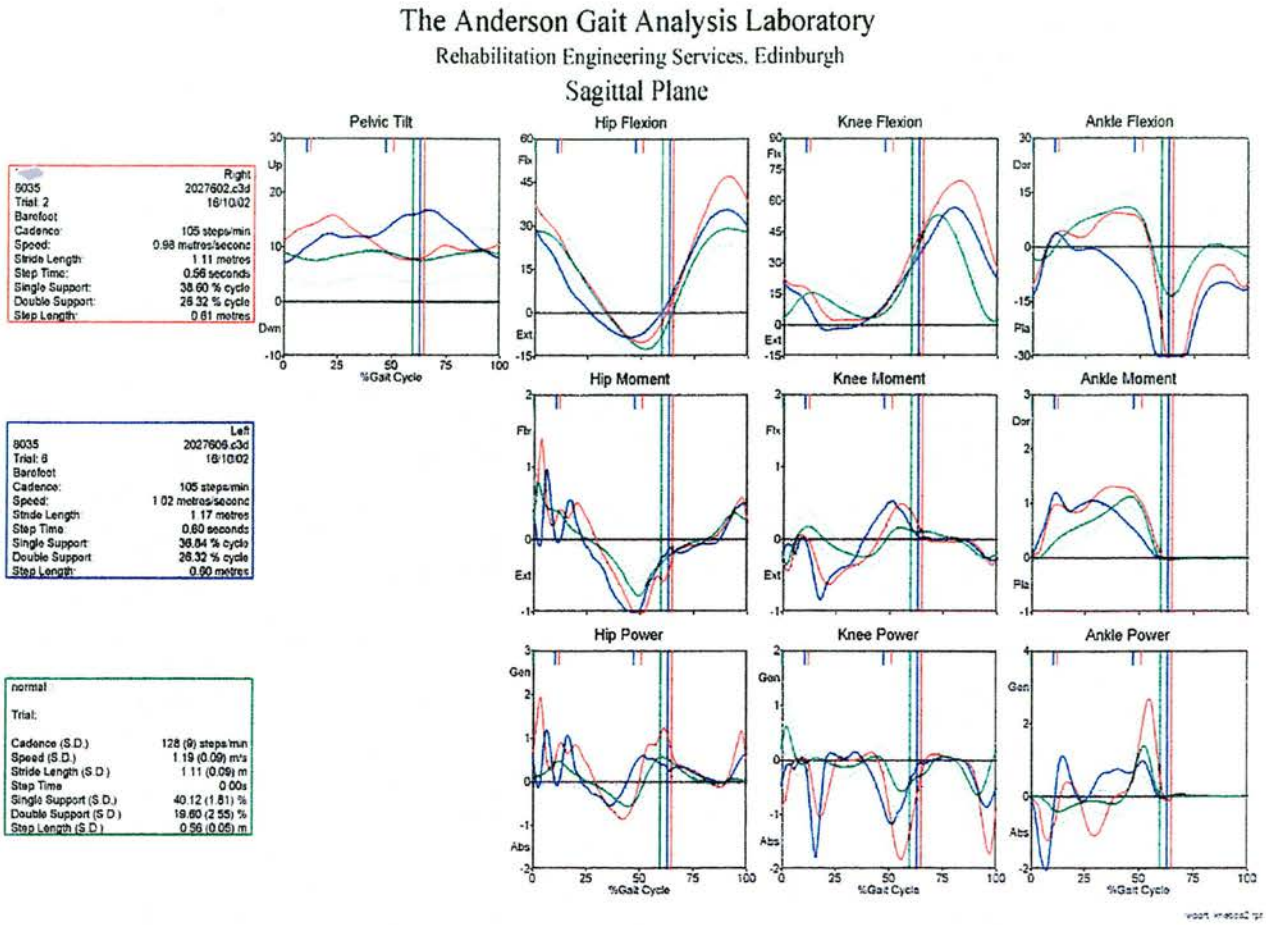

Figure 9a Sagittal plane kinematics and kinetics for a patient who has spastic diplegia. The green traces indicate normal values +/- 1sd from the laboratory's data base; the red trace indicates the patient's right leg and the blue the left.

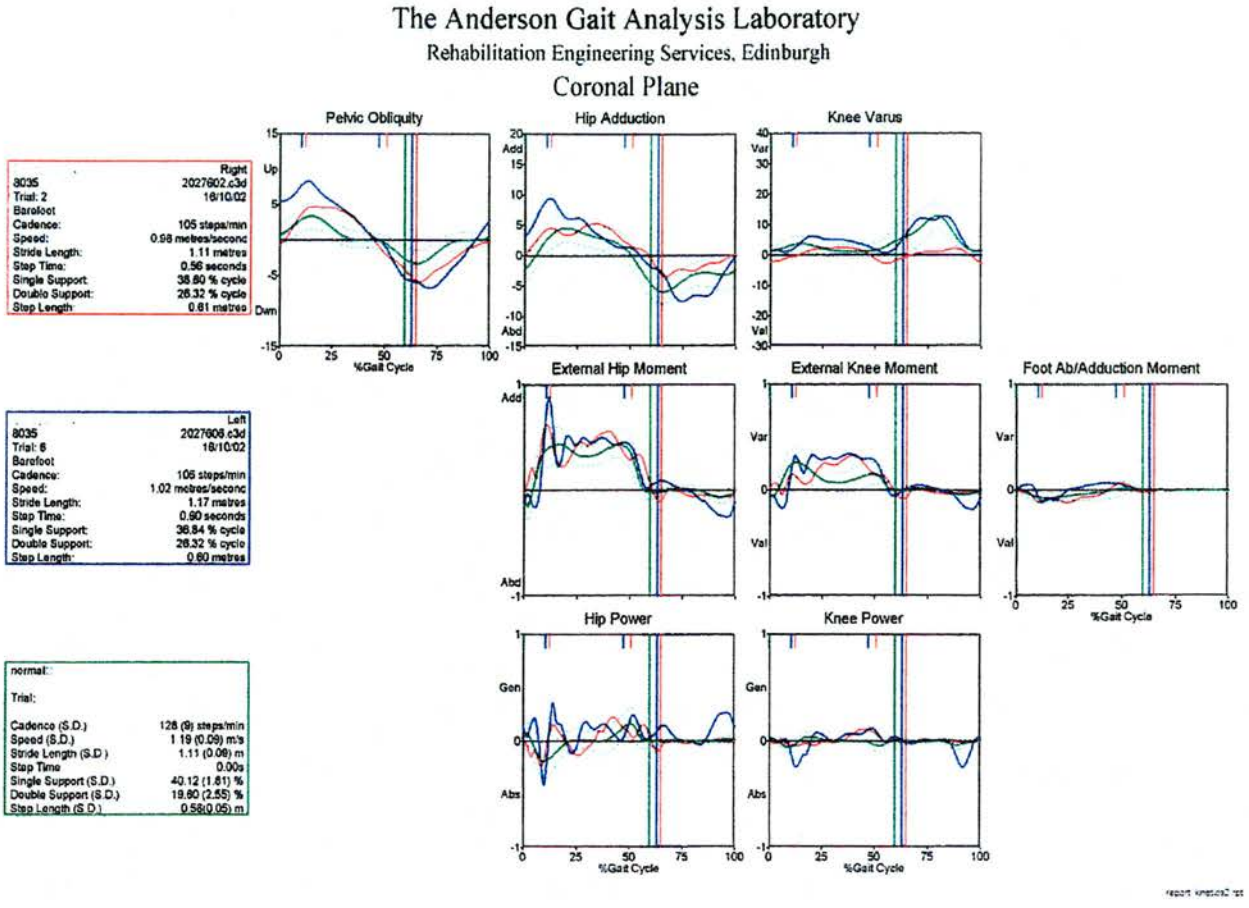

Figure 9b Coronal plane kinematics and kinetics for a patient who has spastic diplegia. The green traces indicate normal values +/- 1sd from the laboratory's data base; the red trace indicates the patient's right leg and the blue the left.

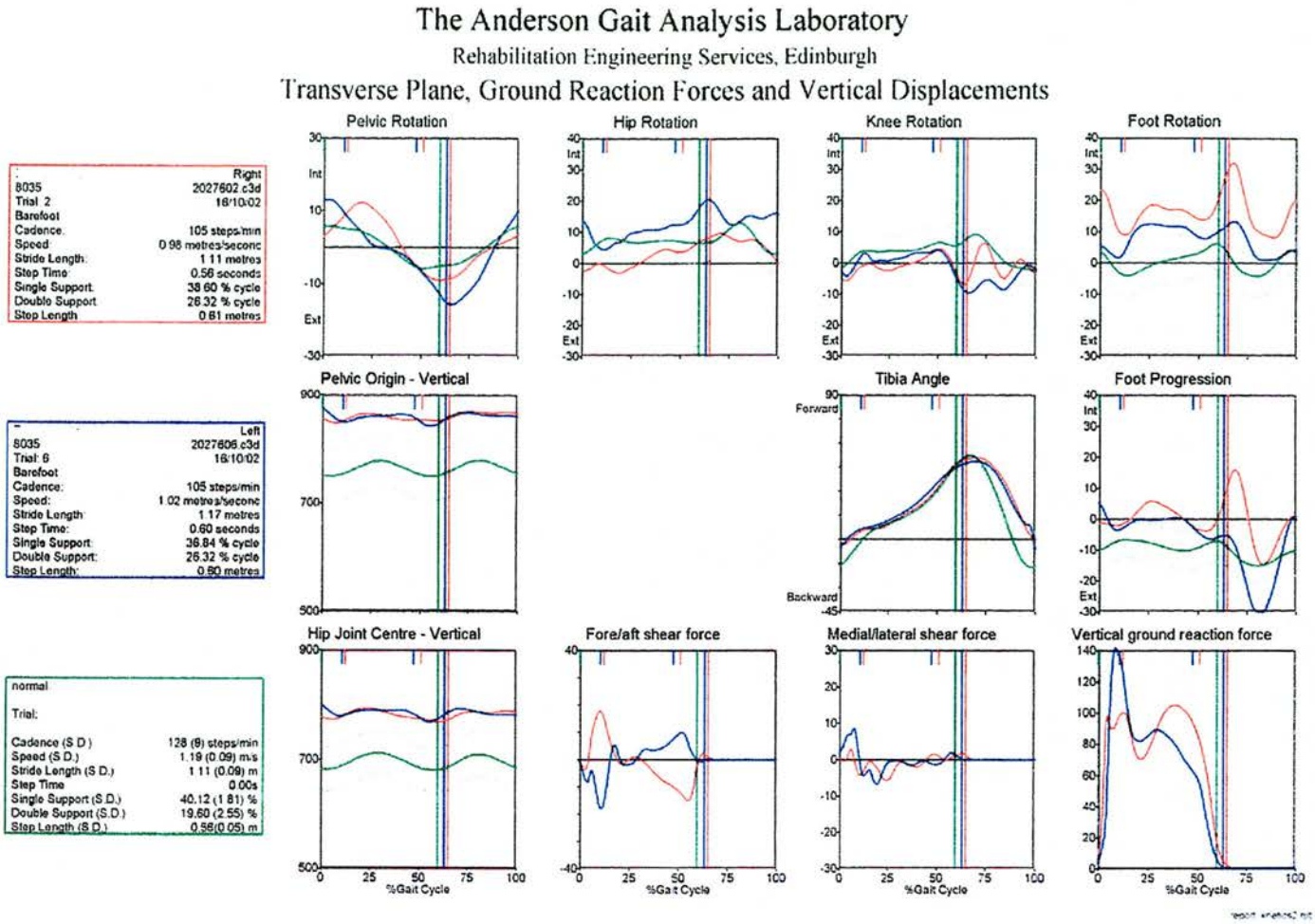

Figure 9c Transverse plane kinematics and kinetics for a patient who has spastic diplegia. The green traces indicate normal values +/- 1sd from the laboratory's data base; the red trace indicates the patient's right leg and blue the left.

Once the kinematic and kinetic data had been collected, two observers used a standard protocol to examine the subjects. Two assessment charts were used; one for patients with cerebral palsy (Figure 10), the other for those with myelomeningocele (Figure 11). Each chart has four sections to record the range of joint motion, tone (Ashworth 1964), muscle strength (Medical Research Council 1943) and anthropometrical data. In our laboratory the interobserver reliability of measuring range of motion varies according to the anatomical location. The best coefficient of reliability was seen for femoral anteversion (3°) and measurements such as hip rotations knee and ankle flexion/extension 10°. The least satisfactory coefficient was 20° for straight leg raising and popliteal angle (Anderson Gait Analysis Laboratory internal data).

Anderson Gait Laboratory

Physical examination chart (with active range)

GA16 (V12 Front

Name:..... DoB: ..... PGA: .....

	RIGHT						LEFT					
Date												
Initials examiner/measurer												
Normality manual goniometry	Mark V against value if vis estimation						Mark V against value if vis estimation					
Start, End of range	S	End	S	End	S	End	S	End	S	End	S	End
Hip flexion knee flexed												
Popliteal angle, opp knee flexed												
Hip flexion knee extended												
Abduction knee flexed												
Abduction knee extended												
Adduction												
Hip extension *Stahell(S) Thomes(T)*												
Internal rotation *prone*												
External rotation *prone*												
Femoral anteversion *prone*												
Duncan Ely -ve/+ve *prone*												
Point of catch in rectus *prone*												
Knee flexion hip extended *prone*												
Knee extension												
Knee flexion												
Dorsiflexion knee flexed												
Dorsiflexion knee extended												
Forefoot equinus/midfoot break												
Plantarflexion knee flexed												
Plantarflexion knee extended												
Hindfoot thigh angle *prone*												
Hindfoot forefoot angle *prone*												
Subtaloid inversion *prone*												
Subtaloid eversion *prone*												
Midfoot inversion												
Midfoot eversion												
Forefoot adduction												
Forefoot abduction												

ACTIVE RANGE (against gravity) and MUSCLE STRENGTH

In column 'R' put range i.e., I= inner, to shortened position; M=mid, O=outer, in lengthened position; nil = no movement. In column '+ r' put range maintained against resistance (i.e., I, M, O, nil). Add MRC 4 or 5 if inner range (i.e., I4).

Hip flexors - supine												
Quads- supine, hip extd, knee fixd												
Quads lag												
Extensors - prone knee flexed												
Extensors - prone knee extended												
Plantarflexors – prone/ SLS tiptoe												
Abductors - side lying												
Dorsiflexors – sitting, voluntary												
Dorsiflexors – sitting, triple flexion												
Confidence in measurements Good/Fair/Poor/see notes												

Name:..... DoB: .....

GA16 (v12) Reverse

Tone

Use Ashworths score, 0 = no increase, 1 = slight catch, 2 = slight catch + slight resistance, 3 = more marked increase but easily moved, 4 = considerable increase, passive movement difficult, 5 = affected part rigid in flexion or extension

	RIGHT			LEFT		
Date						
Hip flexors						
Adductors						
Internal rotators						
Rectus femoris						
Hamstrings (medial/lateral)	/	/	/	/	/	/
Tibialis anterior						
Extensor digitorum/hallucis	/	/	/	/	/	/
Triceps surae						
Tibialis posterior						
Flexor digitorum/ Flexor hallucis	/	/	/	/	/	/
Peroneí						
Clonus						
Type of spasticity						
Confidence in measurements Good/Fair/Poor/see notes						

Indicate with down arrow if muscle bulk is reduced compared to the opposite side

MUSCLE BULK						
Quadriceps						
Calf						
BALANCE						
Sitting: Good/Fair/Poor						
High kneeling: Good/Fair/Poor						
Standing: Good/Fair/Poor						
SLS, secs						
GAIT						
Sitting						
Standing						
Patella alta (Y/N)						
Hallux valgus (Y/N)						
Leg length						
Inter ASIS measurement						
Condyler width						
Malleolar width						
Malleolar width boots						
Transmalleolar axis - footprint						
Transmalleolar axis - jig						
Height						

Notes:

Examiners signature and printed name:

Session 1: ..... Session 2: ..... Session 3: .....

Figure 10 Assessment chart used for patients who have cerebral palsy

Anderson Gait Laboratory

Physical examination chart (with MRC grades)

GA9 (V9) Front

Name:..... DoB: ..... PGA: .....

	RIGHT						LEFT					
Date												
Initials examiner/measurer												
Normality measure goniometry	Mark V against value if via estimation						Mark V against value if via estimation					
Start, End of range	S	End	S	End	S	End	S	End	S	End	S	End
HIP												
Hip Flexion knee flexed												
Popliteal angle, biset												
Hip flexion knee extended												
Abduction knee flexed												
Abduction knee extended												
Adduction												
Hip extension *Staheli(S) Thomas(T)*												
Internal rotation *prone*												
External rotation *prone*												
Femoral anteversion *prone*												
Knee flexion hip extd *prone*												
KNEE												
Knee extension												
Knee flexion												
ANKLE/FOOT												
Dorsiflexion knee flex'd												
Dorsiflexion knee ext'd												
Forefoot equinus/midfoot break												
Plantarflexion knee flx'd												
Plantarflexion knee ext'd												
Hindfoot thigh angle *prone*												
Hindfoot forefoot angle *prone*												
Subtaloid inversion *prone*												
Subtaloid eversion *prone*												
Midtarsal inversion												
Midtarsal eversion												
Forefoot adduction												
Forefoot abduction												
Confidence in measurements Good/Fair/Poor/see notes												

Patella alta (Y/N)						
Hallux valgus (Y/N)						
Leg length						
Inter ASIS						
Condylar width						
Malleolar width						
Malleolar width boots						
Transmalleolar axis -footprint						
Transmalleolar axis - jig						
Height						

Examiners signature and printed name:

Session 1: ..... Session 2: ..... Session 3: .....

Name:..... DoB: ..... PGA: .....

GA9 (V9) Reverse

Muscle strength

Use MRC grades: 1 = flicker, 2 = full range gravity eliminated, 3 = full range against gravity, 4 = full range against gravity and some resistance, 5 = normal power

		RIGHT			LEFT		
Date							
Hip							
Hip flexors	L1,2,3						
Hip extensors knee flexed (gluteus maximus)	L5,S1,2						
Hip extensors knee extended (hamstrings)	L5,S1,S2						
Abductors	L5,S1						
Adductors	L2,3,4						
Internal rotators	L2,3						
External rotators	L3,4,5,S1,2						
KNEE							
Quadriceps	L2,3,4						
Medial hamstrings	L5,S1,2						
Lateral hamstrings	L5,S1,2						
ANKLE/FOOT							
Tibialis anterior	L4,5						
Extensor digitorum longus	L5,S1						
Extensor hallucis longus	L5,S1						
Peroneus brevis	L5,S1,2						
Peroneus longus	L5,S1,2						
Tibialis posterior	L4,5						
Gastrocnemius	S1,2						
Soleus	S1,2						
Flexor digitorum longus	S1,2,3						
Flexor hallucis longus	S1,2,3						
Intrinsics	S2,3						
Confidence in measurements							

Indicate with down arrow if muscle bulk is reduced compared to the opposite side

MUSCLE BULK						
Quadriceps						
Calf						

GAIT						
Sitting						
Standing						

Examiners signature and printed name:

Session 1: ..... Session 2: ..... Session 3: .....

Notes:

Figure 11 Assessment chart used for patients who have myelomeningocele

12. Data interpretation

The results from the analysis were collated and followed by a provisional assessment of problems associated with a pathological gait pattern. A definitive report was issued after the gait laboratory team had considered the provisional report. Inevitably there were staff changes between 1988 and 2002 but the professional composition of the team always included a Bioengineer, a Physiotherapist and the author, an Orthopaedic Surgeon. The statistical methods used in the studies are to be found in the relevant articles.

MATURATION OF GAIT AND A NORMALISATION METHOD FOR USE IN CHILDREN

This section refers to the two following papers which will be considered together because they share the same underlying ideas and methods:
i) Stansfield BW, Hazlewood ME, Hillman SJ, Lawson AM, Loudon IR, Mann AM, Robb JE. Normalised speed not age characterises ground reaction force patterns in 5-12 year old children walking at self-selected speeds. Journal of Pediatric Orthopaedics 2001; 21: 395-402.
ii) Stansfield BW, Hazlewood ME, Hillman SJ, Lawson AM, Loudon IR, Mann AM, Robb JE. Sagittal joint angles, moments and powers are predominantly characterised by speed of progression and not age in 7-12 year old normal children. Journal of Pediatric Orthopaedics 2001; 21: 403-411.

Walking evolves through a series of phases in the first three years of life - newborn stepping, infant supported walking, infant independent walking and child walking (Okamoto, Okamoto and Andrew 2003). Normal children achieve independent walking at about fifteen months of age and gait continues to mature thereafter. Burnett and Johnson (1971) found that pelvic tilt was present three weeks after achievement of independent walking and pelvic rotation occurred about a week later. Knee flexion in mid-stance was present at 15 weeks and a mature base width at 17.5 weeks. Heel strike was present 22.5 weeks after independent walking. They concluded that these features were mature by 55 weeks after the acquisition of independent walking. Statham and Murray (1971) studied walking in seven normal children at two different developmental stages. The first record was made of supported walking when the children were able to make deliberate stepping movements but still required outside support from their mother to progress forward while maintaining their balance in the erect position. The second record was made of the same children once they were able to walk at least 6ft without any support either initially or during walking. Interrupted-light photography was used to record sagittal

kinematic data. Supported walking patterns showed excessive hip, knee and ankle flexion and irregular flexion-extension reversal in stance when compared with normal adult gait. In swing there was excessive hip flexion and toe dragging. When the children achieved independent walking marked differences with adult walking remained.

Grieve and Gear (1966) studied the relationship between stride length, step frequency, time of swing and speed of walking in 50 children and adults. The subjects comprised seventeen children, the youngest being one year old. They concluded that in the early months of walking step frequency bore no relationship to walking speed and that the product of maximum step frequency and the square root of the stature was approximately constant after five years of age. Children acquired a constant time of swing by age of four to five years. They concluded that temporo-spatial parameters of gait in children attained adult values by the age of five years.

Sutherland et al's (1980) account of the development of gait represents a landmark in studies of gait maturation. Cross sectional gait studies were performed on 186 normal children whose ages ranged between one and seven years. Rotations of the lower-extremity joints in the sagittal, frontal, and transverse planes; step length; cadence; walking velocity; and duration of single-limb stance (as percentage of the gait cycle) were all analyzed throughout a walking cycle using high-speed movies, a Graf-Pen sonic digitizer, a computer, a plotter and EMG. The sagittal-plane angular rotations in children from the age of two years onwards were very similar to those of normal adults. Subjects less than two years old had greater knee flexion and more ankle dorsiflexion during stance phase, while their knee-flexion wave (stance-phase knee flexion after foot-strike and subsequent knee extension before toe-off) was diminished. External rotation of the hip in these younger subjects was pronounced.

Reciprocal arm-swing and heel-strike were present in most children by the age of eighteen months. Sutherland et al (1980) went on to classify five important determinants of mature gait: duration of single-limb stance, walking velocity, cadence, step length, and the ratio of pelvic span to ankle spread. As the subjects matured, cadence decreased while walking velocity and step length increased. These authors felt that important factors in the development of a mature pattern of these determinants were increasing limb length and greater limb stability. This is manifested by the increasing duration of single-limb stance, an index of limb stability. A mature gait pattern as determined by these criteria was well established at the age of three years. Sutherland et al also reported that maturation of vertical forces occurred at 5-6 years and both the fore-aft and mediolateral shear forces changed little between 2-7 years. Heel strike, knee flexion wave and reciprocal arm swing appear very early and have been used as a gauge of maturity. Their presence does not necessarily indicate maturation but their absence may indicate pathological gait. Heel strike was present by 18 months, a knee flexion wave at 24 months, and reciprocal arm swing by 48 months.

Subsequent studies of gait in children suggested that gait had not matured as early as Sutherland et al (1980) had suggested. Rose Jacobs (1983) compared the gait at slow, free and fast speeds of three and five year old children using an ink foot print technique and concluded that gait patterns were not yet fully mature at five years. Greer, Hamill and Campbell (1989) studied sagittal kinematics and ground reaction forces in 18 children aged between three to four years. Differences in the kinematics and forces were evident between boys and girls at these ages. These children also showed an earlier transition for braking to propulsion at 37.5% of stance as compared with 50% of stance for adults. Norlin, Odenrick and Sandlund (1981) used foot

switches in 230 normal children aged between 3-16 years to evaluate temporo-spatial parameters. Changes were most pronounced up to 8-10 years but continued up to 16 years. They considered velocity and stride length to be mainly dependent on the age of the child. Menkveld, Knipstein and Quinn (1988) used electrodynagrams, pressure sensitive foot transducers, to measure vertical reaction forces in 60 children aged between 7-16 years. They noted that subtle development of gait patterns continued after the age of seven into adolescence and reported a persistence of a foot-flat position in stance. They felt that this was a residual sign of immaturity leading to a decreased rotation about the longitudinal axis of the foot.

Beck et al (1981) studied temporo-spatial and ground reaction forces in 51 normal children aged between 11 months and 14 years. The children underwent three monthly examinations for one year. They concluded that differences in temporo-spatial and reaction forces are dependent on walking speed and age. The three ground reaction components, standardised to the child's weight, increased between the ages of one and five years and then remained constant. This was the first report of a longitudinal study of ground reaction forces and they concurred with Sutherland et al's (1980) findings. In contrast to Beck et al (1981), Noguchi (1986) reported that the components of the ground reaction force matured at different times. He found that the sagittal component matured by about five years of age, the vertical component by six and the lateral component by nine. Takegami (1992) studied ground reaction forces in 241 children aged between 4-10 years. The wave patterns of the three components were normalised to body weight and time and expressed as a percentage. Significant changes were seen in both the timing and magnitude of the waves to eight years of age. He also noted that velocity steadily increased with age and that there was no change in the step length/height ratio after the age of six years. The step width/height

ratio did continue to decrease with age. These three reports suggest that the kinetics of gait do not necessarily mature synchronously and that there were differences as to when gait was considered to have matured. It should be noted that the studies by Noguchi (1986) and Takegami (1992) were performed on oriental populations which may explain the observed differences between these authors and others.

Katoh, Mochizuki and Moriyama (1993) investigated the changes in ankle-joint kinematics and ground reaction force during gait in normal children aged 4-10 years. Dorsiflexion of the ankle joint becomes slower and better controlled from early to mid stance phase as age increases. The main changes were found at about five years of age. The peak for the fore-aft shear component of the ground reaction force in the deceleration phase occurred later, and was noted during dorsiflexion. However, the peak for plantar flexion, plantar flexion angular velocity and the fore-aft shear component of ground reaction force did not change significantly with increasing age. Their study of a specific segment of the lower limb confirmed Sutherland et al's (1980) earlier findings. Gomez Pellico, Rodriguez Torres, and Dankloff Mora (1995) studied ground reaction forces and temporo-spatial parameters in 62 children aged between 5-6 years old. In contrast, they found that there were still appreciable changes in temporal and spatial parameters and kinetics of walking indicating that gait was not yet fully mature in their subjects. Instead of using force plates, Preis, Klemms and Muller (1997) used instrumented pairs of shoes with eight force transducers sampling at 50 Hz to study ground reaction forces in a younger population of 54 normal children aged between 1-5 years. They found that by the age of 24 months the shape of the ground reaction forces resembled those of an adult pattern. Slight changes were achieved until the age of five years when the pattern became adult. Phases of double support expressed as a percentage of the gait cycle decreased

significantly from the age of 1-5 years, the steepest decrease occurring in the first year of independent walking. There was no left/right asymmetry. Their results are in contrast to those in the studies of Katoh, Mochizuki and Moriyama (1993) and Gomez Pellico, Rodriguez Torres, and Dankloff Mora (1995). In a more recent study of 23 children aged between 4-10 years, Cupp et al (1999) found that children had diminished hip abduction, plantarflexor and knee extensor moments and A2 power generation as compared with five adults. Differences were more noticeable in 4-5 year olds than in 8-10 year olds. They used the Cleveland Marker set normalised to body weight and concluded that gait continued to mature up to and including the age of eight years.

From the foregoing it can be seen that uncertainty remains as to when gait is mature in children and that there are two outstanding problems. Firstly, the evidence presented so far has relied on cross sectional data without the benefits of longitudinal data apart from the brief follow up in Beck et al's (1981) study. Secondly, the effects of both age and speed need to be taken into account as children grow. This is a fundamental concept since two children of the same age walking at the same speed may have very different height and weight. Such differences will affect accelerations that are a key factor in gait and determine inertial forces that, together with gravitational force, determine ground reaction forces, joint forces and moments (Stansfield et al 2001). Walking at self-selected speeds may not be representative of a particular age group while walking at specified speeds may affect gait as well. Some authors have used normalisation to present gait data. For example, Kadaba et al (1989) considered normalisation of joint moments to body weight and height in 53 subjects (age range 4-40 years) to enable comparisons between individuals or groups of subjects. They concluded that normalisation to body weight results in a significant

reduction of intersubject variability whilst normalisation to body height or leg length has a smaller effect. Although Kadaba et al (1989) recorded joint moments they did not consider the effects of walking speed and accelerations. Beck et al (1981) concluded that ground reaction forces in terms of multiples of body weight do not change after the age of five years. In effect they were comparing children moving at different relative speeds as compared with their heights (Stansfield et al 2001). Ounpuu, Gage and Davis (1991) used normalisation in their analysis of joint moments in children, dividing joint moments by the subject's mass. This normalised moment depended on the dimensions of the subjects and could only be justified if the subjects are approximately of the same mass. Children's sizes vary and it is likely that this could have a significant effect on their moment results. Hof (1996) proposed a set of dimensionless variables based on similarities of acceleration to reduce intersubject variability. This approach is appropriate in a paediatric population as it removes differences due to varying heights and/or mass.

Hof's method (1996) of using non-dimensional units appeared an appropriate solution to clarify the difficulties of earlier attempts at normalisation of data. Thus it was decided to study prospectively a cohort of normal children, firstly to establish when gait matured and secondly to provide reference data in children in non-dimensional units. The study took seven years to collect data and has provided unique information that has not been published previously. The normalisation method has shown that age is not a factor in the characterisation of kinematics and kinetics of gait for this population but rather normalised speed is the key factor. We were able to conclude that the ground reaction forces were mature by five years and the sagittal joint kinematics, moments and powers were mature by the age of seven years, thus confirming Sutherland et al's (1980) conclusions from their cross-sectional studies.

The clinical relevance of these two papers is firstly that normalised speed of walking should be considered when comparing normal with pathological gait and secondly that age matched controls do not suffice. Since these two publications the same group from the Anderson Gait Analysis Laboratory (van der Linden et al, 2002) used this method to study healthy children walking at slow speeds commensurate with speeds seen in pathological gaits such as cerebral palsy. The relevance of this particular study was to indicate that the kinematic and kinetic changes occurring at slow speeds of walking may need to be distinguished from the underlying gait pathology.

A VALIDATED VISUAL SCORE FOR USE IN CEREBRAL PALSY

This section refers to the following paper: Read HS, Hazlewood ME, Hillman SJ, Robb JE, Prescott RJ. Edinburgh Visual Gait Score for use in cerebral palsy. Journal of Pediatric Orthopaedics 2003; 23: 296-301.

The visual assessment of gait is integral to clinical practice and gait analysis. One argument in favour of instrumented gait analysis is that the eye cannot perceive events occurring more rapidly than $1/12^{th}$ of a second (Gage 1991) and that rotation out of the plane of progression when walking can be misinterpreted. Torsional anomalies of the femur, tibia and foot secondary to neurological conditions such as cerebral palsy or myelomeningocele are frequently seen in pathological gait. Muscle strength imbalance and abnormal tone, particularly around the hip, can also produce abnormal lower limb rotations in gait even thought there may not be any underlying skeletal torsions. There is often a combination of bony torsions and muscle imbalance.

The effect of rotation on the visual appreciation of an angle can be demonstrated in Figure 12a-c on page 47. In the model of the lower limb the 'knee' angle between the 'femur' and 'tibia' is the same in all three illustrations in Figures 12a-c. The model has been rotated through 60° giving the erroneous visual impression that the angle between the femur and tibia has changed.

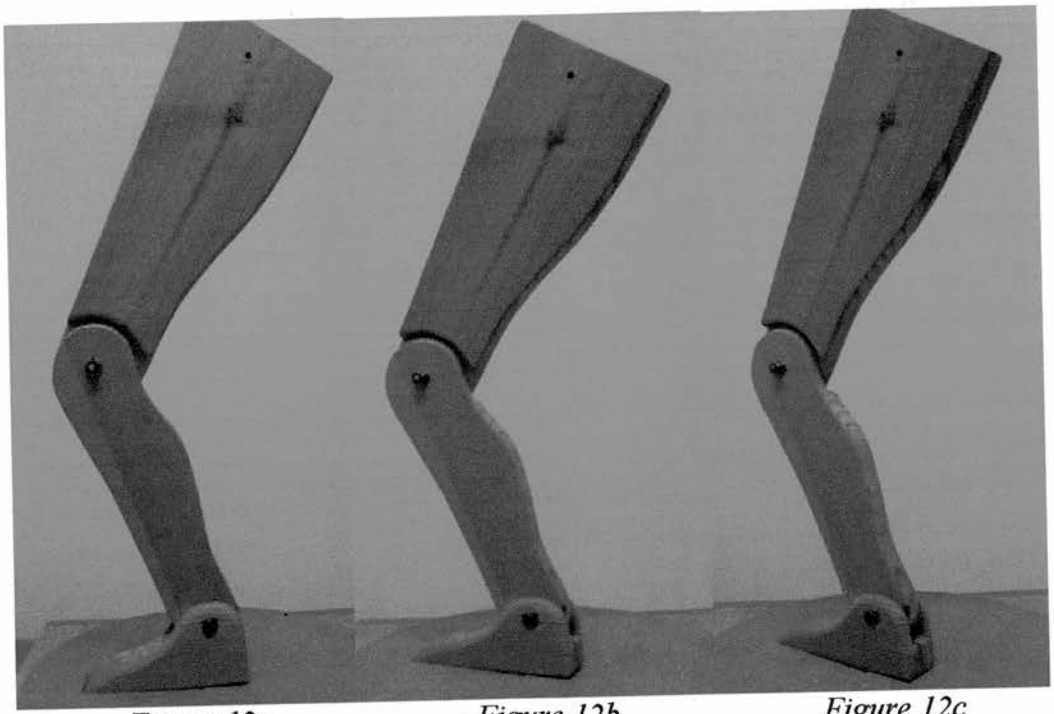

Figure 12a *Figure 12b* *Figure 12c*

Figure 12a-c. The angle between the 'femur' and 'tibia' remains the same; the rotation of the model is 0° (12a), 45° internal (12b) and 60° internal (12c). Increasing internal rotation of the model appears to alter the diminish the 'knee' angle from about 50° (12a) to 40° (12c).

All analytical techniques have limitations and it is important for the observer to be aware of possible sources of error and to make allowances for this in the interpretation of gait. Marker systems used in gait analysis are imprecise at the level of the foot and ankle; the assessment of this segment of the lower limb relies very much on visual assessment.

None of the existing methods of visual assessment of gait in cerebral palsy has been validated specifically for cerebral palsy. Reimers (1972) produced a scoring system for the evaluation of walking in patients with cerebral palsy based on functional ability but this was not validated in any way. Krebs, Edelstein and Fishman (1985), studied videotape recordings of 15 children, 12 of whom had myelomeningocele, two had cerebral palsy and one had a flaccid paraparesis. Three experienced physiotherapists observed gait from videotape as the children walked

when using two different types of knee ankle foot orthosis. Only the lower limbs were analysed, as the raters were unable to agree on the designations for trunk abnormalities during gait. Reliability was poor for the analysis of swing and stance data and so only stance was considered in the study. Rater values of 'normal', 'just noticeably abnormal' and 'very noticeably abnormal' were used without attempting to quantify these values. Data analysis considered the within-rater and between-rater reliability. Identical ratings for both between- and within-raters occurred in 66% of observations. An additional 29% of observations differed by one point. The between-rater intraclass correlation coefficient type 2.1 was 0.73 and within-rater Pearson product moment correlation averaged 0.60. They concluded that specially trained observers are only moderately reliable in assessing the kinematics of gait of disabled children wearing orthoses.

Lord, Halligan and Wade (1998) produced a four-point scale visual gait assessment form for clinical use with patients with neurological deficits. Their study design included a reliability study based on ten patients. Two raters also assessed eight different patients one week apart. Forty-seven patients were assessed and the data was used to examine validity, reliability and sensitivity to change. Other comparative measures used were walking time, stride length, step length asymmetry, balance and a Mobility Index. Inter-rater reliability between multiple raters was considered to be reasonable both for the global scores from the gait assessment form and for the individual items. Complete agreement occurred on 63.8% of all observations and there was a significant correlation between the global score and the various criterion measures and also between changes in the global score and change in walking time in patients who had received treatment. This well designed study gives an indication of the agreement that might be anticipated for inter-rater reliability when

assessing neurological patients. However, it should be noted that the study was undertaken in adults and not in children with cerebral palsy.

Eastlack et al (1991) studied the inter-rater reliability of videotaped observational gait analysis assessments. This study was carried out in adults but used a large number of raters. Fifty-four physiotherapists with varying clinical experience served as raters to assess sagittal and coronal videotapes of three patients with rheumatoid arthritis when walking. Each patient's most severely involved knee was analysed during the four sub phases of stance for the kinematic variables of knee flexion and genu valgum. The raters were asked to determine whether these variables were inadequate, normal, or excessive. Generalized kappa coefficients ranged from 0.11 to 0.52 and intraclass correlation coefficients were slightly higher. The authors concluded that physiotherapists' visual gait assessments were only slightly to moderately reliable.

To overcome problems of assessing rotations in gait, Hillman et al (1998), from the Anderson Gait Analysis Laboratory, devised rotation indicators. These lightweight markers have a half-circular cross-section and a card marked with patterned strips covers the convex surface of each indicator. Each strip subtends an angle of 20°. The principle behind the indicators is that indicators attached to a segment rotating in the laboratory transverse plane will present a varying face to the camera. This could then be used to relate the position of the indicator to other landmarks such as the anterior superior spines of the pelvis thus indicating the relative rotation of one segment to another (Hillman et al 1998).

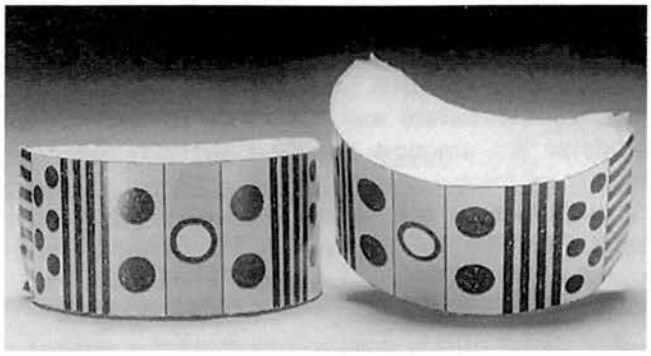

Figure 13 Rotation indicators. Each patterned stripe subtends an angle of 20° (Reproduced with permission)

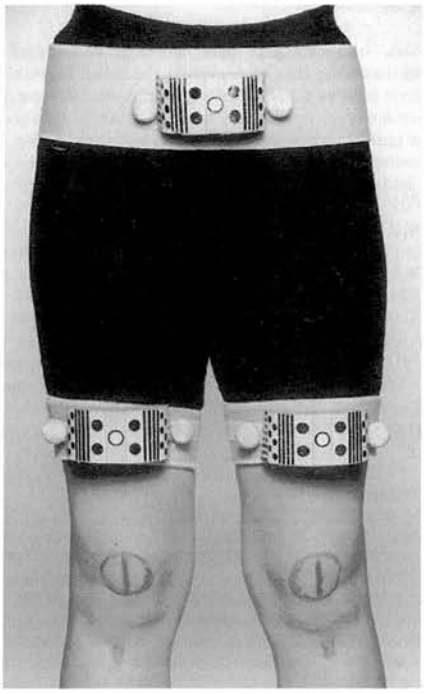

Figure 14 The indicators attached to a subject. An indicator has also been attached to the mid point of the two anterior superior iliac spines to provide a reference point for the thigh indicators. An estimate may be made of the rotation of the thigh relative to the pelvis. (Reproduced with permission)

Instrumented three dimensional gait analysis is not readily available in many parts of the world. This lack of a validated tool for the visual assessment of gait in cerebral palsy prompted Read et al (2003) to produce the Edinburgh Visual Gait Score. Seventeen observations that they considered to be key items of pathological gait in patients with cerebral palsy were selected. Five observers studied video

recordings of five subjects (four patients, one normal control). The observers randomly assessed pre-operative and post-operative video recordings of the four patients. These pre-operative recordings and that of the normal subject were also assessed twice in a varied order by each observer with a minimum of two weeks between assessments to calculate intraobserver variation. The 10 numeric items of the score were also correlated with those obtained from instrumented gait analysis for the same patients. The score used a three-point ordinal scale where normality was considered to be <+/- 1.5 standard deviations (sd), a moderate deviation 1.5-4.5 sd from the mean and marked deviation >4.5 sd from the mean. The score demonstrated good intra-observer reliability and a 70% exact agreement between observers for 2,550 observations. The level of agreement varied between 55% for knee flexion in terminal swing and 96% for initial contact. A comparison of the 10 numeric items with instrumented gait analysis was 64%. This was perfect agreement of two thirds of the 2,550 observations. The score was able to detect change post-operatively and a change of three points was considered to represent real change. The clinical value of this score is that it now makes the quantitative visual assessment of gait possible. Until 2003 there were no validated scores for the visual assessment of gait for cerebral palsy.

Mackey et al (2003) recently investigated the reliability and validity of visual gait assessment in children with spastic diplegia using a modified version of the Physicians Rating Scale (PRS). This was the Observational Gait Scale (OGS). The PRS (Koman 1994) was designed to evaluate the after-effects of injections of botulinum toxin into the calf muscles of children with cerebral palsy. There are six elements to the PRS – knee crouch, equinus, hind foot position, knee recurvatum, speed of gait and gait pattern all of which are based on video assessment. Corry et al

(1998) found that the PRS lacked sensitivity and reliability and they reduced the six elements to three in their study comparing stretching casts with botulinum toxin injections. They also added another category, 'change', to their modification thus giving a fourth element to the modified score. Boyd and Graham (1999) altered the PRS further by expanding it to eight sections, renaming it the OGS. However, it is not known how these changes have affected the reliability of the Scale (Mackey et al 2003). In Mackey et al's (2003) study, two clinicians experienced in the OGS and gait analysis viewed edited split-screen video recordings of 20 children/adolescents made at the time of three-dimensional gait analysis (3-DGA). Using the first seven sections of the OGS, each child's walking ability was scored at initial assessment and reassessed from the same videos three months later. Validity of the OGS score was determined by comparison with 3-DGA. The OGS was found to have acceptable inter-rater and intra-rater reliability for knee and foot position in mid stance, initial foot contact, and heel rise. Comparison with 3-DGA suggested that these sections might also have high validity. Base of support and hind-foot position had lower inter-rater and intra-rater reliabilities and these were not easily validated by 3-DGA. None of the participants had 'significant' transverse plane problems or 'major' mid-foot deformities. They also used a time limit for the evaluation of the video material rather than allowing observers to use as much time as required for visual assessment which would reflect clinical practice more closely.

Mackey et al's (2003) study shares a similar approach to that of Read et al (2003) in which the video material was compared with three dimensional gait data. However, in contrast to Mackey et al's (2003) study, Read et al (2003) did include foot deformity and also considered transverse plane pathology as well as using a more comprehensive validated number of elements to their score: 17 versus six in Mackey

at al's (2003) study . The score developed by Read et al (2003) for use in cerebral palsy therefore appears to be more comprehensive than that of Mackey et al (2003).

The Edinburgh Visual Gait Score (2003) has been in use in the Anderson Gait Analysis Laboratory for two years having evolved from an earlier version (Read et al 1999). The present version is convenient to use and has been shown to be reliable for the 17 selected observations. It is also a reliable indicator of change after surgery (Kerr et al 2002). It is possible that other investigators may prefer the inclusion of other variables or to alter the score in the same way that the PRS has been modified. The Score awaits evaluation by other investigators to determine its general applicability.

CLASSIFYING SPASTIC HEMIPLEGIA USING GAIT ANALYSIS

This section refers to the following paper: Hullin MG, Robb JE, Loudon IR. Gait patterns in children with hemiplegic spastic cerebral palsy. Journal of Pediatric Orthopaedics Part B 1996; 5: 247-251.

Clinicians often rely on classifications for descriptive and prognostic purposes. The classification of cerebral palsy may be neurological, geographical, or based on the characteristics of clinical examination or observation of gait. In the previous section it was shown that gait analysis can be used to assess gait visually. This section illustrates how gait analysis has been applied to the classification of hemiplegic spastic cerebral palsy.

Winters, Gage and Hicks (1987) described gait patterns in spastic hemiplegia in children and young adults. Their observations were based on sagittal plane kinematics of 46 patients who had undergone gait analysis. The population studied consisted of 38 patients with cerebral palsy none of whom had undergone surgery, six who had a traumatic brain injury and two a juvenile cerebrovascular accident. They identified four patterns. Group 1 patients had the mildest deviations from normal, had a drop foot in swing but adequate dorsiflexion of the foot in stance. These patients also had increased flexion of the knee at terminal swing, at initial contact and in loading response. They also showed hyperflexion of the hip in swing as a result of weakness of the anterior tibial muscles and the ensuing dropped foot. Group 2 patients had a plantarflexion that persisted throughout the gait cycle, full extension, or hyperextension, of the knee in stance, and hyperflexion of the hip throughout the gait cycle. These were all due to a static or dynamic contracture of the gastrocnemius and

soleus muscles. Group 3 patients had plantarflexion of the ankle and more limited flexion of the knee in swing. This was felt to be due to more proximal neurological involvement than in Groups 1 and 2. The loss of knee flexion in swing was attributed to loss of co-ordinated contractions of the hamstrings and quadriceps muscles. Group 4 patients had decreased motion at the hip and knee and plantarflexion at the ankle. The difference between Groups 3 and 4 was attributed to the increased activity of the iliopsoas and adductors that prevented the hip from attaining full extension in terminal stance. Interestingly, of the five patients in Group 3 only two patients had cerebral palsy and the kinematic data for this group showed that not only knee hyperextension but also knee flexion were seen in this group. Although these patients may have been part of a clinical spectrum, these are two completely different biomechanical situations as the external ground reaction force would be extensor for those patients who had hyperextension and flexor for those who had knee flexion in stance.

Since gait patterns are likely to result from external forces acting on joints, it seemed desirable to use kinetics to reassess the Winters, Gage and Hicks' (1987) kinematic classification. A pitfall in considering kinematics alone is illustrated below. The illustration in Figure 15 shows a diagrammatic representation of a sagittal plane stick diagram of the lower limb. The angles at the 'knee' and 'ankle' are identical but the knee moments are opposite. On the left there is an external knee flexor moment, and on the right an external knee extensor moment illustrating two completely different biomechanical situations. If using kinematics alone it would be possible to assume that the two situations were identical whereas in this example they are completely different. Knowledge of kinetics aids our understanding of gait patterns. This was the basis of Hullin, Robb and Loudon's (1996) classification of hemiplegia in cerebral palsy.

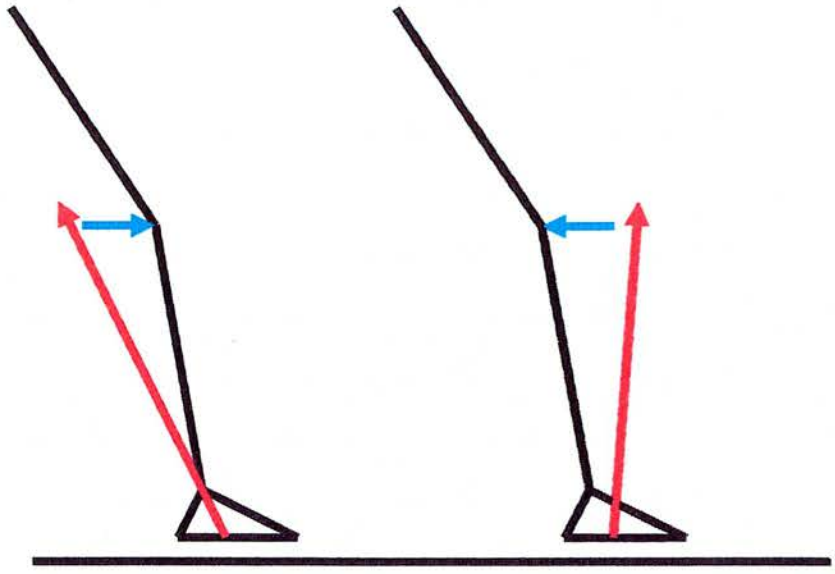

Figure 15 Sagittal stick diagrams showing the femur, tibia and foot. The GRF is in red. The external knee moment, in blue, is flexor on the left and extensor on the right. Although the angles at the knee are identical the biomechanical situation at the two knees is completely different.

Hullin, Robb and Loudon (1996) studied 26 non-operated hemiplegic children who underwent sagittal plane kinematic and kinetic gait analysis. In contrast to the Winters, Gage and Hicks' (1987) study, these patients all had cerebral palsy. Those with traumatic brain injury or juvenile cerebrovascular accidents were not included. Hullin, Robb and Loudon (1996) identified five gait patterns. The first pattern corresponded to Group 1 of the Winters, Gage and Hicks' (1987) classification. However this was the only similarity between the two studies. Hullin, Robb and Loudon (1996) defined loss of hip extension, knee hyperextension, and loss of ankle dorsiflexion in much greater detail. They also introduced the concept of tibial progression by relating this to the tibial angular velocity relative to the floor. The foot/ floor angle was used to define toe contact and heel contact. With the exception of Group 1 the characteristics in Hullin, Robb and Loudon's (1996) groups were as follows: Group 2 had persistent knee flexion and hip extension, Group 3 persistent hip

and knee flexion, Group 4 knee hyperextension and tibial arrest and Group 5 knee hyperextension and persistent ankle dorsiflexion.

Hullin, Robb and Loudon's (1996) study was the first to use kinetics for the classification of hemiplegia. This accounts for some of the differences between their study and that of Winters, Gage and Hicks (1987). Differences in the two populations studied may also account for the different findings. Hullin, Robb and Loudon (1996) described two separate circumstances for knee hyperextension – one associated with tibial arrest and the other associated with persistent ankle dorsiflexion. They concluded that the soleus was functionally tight in the former condition and the gastrocnemius was not functionally tight in the latter though they observed a major horizontal shear force suggesting that these latter patients were 'pulling' the upper body over the stationary foot. Although only two patients exhibited this pattern they felt that the identification of the two different mechanisms was clinically important. In Group 4, where soleus was functionally tight, an Achilles lengthening would have been appropriate, but in Group 5, where the calf muscles were not functionally tight, an Achilles tendon lengthening might risk producing crouch gait. The limitations of both studies were that only events in the sagittal plane were considered, and that the classifications did not distinguish between a primary deformity and a secondary compensation. True shortening of the calf muscles, for instance, might produce a drop foot in swing only (Group 1 Winters, Gage and Hicks) or an equinus attitude of the foot and ankle might be a compensation for a flexed knee and hip (Group 4). The originality, however, of Hullin, Robb and Loudon's (1996) study was to provide a biomechanical explanation for sagittal stance phase patterns in hemiplegic cerebral palsy.

Subsequently, Byrne, Jenkinson and O'Brien (1998) wished to improve the shortcomings of the Winters, Gage and Hicks (1987) study and investigated the sagittal plane kinematics of 55 hemiplegic and 91 diplegic patients. By introducing the concept of the 'plegic limb' they felt able to combine diplegic and hemiplegic patients and reported results on 237 limbs. Patients' kinematics were classified into eight groups. This study did not consider kinetics and the combination of diplegics and hemiplegics into a single group is open to question as their gait characteristics often differs. They also reported, as had Hullin, Robb and Loudon (1996), two groups of patients with knee hyperextension. Those with 'mild recurvatum' had a mean of 1° of hyperextension. As they did not state the measurement error of their optoelectronic system the accuracy of a 1° mean hyperextension may be questionable. Similarly their description of a 'severe recurvatum' group where knee motion ranged from 14° of hyperextension to 64° of flexion did not provide the mean value of knee hyperextension. Although their report is the largest number of patients studied of all series attempting to classify gait patterns using sagittal plane kinematics, methodological shortcomings question the utility of their classification.

Rodda and Graham (2001) used the classification system of Winters, Gage and Hicks (1987) for hemiplegia and that of Sutherland and Davids (1993) for diplegia to provide a management algorithm for cerebral palsy. They did subdivide the Group 2 of Winters, Gage and Hicks' (1987) study into 2a (equinus plus neutral knee and extended hip) and 2b (equinus plus recurvatum knee and extended hip) though they omitted to provide the rationale for this subdivision and kinetics were not studied. However the subdivision into Groups 2a and 2b may have a practical significance for patient management.

An analogous situation occurred in the classification of knee patterns in spastic diplegia. Huk et al (1987) described four gait patterns in the sagittal plane using cinematography though they acknowledged that the degree of deviation of joint angular displacement from normal could not be predicted by physical examination alone. This difference was probably due to muscle tone. One limitation of this study is that they do not appear to have considered the problem of abnormal limb rotation producing an inaccurate assessment of sagittal plane measurements (*vide supra* pp 46-47). It is also questionable whether a classification derived from sagittal plane cinematography, rather than kinematics, is entirely valid. Sutherland and Davids (1993) described patterns of knee problems in spastic diplegia using kinematics only. Seven years later Lin et al (2000) described the kinetic features of the knee in spastic diplegia and confirmed the patterns in three of out four of Sutherland and Davids' (1993) classification. They attributed the differences they found, as did Hullin, Robb and Loudon (1996) for hemiplegia, to the greater insights obtained with kinetics, rather than by kinematics alone.

None of the aforementioned classifications have been evaluated for inter-observer and intra-observer reliability, which is an area that merits further investigation. The classifications of Winters, Gage and Hicks (1987) and Hullin, Robb and Loudon (1996), have described sagittal plane events only. Future research could consider events in the coronal and transverse planes as a part of a more detailed classification. This might engender further complexities in classification though current technologies should enable the creation of functionally useful automated classifications from neural network techniques.

ROCKER FUNCTION WITH A BELOW-KNEE PLASTER CAST

This section refers to the following paper: Hullin MG, Robb JE. Biomechanical effect of rockers on walking in a plaster cast. Journal of Bone and Joint Surgery [Br] 1991; 73B: 92-95.

A rocker is a cam shaped device applied to a shoe or a plaster cast. It can be used to compensate for a stiff first metatarsophalangeal joint allowing the forefoot to rotate in the third rocker phase of gait. A rocker applied to a shoe can equally compensate for the loss of the second rocker in gait, for example when a patient wears an ankle-foot orthosis or a plaster cast. In this case, tibial progression over the stationary foot would be prevented and the shoe rocker would permit a smoother progression of the ground reaction force along the foot. Rockers or plaster boots are used routinely for patients who wear a below or above knee cast.

It is well recognised that walking in a plaster cast increases energy consumption. For example, Waters et al (1982) studied energy consumption in healthy subjects who wore three different types of cast – above knee, below knee and a cylinder. They found the oxygen uptake of healthy subjects increased by 60% when wearing a plaster cast using a non-weight bearing swing-through gait. Rockers applied to a plaster cast permit weight bearing and improve the efficiency of gait by equalising limb length.

Nuzzo (1983) used multiple exposure photography to illustrate walking and sprinting in a healthy subject wearing bilateral below knee casts to which were attached a variety of rockers or shoes. The ankle angle in the casts varied between 5° of equinus and dorsiflexion. This study compared four shoes or rockers – a 'rocker

bottom' shoe, a cushion heel shoe, a custom made cushion shoe and a standard rubber 'door stop' walker. Sagittal plane traces of photographs were permitted a qualitative description of events in stance. There were several limitations to this qualitative study in which the subject acted as their own control. The casts were changed to permit varying ankle angles, which were stated to be 5° of equinus or dorsiflexion but no confirmatory measurements were made. Nuzzo did, however, illustrate the practical use of wedges to position the knee in front of the ankle. He concluded that the rocker shoe performed 'badly', the cushion heel shoe performed better while the custom shoe performed the best in gait.

Pratt (1985) extended Waters et al's (1982) study to consider the energy of gait in subjects who wore below knee casts with different soles. He used a Kistler force plate and tested four different soles and a rocker in 20 volunteers who wore a unilateral below knee plaster cast. Energy consumption was calculated from external work done on the body and not from oxygen uptake. The vertical component of the ground reaction force was measured for normal gait, for the five experimental conditions as well as horizontal and vertical kinetic energy and vertical potential energy. Pratt concluded that the rocker sole was a poor choice for a below knee plaster and the manufactured soles produced a more natural gait and less disruption of energy patterns. Although this study was the first to consider kinetics of walking in a plaster cast, it did not describe either the relationship between the reaction forces and the knee, or the progression of the ground reaction force along the foot.

Peterson, Perry and Montgomery (1985) compared walking in rocker shoes with walking in athletic shoes during free and fast velocities in 15 healthy women. Footswitches, electrogoniometers, surface electromyograms, and a force plate were used for data collection. No significant differences were evident in velocity, cadence,

gait-cycle duration, single-limb support, or swing-stance ratios during free and fast walking. During free walking double support was decreased by 9% with rocker shoes as compared to athletic shoes. This was due to rapid loading of the limb that was twice as fast in the rocker shoes. Impact loading was four times greater in rocker shoes because there was no cushioning from the wooden heel. To compensate, knee flexion increased by 2°. A high degree of hip extensor activity was required to maintain the fast roll off at toe rocker. The rigid sole acted as a restraint to passive advancement of the tibia in pre-swing. These authors commented that the clinical use of rocker shoes should be limited to those with enough hip and knee extensor muscle control to buffer the impact of initial contact. Hullin and Robb (1991) found this to be particularly relevant when they found that stump rockers produce a large external flexor moment and an associated increase in quadriceps activity that is required to prevent a collapsing gait. The boot-type of rockers in Hullin and Robb's study (1991) produced an external knee extension moment and early arrest of tibial progression over the foot.

Strain gauges were used by Pratt et al (1986) to quantify cast stresses in normal volunteers wearing a below knee cast. Strain gauges were placed over the proximal third and middle thirds of the tibia, the ankle and the mid foot. Two different soles were added to the casts – a rocker or a shoe-shaped sole. Ground reaction forces were not measured in this study but they did find that in any given position the rocker sole produced lower contact stresses than the shoe type of rocker. Hellberg et al (1987) measured ground reaction forces from a treadmill in ten normal subjects wearing two casts made from plaster of Paris and a synthetic material. The subjects were also tested wearing three different rockers. The subjects were noted to perform less external work when wearing the synthetic cast and to prefer the rocker heel to a

door-stop type of heel. Hellberg et al (1987) noted that a rocker heel permitted a longer stride than a flat heel, but at the cost of increased work. This study focussed on work performed when wearing the two different plaster casts rather than on the relationship of the ground reaction forces to the lower extremity joints.

In 1991 Hullin and Robb studied the biomechanical effects of rockers on walking in a plaster cast. They extended Nuzzo's qualitative study (1983) and Pratt's kinetic study (1985) to consider rocker function. They developed two criteria not previously used when considering rocker design: tibial floor angular velocity and centre of pressure of progression. In contrast to previous studies they only used one subject who wore the same cast throughout the experiments with ten different commercially available rockers. They also introduced the concept of pelvic high point, or maximum pelvic displacement, being the point in the gait cycle corresponding to maximum potential energy and where there are no shear forces at the ground. Pelvic high point corresponds to the end of mid-stance as defined by Sutherland (1984) where there is the reversal of the fore and aft shear forces. Tibial floor angular velocity was an original concept used to describe the rate at which the tibia progressed over the stationary foot, a key event during the second rocker of stance as described by Perry (1974). Hullin and Robb (1991) further showed the relationship between the centre of pressure progression and changes in the direction of moments at the knee which, until then, had not been described in relation to rocker function. The clinical relevance of their study was to highlight the deleterious effect that many types of rockers had on gait when a subject wore a below knee cast. This would enable clinicians to make a more informed choice of rocker selection for such patients.

Several subsequent studies have considered differing materials for casts and their effects on gait, pressures on the limb within a cast, design criteria for rocker shoes and energy expenditure. For example, White et al (1999) compared the gait of eight volunteers fitted with a below knee plaster made either of rigid glass fibre and a cast shoe or from flexible glass fibre worn inside the subjects own shoes. Temporal and spatial characteristics were no different form normal gait but the subjects had a greater physiological cost index (Butler et al 1984) when walking in a cast. They also found that the first peak of the vertical component of the ground reaction force, Fz1, was significantly greater than for normal gait. However, this aspect was not discussed further.

van Schie et al (2000) tested nine different rocker shoe designs in 17 normal males using plantar pressure measurements. Rocker height and axis location were assessed. Peak plantar pressure was reduced in most forefoot locations by the rocker shoes, but increased at the mid foot and heel. The best axis location for reducing metatarsal head pressure was in the region of 55-60% of shoe length and for the toes at 65%. Four variables were considered to be important in rocker shoe design: the angle of the front part of the rocker to the ground, shoe height, rocker axis position with respect to the long axis of the shoe and the rocker axis angle with respect to the long axis of the shoe. Considerable variability was encountered between subjects in response to the same shoe. Gait training and plantar pressure measurement were deemed to be important adjuncts to shoe prescription. Hullin's work (1990) on experimental rocker design had anticipated this approach by ten years. Hullin had used experimental rockers attached to ankle foot orthoses and varied the pivot of the rocker by placing it at different percentages of foot length. The novelty of van Schie et al's study (2000) is that it considered rockers on shoes rather than with ankle foot

orthoses (Hullin 1990). Hullin and Robb's (1991) study does not appear to have been repeated or modified since its publication and their findings appear to remain relevant to the present day. These studies on footwear modifications for patients with plaster casts illustrate a practical application of gait analysis that has direct relevance to patient management.

GAIT AND ORTHOSIS FUNCTION IN LOW LEVEL MYELOMENINGOCELE

This section refers to the paper: Hullin MG, Robb JE, Loudon IR. Ankle-foot orthosis function in low-level myelomeningocele. Journal of Pediatric Orthopaedics 1992; 12:518-21.

Spina bifida results from a disturbance of the formation of the vertebral arches in the spine and may be associated with deformities of the neural structures within the spinal canal. Myelomeningocele (MMC) is one such neural defect that usually results in a motor and sensory deficit distal to the neural lesion. Sharrard (1964) described patterns of paralysis and their relationship to functioning muscles in 281 patients. Muscle strength is traditionally defined by the Medical Research Council Scale (MRC) (1943) in which 'O' indicates no contraction and '5' normal power. In MMC there is generally a good correlation between the last functioning neurological level and the clinical picture at the level of the foot and knee.

Patients who have low level MMC also have excessive ankle dorsiflexion in stance due to paralysis or weakness of the calf muscles. Typically they also have MRC grade 5 of the quadriceps and grade 4 of the hamstrings. Before 1992 it was widely recognised that these children benefited from the provision of ankle-foot orthoses (AFO) but this was based on clinical, rather than any biomechanical observations. Fulford and Cairns (1978) had suggested that rigid-soles controlled the centre of gravity. Carroll (1974) noted the clinical benefits of thermoplastic AFOs for children with low level MMC but did not consider any biomechanical effects. Linseth and Glancy (1974) reported on the effects of a floor reaction polypropylene rear entry AFO that was adapted from a solid ankle cushion heel orthosis. They studied qualitatively the gait of 47 children who wore these AFOs and who had paralysis from

the third to the fifth lumbar level. Linseth and Glancy (1974) attributed clinical improvements to increased stability, the lightweight of the orthosis and the point of application of the floor reaction force over a large pre-tibial area. They did not, however, investigate the biomechanical explanation of their clinical observations. Thomas et al (1989) reported on the results of six children who underwent kinetic, kinematic and EMG analysis when walking barefoot and in AFOs. Their sample size was similar to that of Hullin, Robb and Loudon's study (1992) and they found a decrease in excess hip, knee, and ankle flexion when children wore the AFOs. A decrease in excess muscle activation time and co-contraction was also noted in their EMG studies during walking when wearing AFOs. It was thought that this would facilitate a decrease in energy expenditure. Although the children were evaluated with a force plate, Thomas et al did not discuss this aspect of the investigation

Hullin, Robb and Loudon (1992) studied the effects of AFOs in low-level MMC using instrumented gait analysis. This work provided the first biomechanical explanation of how kinetics in the sagittal plane are altered in low level MMC by the use of AFOs. It also showed how modifying the AFO can alter ground reaction forces acting on the knee. The practical benefit of this approach to a clinical problem was to provide a biomechanical basis for orthotic modifications that can improve gait in MMC and unload the quadriceps in crouch gait.

One of the shortcomings of Hullin, Robb and Loudon's study (1992) was that only biomechanical events occurring in the sagittal plane were considered and the number of patients studied was small. Subsequent gait analysis studies have extended their concept by investigating events in three dimensions (Thomson et al 1999; Vankoski, Michaud and Dias 2000; Duffy, Graham and Cosgrove 2000).

Thomson et al (1999) studied retrospectively 28 children and adolescents with MMC. Their object was to identify any abnormal patterns that might lead to knee instability in the future. They confirmed earlier observations that there were significant improvements in sagittal plane function, with reductions in excessive ankle dorsiflexion, increases in peak plantarflexor moment and reduction in crouch and knee internal extensor moment (external flexor moment) in the L4 and L5 groups. Thomson et al also noted greater than normal transverse plane knee motion in stance during barefoot walking that also increased significantly when wearing an AFO. Vankoski, Michaud and Dias (2000) examined the effect of tibial torsion on the effectiveness of solid AFOs in 40 patients. Their study confirmed the clinical impression that MMC patients without excessive tibial torsion gained better function when wearing their AFOs than patients with external tibial torsion of more than 20°. Duffy, Graham and Cosgrove (2000) studied the effect of AFOs on gait and energy expenditure in 12 patients. This study also confirmed Hullin, Robb and Loudon's (1992) finding that walking speed and step length improved when wearing AFOs. An interesting result of Duffy, Graham and Cosgrove's study was that the AFOs did not improve the pattern of moments around the knee. They appreciated that limitation of dorsiflexion was not achieved when the AFOs were worn, as these were too flexible. This AFO flexibility would have produced a knee kinematic similar to that of walking barefoot and would also explain why an improvement in knee kinetics was not evident. The patients in Hullin, Robb and Loudon's study (1992), wore rigid AFOs and these authors also showed that a leaf spring AFO, which is less rigid, was of little help for gait in MMC as it allowed excessive dorsiflexion and knee flexion. It is likely that had Duffy, Graham and Cosgrove's patients worn stiffer AFOs, the patients' kinematics and kinetics would have been improved. Bérard et al (1990) found that a

rigid carbon fibre floor reaction AFO which they used in 18 children with MMC proved effective in maintaining an upright posture in gait. However, their study was based on clinical observations and not gait analysis.

Hullin, Robb and Loudon (1992) demonstrated the value of kinetics from instrumented gait analysis and showed that it could be used to improve orthotic prescription for patients with low level MMC. This is another example of the clinical application of gait analysis for patient management.

GAIT ANALYSIS AND TREATMENT PLANNING IN CEREBRAL PALSY

This section refers to the paper: Cook RE, Schneider I, Hazlewood ME, Hillman SJ, Robb JE. Gait analysis alters surgical decision making in cerebral palsy. Journal of Pediatric Orthopaedics 2003; 23: 292-295.

Gait analysis has been shown to be an excellent research tool and for maintaining a record on the clinical progress of individual patients (Robb and Brunner, 2001). The argument for using gait analysis in clinical practice includes the difficulty in assessing complex gait abnormalities visually (Saleh and Murdoch 1985) and the inability to see muscle force. Firstly, it is impossible to form an accurate visual assessment of events occurring at five different anatomical levels - the trunk, pelvis, hip, knee, and ankle/foot - and in three different planes - sagittal, coronal and transverse. The distinction of events occurring at different anatomical levels and planes is conventional but inaccurate as there are interactions across planes and anatomical levels in gait. Secondly, observers cannot see forces in muscles and how they act on joints. Although force, muscle activation, the interaction between joints and skeletal abnormalities cannot be easily visualised, they can be measured using gait analysis. Another advantage of gait analysis is its use to improve the understanding of muscle action from modelling studies. For example, Hoffinger, Rab and Abou-Ghaida (1993) have shown that the hamstrings are rarely short in crouch gait, while Arnold et al (2000) used modelling to show that the medial hamstrings and adductors are not internal rotators in cerebral palsy.

Orthopaedic surgeons have derived important information from gait analysis. The rectus transfer procedure in cerebral palsy, suggested by Perry (1987), was

derived from gait data and this procedure is now standard practice for surgery in cerebral palsy. Lever arm dysfunction, popularised by Gage (1991), has given greater insights into the problems associated with bony torsions. As surgeons have gained more insight into the underlying gait pathologies from gait analysis there has been a trend towards multiple level surgery rather than sequential operations. Tenotomies have generally been replaced by intramuscular tendon lengthening procedures.

As a language, gait analysis has allowed the description of complex three-dimensional movements and a method for teaching the complexities of human movement. It also has provided a communication pathway for individuals from differing professional backgrounds. Gait analysis has provided tools for classification for hemiplegia (Winters, Gage and Hicks 1987; Hullin, Robb and Loudon 1996) and diplegia (Sutherland and Davids 1993).

However, gait analysis has its limitations. One justifiable criticism is that the laboratory is not a natural environment. Gait analysis only describes a patient's gait in a laboratory and is not necessarily relevant to a person's 'normal' activities.

One outstanding question is whether or not the results of treatment of the individual patient are better after gait analysis, which has been used predominantly for complex gait disorders such as those in cerebral palsy and myelomeningocele. There is a wide body of literature that has enhanced the understanding of the underlying mechanisms of these complex gait disorders. It is now becoming increasingly necessary for cerebral palsy patients to have pre-operative gait analysis where it is available. Deluca et al (1997) studied 91 CP patients to compare the surgical recommendations of clinicians experienced in gait analysis who used information from clinical examination and videotape, with those made after the addition of kinematic, kinetic, and EMG data. The video and clinical examination data for each

patient were reviewed by experienced clinicians who then made surgical recommendations. Joint kinematics, kinetics and EMG data were then reviewed, and a second set of surgical recommendations was made. Comparisons between the recommendations showed that the addition of gait analysis data resulted in changes in surgical recommendations in 52% of the patients with an associated reduction in the costs of surgery. These changes were associated with an increase in surgical recommendations for the gastrocnemius (59%) and rectus femoris (65%), and a decrease for the hamstrings (61%), psoas (78%), hip adductors (83%), femur (86%), and tibia (64%). They also commented that the changes in decisions would have avoided inappropriate surgical decisions, which were more likely to be made without gait analysis. Their study involved numerous assessors that may have been a source of inconsistency.

Kay et al (2000) also considered the impact of pre-operative gait analysis on the orthopaedic care of 97 patients (101 gait analyses). Eleven different physicians referred patients for gait analysis though the pre-gait analysis treatment plan was only available for 70 patients. Alterations were made to the treatment plan in 62 (89%) of these 70 following gait analysis. In 10 of the 70 patients with specific treatment plans before the gait study, the referring physician also served as the physician in the gait laboratory; ultimate treatment was changed in nine of these 10 patients. Of the 273 surgical procedures recommended before the gait study for the 70 patients, 106 (39%) of these were not carried out following consideration of gait data. An average of 1.5 procedures per patient planned before the gait study was not deemed necessary after the addition of the gait data. However, an additional 110 procedures (1.6 per patient) were performed after the addition of gait laboratory data. This study showed that the proposed surgical interventions were frequently altered after the addition of gait

laboratory data. A second study from Kay et al (2000) considered the impact of postoperative gait analysis on the care of 38 consecutive patients. Of the 38 postoperative gait analyses, 32 (84%) resulted in recommendations for a change in patient care. Surgery was recommended in 16 of 38 cases (42%), bracing in 20 (53%), and specific physical therapy regimes in eight (21%). Eleven of the 38 patients (29%) had changes recommended in at least two of the three areas (surgery, bracing, and therapy). This study suggested that postoperative gait analysis is not only useful as a measure of treatment outcome, but also as tool in planning continuing care. Both studies clearly demonstrate that gait analysis affects pre- and post-operative decision-making. However, neither study clarifies whether it alters functional outcome. Skaggs et al (2000) assessed the reliability of interpretation of gait analysis data between physicians and institutions. Twelve experienced physicians in gait analysis from six institutions reviewed gait analysis data from seven patients. They identified problems and made treatment recommendations based on the data provided. Agreement for the most commonly diagnosed problems was slight to moderate and they agreed on the identification of soft tissue more than bony problems. The variability of their surgical recommendations for soft-tissue procedures was similar to that for the diagnosis of both soft-tissue and bone problems. There were inter-institutional differences in the frequency of identification of soft-tissue and bony problems, as well in the frequency of recommendations for soft-tissue surgery and osteotomies. One centre recommended bony surgery twice as frequently as all other centres. The authors concluded that although gait analysis data are objective, there is subjectivity in their interpretation.

Noonan et al (2003) studied results of gait analysis and treatment recommendations in 11 patients with cerebral palsy at four different American

centres. Each centre produced kinematic, kinetic and EMG data for these patients. Data from the centres were compared using a discordance method, each leg being considered as an independent data set. The authors justified this approach as they felt that their study was primarily focussed on the accuracy of different measures of gait analysis and not on how they were generated. Almost complete concordance was noted in the comparison of right and left variability measures. Treatment recommendations were chosen from a list of options. Noonan et al (2003) found the static range of motion from physical examination ranged from 25-50°. The range of differences for sagittal motion averaged 28° for the hip, 25° for the knee and 27° for the ankle. In at least 5% of the gait cycle, on average, the difference between the four laboratories might exceed 8-34°. Treatment recommendations also varied greatly. There was agreement for treatment recommendations between three out of four of the centres for two of the three patients with mild CP. However, moderately affected patients had a poorer level of concordance (60%) for treatment and for the severely affected patients the level of concordance was only 33%. Although previous studies (Deluca et al, 1997 and Cook et al 2003) have shown that gait analysis alters decision-making for patient care, Noonan et al's (2003) study is the first to highlight major differences in kinematic values between laboratories. Several possible causes for the variations were considered: patients walking differently on different days, fatigue, or increasing familiarity with the gait laboratory environment. They concluded that the most likely cause for the discrepancies was marker placement. The findings of Noonan et al (2001) and Skaggs et al (2000) are disconcerting as they may suggest that gait analysis is too inaccurate to be of clinical value. Clearly there would be major concerns if centres exchanged data without knowing either the details of the set- up in respective laboratories or the accuracies and reliability of data collection in

individual laboratories. Noonan et al (2003) recommended that laboratories should maintain their own database of normal values implying that laboratories should not rely on commercially available packages, in keeping with good practice guidelines. They were less surprised at the variations in treatment recommendations, which would have reflected the expertise, previous training or clinical judgement available in the four centres. It is important to separate these two issues highlighted in Noonan et al's paper. The variations in the accuracy of data collection need improvement, but the variations in patient management are less surprising for the reasons outlined above.

Two editorials accompanied Noonan et al's (2003) paper. Gage's editorial (2003) accepted that variations demonstrated in Noonan et al's (2003) paper were too large to be acceptable. He also pointed out that the availability of a gait analysis laboratory does not necessarily mean that it will provide knowledge about the treatment of cerebral palsy. He presented evidence to show that at Gillette Children's Hospital inter-therapist, inter-session and inter-trial errors were less than 4% and could be improved further in the future. This statement is probably over optimistic. Accuracy to this level may be found nearer to hand; for example at the Anderson Gait Analysis Laboratory quality assurance tests for the effect of marker placement on hip rotation in gait have coefficients of reliability of less than 6.3° (van der Linden et al 2003). The second editorial (Wright 2003) emphasised the need for clinicians to determine the sources of variability of gait data and to improve its reliability. He questioned the routine use of gait analysis in the management of children with cerebral palsy. Gait analysis undoubtedly changes decision making in cerebral palsy, but at present it is not known if it improves functional outcome.

One of the weaknesses of all these previous studies was that they involved multiple observers which could affect the outcomes. Cook et al (2003) from the Anderson Gait Analysis Laboratory assessed the impact of gait analysis on the management of 102 patients with cerebral palsy. They attempted to recreate a typical clinical scenario in which an orthopaedic surgeon decides on a treatment plan and then requests gait studies. Treatment plans before and after gait analysis were then compared. Unlike the previous studies Cook et al maintained consistency using only one orthopaedic surgeon and the same gait laboratory team to review all the data. They were able to show that the clinical assessment was in close agreement with gait analysis when assessing the need for surgery. However, less agreement existed in identifying the type or level at which surgery might be performed. There was good agreement for bone surgery, suggesting that clinical evaluation for torsions was fairly reliable. There was less reliability for soft tissue procedures probably reflecting the difficulty in assessing tone related problems. Overall, gait analysis altered decisions in 40% of operations. By using the same set of observers both before and after interventions, Cook et al (2003) avoided the limitations of studies in which multiple observer and/or different institutions were use for the gait analyses. While this study did not show that treatment outcomes are necessarily changed by gait analysis, randomized controlled trials would be necessary to show such clinical benefits. Randomisation to treatment with or without gait analysis may not be possible in parts of the world where gait analysis is available. It is likely that the best data might be obtained from comparative, but uncontrolled, clinical outcomes that either do or do not involve gait analysis in treatment planning and postoperative assessment.

GAIT AND HAMSTRING LENGTHENING IN CEREBRAL PALSY

This section refers to the following paper: van der Linden ML, Hazlewood ME, Hillman SJ, Kerr AM, Robb JE. The effects of surgical lengthening of the hamstrings without a concomitant distal rectus femoris transfer. Journal of Pediatric Orthopaedics 2003; 23: 308-313.

Hamstring lengthening alone in the management of cerebral palsy has been criticised for producing knee hyperextension. Egger's procedure (1952) transplanting hamstrings from their insertion on the tibia into the distal femoral condyle, to act as hip extensors, is now obsolete as it tended to result in a knee that had limited flexion and then developed hyperextension. This procedure was subsequently replaced by a tenotomy or transfer of the semitendinosus and fractional lengthening of the semimembranosus (Bleck 1987), and finally by a fractional lengthening of the hamstrings (Green and McDermott 1942). The gracilis is often included in this lengthening and the biceps tendon less often.

Baumann, Ruetsch and Schurmann (1978) used gait analysis to evaluate distal hamstring lengthening and this was one of the first kinematic studies of a postoperative evaluation of surgery in cerebral palsy. The recognition of co-exiting rectus spasticity from gait analysis prompted Perry (1987) to suggest that a distal rectus transfer to the sartorius might help knee flexion in the initial swing phase. This procedure was carried out by Gage et al (1987) and has now become a standard part of the surgical repertoire for cerebral palsy although semitendinosus is usually used now instead of sartorius. The operation suggested by Perry (1987) had its origins in an earlier paper written by the same group (Waters et al 1979) who found that a distal

rectus and vastus intermedius release improved the lack of knee flexion in the initial swing phase of gait in adult hemiplegia.

There have been an increasing number of descriptions of gait following hamstring and rectus surgery since gait analysis has become more readily available. The procedure is usually undertaken as part of multilevel surgery. This causes some difficulty in interpreting data due to the difficulty in isolating the effects of the hamstring procedure from others. Most authors have reported improved extension of the knee during stance following surgery (De Luca et al 1998, Rethlefsen 1999, Thometz, Simon and Rosenthal 1989). Thometz et al studied hamstring lengthening in patients who had additional surgical procedures. Improvement of crouch gait was the main indication for surgery and follow up was by gait analysis. Improvement of knee extension in stance proved to be the main benefit. Velocity, stride length and cadence were evaluated as percentages of age-related normal values and did not increase. The total arc of motion at the knee also remained unchanged postoperatively though patients did gain knee extension in stance but a loss of knee flexion in swing. Damron, Breed and Cook (1993) reported a decrease in passive prone knee flexion from 131° to 117° following combined hamstring tenotomy without rectus femoris transfer in 52 patients. They concluded that this decrease was often not severe enough to limit function. Twenty-three percent of the knees had improved flexion, whereas 6% remained unchanged. Thirteen percent of ambulators eventually required a rectus femoris transfer to correct a "stiff-legged gait". Rethlefsen et al. (1999) evaluated retrospectively the outcome of hamstring lengthening and distal rectus femoris transfer. Postoperatively, both the timing of peak knee flexion during swing and the total arc of knee motion improved significantly. Hamstring range of motion and knee extension at terminal swing also improved significantly, but stride length and gait

velocity did not. Granata, Abel and Damiano (2000) studied 40 children, the majority of whom underwent a combination of multilevel aponeurotomies and tenotomies for restricted range of joint motion. Six patients had isolated distal hamstring lengthening. Improvements were noted in the sagittal joint angles but surgery appeared to have little effect on angular velocities of joints, while hamstrings-quadriceps co-activity around the knee had decreased.

Hoffinger, Rab and Abou-Ghaida (1993) highlighted concerns about increased anterior pelvic tilt following medial hamstring lengthening. They studied the three-dimensional motion of the hip and knee, calculated hamstring muscle length, and evaluated dynamic EMG of the medial hamstrings. Their findings showed that the hamstrings are not necessarily anatomically short in ambulant patients with cerebral palsy. They also considered that the hamstrings may be important hip extensors in some patients with crouch gait and that other causes contributing to crouch such as a hip flexion contracture should be considered before isolated hamstring lengthening is performed.

The pre-operative criteria for a rectus transfer accompanying a hamstring lengthening normally includes increased rectus activity in swing, reduced knee flexion in swing, and a delay in peak knee flexion in swing (Gage et al 1987). Not all patients fulfil these criteria and van der Linden et al (2003) from the Anderson Gait Analysis Laboratory evaluated patients who had undergone hamstring lengthening without a rectus transfer. The aim of this study was twofold: to investigate the effects of hamstring lengthening without rectus transfer and to consider speed-related effects and changes in dimensionless stride parameters following surgery. The latter has not been reported previously and is an important consideration in overcoming the problem of associating changes in gait with changes in height as children grow. The

use of dimensionless quantities enables a more accurate comparison of pre- and post-operative change to be made. The application of dimensionless units extends Stansfield et al's (2001) use of the technique from normal to pathological gait. Post-operatively van der Linden et al (2003) observed a significant increase in knee extension and a decrease in the amount of knee flexion in swing post. The decrease in knee flexion signified a change to more normal speed-related values. Absolute cadence was significantly lower following surgery but the changes in dimensionless cadence were not significant. This difference was attributed to an increase in the body height of patients following surgery illustrating the importance of using dimensionless quantities to obtain a clearer understanding of post-operative change. The clinical significance of van der Linden et al's study is that post-operative effects of surgery in children may be partly be explained by changes in height and not due to the surgery alone.

Delay in the timing of knee flexion in swing is one of the criteria for rectus transfer surgery and this has also been used as an outcome measure to assess its effectiveness. Another study of gait patterns of children walking at clinically relevant speeds from the Anderson Gait Analysis Laboratory (van der Linden et al 2002) showed that the timing of the peak of knee flexion in swing remained constant when calculated as a percentage of the whole gait cycle over the range of speeds studied. However, when calculated as a percentage of the swing phase, peak flexion occurred earlier for slower speeds. This is another indication of the importance of considering walking speed in the interpretation of the post-operative effects of hamstring surgery.

It is recommended that future studies describing post-operative changes of gait in children should use dimensionless quantities and that the insights gained into speed

related effects might also help to clarify the post-operative results surgery for cerebral palsy.

CONCLUSION

This compilation of eight papers has shown that gait analysis has widespread potential for research and clinical applications in the future. The advent of more modern equipment and powerful computation has extended its research potential in ways not possible previously when data collection and processing were so time consuming. Gait analysis is now far less time-consuming than previously and is becoming more applicable to patient care.

The question of the timing of gait maturation in children has probably been answered as far as possible by the results of the two prospective studies from the Anderson Gait Analysis Laboratory (Stansfield et al 2001). It is unlikely that this work would be repeated in the future because of the length of time it took to complete the studies and the substantial volume of data that was generated. Importantly, these two studies illustrate the need to normalise gait data to non-dimensional units in order to overcome the difficulty of changes in gait in children being ascribed to an intervention rather than due to the effects of growth. Many previous studies have used 'age-matched' controls but this is not appropriate in studies of gait where children may be of the same age but of differing height and weight. Moreover, since gait maturation occurs by age of eight years, there is little need to provide age-matched controls for older children and speed-matched controls should be used instead. The concept of using speed-matched controls and dimensionless quantities was extended to pathological gait in the study on hamstring lengthening and has highlighted the need to ensure that comparisons between subjects are appropriate. Some indications for hamstring and rectus transfer surgery and the improved results after surgery may have been based on walking at slower speeds than usual. The hamstring study also

emphasised the need for future studies on surgical outcomes to use non-dimensional units to ensure appropriate comparisons between patients. It is likely that this will happen in the future.

Three-dimensional instrumented gait analysis is not, as yet, readily available throughout the world and the study concerning improving the visual observations of gait is relevant for this very reason. It should provide an advance in the visual observation of gait as it describes a simple method using readily available and inexpensive equipment that enhances the appreciation of one type of pathological gait. It is interesting to note that the derivation of this low technology approach was made possible from knowledge of gait acquired from systems that are very complex.

The classification of hemiplegia based on kinetics is an example of the way in which gait analysis may enhance the understanding of pathological mechanisms in complex gait abnormalities. This classification might profitably be expanded in the future studies to include events in the coronal and transverse planes, and to distinguish between primary gait problems and gait compensations. The advent of neural networks may offer an appropriate tool for recognition and classification of patterns in future studies. However, any useful classification in clinical practice should be reproducible and of prognostic value.

A justifiable criticism of gait analysis is that when applied as a pre-operative investigation for children with cerebral palsy there is no evidence to prove that it alters functional outcomes. This is a challenge for future research. However, there is no doubt that gait analysis has given a better understanding of the pathology of gait and its use as an assessment and audit tool is clear.

One of the continuing challenges for gait analysis concerns the accuracy of marker placement that seems to be quite variable between laboratories. Perhaps the

greatest source of uncertainty is the placement of markers to determine transverse plane movement, particularly of the hip. This is compounded by the difficulty of accurate marker placement in obese subjects. In addition, current commercially available foot models are crude and there is a need for improved foot models. Research into these two problem areas is likely to continue.

It is likely that future innovations will include a greater use of musculo-skeletal modelling and will become part of the routine assessment of patients. It should be possible to use this type of modelling to produce plots of muscle length and to compare normal with pathological data and EMG activity. Preliminary data from muscle modelling suggests that mono-articular muscles behave in a way similar to the joint kinematic curves already seen in routine gait analysis but that bi-articular muscle patterns are different (Baker 2003; personal communication). The problem in modelling bi-articular muscles is that it is an oversimplification to consider their action in a straight line, which is not anatomically correct. Information about skeletal anatomy in pathological situations needs to be factored into any model considering abnormal gait, thus giving a patient specific model. Muscle modelling is a potentially exciting future development and has already been used to confirm that hamstrings are rarely short in crouch gait (Delp, Arnold, Spiers and Moore, 1996). Moment arm modelling is a potential extension to muscle length modelling, and it could render the present M=Fxd method obsolete (*vide supra* pp 16-17). The problem is to consider a three-dimensional moment arm model that would take into account the orientation of muscles as well as their distance from any given joint. Recent studies using moment arm modelling have shown that the hamstrings and adductors rarely act as internal rotators in cerebral palsy (Arnold, Schmidt and Delp 2000) and that internal rotation gait may be a compensatory mechanism to restore abduction capacity decreased by

bony deformity (Arnold, Kommattu and Delp 1999). Centre of mass modelling is another concept likely to give greater insights into gait in myelomeningocele and trunk kinetics. In the future, induced acceleration modelling (Kepple, Siegel and Stanhope 1997; Anderson and Pandy 2003) and forward simulation techniques (Jonkers, Stewart and Spaepen 2003) may offer further insights into individual muscle action in gait. These models are likely to revolutionise our understanding of pathological gait but will require even greater assumptions than the basic models in current use. For this reason, and inherent complexities, it is unlikely that these models will be introduced for routine clinical gait analysis in the near future (Baker 2003; personal communication).

Gait analysis has greatly enhanced our knowledge of normal and pathological gait but perhaps its most important contribution has been to provide a common language to describe human locomotion, which in turn, has allowed a means of communication between professionals from differing disciplines.

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