A STUDY OF BALANCE AND GAIT FOLLOWING STROKE: IMPLICATIONS FOR REHABILITATION

THESIS

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ABSTRACT

This study examined: the features and extent of the gait velocity decrement following stroke; the nature of the balance disorder and; investigated relationships between gait and balance. Twenty, fully-ambulant, hemiplegic subjects (12 men and 8 women), with a mean age of 57.2 year (± 10.7), were compared with 20 age- and sex-matched controls. Spatiotemporal gait kinematics were measured at five walking speeds ranging from "very slow" to "very fast". Balance was then measured while subjects maintained a variety of weight shift postures both with feet parallel and then in diagonal positions similar to those assumed during the double support phases of gait. Location and variability of centre of pressure (CP) were measured and the ranges over which CP could be shifted were calculated. Stroke subjects walked very slowly with their "fastest" walking speed (0.5 stat/s ± 0.23) no different from the "slowest" speed (0.38 stat/s ± 0.11) of the controls ($p > 0.05$). Differences between the other parameters were also found, many of which could be attributed to the differences in walking speeds. However, where the subjects walked at similar velocities, the hemiplegic sample walked with quicker and shorter strides suggesting a "cautious" gait pattern. The single support asymmetry of the strokes decreased with increasing walking speed. The positions of CP in the hemiplegic sample were found to be significantly displaced towards the unaffected leg and deficiencies were found in posterior shifts ($p < 0.05$). Postural sway was significantly greater in the hemiplegic sample implying less stable balance and the ranges over which the hemiplegic sample shifted weight were significantly less than the controls. The diagonal weight shift tests revealed the difficulty the stroke subjects experienced in shifting CP posterolaterally over the affected leg. Significant correlations were detected confirming the presence of relationships between static balance performance and gait. However, these correlation findings left considerable percentages of variance unexplained. These findings suggest that future rehabilitation should address the poverty of range of walking speed possessed by hemiplegic subjects as well as the reduced ability to weight shift over the hemiplegic limb, particularly posterolaterally. Further study to test these proposals is indicated.

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CHAPTER 1

THE PROBLEM

INTRODUCTION

Stroke is a focal neurological disorder of acute onset which follows disruption of the blood supply to the brain and causes damage to the central nervous system often resulting in permanent functional disability (Pryse-Phillips and Murray, 1982). In North America, stroke remains the third most common cause of death after ischemic heart disease and cancer (Hartunian *et al.*, 1980; American Heart Association, 1992). It has been proposed that stroke is the leading cause of disability particularly amongst older people (Nichols, 1979). The incidence of stroke has declined over recent years as a result of more effective management of risk factors such as hypertension (Broderick *et al.*, 1989). However, this decline in incidence has been accompanied by an increased prevalence (Isaacs, 1984). This has occurred because improved medical management has resulted in the survival of more people following the initial insult and longer life expectancy post stroke. Between the years of 1980 and 1990, deaths from stroke decreased by 32.4% (American Heart Association, 1992). However, this advance has had the effect of increasing the numbers of people in the population with significant post stroke disabilities and placing an increased burden on rehabilitation services. An additional factor which contributes to this profile is that there are increasing numbers of older people in the general population. It is known that stroke incidence rises with age, and that therefore, greater numbers of people are at risk of stroke as the population ages (Evans and Caird, 1983). As a result, stroke is, and will continue to be, a significant health care problem.

Once stroke has occurred, its medical treatment has remained limited and rehabilitation of the stroke survivor continues to be the primary therapeutic intervention following the immediate medical emergency (Brocklehurst *et al.*, 1978). Rehabilitation is an established health care process whereby the individual, following the acute stroke, is treated in an attempt to reestablish functional independence (Kottke *et al.*, 1982). A major component of the rehabilitation of the person who has suffered stroke is the re-education of functional gait. This high priority is due to the major importance of gait as a functional competency in humans (Inman *et al.*, 1981) and as an important factor contributing to independent living (Mahoney and Barthel, 1965). Gait disorders resulting from stroke are thought to be due to the loss of normal balance mechanisms in the leg which has been affected by the stroke, muscle weakness, joint instability, alterations of normal muscle function and sensory loss (Musa, 1986). These symptoms are, therefore, the focus of remedial efforts in the rehabilitation setting although the relative contribution of each of these factors in the hemiplegic gait pattern has yet to be clearly established. However, a consistent finding in gait studies following stroke has been that gait velocity following stroke is much slower than normal (Knuttson and Richards, 1979; Wall and Turnbull, 1986).

There have been several studies which have clearly demonstrated the asymmetrical nature of hemiplegic gait (Wall and Ashburn, 1979; Mizrahi *et al.*,1982; Brandstater *et al.*, 1983; Wall and Turnbull, 1986). This asymmetry suggests disordered function in both the support and swing phases of the affected limb (Wall and Turnbull, 1987). In studies which have measured forces beneath the feet, hemiplegics have been shown to bear a greater percentage of body weight on the sound limb when compared to the affected side (Caldwell *et al.*, 1986; Gruendel, 1992). This unequal weight bearing has also been demonstrated in the gait of hemiplegic cerebral palsied children (Seeger *et al.*, 1981). The remedial exercises designed to improve gait asymmetries have the major objective of developing a more efficient gait pattern by increasing the ability of the patient to effectively transfer body weight through the affected leg thus increasing the support phase time of that limb based on the assumption that rectification of this problem will improve the symmetry of the gait pattern particularly in single support (Lane, 1978). This appears to be a sound assumption since it has been shown that an improved symmetry of load bearing is associated with an improvement in the temporal aspects of the gait pattern (Seeger *et al.*, 1981). Concurrently, attempts are made to improve the ability of the patient to move the joints of the lower limb in a selective fashion (as opposed to movement which is dominated by primitive, pathological, reflex motor patterns) thus decreasing asymmetries which are due to swing phase discrepancies. These exercises are based on the physiotherapy model originally proposed by Bobath (1970). As a means to reach this objective, physiotherapy techniques, designed to re-educate gait, concentrate heavily on attempting to retrain the automatic muscle activity which maintains balance with emphasis on the disabled lower limb of the patient. This principle is based upon the premise that normal gait requires the presence of adequate mechanisms of balance in both lower limbs (Bobath, 1970).

In a study which examined the effect of exercise in residual stroke patients, Wall and Turnbull (1987) confirmed the findings of Knutsson and Richards (1979), that even in a relatively homogeneous stroke sample, the "hemiplegic gait" pattern was a complex entity, leading to the conclusion that stroke may affect gait in a number of different ways. However, studies of hemiplegic gait have tended to examine only the preferred walking speed of subjects. While useful, this technique does not address the capability of these subjects to alter walking speed, an important functional characteristic of normal gait. Little is known about the adaptability of hemiplegic gait and how altered walking speeds may affect clinically important variables such as gait asymmetry. In addition, it is unclear how this limited walking speed influences the temporal and spatial gait kinematics and the characteristic asymmetry of the hemiplegic gait pattern.

Normal gait has been described as "a series of catastrophes narrowly averted" (Steindler, 1955) and it is logical to assume that normal gait is dependent on a fully functional balance system. A number of disorders which affect the balance system are associated with gait deviations. In cerebellar disorders, motor ataxia causes balance inadequacies and such patients demonstrate a characteristic "staggering" gait pattern (Gilman, 1989). Similarly, elderly people, who fall or who have a fear of falling, walk in a "cautious" manner a pattern characterised by decreased single support durations and an increased time spent in double support. (Hogan *et al.*, 1987). Wall and his co-workers also identified a relationship between abnormal balance, for which the patient sought medical advice, and the pattern of gait adopted by these patients (Wall *et al.*, 1991). Thus, gait and balance appear to be inextricably linked. However, the relationship between balance performance and gait following stroke remains to be fully investigated despite the research conducted by Murray *et al.*, 1975 and Dettman *et al.*, 1987. These researchers showed a relationship between standing balance and gait but examined balance with the feet in the same plane in a typical relaxed standing position; a posture which does not resemble any part of the gait pattern. It has, therefore, been proposed that deficient balance on the hemiplegic limb contributes to asymmetries in the single support phase of the gait pattern (Wall and Turnbull, 1987). Treatment, therefore, has addressed this balance discrepancy with a view to improving the gait asymmetry (Lane, 1978). In particular, these remedial techniques have largely consisted of retraining the ability of stroke patients to shift and maintain their centre of mass over the affected limbs, motor scenarios which require the hemiplegic subjects to load their affected lower limbs. However, Wall and Turnbull (1987) found that this treatment direction was not effective in improving single support asymmetry in gait. These authors proposed, along with other possible reasons for this result, that there was some doubt as to whether the subjects in the study were actually transferring weight through the affected leg in the manner required. There is a need therefore, to examine this hypothesis.

Little is known about the precise relationship between balance performance and gait following stroke, a situation which places the efficacy of the widespread rehabilitation practice of attempting to re-educate balance as a means of improving gait in doubt. As a result, there is a need to study, more closely, the extent of the gait discrepancy following stroke, the balance performance of hemiplegic subjects as reflected in balance tests which are more relevant to gait than previous studies and the relationship of the gait and balance variables to each other. Study of these aspects of hemiplegic performance may help to confirm or refute current clinical practice and identify avenues of deficient performance, remediation of which may result in more satisfactory gait rehabilitation in this important clinical population.

STATEMENT OF THE PROBLEM

The major purposes of this study were: to investigate the extent to which balance and gait are affected in stroke by comparing patients with controls and; to investigate the relationship, if any, between balance performance and gait in stroke. The long term objective of this work is to propose more efficient methods of re-training a more normal gait pattern in this clinical population.

The specific aims of the study were as follows:

1. To compare the temporal and spatial parameters of gait of fully ambulant hemiplegic subjects with those of normal controls, matched for age and sex, at a variety of walking speeds.

2. To investigate and compare the adaptability of hemiplegic gait with normal gait based upon ranges of velocity and temporal and spatial symmetry and to establish the capacity of the hemiplegic subjects to consciously modify their gait pattern.

3. To compare the balance performance of hemiplegic subjects with normals based upon the position and variability of the position of the centre of pressure (CP) in standing, the ability to voluntarily shift and maintain the CP relative to the base of support.

4. To identify relationships between the temporal and spatial gait kinematics of hemiplegic subjects with balance performance and discuss the relative values of balance tests conducted with the feet parallel and with the feet in positions which represent the double support phases of the gait cycle.

RESEARCH HYPOTHESES

Two major hypotheses were formulated with a view to studying gait, balance and their relationship in people disabled by stroke.

Hypothesis 1

The temporal and spatial parameters of gait of fully ambulant hemiplegic subjects differ significantly from those of normal age- and sex-matched controls at a variety of psychometrically derived walking speeds designed to identify the extremes of range of gait velocity. Parameters compared included velocity, stride time, stride length, single support asymmetries, double support asymmetries, step length asymmetries, the durations of total support, single support and double support and the ranges of walking speed of which the subjects were capable. The walking speeds from which the comparisons were made included the self-selected speeds of "slowest", "slow", "free", "fast" and "fastest" and an additional "corrected" walk for the hemiplegic sample.

Stated statistically, the appropriate null hypothesis was:

H_0: $\mu H_{gk} = \mu N_{gk}$

Where μ represents the population mean, H represents data from the hemiplegic sample, N represents data from the normal sample and $_{gk}$ represents the gait kinematic variables.

The alternative hypothesis was:

H_a: $\mu H_{gk} \neq \mu N_{gk}$

Hypothesis 2

The balance performance of fully ambulant hemiplegic subjects significantly differs from that of normal subjects based upon: the location and variability of the position of CP while attempting to maintain predetermined postures in a number of directions and with different bases of support; the ability to voluntarily shift and/or maintain CP in these directions; and the ranges over which CP can be shifted in these directions relative to the base of support.

Stated statistically, the null hypothesis tested was:

H_o: $\mu H_{bp} = \mu N_{bp}$

Where μ represents the population mean, H represents the data from the hemiplegic sample, N represents data from the normal sample and $_{bp}$ represents the balance performance variables.

Balance performance parameters compared included the mean position of the CP on the X and Y axes (CPX and CPY) and the magnitude of postural sway as indicated by the variance of the mean position of the CP (dispersion index and the % of test time within 5%, 10%, 20%, 40%, 60% and 80% of body weight displaced from the mean CP). These measures were obtained while the subject stood in quiet standing and while maintaining shifts in four different cardinal plane postures (anterior, posterior, right and left) with the feet situated beside each other. Subjects were also compared in the maintenance of six postures where the feet were placed in positions which simulated both phases of double support of the gait cycle. Direction one (D1) (left foot back, right forward) and Direction two (D2) (right foot back, left forward) were compared with weight maintained over the posteriorly placed foot, then symmetrically between both feet and, finally, with weight maintained over the anteriorly placed foot. A dynamic balance test in both diagonal positions, where subjects were required to shift weight from the posteriorly placed foot to the front foot during data collection, was also included.

The alternative hypothesis was:

H_o: $\mu H_{bp} \neq \mu N_{bp}$

With these central hypotheses under test further statistical treatments explored the variability of the data in the hemiplegic sample by determining correlation coefficients between the various balance and gait measurement parameters. The research plan anticipated the possibility of making definitive statements about the strength of these relationships which, prior to the study, were expected to exist between gait kinematic factors and balance performance and which would be revealed once the two central test hypotheses were explored. In addition, a basis for discussion would be provided concerning the relevance to gait and the relative merits of the balance tests conducted with the feet parallel compared to the tests undertaken with the feet in positions which simulate both double support phases of the gait pattern.

DELIMITATIONS

Stroke is known to cause a wide variety of disabilities and can result in diverse symptoms. In an attempt to standardize the sample studied, inclusion and exclusion criteria were applied so that a relatively homogeneous group of post stroke patients were studied. Many of the exclusion criteria, specifically serious or unstable medical conditions, unilateral neglect, major perceptual dysfunction, significant peripheral sensory loss, severe homonymous hemianopsia, cognitive disturbance including memory loss, severe intractable pain, incontinence of bowel or bladder, are known to be negative rehabilitation prognosticators as described by Stonnington (1980). Thus, application of these criteria served to standardize the sample and permit study of those subjects who share the characteristics with those found in rehabilitation programmes. In addition, the common requirement that all stroke subjects in the study had qualified for, and received, rehabilitation further ensured standardization of this sample. All subjects walked without the assistance of a cane, tripod or quadripod further limiting the diversity of the group. The stroke subjects could, therefore, be considered elite in that ambulation performance was such that gait was fully independent without the assistance of any ambulatory aids.

The hemiplegic subjects who were examined in this study were people who had suffered a stroke some time ago. Their symptoms, therefore, were residual and stable. The reason for choosing this group as opposed to subjects who had suffered stroke more recently was that residual stroke subjects are more likely to be stable from a motor viewpoint. Many of the motor strategies of these subjects have become habituated thus ensuring a consistent performance in gait and balance performance during testing. This is in contrast to the performance of those whose stroke occurred more recently. In addition, recent stroke subjects can be considered to be somewhat frail from a medical viewpoint and it would be inappropriate to risk causing a medical crisis in the research laboratory. Further, the chronic subjects were more likely to undertake the tests with less risk than those with more recent strokes.

The purpose of studying only subjects who exhibited no disabilities from other disorders was designed to control the confounding variable of the gait deviation being embellished by some other pathology such as osteoarthrosis of the hip. Although this reduced the numbers available to be studied, this criterion was critical to include to ensure that the data reflected performance resulting only from stroke.

Those subjects who demonstrated receptive aphasia, one sequela of stroke, were excluded from the study as they may have had difficulty in fully comprehending instructions during both the gait and balance components of the testing.

An important objective of this study was to ensure that the findings were relevant to those patients undertaking rehabilitation. As indicated earlier, subjects were eliminated if they demonstrated symptoms which were associated with a negative rehabilitation prognosis. In addition, the incomplete motor recovery of the subjects studied, ensured by the inclusion/exclusion criteria, permitted some generalizations to be made to the more recent stroke population with some degree of confidence.

Due to the nature of this study, it was necessary to select subjects who were able to walk on the walkway without a cane or any other type of hand held walking aid. Though restricting the study to those who had regained good function following the stroke, the use of a walking aid during gait analysis would have created misleading data. Similarly, those who used a cane regularly were not included because the gait pattern would have become habituated to its use. The data collected without a cane, therefore, would not have been representative of the typical walking pattern of the subject.

LIMITATIONS

As has been pointed out in the literature, the study of stroke is typically complicated with difficulties in controlling certain variables (Bach-y-Rita, 1981). The following limitations should be considered when considering the implications of this study.

1. Stroke incidence increases proportionally with age and is less common in younger age groups. However, many of the older stroke sufferers, considered for inclusion in this study, were unable to walk without an assistive device. This would have made data collection difficult and hazardous. In the younger age groups, however, many of the stroke victims were in possession of a functional gait pattern and could walk without the assistance of a cane or quadripod despite showing clear subjective evidence of an asymmetrical hemiplegic gait pattern. The inclusion criterion that the stroke subjects would possess a confident but asymmetrical gait resulted in the sample studied being younger than originally anticipated. Thus, the mean age groups of the subjects in both samples were lower than that typically associated with stroke.

2. Despite elaborate attempts to standardize the stroke sample being studied, there was still variability of symptoms. This feature has been described before in studies by Knutsson and Richards (1979) and Wall and Turnbull (1987) and has been identified as a complicating factor in stroke related research (Bach-y-Rita, 1981). However, this variability permitted study of a range of effects caused by the severity of the stroke. A negative effect of these actions was that the results can only be confidently applied to patients with similar clinical profiles.

3. Subjects who made up the normal sample in this study may not be typical of a "normal, healthy population." This was due to the fact that universally acceptable criteria, which define a normal population, particularly those in the older age groups, are not clearly

established. The subjects studied in this project, may be different from a similar group of another country or from another region of Canada, thus jeopardizing widespread generalization from the study. However, this limitation is offset by the fact that the major role of the healthy sample was that of controls. Comparisons between the stroke sample and the control sample were, therefore, valid because the subjects in both groups shared age, sex and demographic characteristics.

4. The determination of "normality" was reached simply by asking the subjects if they suffered from any of the conditions indicated in the exclusion criteria. It was possible that some of the normal subjects were "abnormal" in that they may have failed to report a condition or were suffering from a disorder at a subclinical level.

5. Both groups studied in this project were not screened to ascertain activity levels. It is probable that the stroke group was more sedentary than the controls and this could have accentuated the differences detected. This limitation, however, is somewhat typical of the real situation in that the stroke population is likely to be less able from a cardiovascular and musculoskeletal viewpoint than the normal population.

6. Some clinical considerations such as the precise location of the lesion and the cause of the stroke were not controlled. Such information was not available. However, it was felt that the similarity of symptoms exhibited by the stroke subjects and the nature of the study rendered this limitation relatively inconsequential.

7. Both males and females and right sided and left sided hemiplegics were included in this study. A possibility exists that the hemisphere involved and the sex of the subjects may have introduced unforeseen confounding variables. Ideally, selecting the subjects from the same sex who display motor symptoms on the same side of the body would have been desirable. However, attention to these factors would have adversely affected the number of subjects available for the study thus threatening its internal validity.

8. The method of gait analysis examined the temporal and spatial kinematics of the gait pattern which can be considered to be measures of motor outcome. Similarly, the Chattecx Balance System™, used to determine the location and variability of the CP of the subject, was also an outcome measure. These measurement techniques provide no indication of alterations in muscle or nervous system physiology. However, these methods were appropriate to reach the objectives of this study and to provide useful, meaningful information concerning balance, gait and their relationship.

9. The subjects studied were fewer in number than had originally been anticipated. This was a function of the demanding inclusion/exclusion criteria which were applied to standardize the sample. This, together with the lack of a local register of stroke victims, made the locating and recruiting of qualified subjects extremely difficult.

10. The method of varying walking speed used in the study attempted to determine the velocities of which the subjects were capable. The technique required that subjects select their "slowest", "slow", "normal", "fast" and "fastest" speeds as well as the "corrected" condition in the stroke sample. As a result of its subjective nature, some variability of walking speed undoubtedly resulted from this method. However, the objective of studying a variety of walking speeds was attained.

11. There was a considerable range of age in the stroke subjects recruited for this study. This resulted from the demands of the inclusion/exclusion criteria and the relatively small population base from which subjects were selected (< 200,000). Although of some concern, stroke affects all age groups in similar ways in terms of motor function. Therefore, the impact of this limitation was not considered to be a major factor.

12. Each subject was tested only once at each gait test condition and once at each balance position. Such a procedure was undertaken because the length of each data collection period was considerable and it was felt that testing subjects more than once may have put the subjects at risk or have altered the results due to fatigue. This was a particular

concern in the hemiplegic group. Given that the data obtained for each walk and balance test was the mean of the entire test, it was felt that the benefits of having subjects undertake only one trial far outweighed the risks involved.

CHAPTER 2

REVIEW OF THE RELATED LITERATURE

DEFINITION OF STROKE

Stroke (or cerebrovascular accident) is an abrupt event resulting from a disruption of the blood supply to an area of the brain. This can be due to either hemorrhage of intracranial blood vessels, with subsequent damage to adjacent brain tissue, or to ischemia, resulting from a thrombus or embolism, which causes cerebral infarction (Toole, 1984). The subsequent neurological deficit results in a variety of disorders of function which are dictated by the particular areas of the brain which have been compromised and the extent of this damage (Brust, 1989). Included in these sequelae are sensory disruption, visual field defects, motor disorders characterized by abnormal muscle tone and the appearance of primitive spinal and brainstem mediated motor patterns, dysphasia, loss of perception of body image and spatial relations and loss of functional independence (Stonnington, 1980, Brust, 1989).

EPIDEMIOLOGY OF STROKE

It has been frequently reported that residual functional deficit due to stroke has far reaching implications economically, socially and in terms of quality of life of the victims (Hartunian *et al.*, 1980; Brocklehurst *et al.*, 1981). Stroke is the third leading cause of death in North America and is the primary cause of disability in the elderly (McCann and Culbertson, 1976). It has been estimated that in Canada each year, roughly 40,000 people will sustain stroke. This compares with 575,000 new cases per year in the United States (Goodstein, 1983) and approximately 110,000 in the United Kingdom (Evans and Caird, 1983). In the United States, recent statistics released by the American Heart Association (1992) estimated that there were 3,020,000 stroke victims alive in the United States in 1992. It is known that stroke afflicts between 1 and 2 people per one thousand of the population in the western world (McCormick, 1983). It is also known that the incidence of stroke increases with age and that amongst the elderly, this rate rises to 25 per one thousand (Evans and Caird, 1983). The incidence of stroke also varies with geographical region, with higher rates reported in China (Shi *et al.*, 1989); sex, with the incidence of stroke in men double that in women (Ward *et al.*, 1988); and race, blacks being more likely to suffer stroke than whites (Gillam, 1988).

The prognosis of those who have suffered stroke has been reported in the literature but the results appear to be somewhat variable. In some instances, it has been shown that between 25% and 35% of all people who sustain stroke die at the time or shortly after the stroke (Christie, 1983). Of those survivors, the majority suffer neurological dysfunction and physical disability of varying degrees (McDowell, 1976). Adams (1974), in a study conducted in Ireland between 1964 and 1968, found that 60% of patients recovered function from stroke while a further 30% remained incapacitated to the extent that they required institutional care. The remaining 10% died within two months. Of the 60 % who recovered to some degree, Adams found that more than one third became fully independent, regained intellectual clarity, had some effective use of the disabled hand and possessed a confident gait pattern without any appliances or ambulatory aids. The remaining two thirds of the "recovered" group were found to be more disabled and possessed a mental and/or sensory deficit, had no function in the affected upper limb and required an orthosis and an ambulatory aid, such as a tripod cane, to allow them to walk independently.

Over the last thirty years, although there has been a decline in the incidence of stroke in North America and Western Europe, the prevalence has increased (Isaacs, 1984). The reduction in incidence has been attributed to the improved management of known risk factors. More recently, however, a study in Rochester, Minnesota demonstrated that the incidence of stroke was again increasing; a phenomenon which is thought to be associated with the development of computed tomography. It has been proposed that this new technology has resulted in the diagnosis of stroke in cases which were previously undetectable (Broderick *et al.*, 1989). The rise in prevalence has also been due to a reduction in fatalities resulting from stroke, longer post-stroke survival periods and an increase in the numbers of elderly, within the population, who are the group at highest risk (Isaacs, 1984). Given that it is projected that the number of elderly in Canada and the United States will rise by 12% by the year 2000, disability as a result of cerebrovascular disease will continue to exert a considerable demand on health care systems for some time to come (Lane, 1978; Levy, 1979; Lane, 1981; Craik, 1990).

RECOVERY OF FUNCTION FOLLOWING STROKE

Recovery of function following stroke has been shown to be dependent upon the extent of brain damage sustained at the time of the acute episode (Matenga *et al.*, 1986). The least severe form of stroke is referred to as transient ischemic attack (TIA). These "mini strokes" are characterized by the appearance of stroke-like symptoms, which completely reverse within 24 hours of their appearance (Caronna, 1976). Recovery is, therefore, complete. Reversible ischemic neurological deficit (RIND) demonstrates a similar picture to TIA, but complete recovery occurs in excess of the 24 hour period (Buonna and Toole, 1981). All other cerebrovascular accidents, referred to as completed strokes, result in loss of function which will recover, to varying degrees, from days to years following the lesion (Newman, 1972: Bach-y-Rita, 1981). It is thought that rehabilitation enhances recovery in terms of speed and extent (Lehmann *et al.*,1975) but, this has not been unequivocally demonstrated. This situation has, at least in part, been caused by well known difficulties in researching this topic (Isaacs, 1979; Bach-y-Rita, 1981; Tallis, 1984). A number of factors have complicated research in the rehabilitation management of stroke and have been described by Bach-y-Rita (1981) as: difficulties in identifying the precise location and extent of brain damage which leads to an inability of researchers to interpret the exact mechanism of recovery of function; difficulties in obtaining sufficient numbers of patients with similar lesions and other characteristics; problems in accurately quantifying progress; the long term nature of the CNS recovery process; and the high cost of long term clinical research programmes. In addition, another factor which has compromised research of the effect of rehabilitation in stroke has been the difficulty in separating the effect of rehabilitation from functional return due to spontaneous recovery of CNS activity (Lehmann *et al.*, 1975; Bishop, 1982a), a problem which has been recognized for over a century (Gowers, 1888). It is often assumed that spontaneous recovery occurs in the first six months following the stroke and is virtually complete after one year. Lehmann *et al.*, (1975) hold this view, but they have also produced evidence to strongly suggest that rehabilitation training produced gains that were not attributable to spontaneous recovery alone. Similar observations have been made by Skilbeck *et al.*, (1983) who also proposed that many of the tools used to assess progress are insensitive to change following the initial stages of recovery, thus masking the detection of later functional return. Bach-y-Rita (1983) has reported long term recovery occurring in an elderly male for seven years post stroke as have Wall and Turnbull (1987) who showed that some stabilized stroke patients were capable of improving certain gait parameters with specific training. The subjects in the latter study were all at least eighteen months post stroke. Wade *et al.*, (1992) also demonstrated functional improvement in a group of chronic stroke patients although the improvements were not maintained upon cessation of active physiotherapy treatment.

Recovery following CNS damage is a complex process. For many years it was thought that the CNS possessed little potential for recovery following damage from such insults as stroke (Bishop, 1982b). Recovery of function was attributed, therefore, to the ability of the patient to develop compensatory strategies whereby functional goals could be achieved by using parts of the motor system which had not been damaged (Tallis, 1984). More recently, however, a deeper understanding of the capability of the CNS to recover function has developed (Held, 1993). It has been proposed that recovery is due, in part, to the re-establishment of CNS function particularly by neurons which may have been damaged at the time of the stroke but not irreversibly (Wall, 1980). As these neurons recover and recommence functioning, an associated improvement in functional ability occurs. Another process of recovery, referred to as re-organization, may also contribute to CNS recovery (Yu, 1980). This theory proposes that the functions of a permanently damaged component of the CNS can be taken over by an adjacent but previously functionally distinct part of the brain. It is unclear which of these mechanisms predominates and the relative contribution of each to the recovery of function (Turnbull, 1986). Additional mechanisms such as collateral sprouting and the activation of previously non-functional synaptic connections have also been proposed as reasons for the recovery of CNS function following brain damage (Laurence and Stein, 1978). From these findings, therefore, it can be proposed that early recovery following stroke is a result of the re-establishment of neural activity whereas later recovery is probably due to CNS reorganisation.

STROKE REHABILITATION

Brocklehurst *et al.* (1978) have pointed out that the medical and surgical treatment of completed stroke is limited and as a result the primary therapeutic approach has been directed towards the rehabilitation of the survivor. Rehabilitation has been described by Rusk (1949), in general terms, as "...the third phase of medical care to be instituted following the first phase, preventive medicine, and the second phase, curative medicine and surgery". Nichols (1979) has defined rehabilitation as implying the restoration of patients to their fullest physical, mental and social capabilities but proposes that, in actual practice, it usually refers to the physical treatment of the resultant disability. A variety of studies have attempted to evaluate the process of stroke rehabilitation with generally favourable results (Gillner *et al.*, 1969; Lehmann *et al.*, 1975; Anderson *et al.*, 1977; Feigenson *et al.*, 1978; Shiavi *et al.*, 1979; Smith *et al.*, 1981) but these studies have tended to use global outcome measures such as degrees of dependence and independence (Garraway *et al.*, 1980), indices of activities of daily living (Smith *et al.*, 1981), vocational return (Fugl-Meyer *et al.*, 1975) and outcome and cost (Feigenson *et al.*, 1978). Lehmann *et al.*, (1975), however, have cited a number of studies which demonstrate contradictory conclusions in assessing the effectiveness of stroke rehabilitation due to wide variations in measuring outcomes. Additionally, it is well known that stroke rehabilitation is an expensive process (Feigenson *et al.*, 1978) and that consideration of the cost/benefit is an important factor in maximising the resources of health care delivery systems (Wylie, 1966; Lane, 1978; Feigenson, 1979). In attempts to maximise available resources, it has been proposed that attention should be paid to prognostic factors in selecting patients for realistic rehabilitation (Nickel, 1976; Brocklehurst *et al.*, 1978; Gowland, 1986). It has been shown that stroke survivors with particular prognostic signs and symptoms do not respond well to the rehabilitation process (Adams, 1974; Andrews *et al.*, 1980). Stonnington (1980) has listed the persistence of incontinence of bowel and bladder, cognitive disorders, behavioural disturbances, serious concomitant illness, persistence of central sensory disturbances, aphasia, serious motor deficiencies, pain and advanced age as negative prognosticators. The presence and severity of many of these symptoms are likely a function of the amount of CNS tissue compromised as a result of the stroke (Matenga *et al.*, 1986).

Early models of rehabilitation were predicated on the belief that the CNS would recover little function (Tallis, 1984). Although recovery was maximised in a reactive sense many of the rehabilitation efforts were designed to train the patient to compensate for the functional deficit which resulted from the CNS lesion (Swenson, 1978). In the stroke patient, emphasis was placed on teaching the patient to use intact motor function on the non-hemiplegic side of the body (Swenson, 1978). Functional movement such as walking, was asymmetrical, and, in the case of gait, the patient was supplied with orthoses to control the position of the hemiplegic foot and a cane used in the uninvolved hand, to assist in ambulation (Swenson, 1984). In recent years, given that it is now well accepted that the CNS has a much greater potential for recovery than previously thought, rehabilitation attempts to normalise disordered movement proactively, thus aiming to improve the functional ability of the patient (Ryerson, 1990). This change in emphasis highlights the need to identify further effective re-education processes and, once developed, research the effectiveness of such procedures.

THE ROLE OF PHYSIOTHERAPY IN STROKE REHABILITATION

Rehabilitation is carried out by a team of professional personnel and the physiotherapist is considered to be a fundamental member of the stroke rehabilitation team (Isaacs, 1978; Redford and Harris, 1980; Stonnington, 1980). The role of the physiotherapist in the rehabilitation effort, in general terms, has the objective of restoring more normal function and is particularly involved in the assessment, treatment and management of movement dysfunction (Bobath, 1978; Carr and Shepherd, 1983; Banks, 1986; Turnbull and Wall, 1989).

A major component of physiotherapy in rehabilitation is the application of therapeutic exercise procedures. Historically, therapeutic exercise evolved from the framework of recreational exercise and sport (Licht, 1985). The same author has defined therapeutic exercise as "the motion of the body or its parts to relieve symptoms or improve function". Indeed, there is evidence that exercise was used as a treatment modality by the Ancient Chinese and the Ancient Hindus around 1000 BC (Drabkin, 1950). Specific treatment of stroke using therapeutic exercise was documented as early as 200 BC by Soranus of Ephesus who advocated the use of hydrotherapy, heat, progressive resistance exercise and gait re-education using devices to assist ambulation (Adams, 1974). This regime was surprisingly enlightened and went as far as recognising abnormal muscle tone (spasticity) as a symptom which interfered with motor function. However, at that time, it was believed that the underlying cause of spasticity was *rigor mortis* which led to stroke being referred to as "the half dead disease" (Drabkin, 1950).

The use of modern-day exercise techniques by physiotherapists began, in Britain, in 1910 when the recently formed physiotherapy profession incorporated the Swedish Remedial Exercises into its curriculum (Young, 1967). These exercises had been developed at the Central Institute of Gymnastics in Stockholm by Pehr Henri Ling (Hirt, 1967). These procedures heavily favoured an analytical and kinesiological foundation, with Ling proposing a series of guidelines to govern the application of movement procedures to patients. However, the jurisdictional threat to medicine caused physicians of the day to decry this approach to rehabilitation, thus inhibiting its widespread adoption.

A number of forces combined to lay the foundation for the development of the types of physiotherapy applied today in the rehabilitation of the stroke patient. The dramatic neurophysiological progress in the early Twentieth Century resulting from the work of Von

Uexkull (1905), Sherrington (1906) and Magnus (1909; 1926) stimulated the development of a variety of innovative approaches to the treatment of disability following damage to the CNS. The significant features of these new methods were that they were directed towards central and peripheral nervous system activity, thereby attempting to influence the functioning of these systems. This was in contradiction to previous methods which were restricted to muscle strengthening and increasing joint range of movement as the predominant methods to improve motor performance (Pinkston, 1967). Frenkel, in 1889, was the first to propose that disordered CNS function could be improved through repetitive exercise. In his programme of treatment for patients with tabes dorsalis, in which sensory ataxia resulted from progressive destruction of the dorsal columns of the spinal cord, he suggested that repetitive, augmented feedback exercises could improve the marked incoordination associated with this debilitating condition thus improving gait (Frenkel, 1902).

These events were taking place at the same time as the ravages of war and pandemic diseases, such as poliomyelitis, were placing a major demand on societies to cope with long term disability. Similarly, the explosion of technology, in particular the mechanisation of farming and the advent of the automobile were associated with an increasing incidence of accidents accompanied by serious trauma and so increasing further the prevalence of serious permanent disability. Medical care also advanced with the advent of the successful management of bacteria with resulting increases in survival rates from severe illness (Licht, 1975). People with chronic, long term disabilities, therefore, began to place a significant demand on health care resources.

The late 1950s and early '60s heralded the appearance of a number of new habilitation and rehabilitation approaches which were designed to deal more effectively with the patient with neurological disease. Doman *et al.*, (1960) proposed an integrated approach to the cognitive and motor development of children with cerebral palsy. Knott and Voss (1968) developed the earlier theoretical framework of Kabat (1952) and introduced the proprioceptive neuromuscular facilitation (PNF) approach which was specifically targeted toward the treatment of acute anterior poliomyelitis. Rood (Stockmeyer, 1967), working from the premise that a motor act was preceded by a sensory experience, extrapolated her techniques from the field of cerebral palsy to the treatment of the adult neurological patient which was based on an ontogenetic neurodevelopmental model. In addition, Rood proposed the use of sensory stimulation by such techniques as icing, vibration and brushing to improve motor function (Goff, 1969). Brunnstrom (1970) extended the work of Twitchell (1951) and Fay (1954) in developing a treatment framework specific to adult hemiplegic patients. Founded upon phylogenetic principles, Brunnstrom argued that recovery of function following stroke progressed through a series of predictable stages from total absence of muscle function through primitive reflexive motor behaviour, which she referred to as the basic synergies, to the sophisticated highly specialised movement patterns characteristic of humans with the resultant disability being determined by the stage at which recovery plateaued. Brunnstrom (1970), through systematic observation of a large number of stroke subjects, described the characteristics of each stage of recovery which later formed the basis of her assessment system. The first stage of recovery, scored as a 1, described the early post stroke patient who was in the phase of cerebral shock. No motor activity on the hemiplegic side of the body was evident and the patient was designated as "flaccid". Stage 2 was characterised by the appearance, on the affected side, of the basic synergies presenting as increasing spasticity and associated reactions, this latter symptom being defined as the appearance of abnormal synergies upon the application of appropriate facilitatory stimulation by an assessor. Stage 3 was reached when the primitive synergies became clearly apparent, spasticity was severe and the patient could initiate the synergies voluntarily. Stage 4 was reached when levels of spasticity began to decline and the patient developed the ability to produce movements which, although still dominated by the basic synergies, possessed elements which were not part of the synergistic patterns. Stage 5 was attained when spasticity was found to be minimal, an enhanced repertoire of sophisticated movement was possible in the affected limbs and some use of the affected hand was possible. The final stage, Stage 6, was reached when normal or near normal motor activity was acquired. It was Brunnstrom's contention that all stroke patients passed through these levels of recovery and treatment of the patient was designed to attempt to reach the highest hierarchical motor level.

Bobath, basing her ideas on the work of the neurophysiologists Magnus and Von Uexkull and heavily depending upon an ontogenetic neurodevelopmental framework, developed a series of principles of treatment along with accompanying exercise procedures to treat spastic, cerebral-palsied children (Bobath, 1963). Later, armed with anecdotal success and widespread subscription to her ideas, Bobath (1970) extrapolated her techniques to the treatment of the adult hemiplegic patient.

A negative effect of these rapid developments in the approaches to the treatment of the neurological patient was considerable confusion amongst rehabilitation professionals concerning the legitimacy of these techniques. In an attempt to clarify the situation these physiotherapy approaches, some similar but others contradictory, were subjected to intense examination in a landmark conference in 1966 before many of the techniques were fully documented (NUSTEP, 1967). Leading practitioners of the various approaches were invited to describe, demonstrate and submit their work for peer review. At the end of the conference, Wood (1967) concluded that all the techniques appeared to share five common denominators. These common elements were: a recognition that sensation and motion were intimately related; a neurophysiological basis through an understanding of the sequence of normal motor development; a recognition of the fact that early motor behaviour is reflexive and, in the treatment of damage to the CNS, these reflexive pathways can be used to influence voluntary effort; the employment of important concepts of motor learning particularly repetition, frequency of stimulation and the employment of sensory cues to facilitate learning and; the establishment of a close personal interaction between patient and physiotherapist with the physiotherapist developing a delicate sensitivity to the response and movement of the patient.

More recently, extensions of these approaches have been proposed, particularly that of Bobath, to deal more effectively with the rehabilitation of people suffering from neurological disorders including stroke (Johnstone, 1980; Carr and Shepherd, 1983; Cotton and Kinsmen, 1983; Davies, 1985; Turnbull, 1986). These more recent approaches have concentrated more on principles of psychomotor learning accessing the considerable body of knowledge in this area (Turnbull and Wall, 1989). In addition, the use of technology has become more apparent with the appearance of functional electrical stimulation, electromyographic biofeedback and force sensitive platforms which provide performance feedback of attempted motor activities to both the operator and the patient (Turnbull and Wall, 1989).

Although physiotherapy is applied widely in the rehabilitation of stroke, many of the techniques have never been subject to carefully controlled scientific scrutiny (Peat, 1981: Rothstein, 1991). Views of the effectiveness of physiotherapy, therefore, have ranged from enthusiastic intuitive support (Ericsson, 1978; Harris, 1978) through critical suspicion (Basmajian, 1975) to outright cynicism (Brocklehurst *et al.*, 1978) with the latter group proposing a redefinition of the objectives of physiotherapy and suggesting that alternative treatment, carried out by volunteers or more simply trained personnel, merited consideration, a view shared by McDowell (1976). Another major complicating factor in researching the effectiveness of physiotherapy in enhancing recovery following stroke, is the precise definition of what constitutes "physiotherapy". In many reported studies, the physiotherapy procedures are not described (McCann and Culbertson, 1976; Brocklehurst *et al.*, 1978; Smith *et al.*, 1981). In other instances, physiotherapy is referred to in general terms with little attention being paid to specific details of technique, frequency and duration of application (Gillner *et al.*, 1969, McDowell, 1976: Smith *et al.*, 1976; Anderson and Kottke, 1978; Nichols, 1979; Johnston, 1983). As a result of these diverse opinions concerning the validity of physiotherapy as a means to enhance recovery following stroke, there is an urgent need to evaluate the effectiveness of physiotherapy procedures and validate the theoretical assumptions which underlie the techniques of treatment.

CURRENT PHYSIOTHERAPY APPROACHES

The approach to the rehabilitation of the neurological patient developed by Bobath (1970), has received wide subscription and has significantly influenced thought which is evidenced in the neurological physiotherapy and related literature (Todd, 1974; Lane, 1978; Wall and

Ashburn, 1979; Stonnington, 1980; Carr and Shepherd, 1980; Ruskin, 1982; Todd and Davies, 1982; Carr and Shepherd, 1983; Davies, 1985; Banks, 1986; Umphred, 1990). In the rehabilitation of stroke, the principles proposed by Bobath emphasised the re-education of the affected, hemiplegic side of the body by the suppression of abnormal patterns of movement which included the reduction and control of levels of spasticity (Bobath, 1978). During the achievement of this objective, procedures were applied which were designed to encourage more sophisticated patterns of movement including the elicitation and retraining of normal postural reactions such as righting, balance and weight bearing responses (Lane, 1978; Todd and Davies, 1982). Those principles were applied in the rehabilitation of gait, an activity to which the highest proportion of rehabilitation time is devoted (Nichols (1979). Strategies designed to improve other complicating factors such as sensory disturbances and loss of sophisticated selective movement patterns were also applied during rehabilitation (Gordon, 1990; Carr and Shepherd, 1990).

The Bobath approach specifically addresses the clinical symptom of abnormal muscle tone which, she contended, interfered with the production of normal functional movement. The concept of tone, as applied to striated muscle, was first introduced by Müller in 1838 (Wyke, 1976) and was defined as "the slight contractile tension characteristic of normal skeletal muscle when at rest". Later, Wyke (1976) defined muscle tone as "the tension obtaining at any moment between the origin and insertion of each muscle..." Wyke (1969) and Basmajian (1974) attributed muscle tone to three underlying factors: the elastic properties of the connective tissue stroma and tendons of the muscle; the visco-elastic properties of the fibrillary proteins contained within each muscle fibre; and the degree of motor unit activity obtaining at the time throughout the muscle. It is the latter factor which gives rise to the phenomenon of spasticity, at least initially, because it is the only one which is influenced by the CNS (Wyke, 1976). A disruption of the excitatory/inhibitory balance, caused by the damage to higher control centers within the CNS, results in abnormally high levels of resting motor unit activity (Moore, 1981). It is believed that more sophisticated levels of motor activity, which are mediated at higher levels of the CNS, are replaced by primitive, reflex motor patterns which become released as a consequence of reduced control from higher centers in the CNS. This reflex motor behaviour (defined as stereotypical motor responses resulting from given sensory stimuli) are not abnormal in the neonate but, as the CNS develops and matures, these primitive movement patterns are replaced by more sophisticated righting responses (defined as motor acts which maintain the relationship of the head, neck and body in normal alignment) and equilibrium responses (Fiorentino, 1977). This latter group of automatic motor behaviours are necessary to maintain the relationship of the centre of gravity of the individual with the base of support. Thus, these responses are essential to maintain balance in sitting, standing and while moving. In the stroke patient, the mature automatic motor responses are replaced by the reappearance of primitive motor patterns on the contralateral side of the body to the damaged hemisphere. According to Bobath (1970), treatment of such patients commences with attempts to inhibit abnormal muscle tone. Inhibitory positioning, which usually places the hypertonic muscles in a lengthened position, is used to decrease tone along with the influencing of "key points of control". These key points are specific triggers, usually situated proximally, which, when influenced, reduce tone in the whole limb. For example, the hemiplegic scapula is typically retracted and rotated so that the glenoid cavity faces downwards mainly due to increased resting tone in the rhomboids. Protraction of the scapula, either actively or passively, has the effect of exerting a sustained stretch on the hypertonic rhomboid muscles thus causing them to relax, an event associated with a reduction in tone in the hypertonic muscles of the whole upper limb. When inhibition has been achieved, normal righting and equilibrium reactions are encouraged and repeated in attempts to render them more permanent.

According to Bobath (1978) an important concept pertaining to the understanding of the abnormal motor condition of hypertonicity involves the basic synergies. Synergies can be defined as the basic primitive movement patterns where a number of body segments are involved in a predictable, relatively uncontrollable and reflexive manner. They are triggered by a specific sensory input and, as a result, satisfy the definition of a reflex. These synergies can be divided into those which are mediated at the spinal level and those which are mediated in the brain stem. Examples of spinal level synergies include the flexor withdrawal, the extensor thrust and the crossed extensor reflexes (Rothstein *et al.*, 1991). They are phasic reflexes and, in reaction to a sensory stimulus, produce stereotypical movement simultaneously at the proximal, intermediate and distal joints at the same time. The brainstem reflexes are tonic in nature and are characterised by alterations in muscle tone. Examples of these include the tonic neck reflexes, both asymmetrical and symmetrical, and the tonic labyrinthine reflexes (Fiorentino, 1977). Synergies, therefore, are mass movement patterns which are not selective and, therefore, interfere with normal, sophisticated functional movement. It is these synergies which Bobath believes must be inhibited before more normal movement can be elicited. Based on this theory, a wide range of specific techniques have been developed and documented in attempts to re-educate diverse functional activities (Bobath, 1978).

Fundamental to the whole of the Bobath approach is the foundation of the ontogenetic sequence of motor development. According to Bobath (1978), re-education of motor function follows normal developmental principles. Proximal function such as control of the head and shoulder girdle are attempted before distal function. Thus, the foundation of cephalocaudal motor development is maintained. Similarly motor responses which permit weight bearing, such as those seen in the upper limb during crawling in the child, are attempted before free movement is produced. Bobath has referred to these free movements as "placings" and they can be defined as the ability to maintain a posture against the force of gravity. Considerable attention is also paid to the achievement of normal postural alignment of body segments. For example, the stroke patient tends to exhibit an abnormal alignment of the lower limb which likely contributes substantially to an abnormal gait pattern. The pelvis is rotated so that the affected side is posterior to the unaffected side in the transverse plane. To compensate for this, the patient maintains the affected hip joint in slight flexion. The line of gravity of the body is altered as a result of these discrepancies and passes anterior to the knee joint and posterior to the ankle joint. The patient, therefore, walks with the knee in hyperextension and a plantarflexed ankle joint on the hemiplegic side. In addition to the abnormal alignment of the leg, this position tends to aggravate the already increased tone in the extensor pattern of the lower limb. This will occur as a result of the the stimulation of the ball of the foot as the foot first touches the ground eliciting the extensor thrust synergy and further hindering the achievement of a normal gait pattern.

There have also been a number of elaborations on the work of Bobath and others. Johnston (1980) used inflatable air splints to control primitive movement patterns while attempting to stimulate normal movement. Her first publication in 1976 advocated the proprioceptive neuromuscular facilitation approach. However, in subsequent writings, she acknowledged the influence of Lane who convinced her to subscribe to a Bobathian framework. Johnston (1980) also recognized the importance of repetition of the inhibition of abnormal movement and the production of normal movement by titling a book "Living in Pattern" which, as the title implies, argues strongly in favour of constant repetition of normal movement so that it could be learned and, therefore,become more permanent. Carr and Shepherd also recognized the importance of psychomotor learning which they proposed was the foundation of a "novel" approach to the treatment of stroke (Carr and Shepherd, 1983). Their "Motor Re-learning Program", upon close examination, bore considerable resemblance to the Bobath approach but applied it in a psychomotor learning framework, citing the work of such motor learning researchers as Singer, Marteniuk, Fitts and Bilodeau. An interesting characteristic of the Carr and Shepherd model was that the usual plethora of neurophysiological justification of the treatment basis was downplayed. Instead, these authors used the type of models used by theoreticians in the motor learning area and considered environmental and functional variables. In 1987, these same authors presented a compilation of their thoughts on this "new" approach to the physiotherapy treatment of the patient suffering from neurological dysfunction (Carr and Shepherd, 1987). This book, in addition to promoting the motor learning model for rehabilitation, argued that the time had come for a move away from the exclusive subscription to neurophysiological based theories to justify physiotherapy procedures and focussed instead on the theory and practice of skill acquisition and the process of recovery of the CNS following brain damage. This proposed "movement science" had its foundation in such fields as biomechanics, motor learning, muscle biology, neurophysiology, cognitive psychology and human ecology.

In a separate development, Cotton and Kinsman (1983) published a book in which they proposed "conductive education" as an option in treating stroke patients. Conductive education was developed in Budapest by Petö and, like the Bobath approach, was developed to treat children with cerebral palsy and was later extrapolated to treat adults with CNS disorders. Key elements of the conductive education approach included the philosophy of the use of "the patient's own active participation and initiative" rather than the direct hands-on intervention by a therapist. Building upon the work of Pavlov, Luria, Vygotsky and Bernstein, Petö applied such concepts as: rhythmical intention, where attempts were made to improve movement quality through the use of language; intrinsic motivation; continuity, which described the requirement to ensure that movement patterns were practiced in every day living situations; techniques of facilitation and; the use of supervisory personnel referred to as conductors. The conductors functioned in a manner similar to specialized physiotherapists and their role was to assess the patient both in the execution of activities of daily living as well as during group treatment, to organize and direct the patient's daily routine, to initiate the patient into the group, to direct the group, to make up the task-series and to create a pleasant working atmosphere. Conductive education advocates, as a central component, frequent repetition of normal functional movement patterns with the conductor providing feedback.

As a follow up to the landmark NUSTEP conference in 1966, IISTEP was held in 1990 (Physical Therapy Research Foundation, 1990) at the University of Oklahoma, Norman, Oklahoma. This conference had similar objectives to the first conference which was to scrutinize and rationalize current practice in the treatment of the patient with neurological disorder. The central themes of this conference related to motor control, balance, motor skills acquisition and motor development. Many of the traditional physiotherapy practices were claimed to be somewhat mystical and calls were issued for the undertaking of sound, well-controlled clinical research to justify current and future practice. The position was clearly established that control of movement rather than the promotion of muscle strength was a key element of physiotherapy in the treatment of the neurological patient along with essentially the same ingredients as those proposed earlier by Carr and Shepherd (1987).

The further recognition of motor control as a concept was repeatedly emphasised. From a critical viewpoint, IISTEP was unlike the NUSTEP conference in that it was organized, funded and presented by physiotherapists for physiotherapists. The potential, therefore, for unintentional professional bias to colour discussion was greater than the NUSTEP conference which was attended by a greater diversity of health care professionals. Further, those presenting and attending were almost exclusively from the United States which added a parochial element to the proceedings. It is interesting to note that motor control, which was advocated as a principal new direction, has been the corner stone of the Bobath approach for as many as twenty years prior to this event. Finally, the content of the IISTEP conference bore some resemblance to the "common denominators" identified at the 1966 NUSTEP conference some 24 years earlier namely the application of neurodevelopmental concepts and motor learning theories. Despite these concerns, the IISTEP conference permitted debate and documentation of the direction of neurological and pediatric physiotherapy at least in the North American continent.

A trend, therefore, has emerged in that the application of the body of knowledge relating to psychomotor learning has increasingly become claimed to be a critical element of the physiotherapy treatment of the neurological patient. The significance of this trend has been that the traditional physiotherapy treatment paradigm (short periods of motor practice several times weekly) is being questioned and attempts to resolve this conflict are being proposed in an attempt to create a greater intensity of relevant practice (Turnbull, 1992).

There has been an increasing interest, over the recent past, in attempting to conceptualize the functioning of the CNS and the effect of disturbances of function caused by lesions of the CNS in the rehabilitation context (Marteniuk, 1979; Carr and Shepherd, 1983; Turnbull, 1986). These authors have attempted to develop theoretical constructs which focussed on the function of the CNS rather than the structures and processes which mediated those functions. Although these ideas have been extensively developed and tested in the psychomotor area, it is only recently that attention has become increasingly focussed on this approach in the rehabilitation area. The advantage of this approach is that the effect of a lesion on motor performance can be much more clearly identified and remedial strategies developed.

Therefore, it can be seen that there has been an upsurge of interest in expanding the notion of motor control away from the traditional neuroanatomical and neurophysiological frameworks usually associated with the health related professions and to consider rehabilitation of function in a motor learning context. For example, the therapeutic exercises provided by physiotherapists and designed to improve the hemiplegic gait pattern can be considered to be part practice of components of the gait pattern. Turnbull and Wall (1989) have identified the support and the swing phases as two parts of the gait cycle which are functionally different and have proposed that part practice of both is appropriate with attempts later to combine the two into one functional motor entity. Others have contended that the gait pattern is a highly organised, integrated skill which should be practiced in its entirety (Winstein *et al.*, 1989; Waagfjörd *et al.*, 1990). Nonetheless, these opinions, which are samples of thought in physiotherapy, demonstrate the utilisation of motor learning principles into therapeutic exercise. Given that balance reeducation is thought to be a necessary precursor to gait training following stroke, it is necessary to establish more precisely the role of balance in gait so that it can be improved through practice. Identifying this linkage with a view to improving therapeutic exercise is a long term objective of this study.

HEMIPLEGIC GAIT

A variety of methods have been used to describe hemiplegic gait ranging from the observation of discrepancies from normal to the use of highly sophisticated technology. Although frequently described as a generic entity, it is known that there are a number of subtypes of this gait pattern. Colaso *et al.*, (1971) identified 29 pattern subtypes in a group of 50 hemiplegic subjects while 3 were proposed by Knutsson and Richards (1979). Wall and Turnbull (1986) also described a variety of different patterns based upon the profiles of the temporal kinematics of a sample of 24 residual stroke subjects.

Using observational techniques, Perry (1969) described the swing phase of the hemiplegic gait pattern as being characterised by hip flexion, knee extension, ankle plantarflexion and overall circumduction of the lower limb and the support phase as demonstrating plantarflexion upon loading, hyperextension of the knee at mid support and deficient roll-off at toe off. The same author proposed that these abnormalities were due to: poor single limb balance; difficulties in controlling forward progression; lack of adequate shock absorption at heel strike; lack of control of momentum during stance; lack of ability to generate force for push off to maintain forward propulsion and; lack of quick, adequate excursions of the hemiplegic leg during swing (Perry, 1969).

Clinical descriptions of hemiplegic gait have also been made considering the effect of abnormal movement patterns on normal walking (Brunnstrom, 1970; Bobath, 1978; Musa, 1986). From this perspective, the following observations can be made. The affected lower limb of the stroke patient moves in a manner which is dominated by the mass primitive movement patterns of flexion and extension rather than the skilled selective movements, which normally occur in a coordinated sequence between the joints of the lower limb (Brunnstrom, 1970). The hemiplegic subject is unable to break up these mass synergies and movements of the hemiplegic leg are frequently dominated by extensor hypertonicity. This extensor synergy is a primitive motor pattern which has been released from higher central nervous system control. According to Musa (1986), this pattern consists of extension, adduction and internal rotation of the hip, extension of the knee joint and plantarflexion and inversion of the foot. During movement in the upright position, the hip joint does not fully extend: it remains in slight flexion while the knee joint hyperextends and the foot assumes a plantarflexed and inverted posture. The patient is unable to move the joints of the lower limb selectively with movements dominated by stereotypical, primitive reflexive actions. The pelvis of the patient is habitually rotated to the affected side in the transverse plane, a posture often referred to as pelvic retraction, and is elevated on the affected side of the body. The flexion synergy in the hemiplegic lower limb consists of flexion of the hip and knee joints with dorsiflexion of the ankle joint. As with the extensor synergy, the patient is unable to move the joints selectively. Any attempt to flex the hip joint is accompanied by hip abduction, associated flexion of the knee and dorsiflexion and inversion of the foot (Sawner and LaVigne, 1992). The extensor synergy dominates the movement pattern particularly during the support phase and the hemiplegic patient will often initiate the flexion synergy to advance the foot during the swing phase. Lehmann *et al.*, (1987) have reported that the swing phase is initiated by flexion of the entire lower limb instead of the normal coordinated sequence. The characteristic gait pattern of the hemiplegic person is the result of the influence of these abnormal motor patterns.

During the support phase of gait on the hemiplegic leg, initial contact is often made with the lateral border of the foot in the region of the base of the fifth metatarsal (Brunnstrom, 1970; Colaso *et al.*, 1971). One cause of this discrepancy is weakness of the pretibial muscles (Stanic *et al.*, 1978). However, it may also be due to the predominance of the extensor synergy (Musa, 1986). Perry (1969) and Knutsson and Richards (1979) have drawn attention to the inability of the hemiplegic subject to dorsiflex the ankle joint on the affected side during the initial part of the support phase on the affected leg, a motor pattern which normally permits the tibia to rotate forwards on the fixed foot. These authors proposed that this abnormality was due to excessive tone in the posterior tibial muscles. Knutsson and Richards (1979) have also postulated that the inability of the calf muscles to relax and elongate was a major factor which contributed to the hyperextension of the knee joint during the support phase thus giving rise to one of the characteristic features of the hemiplegic gait pattern.

As reported earlier, the maintenance of extension at the knee joint during support is often dominated by the extensor synergy or can be created by forward leaning of the trunk which causes the knee joint to assume the mechanically, more stable position of extension during weight bearing. This is of some significance because this abnormal motor pattern results in the characteristically deviant gait. The production of the extensor moment by the affected limb during weight bearing, therefore, if produced or dominated by a primitive extensor synergy, is inadequate to produce a normal support phase in the gait pattern. Normal balance mechanisms, which are sophisticated automatic motor patterns and which are required to maintain the relationship between the base of support and CP of the walker, are deficient as a result of the hemiplegia (Bobath, 1978). Furthermore, extension of the hip is limited possibly because the rectus femoris muscle is unable to relax sufficiently to permit its full elongation (Perry, 1969). The affected lower limb, therefore, has little ability to produce the normal upward and anterior force during the latter part of support and the subsequent step by the unaffected leg is shortened, resulting in a step length asymmetry (Wall and Turnbull, 1987).

In cases where the extensor synergy is excessively predominant, the patient demonstrates an inability to shorten the limb during the swing phase. The hip joint remains in slight flexion while the knee joint is extended and the ankle joint is plantarflexed and inverted. The patient, therefore, is forced to abduct the hip joint so that the foot can clear the ground as it progresses anteriorly. In addition, this movement may be accompanied by compensatory trunk movements (Perry, 1969; Bobath, 1978; Musa, 1986). In patients who are capable of initiating the primitive flexor synergy to execute the swing phase, extension of the knee in the latter part of the swing phase is associated with hip extension and ankle plantarflexion as the extensor synergy is initiated. This results in a shortened step of the affected leg and initial contact discrepancies (Perry, 1969). Similarly, the normal rotation of the pelvis is deficient which compounds the gait abnormality.

Another common swing phase abnormality is caused by "functional" weakness of the hemiplegic leg (Bohannon, 1991). The patient tends to "drag" the affected leg behind them, a problem which is accentuated by the presence of a retracted pelvis on the affected side. This gives rise to the "crab-like" gait pattern as described by Wall *et al.*, (1987). The hemiplegic subject may also spend abnormally long periods of time in thrusting double support on the affected side as a result of this functional weakness.

The slow speed of the hemiplegic gait, in comparison to normal, has been reported repeatedly along with associated limitations in stride time and stride length (Peat *et al.*, 1976; Wall and Ashburn, 1979; Brandstater *et al.*, 1983). Particular reasons for these deficiencies have been proposed to be slowness in advancing the affected leg in swing and inadequate shifting of weight over the affected leg in support (Wall and Ashburn, 1979; Knutsson and Richards, 1979). Giuliani (1990) reported that hemiplegic subjects demonstrate spatiotemporal asymmetries and experience difficulties in single limb balance between limbs. Altered temporal parameters of gait have also been identified with increased time spent in double support phases and reduced time spent in single support (Dewar and Judge, 1980). This finding is similar to that found in elderly fallers by Wall *et al.*, (1991) who proposed that this gait profile was associated with a "cautious" gait pattern where balance was compromised.

Hemiplegic subjects have been shown to produce knee extension in early support in contrast to the normal feature of energy absorbing flexion (Richards and Knutsson, 1974; Knutsson and Richards, 1979; Lehmann *et al.*, 1987). In addition, ankle dorsiflexion has been found to be reduced during loading of the affected limb and plantarflexion decreased during push off at the termination of support (Giuliani, 1990). Murray *et al.*, (1966) has drawn attention to the reduced amplitude of lower limb movement during swing which is accentuated at higher walking speeds. Increased movement speed has also been proposed to exaggerate motor dysfunction (Sahrmann and Norton, 1977; Gowland, 1986). Deficiencies in motor control in hemiplegic patients have been described by Knutsson and Martenson, (1980) particularly when phase transitions are occurring in the affected lower limb. These difficulties occur when a transition movement from flexion to extension or extension to flexion is attempted. In particular, these phenomena occur during the initial act of loading the limb, at toe off and midswing as the extensors are activated to brake the flexion pattern of the swing. It is also known that hemiplegic subjects have difficulty performing rapid reciprocal movements as a result of antagonist cocontraction (Knutsson and Martenson, 1980).

From a kinetic perspective, hemiplegic subjects have been shown to have difficulty controlling force parameters particularly during weight acceptance on to the affected leg and in transferring weight to the affected limb (Giuliani, 1990). Studies of joint moments and ground reaction forces have shown that hemiplegic subjects differed from normal in terms of the magnitude of the forces and their profiles (Carlsöö *et al.*, 1974). Lack of motor control has also been demonstrated from the comparison of muscle recruitment patterns in the hemiplegic leg (Giuliani, 1990). This same author proposes that these problems may be due to alterations of the viscoelastic properties of muscle and tendon and poor regulation of motor unit activity. Similarly, atrophy of fast twitch muscle fibres and hypertrophy of slow twitch fibres have been found by Chokroverty *et al.*, (1976) which probably contribute to the disordered ambulatory function.

Electromyographic anomalies of the muscles of the lower limb in hemiplegic subjects have also been reported and consist of: less muscle activity in the hemiplegic limb; prolonged burst durations; tonic rather than phasic activity at gait transitions and; periods of peak muscle activity that differ with those seen in normal gait (Peat *et al.*, 1976; Knutsson and Richards, 1979). It has also been proposed that these findings are highly variable between subjects but are consistent within subjects (Knutsson and Richards, 1979). The electromyographic activity of selected, key muscle groups were studied by Knutsson and Richards, 1979). These researchers measured the activation patterns of six muscle groups, in a group of hemiplegic subjects, by monitoring the electromyographic activity during various components of the gait cycle. The muscles were the hip adductors and abductors, quadriceps, hamstrings, triceps surae and tibialis anterior. Based upon these patterns of activation, they described three classifications of hemiplegic gait. A Type I activation pattern was characterised by premature activation of the calf muscles which was combined with low levels of activity in the anterior tibial group. The other muscles in the affected lower limb demonstrated normal or near normal activation patterns. As a result of this, stretch was imposed upon the calf muscles during walking particularly during foot to floor contact which was proposed to contribute to hyperextension of the knee joint during support. A Type II activation pattern was characterised by an abolition or a marked decrease in electromyographic activity in two of the recorded muscles in the hemiparetic limb. This resulted in poor knee flexion during swing, foot/floor discrepancies and hyperextension during the support phase. The Type III activation pattern was marked by abnormal co-activation of several of the recorded muscles caused by high levels of spasticity and dominance of primitive synergistic motor activity. In terms of functional gait, patients with a Type II activation gait pattern walked at a velocity of 0.47 m/s (range 0.18 - 0.78) implying that this pattern may have been the most disabling if walking speed is considered. The subjects with a Type III pattern appeared less disabled than the Type II, walking at a velocity of 0.62 m/s (range 0.59 - 0.71). Subjects with a Type I pattern were the quickest walkers of the three groups ambulating at 0.72 m/s (range 0.53 - 1.03). Nonetheless all three groups walked slowly compared with normal. The authors also ranked the subjects further stating that those with Type I patterns had a slight problem while those with Type II patterns demonstrated paralysis as a major feature of the gait pattern.

A major characteristic of the hemiplegic gait pattern is motor asymmetry (Wall and Ashburn, 1979; Mizrahi *et al.*, 1982, Brandstater *et al.*, 1983; Wall and Turnbull, 1986). The magnitude of the asymmetry of the hemiplegic gait pattern has been proposed to be inversely related to the degree of recovery and positively related to walking speed (Carlsöö *et al.*, 1974: Dewar and Judge, 1980: Brandstater *et al.*, 1983). However, Harro and Guiliani (1987) found that asymmetries were invariant across walking speeds. Wall and Turnbull (1986), who studied both temporal and spatial asymmetries, also proposed subtypes of the hemiplegic gait pattern. This study, conducted on stabilised, residual stroke patients, identified a number of variations of the gait patterns based upon the location of temporal asymmetries. In some cases, patients demonstrated marked asymmetries in certain phases of the gait cycle more so than in others. In those with predominately braking double support (BDS) asymmetries, where the patient spent longer in this phase on the unaffected leg when compared with the affected leg, scrutiny of the videotapes of the walk for which data were collected, showed that, in many cases, this pattern was associated with the use of a walking aid or was a feature of an "interrupted" walk. This latter pattern was characterized by the patient completing one cycle of the gait pattern and then hesitating before continuing with the next cycle. Attention has also been drawn to this hemiplegic sub-pattern by Miller and Musa (1982). Similarly, the BDS asymmetries seemed to be associated with a lack of ability to transfer body weight through the affected leg. A balance discrepancy in the affected leg was proposed as a possible cause. The same asymmetry was produced when the patient appeared to have difficulty initiating the swing phase of the affected leg. Subjects spent an abnormally long time in this phase of the gait cycle while attempting to swing the affected leg forward. The initiation of the swing phase on the affected side appeared to be made difficult by muscle weakness or high levels of spasticity which resulted in the foot "sticking" to the floor at toe-off. Another type of double support asymmetry was also identified. In cases where the patient spent longer in the phase of BDS on the affected leg when compared to the unaffected lower limb, a hesitation was observed immediately following initial foot/floor contact before an attempt to transfer weight through the affected leg was made. This appeared to be associated with significant hypertonicity which had the effect of causing the foot to invert during weight bearing. Patients, therefore, were reluctant to complete this part of the gait cycle due to the potential for damage to the ankle which resulted in this type of asymmetry. In patients with single support asymmetries as the predominant feature, study of the videotapes revealed that these patients tended to rush through the single support phase on the affected side. This again, in most instances, appeared to be due to a reluctance to transfer weight through the affected limb possibly as a result of a balance deficit. An additional reason for this asymmetry was an abnormally long swing phase on the affected side which appeared to be caused by difficulties in shortening the hemiplegic leg either because of excess spasticity or apparent weakness. A recurring feature of these hemiplegic gait subtypes was a suspicion that normal balance, in the affected leg, appeared compromised.

It is likely that weakness of the affected lower limb and levels of spasticity as well as abnormal balance contribute to the asymmetrical nature of the gait pattern (Murray *et al.*, 1975; Wall and Ashburn, 1979; Friedman, 1990; Bohannon, 1991). However, although a major assumption used in the treatment of gait disorder following stroke, it is unclear as to how balance deficiency in the affected lower limb affects hemiplegic gait. As a result, the precise method of retraining balance with a view to improving gait in the rehabilitation setting remains obscure.

THE ADAPTABILITY OF HEMIPLEGIC GAIT

In a study conducted by Wall and Turnbull (1986), the self-selected, preferred speed of the hemiplegic sample studied was extremely slow with a mean velocity of 0.25 stat/s. Only 2 subjects out of 25 were able to reach velocities of 0.4 stat/s, a speed regarded as being in the slow range for normal elderly people (O'Brien *et al.*, 1983). This velocity decrement has potentially important functional implications. For example, many environmental factors, such as signals at cross-walk intersections, are geared towards a much faster walking speed (Hogan *et al.*, 1987). Further, it has been proposed that increases in movement velocity results in deterioration of motor performance (Gowland, 1986). Gait studies related to hemiplegia tend to examine gait at the subject's preferred walking speed. Few studies have examined the range of speeds of which hemiplegic subjects are capable and, therefore, little is known of the adaptability of hemiplegic gait. Although it has been repeatedly shown that hemiplegic subjects walk slowly, Giuliani (1990) has drawn attention to the fact that very little research has been conducted on the speed related changes in the hemiplegic gait patterns. One study which did examine different walking speeds was that conducted by Bohannon (1992). In an attempt to examine the functional implications of walking speed, he found a significant difference in velocity between a group of hemiplegic subjects "comfortable" walking speed and their "maximum safe speed". In addition, this study showed that there was a relationship between the "comfortable" and "maximum" speeds. However, this study investigated no gait parameters other than velocity. Harro and Guiliani (1987) found that the self-selected walking speeds of hemiplegic subjects were considerably slower than those of controls and reported deficiencies in the ability of hemiplegic subjects to increase walking speed compared to controls. This poverty of walking speed has been attributed to decreased joint amplitudes and step lengths as well as an inability to produce selective movement in the joints of the lower limb and poor balance (Perry, 1969; Brunnstrom, 1970). Attempts to walk fast, deviating from a "preferred" speed of walking, may result in an accentuation of the hemiplegic gait pattern because of the direct relationship between asymmetry and speed (Carlsöö *et al.*, 1974; Dewar and Judge, 1980; Brandstater *et al.*, 1983). This could result from increases in levels of spasticity with an associated increase in gait asymmetry as a result of effort, a phenomenon that Bobath (1978) has referred to as "associated reactions". This deterioration of performance, which remains to be systematically described in relation to hemiplegic gait, appears to be worthy of consideration as an important rehabilitation concern. However, at this time, minimal attention is given to addressing this velocity discrepancy and attempting to re-educate gait at different speeds. If deterioration of the walking pattern does occur at different speeds and gait symmetry is affected by this variable then there is a need to determine the range of speeds of which stroke sufferers are capable and to see if this is related to the efficiency of the balance performance of the patient.

In the interests of energy efficiency and function, it is likely that the hemiplegic patient will adopt a gait pattern which may not be cosmetically appealing and will thus appear abnormal (Olney, 1987; Bohannen, 1988). Over time, this pattern would likely become habituated. This "learned" abnormality has been alluded to by Carr and Shepherd (1983) and Turnbull and Wall (1989). Correction of perceived gait abnormalities is a fundamental part of any gait re-education programme, however, this approach is based upon the assumption that a hemiplegic gait pattern is correctable. There is a need to examine the degree to which the stroke patient can adapt their gait pattern to walk in a manner which appears more normal.

ASSESSMENT OF HEMIPLEGIC GAIT

The most commonly used method of gait assessment in physiotherapy is subjective and, therefore, has been shown to be heavily dependent upon the training, experience and personality of the the therapist (Goodkin and Diller, 1973). Many systems, however, have been used to objectively measure gait following stroke, thus overcoming the problems of reliability and variability. Electromyographic (EMG) techniques have been used to demonstrate abnormal muscle activity and activation patterns following stroke (Peat *et al.*, 1976; Perry *et al.*, 1978; Knutsson and Richards; 1979). Knutsson and Richards (1979) also measured angular displacements in the lower limbs to assist in the interpretation of

EMG data. Analysis of power patterns and work during hemiplegic gait have been used by Olney *et al.*, (1991) and several workers have studied the spatiotemporal kinematics of hemiplegic gait (Mizrahi *et al.*, 1982; Wall and Turnbull, 1987). However, Giuliani (1990) has drawn attention to the fact that there are few comprehensive studies which have examined the temporal/distance gait parameters in a homogeneous diagnostic group of stroke patients and that spatiotemporal gait parameters are of particular clinical value for assisting in the formulation of appropriate treatment aims and for documenting change over time.

Gait measurement systems have also been used in studies on hemiplegia to assess the effectiveness therapeutic exercise regimes (Böghardh and Richards, 1981; Wall and Turnbull, 1987), monitoring progress after treatment (Wall and Ashburn, 1979) and in assessing the effectiveness of biofeedback (Takebe and Basmajian, 1976; Krajnik *et al.*, 1981; Seeger *et al.*, 1981). These studies have all clearly demonstrated the asymmetrical nature of the hemiplegic gait pattern. These asymmetries suggest a "favouring" of the hemiplegic leg. In studies which have measured forces beneath the feet, hemiplegic subjects have been shown to bear a greater percentage of body weight on the sound limb when compared to the affected leg (Murray *et al.*, 1975; Seeger *et al.*, 1981; Dettman *et al.*, 1987; Gruendel, 1992). During gait, the support phase of the gait cycle on the affected leg becomes rushed and is consequently shorter than the corresponding phase on the sound limb. This results in an asymmetrical gait pattern (Wall and Turnbull, 1987). In an investigation of this phenomenon, Dettman *et al.*, (1987) found that the hemiplegic group bore weight unevenly (36% of body weight through the hemiplegic leg, compared with 50 % in the normal group). These findings are similar to those found by Caldwell *et al.*, (1986) who showed that, in quiet standing, hemiplegic subjects transferred much smaller percentages of total body weight (35.8%) through the affected leg when compared to both young and elderly normals. This unequal weight bearing was also described by Arcan *et al.*, (1977) who found that hemiplegic subjects bore a greater percentage of weight on the sound limb than the affected leg. Unequal weight bearing has also been demonstrated in the gait of hemiplegic, cerebral-palsied children (Seeger *et al.*, 1981). In addition, there is some evidence from the same study that an improvement in the temporal aspects of the gait cycle is associated with improved ability to load the hemiplegic leg.

The asymmetrical nature of hemiplegic gait has been measured in a variety of ways using the temporal aspects of gait. Wall and Ashburn (1979) selected the percentage of the stride that each foot was in contact with the ground to reflect asymmetry. Step time as a percentage of stride was used by Dewar and Judge (1980) to demonstrate the asymmetrical nature of the hemiplegic gait pattern while Dettmann *et al.*, (1987) compared the duration of swing time between the hemiplegic and unaffected limbs (swing ratio) to obtain a measure of gait asymmetry. Brandstater *et al.*, (1983) also used comparison of swing times to reflect asymmetry after finding that this measure appeared to reflect functional recovery. Wall and Turnbull (1986) used an asymmetry ratio (AR) to determine the magnitude and the location of support phase asymmetries within the gait cycle in an attempt to assist with clinical decision making. This technique identified the location of the asymmetries, so that specific remedial strategies could be identified which were designed to reduce the asymmetry. In this study, the AR for a given phase was calculated by dividing the duration of the phase on the unaffected side by the duration of the same phase on the affected, hemiplegic side. These authors attempted to describe the magnitude of asymmetry by comparing the duration of a given component of the gait cycle of one leg with the same value for the other leg. However, they drew attention to the fact that simply dividing the value for the unaffected leg by that for the affected limb could result in inaccurate conclusions as illustrated by the following scenario. If the value for the unaffected limb was 2 and that for the unaffected limb was 1, then an asymmetry ratio of 2 resulted. However, if the value for the affected leg was 2 and that for the unaffected limb was 1 then the asymmetry ratio would be calculated as 0.5, implying that this latter AR was one quarter of the magnitude than the previous example despite the fact that the asymmetries were the same. These authors, therefore, proposed that the greater value be divided by the smaller and in instances where the affected limb value was greater than the unaffected, the reciprocal would be used and assigned a negative value. This technique, therefore, would indicate the magnitude and the location of the asymmetry. Thus, a measure of the magnitude of the asymmetry was derived. In a perfectly symmetrical walking pattern, this equation would yield a value of 1.

GAIT RE-EDUCATION FOLLOWING STROKE

From a descriptive viewpoint, re-education of gait following stroke is somewhat difficult to isolate from procedures directed towards other motor functions. It has been proposed that techniques which will ultimately impact upon the gait performance of the stroke patient commence immediately post stroke when the patient is still on bed rest (Todd, 1974; Todd and Davies, 1982; Carr and Shepherd, 1983). Procedures derived from Bobath's work are being extensively utilised to minimize the development of excessive muscle tone in early stroke sufferers (Todd, 1974; Dardier, 1983; Turnbull and Bell, 1985). Encouragement to move the affected motor components is also commenced early in attempt to control the future readjustment of the CNS in reaction to the disordered function caused by the stroke (Carr and Shepherd, 1983). Amongst these movements are procedures which will later influence the gait pattern. For example, bridging is a movement procedure performed in lying with both hips and knees flexed so that the soles of the feet are flat on the supporting surface. The subject then extends the hip joints so that the pelvis is raised up from the supporting surface and the subject bears weight through the shoulders and feet. Early, performance of bridging, during which the patient is helped and encouraged to bear weight equally through both lower limbs in a supine position which inhibits spasticity, has been advocated as an essential strategy (Turnbull and Bell, 1985). The long term purpose of this procedure is to encourage the patient to load the affected lower limb in a manner which mimics the support phase of the gait cycle and is consistent with the normal, developmental sequence. Similarly, attempts are made by the patient, again as early as possible following the stroke, to move the individual lower limb joints in a selective manner (placings) and not as a total lower limb primitive reflex. This is done as preparation for the swing phase of the gait cycle which is a skilled, functional and complex movement (Bobath, 1978). Again, consistent with the developmental sequence, control is reeducated from proximal to distal commencing with control of the hip joint, then the knee and finally the foot complex (ankle and mid tarsal joints). Control is gained first at one joint (the proximal joint) then two and finally all three (Lane, 1987). However, the target movement repertoire is characterised by the ability of each joint to move selectively and not as a total primitive synergy.

As the patient improves, transfers, such as from bed to chair, are carried out in a manner which promotes weight bearing through the affected, hemiplegic leg (Davies, 1985). As the patient progresses, more specific exercises are added to the patient's treatment regime in lying, sitting, standing and then while walking. Utilizing a Bobathian framework, Lane (1978) has developed and described a series of therapeutic exercises designed to promote reduction of movement asymmetry caused by hemiplegia. It is proposed that this objective will be reached through the reduction of spasticity, which is thought to interfere with the production of normal movement patterns, improve the ability of the patient to control the affected limbs in space and promote weight transfer through the affected lower limb through the retraining of normal balance reactions. The underlying theory of the provision of these weight transfer exercises is that the hemiplegic patient is unable to load the hemiplegic lower limb while walking. This is proposed to result from the patient having lost the automatic motor balance mechanisms in the affected leg. As a result, the patient cannot adequately place body weight through the affected leg and, during walking, this feature is accompanied by a reduction in single support time on the hemiplegic leg. This deficiency gives rise to gait asymmetries particularly in single support (Wall and Turnbull, 1986). The ultimate objective of these exercises is to improve this support phase gait asymmetry by improving balance on the affected leg. These balance reactions have been described as automatic, sophisticated adaptations of muscle activity carried out subconsciously by the individual in order to maintain the CP of the body over the base of support of the individual or to establish a new base within which the CP is located (Bobath, 1978). However, a study by Wall and Turnbull (1987) questioned the ability of stroke patients to load the hemiplegic leg. even when deliberately attempted.

Although the strict application of the neurodevelopmental sequence to progress the motor recovery of the adult neurological patient has been questioned recently by Van Sant (1991), adherence to developmental principles as a framework to guide motor rehabilitation remains widely practiced (Davies, 1985). For example, attempts to improve balance are common first in lying, sitting and then in standing as precursors to walking as is the case in the developing child. Further, stability of proximally placed joints are believed to precede the development of distal, dextrous motor activities (Davies, 1985).

Another tenet of the neurodevelopmental treatment (NDT) method, the current North American label for procedures derived from the Bobath approach, is the re-establishment of normal musculoskeletal alignment. The stroke patient has a tendency to develop a flexed spine in that the normal secondary curves at the cervical and lumbar region become diminished (Bobath, 1978). The resultant postural abnormality creates compensatory changes at the hip and knee joints and interferes with the production of normal movement (Atkinson, 1982). Rectification of this abnormal alignment is, therefore, required before normal movement patterns can be expected. Another typical example of this principle as it affects gait pertains to the effect of the retracted hemiplegic pelvis and has been described earlier. It has been proposed that hypertonicity of the lateral trunk flexors raise the pelvis on the hemiplegic side of the body. Similarly, the hip joint maintains a posture of slight flexion which is part of the primitive motor pattern (Rothstein *et al.*, 1991). To compensate for this alignment disturbance, it is common for many hemiplegic patients to fully extend the knee joint particularly during initial weight bearing. The hemiplegic foot therefore is maintained in a plantarflexed position during weight bearing and the lower leg does not rotate forward on the fixed foot in the early component of the support phase. Böghardh and Richards (1974), Davies (1985) and Turnbull and Wall (1989) have proposed that correcting the pelvic postural abnormality is likely to improve the alignment of the whole lower limb thus improving the quality of the gait pattern. Thus, re-educating more normal postural alignment is a commonly utilised method of improving hemiplegic gait.

Biofeedback has also been proposed by a number of authors as a method of improving hemiplegic gait. Some workers have supplied patients with auditory signals amplified from the electromyographic (EMG) activity of inactive muscle groups with a view to improving the magnitude of the voluntary contraction (Basmajian *et al.*, 1975; Binder *et al.*, 1981; Wolf and Binder-Macleod, 1983). Following training, subjects then attempt to incorporate this improved function into the overall gait pattern. Others have used EMG biofeedback to reduce excessive tone in spastic muscle groups (Woolley-Hart *et al.*, 1978). In addition to EMG biofeedback, attempts have been made to use positional feedback provided by transduced goniometers to improve the gait pattern by assisting the patient to attain a target joint position (Koheil and Mandel, 1980). More recently, devices have been developed to provide feedback to the patient concerning limb loading on the hemiplegic leg (De Weerdt *et al.*, 1988; Turnbull and Wall, 1989). Biofeedback techniques have also recently been enhanced by the development of computer technology with Olney *et al.*, (1989) describing computer assisted visual and auditory feedback treatments of stroke and Mandel *et al.*, (1990) using computer enhanced, electromyographic and positional biofeedback in hemiplegic gait re-education.

A recent trend in the treatment of hemiplegic gait was as a direct result of the trend in physical therapy to apply a psychomotor learning perspective to treatment decision making. Rather than training components of the gait pattern (part practice) and later combining these parts into the whole pattern, it was proposed that whole practice of the gait pattern may be more beneficial . A study by Waagfjörd *et al.*,(1990) in which a treadmill was used to retrain gait following stroke appeared to show that this method of treatment was associated with some success. Recently, Malouin *et al.*, (1992) described a task-specific gait re-education programme for stroke patients. Richards *et al.*, (1993) reported encouraging results from this approach conducted on 27 hemiplegic subjects, all of whom had middle cerebral artery lesions confirmed by computed axial tomography. The early intervention, gait-specific training was provided by means of a treadmill and was found to significantly improve walking velocity compared with more traditional treatment approaches. The rationale for introducing such a task-specific procedure was based on the work of Winstein *et al.*, (1989) who demonstrated that although standing balance and locomotor control may be highly interrelated, improvement in standing balance asymmetry did not necessarily lead to reduced asymmetry in walking. These authors concluded that gait may be more effectively rehabilitated if the whole task was practiced during treatment.

At present, a number of investigators are studying the efficacy of a gait re-education technique in which whole gait practice is undertaken but with a percentage of body weight supported by an overhead harness (Barbeau *et al.*, 1988). The rationale underlying this technique is that deficiencies in weight bearing through the hemiplegic limb following stroke cause gait deviations in other parts of the gait cycle. By reducing the weight bearing demand in the early stages post stroke by supporting a percentage of body weight externally, it has been proposed that gait re-education will be enhanced (Pillar *et al.*, 1991).

It can, therefore, be stated that although gait re-education following stroke has been widely acknowledged as being of great importance, solutions to its successful rehabilitation are being pursued vigorously. It is interesting to note from the literature that many of the treatment procedures proposed have been based upon empirical observation. Rigorous scientific testing of these clinically applied techniques, which are essentially hypotheses, has not taken place in a systematic manner. There is a need, therefore, to subject some of the elements of these treatment approaches to carefully controlled experimentation. This current study is designed to contribute to the attainment of that objective.

BALANCE

Balance reactions are discrete, automatic, motor responses which serve to maintain the CP of the individual within the base of support. Centre of pressure, in this context, is defined as the centre of distribution of the total force of the body applied to the supporting surface (Murray *et al.*, 1975). In the human, balance is maintained by means of the interaction of a group of discrete, automatic muscle contractions referred to as balance reactions. These reactions appear around six months of age and continue to develop until the individual is capable of producing a wide variety of functional movement patterns (Fiorentino, 1977). Balance is developed in sitting, kneeling and standing with each developmental milestone being characterised by greater independence (Thelen *et al.*, 1989). Walking begins around the twelfth month and coincides with the appearance and perfecting of balance reactions in the lower limbs (Banus, 1971). Initially, early gait is characterised by frequent falls.

However, as balance becomes more sophisticated, these falls decrease until, under normal circumstances, they are eliminated. However, the incidence of falls increases again with advancing age (Hinchcliffe, 1983).

The maintenance of balance requires the integrated interaction of the vestibular, somatosensory and visual systems. The vestibular system is thought to provide information concerning the position of the head in relation to gravity and the motion of the head through the detection of linear and angular accelerations of the head (Nashner *et al.*, 1982). The vestibular system is also thought to resolve situations of sensory conflict (Black *et al.*, 1983). The somatosensory system, which is comprised of proprioceptors, cutaneous receptors and joint receptors, provides information concerning the movement of the body segments in relation to each other. The visual system orientates the position of the body in relation to the environment (Berg, 1989). Disturbances of balance, when detected, are corrected by discrete automatic motor strategies which can be classified into ankle strategies, if the displacement is minor and hip strategies, if the displacement is more vigorous (Berg, 1989). Combinations of these strategies can also be used concurrently (Horak and Nashner, 1986). Extreme perturbations can be counteracted by altering the position of the base of support in a stepping response (Berg, 1989). Balance responses, which take approximately 100 ms to be initiated, are utilized to maintain the position of the body, to permit functional movement through postural adjustments and to resist the impact of external forces (Berg, 1989). In addition, displacements necessitating a postural response can be self initiated or can be imposed by environmental events (Hinchcliffe, 1983).

Isaacs (1983) has described balance mechanisms as consisting of three components. First, the individual has to adequately detect that a displacement is taking place. This detection would include information concerning the magnitude, speed and direction of the displacement. Secondly, the detected sensory information is processed by the central nervous system and an appropriate motor response is planned and formulated. Finally, a "custom-made" motor strategy is effected which serves to preserve balance. Following detection, several parts of the central nervous system process this information and the resultant motor response occurs as a result of extensive motor activity and feedback systems

(Granit, 1972). Hinchcliffe (1983) has stated that should the mechanisms which maintain upright posture and normal gait be challenged, such as by some external force, rescue reactions are then brought into play. The same author has identified "sway", where the individual opposes the displacing force; "staggering", which includes a series of steps or hops; and "sweeping" reactions where the limbs are used as inertial paddles. Lane (1969) has described these same balance mechanisms under the classifications of "propping", where the individual resists displacement of the CP; "stepping", where a new base of support is established to "catch-up" with the displaced CP; and "magnet" responses, where the CP is maintained within the base by the re-arrangement of body segments which in extreme cases are extensive. All of such activities are the result of automatic, subcortical motor mechanisms designed to control and maintain the line of gravity of the individual within the base of support.

If these rescue reactions should prove inadequate, a further series of automatic motor responses are brought into play (Hinchcliffe, 1983). These responses, which have been referred to as "fall-breaking" reactions, serve to minimize injury from the subsequent fall and include the use of the upper limbs either to protect the face and head or to break the fall (Hinchcliffe, 1983).

It must be re-emphasised that these balance reactions are carried out at an automatic, subcortical level of the central nervous system and place little demand on the attention of the individual under normal circumstances. However, they are not simply reflex motor behaviours and can be cognitively modified at any time should this become necessary.

From a motor developmental viewpoint, improvements in the ability of a child to balance are paralleled by the appearance of a broader functional repertoire. All normal children acquire the ability to balance, first in sitting, later in standing and then while walking. Much of the improvement in balance in the young child results from the maturation of the CNS (Odenrick and Sandstedt, 1984). However, these new motor competencies appear, at least to some degree, to be acquired as opposed to being entirely innate (Turnbull and Wall, 1989). The young child will attempt a movement which will displace its centre of mass. Frequently, errors will appear and the child will fall. Gradually, however, performance improves with greater feats of balance being achieved. This trial and error process, which appears to be consistent and thorough, contains many of the characteristics of the learning of a novel motor skill.

Balance is an important ingredient of functional movement particularly transfers from one posture to another and while standing and walking (Berg, 1989). When balance becomes disordered, functional inadequacies arise including the appearance of abnormal gait. In fact, these balance disorders have given rise to descriptive terms of the resultant gait disturbance. For example, in multiple sclerosis of the cerebellar variety, the term "ataxic gait pattern" has been applied and in Parkinson's disease, descriptions such as "festinating gait" and "marche a petit pas" arise In older populations, idiopathic gait disorder of the elderly (IGDE) is probably caused by diminishing balance ability (Hogan *et al.*, 1987). The association of falls with advancing age may, to some extent, be due to a deterioration in the complex systems contributing to the maintenance of balance (Isaacs, 1983). Slowing of reaction time and speed of response, which may delay the implementation of "saving" reactions, have been found in older populations (Botwinick, 1973). However, it should be noted that active elderly may not demonstrate the same degree of slowing (Agruso, 1978).

A number of medical conditions are known to affect balance which can give rise to unsteadiness and, in extreme circumstances, falls. There also appears to be a relationship between abnormal balance and the pattern of gait adopted by the patient (Wall *et al.*, 1991). In attempting to remedy these types of balance deficits, it is important that a number of medical conditions known to result in unstable gait are eliminated as the underlying cause. Examples of these medical conditions are neurological disorders, vestibular or inner ear disorders, severe visual disability, significant peripheral sensory loss, severe degenerative osteoarthritis, cognitive disturbance including memory loss, marked skeletal deformity, postural hypotension, chronic alcohol abuse, advanced cervical myelopathy, normal pressure hydrocephalus and multiple sensory disorder (Hogan *et al.*, 1987).

ASSESSMENT OF BALANCE

The maintenance of the upright standing posture is a dynamic and continuous process characterised by short bursts of automatic muscle activity particularly around the ankle joints (Era and Heikkinen, 1985). This activity gives rise to discrete movements which are commonly referred to as postural sway. Postural sway has been defined by Sheldon (1963) as the constant small corrective deviation from the vertical when standing upright. This system has been referred to as the inverted pendulum model (Winter, 1992). Damage to the CNS can interfere with this mechanism giving rise to changes of the normal characteristics of this sway (Murray *et al.*, 1975; Jarman and Lord, 1981, Dettman *et al.*, 1987). Rogers (1990) has differentiated between the terms "balance" and "posture" because of the tendency for the terms to be used interchangeably. He defined balance as: "the maintenance of the projected location or trajectory of the body centre of mass within the area defined by the base of support". Posture was defined as: "the relative orientation of the body segments to each other during quasi static (when minimal accelerations are occurring in the system) and dynamic conditions".

Clinically, postural sway, in an anteroposterior direction, is frequently assessed subjectively by the application of the Romberg test, first described in the literature in 1853 (Romberg, 1853). This test assesses the sway of the subject in quiet standing with eyes open and closed (Odenrick and Sandstedt, 1984). Both conditions are applied during this test to examine the relative contributions of the proprioceptive system and the visual systems. In addition to being highly subjective, this procedure is known to be somewhat insensitive despite its widespread use (Van Allen, 1976). Briggs *et al.*, (1989) have modified this traditional test by assessing sway with one foot directly in front of the other. This has been termed the "Sharpened" or the "Tandem" Romberg test. This test destabilizes the patient laterally and thus increases the lateral sway of the subject. A number of other workers (Potvin *et al.*, 1980; Stones and Kozma, 1987) have further extended this test by having the subject stand on one leg during testing although Bohannon *et al.*, (1984) have demonstrated that normal elderly subjects may have extreme difficulty in performing this test. Shumway-Cook and Horak (1986) have developed a series of tests, based on the work of Nashner, during which the relative contributions of the proprioceptive, visual and vestibular systems are assessed in maintaining upright balance. Six test positions are scored on a four-point scale which range from minimal sway (1) through mild sway (2), moderate sway (3) and finally a fall (4). The subject is first tested in a quiet standing position and then with vision occluded to assess the relative contribution of vision. The subject is then fitted with a device referred to as a "visual conflict dome" (VCD). The VCD is placed around the head of the subject and has non parallel vertical and horizontal lines painted on the inside in such a way as to provide the subject with a consistent visual image irrespective of the position of the head but which provides no vertical or horizontal terms of reference to the subject. The purpose of this test condition is to examine the contribution of the vestibular system in maintaining balance. The fourth, fifth and sixth test positions repeat the previous three except that the subject stands on a foam rubber mat which minimises the contribution of the exteroceptive input thus allowing assessment of the role of these variables. The application of this subjective test has the advantage of being able to provide investigators with an indication of the probable location of any balance defect by isolating, in sequence, each of the three factors known to regulate standing balance.

Global tests of subject balance performance have also been developed. In this testing paradigm, a number of activities of daily living, the successful completion of which are dependent on functional balance, are scored. Examples of such testing formats are the "Get Up and Go Test" proposed by Mathias *et al.*, (1986), Balance Coding suggested by Gabell and Simons (1982) and the functional performance tests developed by Berg *et al.*, (1989).

A number of objective techniques have been developed and have been used over the years to attempt to measure balance performance. Wright (1971) measured postural sway by means of an ataxia-meter which was able to monitor the excursion of a point on the body in the sagittal plane. Sway was expressed as the total angular movement summed regardless of sign. This device was cheap, simple to apply and has been used in later studies by Overstall *et al.*, (1977) and Brocklehurst *et al.*, (1982). Fernie *et al.*, (1982) adapted this device to measure sway in both the frontal and sagittal planes.

Ring *et al.*, (1988) used a "visual push" to assess balance in elderly subjects. This ingenious system measured postural sway on a force platform while the subject was presented with an image on a screen which provided the illusion that the image on the screen was moving both towards and away from the subject. The resultant effect on postural sway was used as a measure of postural control.

The maximum load test was used by Lee at al (1988). This procedure called for a harness to be placed around the waist of the subject to which was attached weights via a pulley system thus displacing the balance of the subject in both the frontal and sagittal planes but at different times. The balance performance was assessed as the amount of weight expressed as a percentage of body weight, which the subject could tolerate without losing balance. Similarly, the Postural Stress Test used a single given weight calculated as a percentage of the body weight of the subject, which was dropped from a height, to produce an unexpected posterior displacement at the pelvis while the subject was being videotaped. The videotape was then scored to determine the type of postural response evoked (Wolfson *et al.*, 1986). The Sternal Thrust test was used by Wild *et al.*, (1981) and Weiner *et al.*, (1984). This test consisted of a "modest thrust" force being applied by the tester to the sternum of the subject so that the resulting displacement was posterior in direction. The postural response of the subject was then scored as; "normal", if the subject took no steps or one backwards; "moderately impaired", if two to three steps of retropulsion was observed; and "severely impaired" if four or more steps or a fall occurred.

The use of force platforms to obtain information concerning postural control and balance has been known for some time and is widely used in balance-related research (Patla *et al.*, 1990). Measures of postural sway, using force platforms, have been used to demonstrate various balance characteristics of normal young (Riach and Hayes, 1987) and elderly subjects (Hasen *et al.*, 1990). Similarly, the performance of people with such locomotor disorders as lower limb amputation (Fernie and Holliday, 1978), Alzheimer's Disease (Visser, 1983), cerebellar disease (Dichgans and Mauritz, 1983) and elderly fallers (Nayak *et al.*, 1982) have been studied. In the study of hemiplegic balance, Murray *et al.*, (1975) and Dettmann *et al.*, (1987) used force platforms to measure postural sway and its consistency in stroke patients.

More recently, the validity of using only postural sway has been challenged as a method of reaching conclusions concerning balance (Berg *et al.*, 1989; Patla *et al.*, 1990). These authors propose that an increase in postural sway need not necessarily be associated with inferior balance. Patla and his co-workers use the analogy of placing a mannequin on a force platform. The absence of postural sway does not mean that the mannequin has efficient balance. Thus, the paradigm of studying balance in response to a displacing stimulus has recently evolved and the use of moveable platforms, first proposed by Nashner (1976), have become more common in balance research. This research model measures the postural reaction to sudden unexpected perturbations, created by movement of the supporting surface of the subject. The excursion of the CP of the subject and/or the electromyographic activity of various postural muscles and their sequence of recruitment has thus been measured (Badke and Duncan, 1983: DiFabio *et al.*, 1986).

In addition to the force platforms which are commonly used to measure balance in the research setting, a number of devices designed to yield clinical assessments have recently been developed. One such device is the Equitest™ System (Neurocom International Inc., Clackamas, Oregon). This sophisticated clinical device was built with the testing of vestibular dysfunction as a primary objective. The subjective test, previously described, and designed by Shumaway-Cook and Horak (1986) was developed based upon the protocol of the Equitest™ system. As with the subjective test, the protocol for this machine sequentially isolates each of the three subsystems which contribute to balance thus pinpointing the source of a particular balance disorder. A critical component of the operation of this device is that the platform upon which the subject stands can move at the same frequency as the postural sway of the subject thus eliminating input from the ankle proprioceptors. In addition, the frame which the subject faces can also be sway-referenced thus interfering with normal visual input. The supporting surface is moveable and can provide the subject with precise perturbations in a variety of directions. Characteristics of the location, variability and movement of the CP of the subject can be measured under any of the test conditions available.

The Chattecx BalanceSystem™ (Chattanooga, Tennessee) is another of these recently developed clinical devices designed to measure balance. Like the Equitest™ system, this device provides information concerning the location, variability and movement of the CP of subjects who stand on two transduced footplates. In addition, this equipment has a training mode which provides subjects with visual feedback concerning location of CP. Two recent studies have shown this device to be credible and valid as a means of measuring balance performance (Dickstein and Dvir, 1993; Grabiner *et al.*, 1993).

HEMIPLEGIC BALANCE

Balance studies examining hemiplegic subjects have shown increased postural sway, decreased area of stability and uneven weight bearing in quiet standing with a decreased percentage of body weight transferred through the hemiplegic leg (Murray *et al.*, 1975; Arcan *et al.*, 1977; Caldwell *et al.*, 1986; Dettman *et al.*, 1987; Shumway-Cook *et al.*, 1988; Gruendal, 1992).

Murray *et al.*, (1975), compared normative data with those of a hemiparetic subject and found that there were marked differences in performance. The CP of the hemiplegic subject was displaced away from the hemiplegic side indicating reduced weight supporting activity through the affected leg. In addition, the total excursion of the CP was ten times that of the normal subjects indicating increased postural sway (decreased postural steadiness). Further, the area of stability over which the subjects could control the CP was significantly reduced in the hemiplegic subject compared with normals. Dettman *et al.*, (1987) used a force platform to quantify the area of stability over which weight could be shifted and maintained and the steadiness of the CP during these shifts in normal males. Each subject was required to shift weight as far as was comfortably possible to the anterior, posterior, left and right. Postural steadiness was considered as a measure of the total excursion of the CP during unsupported standing for a period of 30 s. Area of stability was derived by measuring the area circumscribed by the mean points reached by the CP of subjects in each of the positions to which the subject was required to shift weight. This study found that the postural steadiness and the area of stability in hemiplegic males was substantially less than that of normal men and was displaced towards the non-hemiplegic side. In addition, the hemiplegic group bore a smaller percentage of body weight through the non hemiplegic leg (36% of body weight through the hemiplegic leg, compared with 50 % in the normal group). These findings are similar to those found by Caldwell *et al.*, (1986) and Shumway-Cook *et al.*, (1988). Dettman *et al.*, (1987) also identified relationships between the balance measures of postural steadiness, ability to shift weight forward and area of stability with certain gait measures were also detected although these relationships, though significant, were not particularly strong. These gait parameters were speed, measured as m/s, stride length, measured in cm, paretic step length, also measured in cm, and swing ratio.

Badke and Duncan (1983) studied the electromyographic activity in key postural muscles in a group of hemiplegic subjects. Perturbation of balance was achieved by sudden displacement of the surface upon which subjects stood. Compared with a group of younger controls, the hemiplegic subjects demonstrated longer and more variable response latencies and a tendency to produce co-contraction of all muscles being recorded. In addition patterns of recruitment were abnormal and did not display the normal, distal to proximal sequencing. DiFabio *et al.*, (1986) found that hemiplegic subjects relied upon the postural responses in the unaffected leg when perturbed in contrast to controls who showed no difference between the lower limbs. However, these balance findings appear to be independent of weight bearing. Dickstein *et al.*, (1989) detected similar EMG abnormalities despite placing the hemiplegic leg on a step which significantly increased loading of the affected leg. Thus, the asymmetry of motor function, described earlier in relation to walking, was present in balance testing.

Wolfson *et al.*, (1986) found that balance in hemiplegic subjects was more easily disturbed when a sudden directional, external force was applied. In particular, the hemiplegics performed worst when exposed to a force directed posteriorly. A similar test was used by Wing *et al.*, (1993). In this study, a sudden force was applied laterally to the hips of hemiplegic subjects and suddenly removed. This balance disturbance caused the hemiplegic subjects to sway more and they took longer to stabilise after the application and removal of the force than controls. In addition, the hemiplegic sample had more difficulty recovering from the "release" than the initial application of the force.

Rogers *et al.*, (1993) moved away from the perturbation model and examined the dynamic characteristics of balance while a group of hemiplegic subjects raised one leg while standing on a force platform. The test was then repeated with the other leg bearing weight while the contralateral leg was raised. The procedure was designed to study the transition to single support and swing characteristic of gait. It was found that the dynamic weight transfer on the affected leg was deficient compared to the normal leg.

Ample evidence has been produced to demonstrate that the balance of the hemiplegic subject is abnormal. A similar asymmetry of function exists in balance as occurs in gait. However, the nature of this relationship is not clear. Although Dettmann's study was useful in drawing attention to the reduced balance performance by the hemiplegic group and that a relationship existed between balance and certain gait parameters, the results raised a number of additional questions. These data were obtained from a clearly heterogeneous stroke sample. The time post-stroke was between 2 months and 11 years and the gait kinematics corrected for the height of each subject were not addressed. Similarly, the performance of the subjects walking at a range of speeds was not included. Only preferred walking speed was measured which provides no information concerning the speeds of which the subjects were capable. The protocol of having the subjects produce postural responses would seem to provide information as to the dynamic balance capabilities of the subject and, thus, may be more reflective of the balance strategies required when walking than simply measuring postural sway in quiet standing. In addition, this framework does not invite danger to the subjects as is the case with the moveable platforms. Further, harnessing the subjects for safety, as is required when externally imposed perturbations are used, was not necessary. It could be argued that harnessing procedures may lead to the generation of misleading data concerning balance. Another improvement which could be suggested pertains to the relevance of the balance tests for gait. The research protocol used by both Murray *et al.*, (1975) and Dettman *et al.*, (1987) placed the feet in the same plane a comfortable distance apart. The types of weight shifts required were cardinal plane movements (anterior, posterior, lateral to the left and lateral to the right). These movement strategies did not appear to resemble any part of the the gait cycle. By placing the subject's feet in a position which mimics a step in the gait cycle (one foot placed anteriorly and the other posteriorly) and studying the abilities of the subjects to bear weight in these positions first over the rear foot and then over the front foot, which resembles the transfer of weight required while walking, more relevant information concerning balance and its relationship to gait may be generated. In other balance studies, a variety of foot positions have been used when balance was being tested. Preset foot positions were used by Tropp *et al.*, (1984) and Mahar *et al.*, (1985) while other researchers allowed the subjects to assume a natural standing posture (Murray *et al.*, 1975, Harris *et al.*, 1982). The protocol with the feet positioned diagonally was studied by Kirby *et al.*, (1987) in a group of normal subjects in quiet standing and showed that postural sway increased when the feet were placed in this position and that the CP was situated toward the rear placed foot. However, no weight shifts were conducted in this study. Goldie *et al.*, (1990), however, used a "step" position in which hemiplegic subjects were required to stand with the hemiplegic leg forward. The subject was then requested to load that limb while a tester timed the period that the subject could maintain the posture. The protocol of testing balance with the feet placed in positions resembling the double support phases of the gait pattern may, therefore, be more relevant to gait than the more traditional models of foot placement.

BALANCE RE-EDUCATION FOLLOWING STROKE

In the clinical setting, re-education of balance in the adult is achieved by displacing the centre of mass while the patient attempts to bring about an appropriate postural correction. Through repetition with feedback and progression of these techniques, it is believed that the patient will register a subsequent increase in functional ability. These treatments have been carried out using weight shifting exercises (Bobath, 1978), biofeedback (Turnbull and Wall, 1989) and exercises on an unstable platform (Davies, 1986). It would appear, however, that not much is known about the capability of people, particularly the elderly, to improve balance in the rehabilitation environment. However, among pathological groups of individuals, balance has been shown to improve with practice. For example, Hocherman *et al.* (1984) demonstrated that subjects, who had recently sustained a stroke, improved the ability to balance on a moving platform with training. However, in that study, the possibility that this improvement may have resulted from spontaneous recovery of CNS function rather than an improvement in balance as a result of training cannot be discounted. Although, balance in the normal individual is an automatic activity, rehabilitation requires that, initially, the appropriate movement be voluntarily produced. With practice, the skill becomes more automatic and, as a result, can be integrated into functional activities. Wall and Turnbull (1987) reported that they were suspicious that hemiplegic subjects could not consciously load the hemiplegic leg. However, Bohannon and Tinti-Wald (1991) and Engardt and Olsson (1992) reported studies which suggested that hemiplegic subjects could voluntarily load the hemiplegic leg, when requested, during functional activities.

Re-education of balance, therefore, particularly in standing, is a common physiotherapeutic procedure and is a central component of the functional rehabilitation in many conditions including stroke and has been proposed to be a fundamental prerequisite to the restoration of a normal gait pattern (Wall *et al.*, 1991).

In summary, disability as a result of stroke is, and will continue to be, a significant health care problem. Rehabilitation of those who survive is of critical importance particularly the re-education of a functional gait pattern. Many procedures are in place to attempt to reach this objective but many remain unproven from a research viewpoint. The recent trends of improving balance as a means of enhancing the quality of the gait pattern appears to be a promising direction but more requires to be known about the relationship between those two variables in the stroke survivor. This is particularly true when proposing effective exercise procedures through which the patients will attempt to acquire functional gait.

In the light of this study, the literature review provided a number of critical pieces of information which assisted in the generation of a research design to maximise the likelihood of answering the questions posed.

1. Although attempts were made to render the study group homogeneous, this has been shown to be a difficult objective.

2. No attempt had been made in the past to study the range of walking speeds of which hemiplegic subjects are capable. This is an important consideration because of its functional implications and because it is known that people with questionable balance walk slowly.

3. Many of the research paradigms which measure postural control perturb balance mechanically by moving the surface on which the subject stands or by subjecting the subject to an unexpected external force. In older people or in abnormal populations with compromised balance this is a questionable practice and is unrealistic in terms of function except in accident situations. Active movement of the CP by the subject appeared a much more realistic option and is infinitely safer. Similarly, using a patient-controlled perturbation makes it unnecessary to harness the subject, a practice which could influence the balance performance of the subject.

4. In the vast majority of studies which have examined postural control, balance has been perturbed with the feet situated in the same plane. This does not represent the position of the feet while moving in a functional capacity particularly during ambulation. As a result it

appeared logical to examine balance while the feet of each subject were in positions which were representative of the gait cycle.

5. It is well known that many of the standardised balance measurement techniques are static in nature and thus their relevance to dynamic function such as gait can be questioned. Therefore, it appeared important to include the measurement of a dynamic element in the testing procedure even although this involved the maintenance of extreme balance postures.

6. Rehabilitation of the stroke survivor includes high percentages of time devoted to gait re-education. Much of this includes practicing components of the gait cycle particularly the repetition of the patient transferring weight through the hemiplegic leg during the support phase of the gait cycle. There is a need to test this assumption by investigating the relationship between gait performance and the ability of the subject to transfer weight in a variety of ways through the hemiplegic leg.

7. Normative data concerning balance performance and the range of speeds at which older people can walk remains incomplete. To make definitive statements about the hemiplegic sample, it was necessary to compare performance with that of a control group.

Consideration of these elements influenced the design of the present study.

CHAPTER 3

EXPERIMENTAL METHODS AND PROCEDURES

SUBJECT SELECTION

A group of stabilised stroke patients (n=20) was studied and compared with an age-matched and sex-matched control group. The study sample included 12 males and 8 females between the ages of 32 and 73 years inclusive (Mean = 57.2 $\pm$ 10.65) who demonstrated residual hemiplegia from a single stroke suffered between 16 months and 20 years previously. This duration post stroke ensured that the neurological status of the subjects was stable. In order to ensure that a relatively homogeneous group of stroke patients was studied, exclusion and inclusion criteria were applied. These criteria are commonly used to select patients as candidates for rehabilitation. All subjects had undertaken formal rehabilitation and had been discharged from this process. Excluded from the study were patients with symptoms which were associated with a negative rehabilitation prognosis as described by Stonnington (1980). These prognosticators included serious or unstable medical conditions, such as heart disease or uncontrolled hypertension, major perceptual disturbances including unilateral neglect, significant peripheral sensory loss, visual field defects including homonymous hemianopsia, marked cognitive disturbances including memory defects, severe intractable pain and incontinence of bowel or bladder. Subjects were all capable of comfortably walking at least 50 m (which was in excess of the distance over which data were collected) without the assistance of an ambulatory aid, such as a cane, but all demonstrated, subjectively, an "asymmetrical, hemiplegic gait pattern". Both right and left sided hemiplegics were included. Subjects with disabilities from other pathologies, such as osteoarthrosis of the hip, which would have potentially interfered with a normal gait pattern were also excluded, as were those who suffered from receptive aphasia which would have compromised the giving of instructions during data collection. Stroke subjects were recruited from local stroke clubs in the adjacent metropolitan area.

The control group was comprised of twenty healthy individuals (12 men and 8 women), aged between 33 and 84 years (Mean = 61.5 ± 12.98) who were functionally and socially independent. Specific exclusion criteria were applied to ensure that no known factors which would affect balance and gait were present. These criteria, proposed by Hogan *et al.*, (1987), consisted of a history of neurological disease, vestibular or inner ear disorders, severe visual disability, significant peripheral sensory loss, severe degenerative osteoarthritis, cognitive disturbance including memory loss, marked skeletal deformity, postural hypotension, chronic alcohol abuse, advanced cervical myelopathy, normal pressure hydrocephalus and multiple sensory disorder. These subjects were recruited from local, community-based groups located in the same metropolitan area from which the stroke subjects were recruited.

MEASUREMENT OF GAIT

The temporal and distance kinematics of the gait cycle were measured using an automatic, computerised, resistive grid walkway originally developed by Wall *et al.* (1976) and later modified by Crouse *et al.* (1987). The walkway consisted of a series of mats into which a grid was set. The grid was made up of copper-clad steel welding rods embedded in ribbed rubber mats and which were situated 0.78 cm apart. Each mat had two grids so that the left and right feet were measured separately. The walkway was 10.4 m long with the central 7.2 m transduced. Dummy mats were placed at the beginning and at the end of this transduced area to eliminate the effects of the accelerations and decelerations which occurred at the start and conclusion of each walking trial (Figure 3.1). Thus, constant velocity gait trials were measured. A stabilised voltage source drove the walkway. Each subject had a strip of self-adhesive, aluminum tape attached to the sole of his/her own shoes which served to complete a current path to ground, when the tape was in contact with the mat, through the otherwise electrically isolated rods. A linear voltage/position relationship was established where the voltage was measured at the output of the current source and the position alternated between the most proximal and the most distal part of the foot in contact with the walkway. These signals were processed through a control box and data were collected and stored in an Apple IIe microcomputer (Apple Computer, Cupertino, California) through an A/D and D/A convertor. Data, in graphic and tabular format, were later printed out for analysis (Appendix 1).

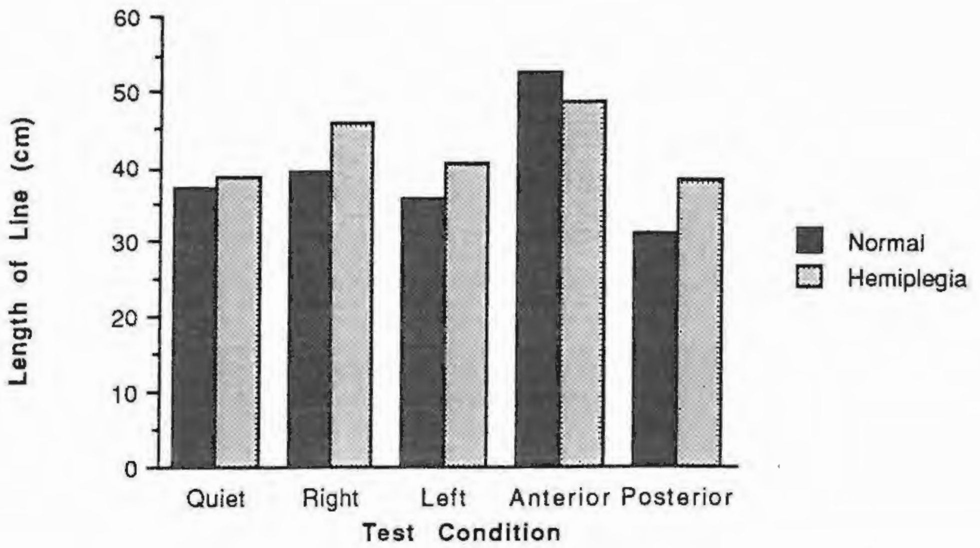

Figure 3.13: Comparison of length of line traced by both subjects CP during all test conditions.

With regard to the mean position of the CP on the Y axis (Figure 3.15), the normal subject was able to displace his CP further posteriorly than the hemiplegic subject. Similarly, the normal subject was able to displace his CP much further anteriorly than the hemiplegic subject. Therefore, the range of weight shift in both axes was much greater in the normal subject in both planes.

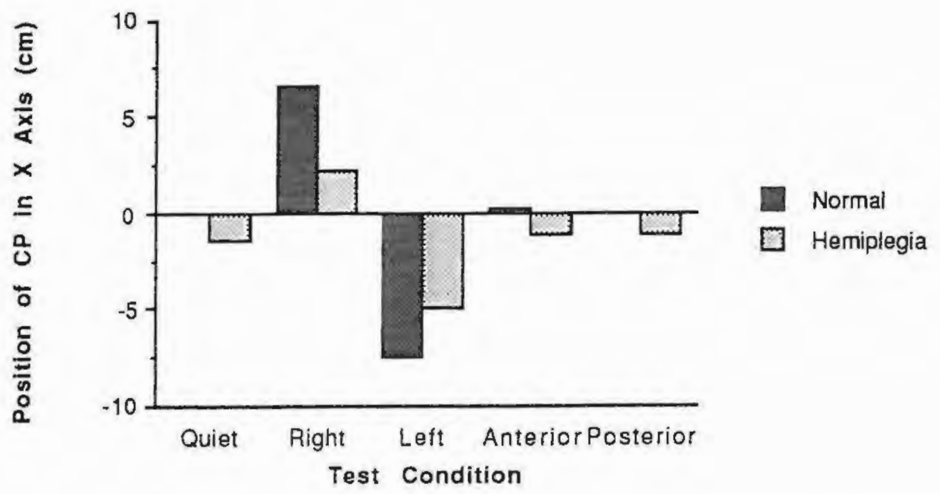

Figure 3.14: Comparison of mean position of the CP on X axis during all test conditions.

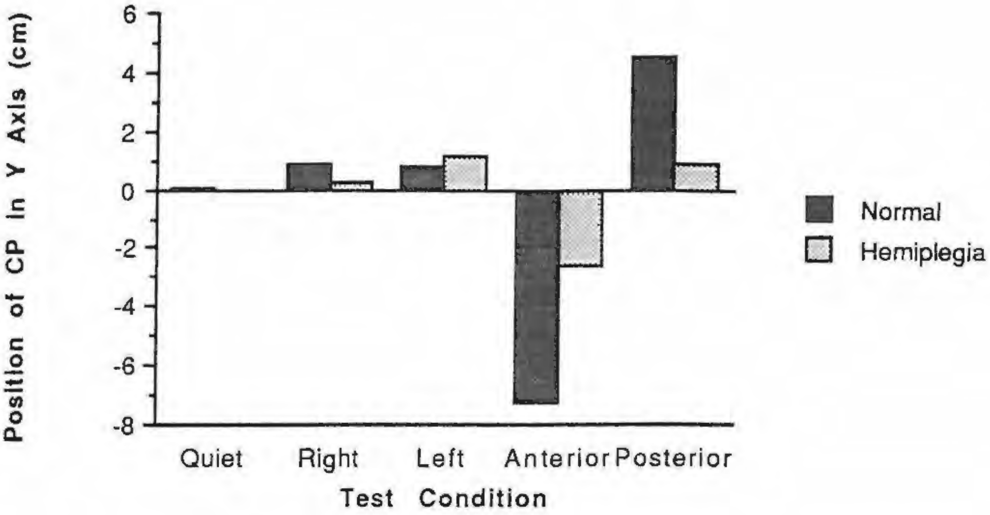

Figure 3.15: Comparison of mean position of CP on Y axis for all test positions.

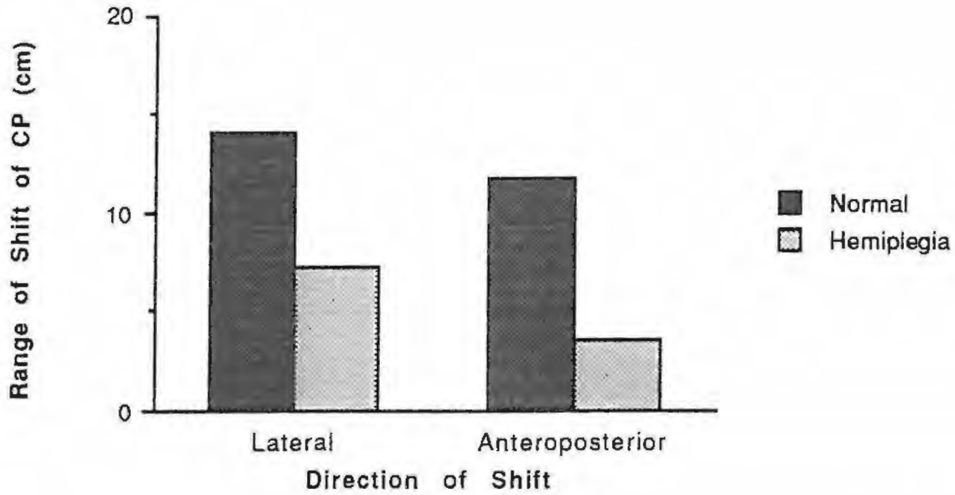

Figure 3.16: Comparison of the ranges of shift of CP in both lateral and anteroposterior directions.

Figure 3.16 shows that the normal subject was capable of shifting his weight over a much greater range in both lateral and anteroposterior directions than the hemiplegic subject.

Figure 3.17 shows that the standard deviation of the mean position of the CP on the X axis is greater in the hemiplegic patient with the exception of the trial in which the normal subject almost fell. This would tend to suggest that magnitude of the postural sway of the hemiplegic subject was greater than that of the normal subject. The same profile appears to be the case in the variability of the mean position of the CP in the Y axis with the exception of the trial where the normal subject almost lost balance (Figure 3.18).

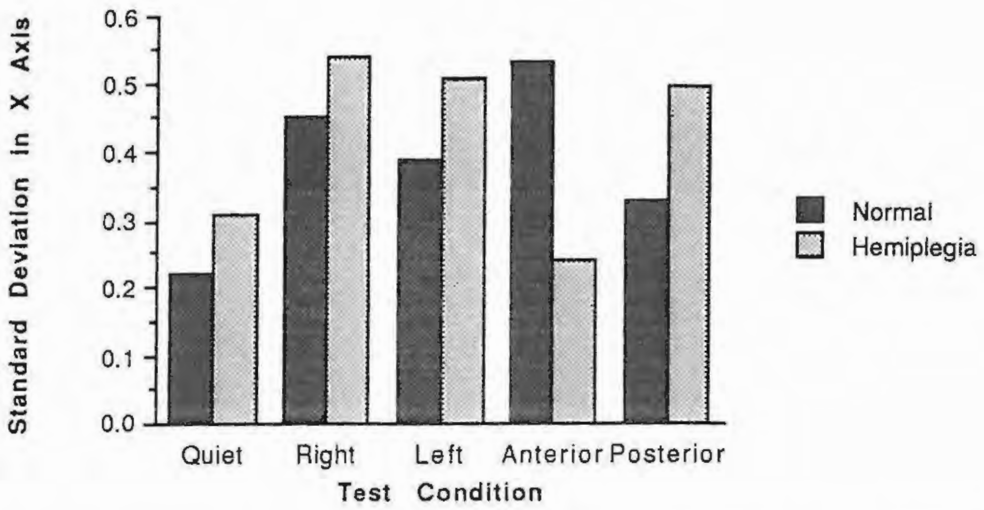

Figure 3.17: Comparison of the standard deviations from the mean on the X axis.

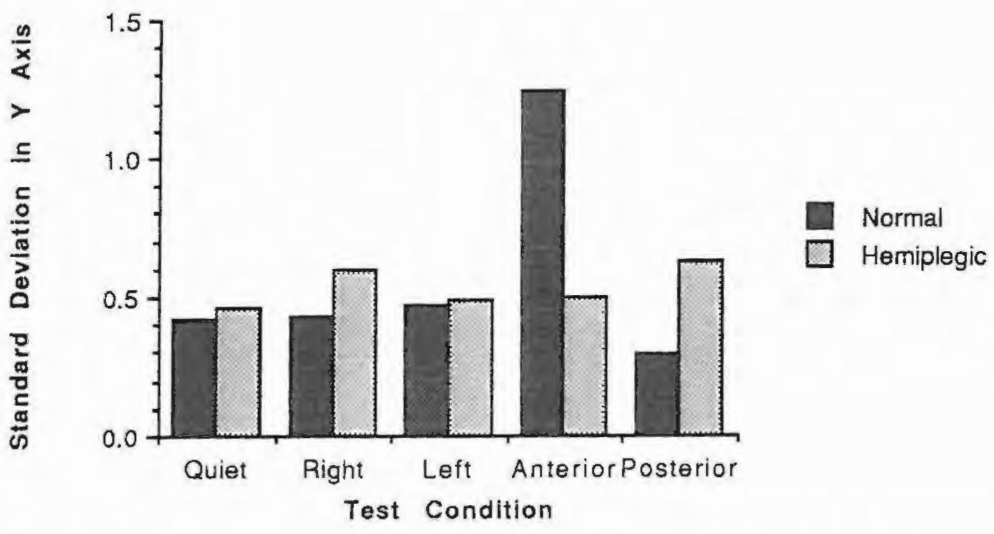

Figure 3.18: Comparison of the standard deviations from the mean on the Y axis.

Conclusions of first pilot study:

The pilot study showed that the protocol of the proposed study appeared sound, achievable and appropriate to answer the research questions. A few problems, however, existed, all related to the performance of the sway platform (SP). While collecting data from the SP, the healthy individual consistently moved his CP beyond the limits of the equipment. In order to overcome this, the normal subject was required to stand with his feet closer together. This factor had the effect of reducing the differences between the subjects. This problem could have been resolved by having both subjects stand with their feet together. However, this would have destabilized the already fragile balance of the hemiplegic subject.

An additional problem was that the position of the feet on the SP had to be traced on to a template situated on the surface of the platform so that manual calculation of the geometric centre of the base of support could be made. This was the only way to identify the position of the CP of the subject relative to the centre of his base of support. This was time consuming and reduced the consistency of the testing procedure creating problems if the subject had to step off the platform for any reason. Another issue which became obvious in this pilot study was that the weight shifts requested appeared unrelated to function and did not resemble any component of the gait pattern. The weight shifts requested, although typical of those required during rehabilitation and the same as those used by Murray *et al.* (1975) and Dettman *et al.* (1987), were cardinal plane shifts, unlike those observed in the gait cycle. Thus, it appeared that it would be wise to consider weight shifts with one foot in front of the other such as those which occur during the double support phases of the gait cycle. Attempts to measure diagonal shifts on the SP were unsuccessful mainly as a result of the limited size of the supporting surface. The testing period of 30 s also appeared to be excessive in that it was clear that fatigue became a factor towards the end of the test thus biasing the balance data.

The walkway performed appropriately without incident.

A second pilot procedure tested a revised protocol for measuring the balance parameters.

Data were obtained from the same two subjects. Gait was not measured in this pilot. To test the balance parameters, the commercially available Chattecx Balance System® (Chattanooga, Tennessee) was used. This system measured similar parameters to the sway platform but the subject stood on two footplates each of which had anterior and posterior components. As a result, the footplates could be adjusted to accommodate the foot length of each subject. The position of the footplates, which represented the base of support of the subject, were input to the computer which calculated the geometric centre of the base of support. Thus, the position of the CP of the subject could be considered in relation to the base of support of the subject. Subjects were tested in quiet standing (Q), with CP maintained anteriorly (A), posteriorly (P), to the right (R) and the left (L). In addition, subjects were required to stand first with the left foot back and the right forward (D1) in a manner which resembled the foot placements during the double support phase of the gait cycle. Subjects were asked to weight bear on the posteriorly positioned foot (B), then symmetrically with weight through both legs (S), and then through the anteriorly placed foot (F). The procedure was repeated but with the feet placed in opposite positions in a manner similar to that assumed during the other double support phase of gait (D2). Data were collected for 10 s under each condition. The results of this pilot are shown graphically.

The results on the X axis were largely consistent with a picture of reduced ability of the hemiplegic subject to shift his CP. Results which were somewhat unexpected, however, were evident in that the hemiplegic subject appeared to place more weight through the hemiplegic leg (left) during quiet standing and during anterior and posterior weight shifts (Figure 3.19). This feature was also detected in the earlier pilot study. In addition, in the D2 position (left foot forward, right back), the hemiplegic subject was able to extend his CP further than the normal subject when leaning forward over his hemiplegic leg.

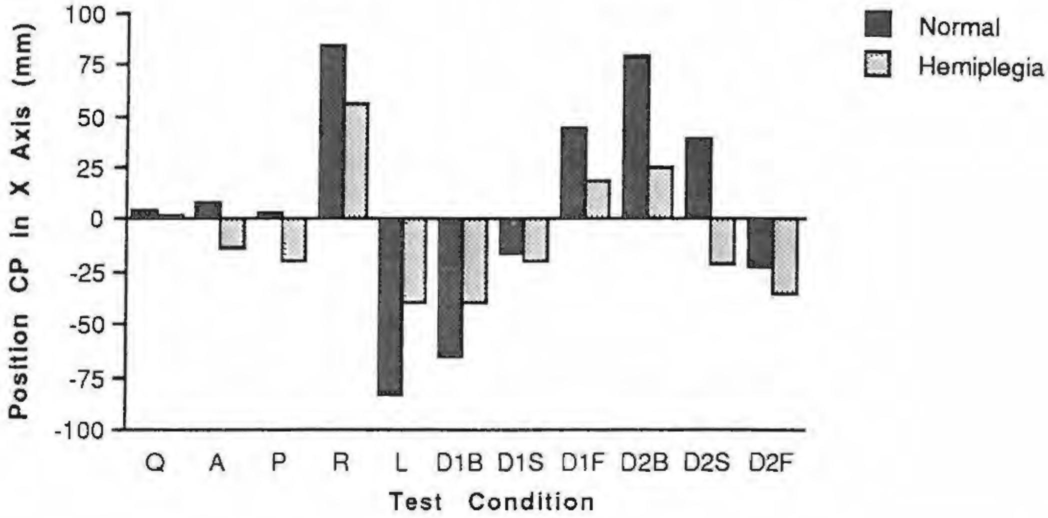

Figure 3.19: Comparison of mean positions of the CP on X axis in each test position between both subjects.

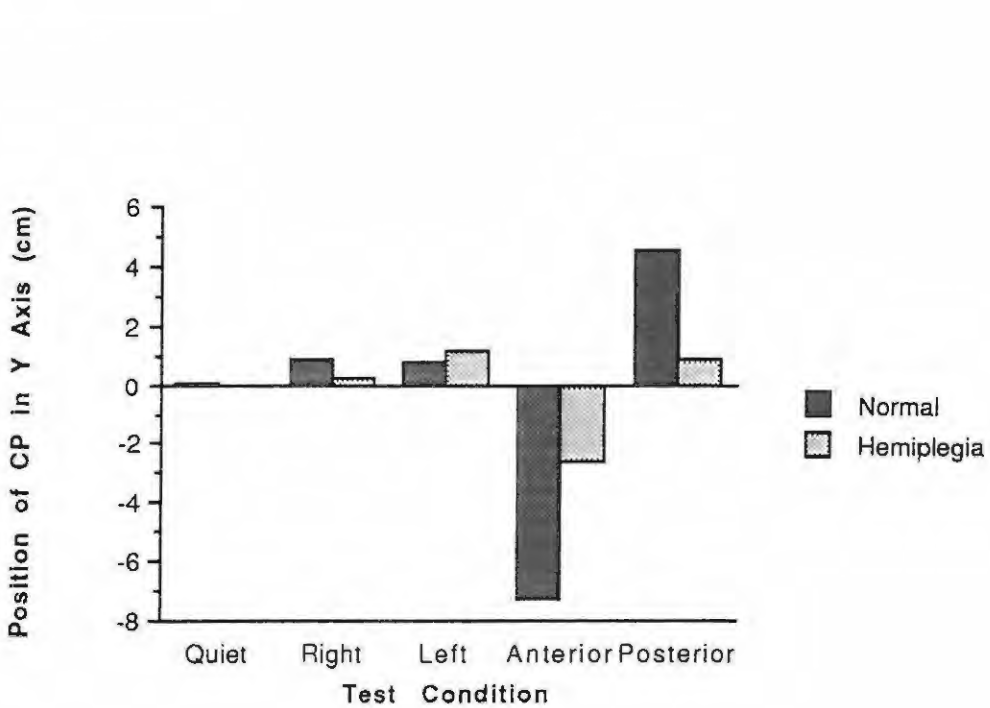

Figure 3.20: Mean positions of the CP on Y axis in each of the test positions.

On the Y axis, the results were indicative of a reduced ability of the hemiplegic subject to shift his CP (Figure 3.20). However, there was little difference in anterior shift performance but the difference was marked when the posterior shift was examined. It was interesting to see that in the D2 position (left foot forward, right back), which was commented upon earlier, the normal subject was able to extend his CP further than the stroke subject when leaning forward over the left leg when considering it in this axis. Perhaps the reason that the hemiplegic shift appeared impressive when considering the X axis, was that the shift was predominantly lateral rather than anterior, which may be an important finding when considering the CP shifts during gait.

Data obtained for the measures of postural stability, the Disp. and the percentage of test time spent within the various % of body weight rings, are shown in Figures 3.21 - 3.24. These data were indicative of a greater postural sway in the hemiplegic subject in all of the test positions, suggesting decreased postural stability. The findings of the Disp. measures and the percentage of test time within the % of body weight rings were consistent with each other.

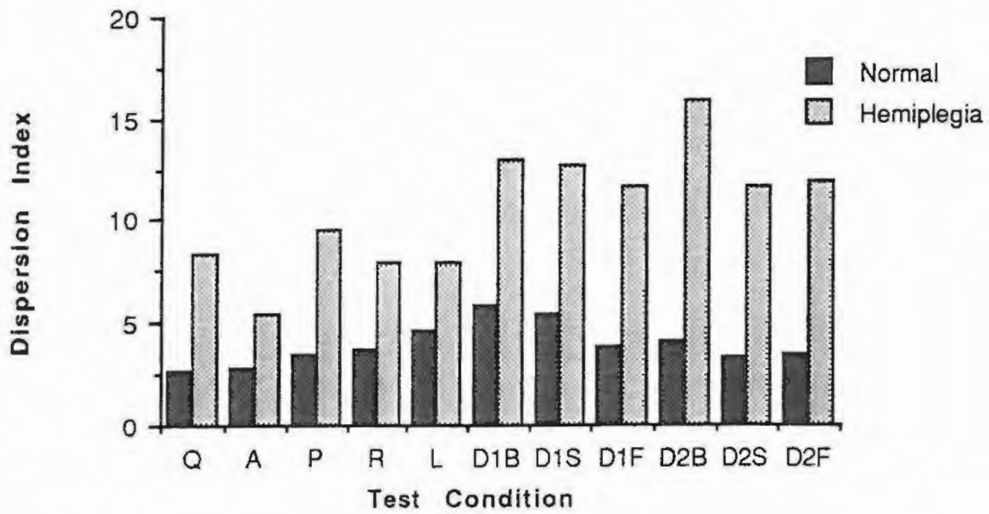

Figure 3.21: Postural steadiness as measured by Disp. values in each test position.

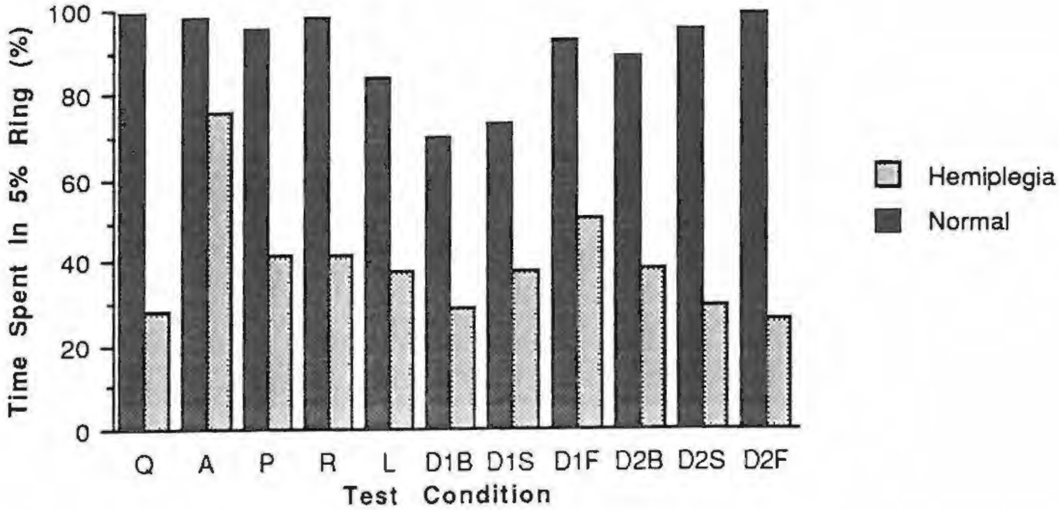

Figure 3.22: Comparison of postural steadiness between subjects measured as the % of test time spent within the 5% body weight ring.

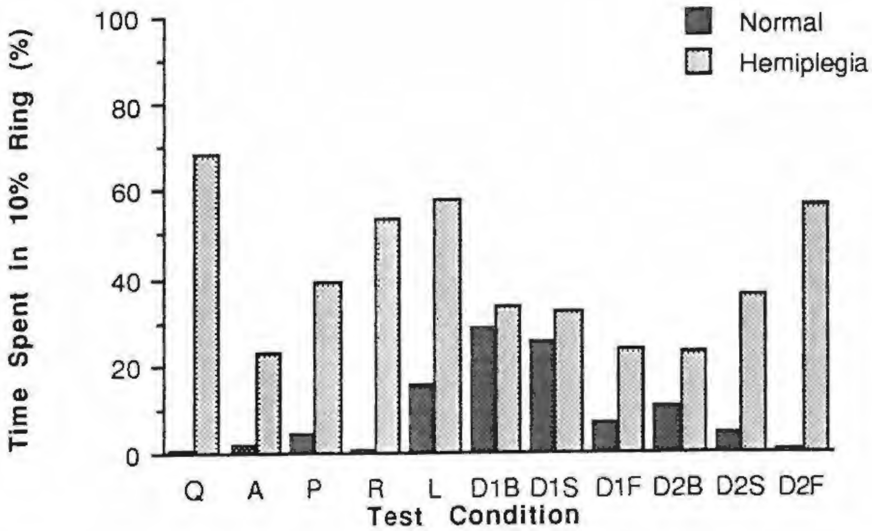

Figure 3.23: Comparison of postural steadiness between subjects measured as the % of test time spent within the 10% body weight ring.

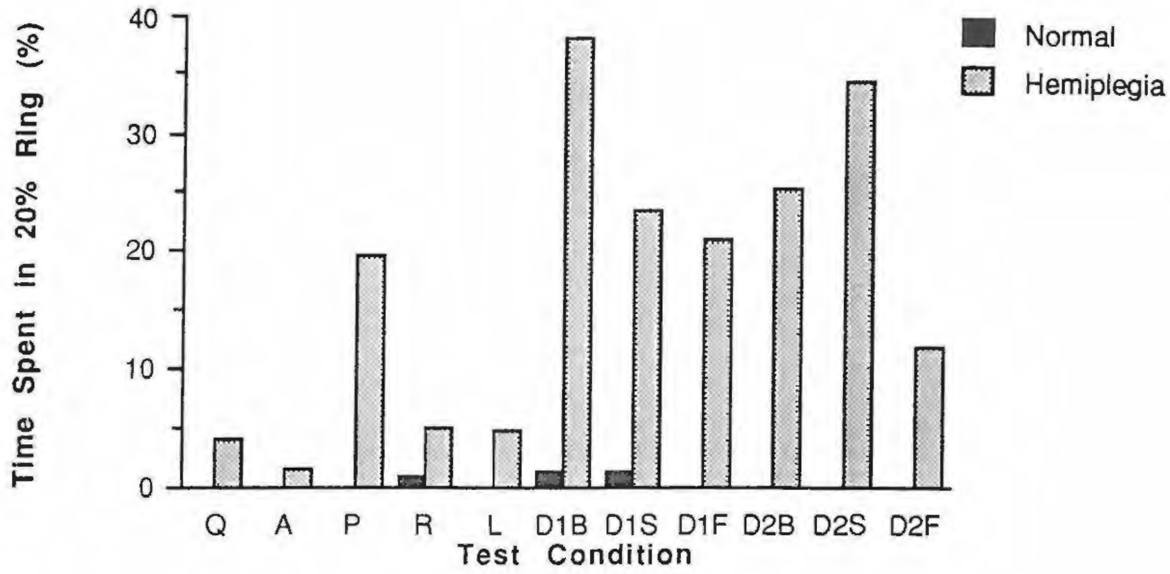

Figure 3.24: Comparison of postural steadiness between subjects measured as the % of test time spent within the 20% body weight ring.

Figure 3.25 compares the range of weight shifts in both the cardinal plane directions and the diagonal directions. Clear differences existed between the performances of the subjects in shifting CP anteroposteriorly and laterally. This finding was consistent with those in the previous pilot. However, these differences were accentuated when the diagonal shift was examined. In particular, the ability of the hemiplegic subject to shift his CP when his affected leg was situated posteriorly was markedly compromised when compared with the normal subject. It is possible that measuring balance under these diagonal conditions may be more relevant when considering gait than cardinal plane shifts because the positions are representative of the double support phases of the gait cycle.

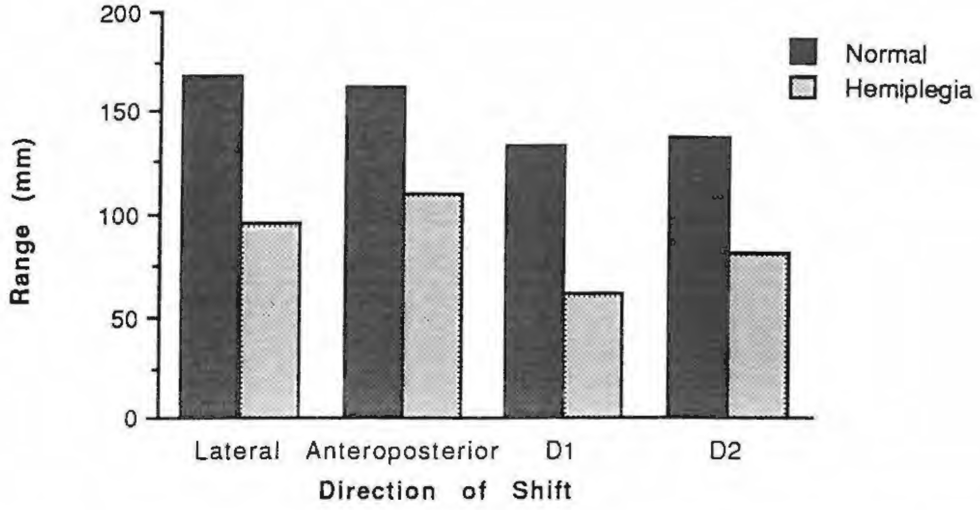

Figure 3.25. Comparison of range of shift of CP between subjects in the lateral, anteroposterior and both diagonal directions (D1 and D2).

This pilot study confirmed that the Chattecx BalanceSystem™ resolved all of the shortcomings of the sway platform used initially, particularly in relation to calculating the position of the geometric centre of the base of support and dealing with the foot placement issue. The problem of the normal subject exceeding the bounds of the equipment was also resolved. Similarly, the use of this system facilitated the measurement of diagonal weight shifts which may be more strongly correlated to gait performance than the cardinal plane protocols which have been typically used prior to this study. Thus, it was decided to proceed with the use of this balance measurement system for the study.

THE EXPERIMENTAL PROTOCOL

A proposal describing the project was approved by the Ethical Review Committee, Faculty of Health Professions, Dalhousie University (Appendix 3). Hemiplegic subjects were recruited from local stroke clubs. These clubs were community-based, social support groups which met regularly. Volunteers were individually assessed by the investigator, an experienced neurological physiotherapist, to determine their suitability to be included in the study. This was done by assessing the stroke subjects and completing a checklist

(Appendix 4). A similar checklist (Appendix 5) was also completed for each of the control subjects to identify any variables which could potentially interfere with the gait pattern of the subject and which ensured that all subjects met inclusion/exclusion criteria.

Subjects visited the Gait Laboratory at the School of Physiotherapy, Dalhousie University, Halifax, Nova Scotia, Canada. Each subject completed an Informed Consent procedure and signed a related document (Appendix 3).

Following these procedures, all subjects were measured for stature using a calibrated stadiometer (Healthometer, Continental Scale Corporation, Bridgeview, Illinois) to the nearest cm for the purpose of calculating relative walking velocity and relative stride length. Weight was determined by weighing each subject on a calibrated medical weigh scale (Healthometer) and was collected to input to the Chattecx Balance System® thus permitting the calculation of output regarding characteristics of balance performance.

Protocol for Measurement of Balance: The Chattecx Balance System® was calibrated prior to data collection for each subject. Ten different tests, each lasting 10s and each measuring a different balance condition, were undertaken by each subject. Subjects stood with the feet parallel and 8 cm apart with eyes open with vision fixed upon a focal point situated on an adjacent wall. This is a modification of the protocol established by Mahar *et al.*, (1985). The same footplate coordinates were used for each subject except for minor adjustments to accommodate different foot lengths (Figure 3.26). Data were collected during the following conditions:

i) subjects were requested to stand in a quiet manner

ii) subjects were asked to maintain their weight as far forwards as possible.

iii) subjects were then asked to maintain their weight as far backwards as possible.

iv) subjects were asked to maintain as much weight as possible through the right leg.

v) subjects were asked to maintain as much weight as possible through the left lower limb.

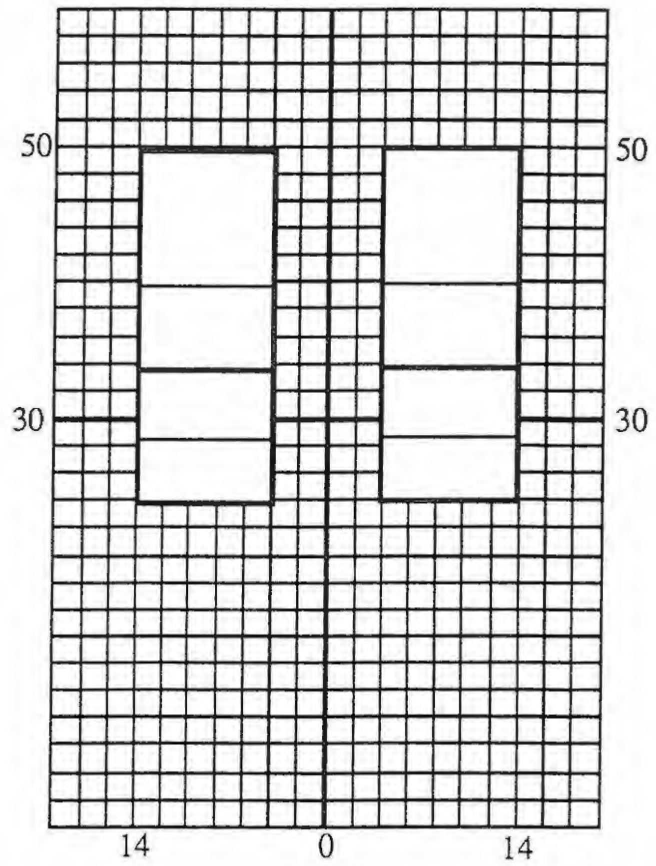

Figure 3.26: Location of the footplates for the quiet standing and all cardinal plane weight shift conditions (anterior, posterior, left and right). The top left corner on the left footplate was positioned at coordinates 50, 14 Left on the supporting surface while the top right corner of the right footplate was positioned at 50, 14 Right.

One footplate was then moved backwards while the other was moved forwards (Figure 3.27). The subjects then stood on the footplates in a manner which was similar to the position of the feet in the double support phase of the gait cycle with the left foot back and the right forward (diagonal position D1).

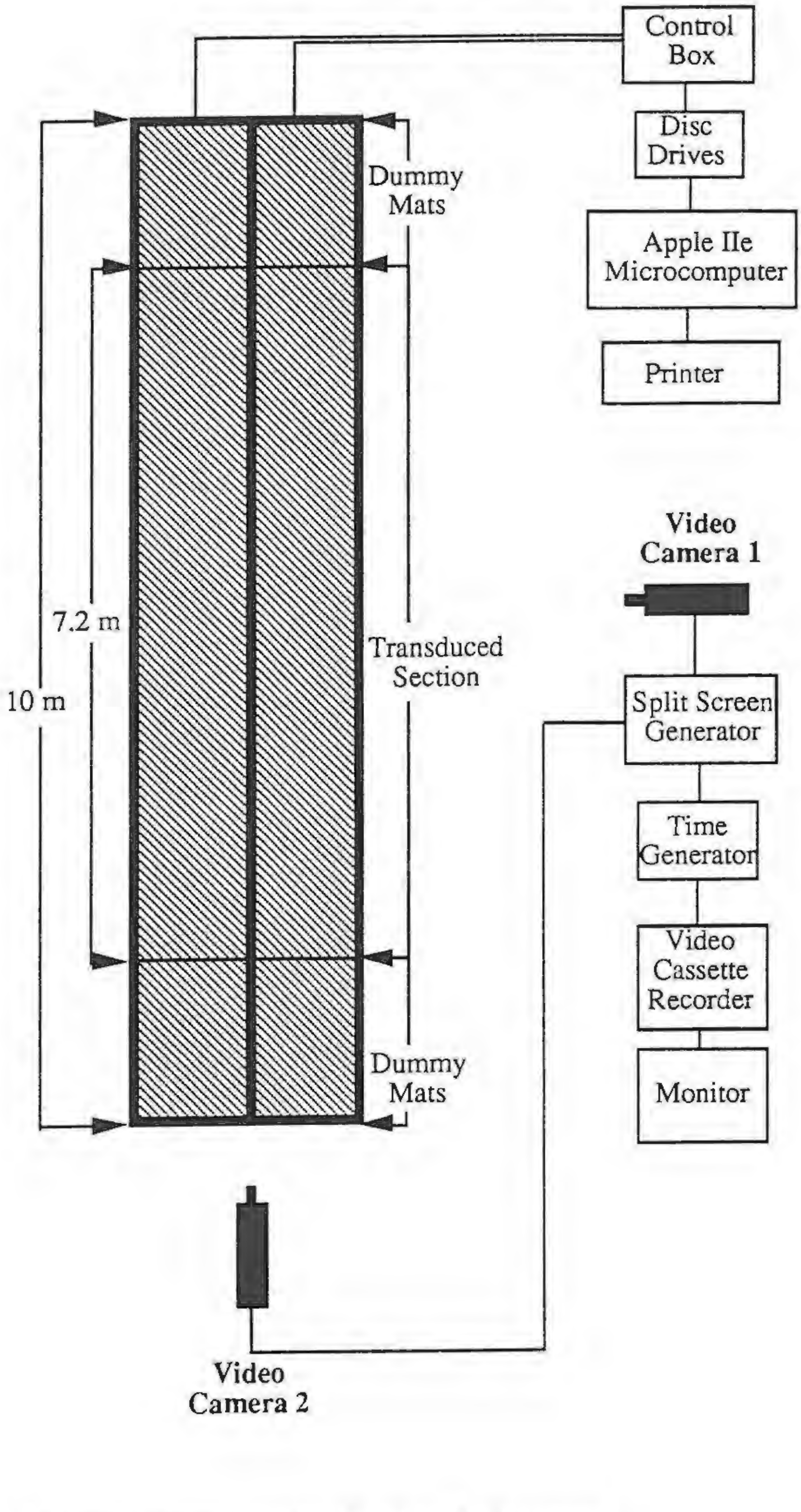

Figure 3.1: Schematic representation of gait analysis system.

Information concerning the temporal (i.e. stride time, step time as well as the duration of each phase of the gait cycle) and spatial (stride length and step length) characteristics of the gait cycle were thus obtained. From these, velocity was calculated in both absolute (m/s) and relative (st/s) terms by calculating the average values for each stride taken during each trial. The unit of velocity referred to as relative speed was proposed by Grieve and Gear (1966) as a measure of gait velocity which accounted for differences in walking speed as a result of subject height. To calculate this measure, velocity was divided by stature and was expressed as the number of times body height is covered in overground walking in 1s. This measure has been shown to be better correlated to other gait parameters than velocity expressed in terms of m/s (Rosenrot *et al.*, 1980).

The magnitude of asymmetries of both temporal and spatial parameters was also calculated using a modification of the asymmetry ratio (AR) described by Wall and Turnbull (1986). For the purposes of this study, only a measure of the magnitude of the asymmetry was required, therefore, the greater value was divided by the lesser value. To express the magnitude of asymmetry from a baseline of zero (0), the following equation was used:

$$\mathbf{AR} = \left(\frac{\text{Lower Limb with Greater Value}}{\text{Lower Limb with Lesser Value}}\right) -1$$

where AR represents the asymmetry ratio; "Lower Limb Parameter with Greater Value" denotes the value of gait parameter on the leg which has the greater value of the two and "Lower Limb Parameter with Lesser Value" denotes the value of gait parameter on the leg which has the lesser value of the two·

For example, if the unaffected leg had a single support value of 30% of stride time compared with 10% for the affected leg then the asymmetry ratio would be (30/10) - 1 = 2. The AR, therefore, compared the value of one limb with that of the other and so provides an indication of the magnitude of the asymmetry with the value of 0 indicating that no asymmetry exists. With regard to the normal subjects, the same procedure was followed as that for the hemiplegic subjects

The stroke subjects were also videotaped, in both sagittal and frontal planes, while data were being collected from the walkway. This was done to check and facilitate interpretation of the data by enabling the investigator to view the videotapes in the light of the data obtained for those walks. The videosystem consisted of two colour cameras (JVC S-62U) connected through a split screen generator (Vicon V270 SP) and a time generator (Vicon V 240 TW) to 1.25 cm VHS videocassette recorder (JVC BR - 6400U). The output from the recorder was viewed on a 30 cm color monitor (Panasonic CT - 110 MCA). The cameras were mounted on wall brackets thus standardizing the images from subject to subject. One camera was mounted at the end of the walkway to permit filming in the frontal plane while the other was placed at the side of the walkway to allow for sagittal plane filming. The location of the cameras during data collection is shown in Figure 3.1.

MEASUREMENT OF BALANCE

Balance performance was measured using the commercially available Chattecx BalanceSystem™, a schematic representation of which is shown in Figure 3.2. This measurement device has only recently appeared on the market and two recent studies have shown the device to be credible and valid as a means of measuring balance performance (Dickstein and Dvir, 1993; Grabiner *et al.*, 1993).

The principle of operation of this device was the measurement of body weight distribution while the subject stood on four footplates installed with electronic pressure transducers.

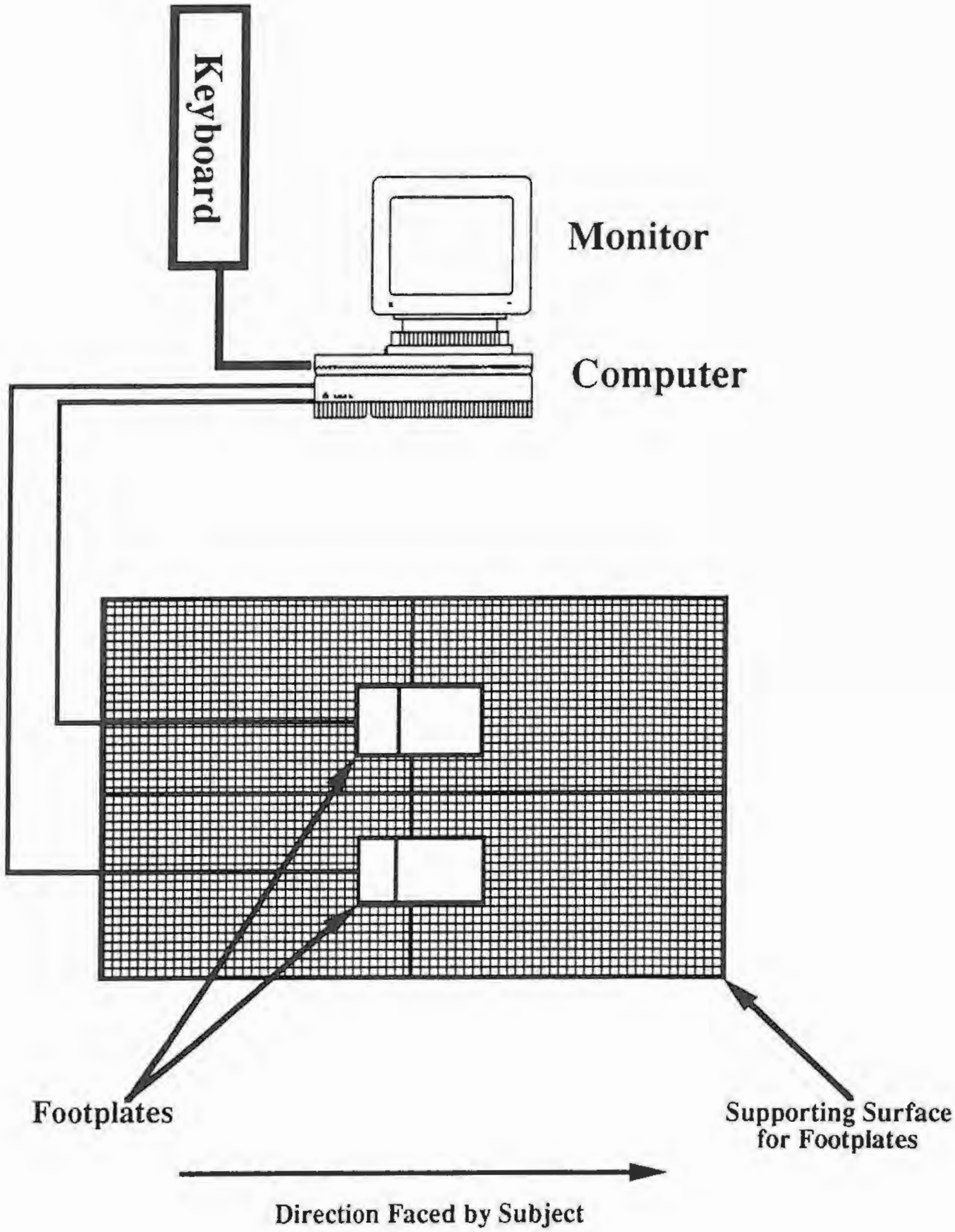

Figure 3.2: Schematic of balance system configuration (footplates set for cardinal plane weight shifts upon the calibrated supporting surface).

The footplates functioned as pairs with the anterior plate measuring the pressure exerted through the forefoot while the posteriorly-positioned footplate measured the force exerted through the heel. The heel and toe plates were connected by two metal rods which permitted the footplates to be adjusted for varying foot sizes (Figure 3.3).

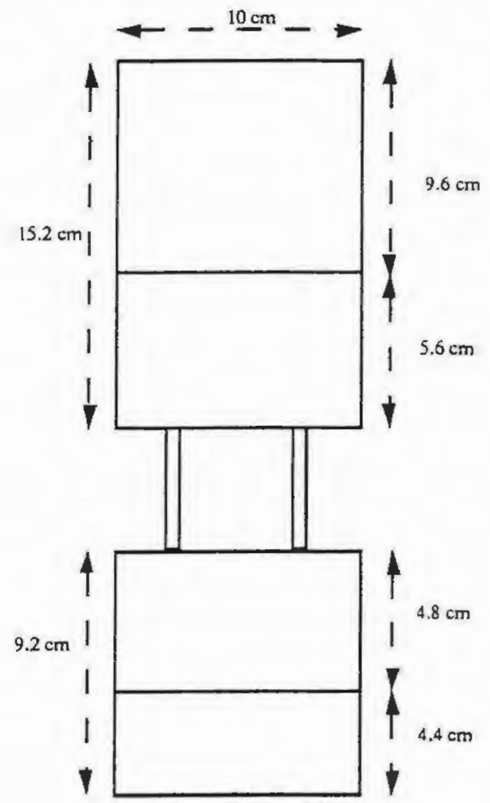

Figure 3.3: Dimensions of the footplates. Each pair of footplates was joined by two adjustable rods to permit adjustment to compensate for different foot lengths. The transverse lines on each force transducer were designed to standardize the position of the foot on the footplate. The calcaneus was placed on the line on the posteriorly situated transducer while the ball of the foot was placed on the line on the anteriorly placed transducer. The height of each footplate was 3.2 cm.

The location of the footplates relative to the supporting surface was altered for the two diagonal tests. Consistency of footplate position between subjects was achieved by aligning them with a grid etched into the surface of the supporting platform (Figure 3.4). Fluctuations in the distribution of pressure over the four footplates were measured and used to determine both the amount and direction of the deflection of the CP.

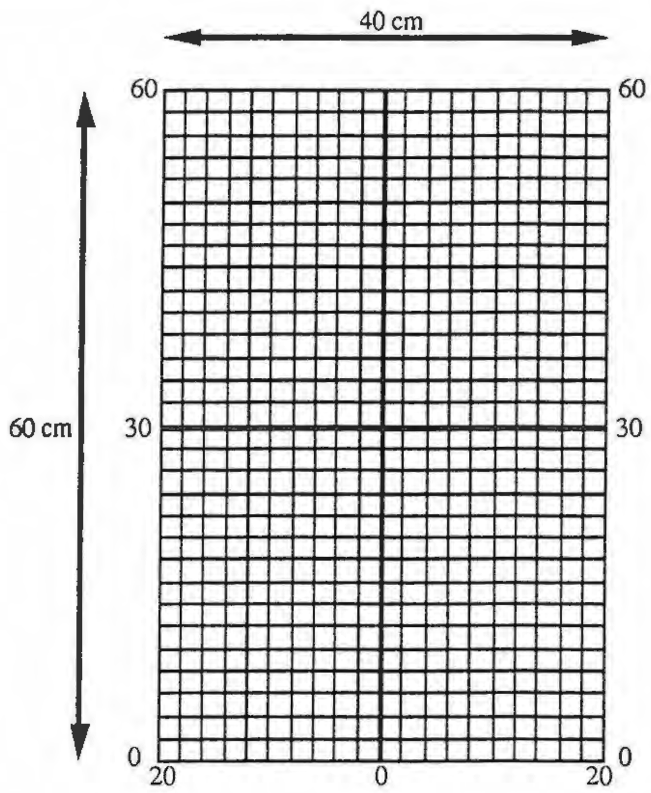

Figure 3.4: The dimensions of the calibrated supporting surface on which the footplates were set. Each square was 2 cm^2.

Signals from the footplates were input to an IBM PC computer where they were processed and stored to provide information concerning the balance performance of subjects. Computer software calculated three measurements of balance performance: 1) the location of the mean CP expressed in terms of its mediolateral (X) and anteroposterior (Y) coordinates. 2) postural sway reflected as the degree of scatter of the data about the mean CP and expressed as the Dispersion index (Disp.); and 3) the limits of postural stability expressed as the percentage of time of the test spent away from the mean CP. The positions of the footplates in relation to their position on the calibrated supporting surface for each series of tests are shown in Figures 3.26 and 3.27. To ensure a degree of consistency regarding the base of support of all subjects the following procedure was applied when adjusting the footplates to compensate for the differing foot sizes of the subjects. In the test position with the feet parallel, the rear force transducer was adjusted only while the front transducer remained in the same position. In the diagonal placement the rear transducer of the anteriorly positioned footplate was adjusted as was the front transducer for the posteriorly placed footplate. As a result, in the diagonal weight shifts, the base of support remained consistent between subjects.

Data were collected at a frequency of 100 Hz and the duration of each test was 10 s. The mean CP was calculated automatically by the computer system in both the X and the Y axes and was an expression of the average value of all data points collected during the testing period in relation to the geometric centre of the base of support. The Disp. was also determined automatically and consisted of a single number calculated using the standard deviations from the data collected on both the X and Y axes during the test. This measure of variance was calculated using the following formula:

$$\mathrm{Disp}^2 = \frac{(\Sigma\, \mathrm{CP\,X} - \mathrm{Xi})^2 + (\Sigma\, \mathrm{CP\,Y} - \mathrm{Yi})^2}{\text{Time of Test (s)} \times 100}$$

where Xi denotes a given value obtained during data collection on the X axis (10 s at 100Hz). Thus, there were 1,000 Xi points during the 10 s data collection period. Yi represents a given value obtained during data collection on the Y axis (10 s at 100 Hz). Therefore, there were also 1,000 Yi points during the 10 s data collection period. Disp. was, therefore, a coefficient of variation and provided an indication of variability about the mean value in both the X and Y axes. From this variability, a measure of postural sway was provided.

An alternative measure of postural sway was also produced from this apparatus and described by the manufacturers of the device as "the limits of postural stability". These values expressed the percentage of test time spent in a series of concentric rings representing percentages of body weight. Figure 3.7 illustrates the concentric rings represented 5% (innermost ring), 10%, 20%, 40%, 60%, and 80% of body weight (outermost ring). These rings were generated automatically on an individual basis and were based on the outputs of the footplates. Thus, for a subject weighing 100kg, the innermost ring would represented 5% of that body weight (5kg.) while the outermost ring (80%) would represent 80kg. Therefore the data in each ring represented the percentage of test time that the instantaneous CP fell within that ring. Thus, a subject with a small postural sway would demonstrate 100 % of test time within the 5% of body weight ring. A subject with a greater sway would spend higher percentages of time further from the mean CP perhaps in the 20 % or 40 % of body weight rings. Thus, an additional measurement of the postural steadiness of the subject was provided. The advantage of this method of assessing postural sway was that it considered the individual body weight of each subject and minimized differences in the magnitude of sway which might result from stature (Medeiros, 1992).

Results were displayed both numerically and graphically and were later printed out for analysis (Appendix 2).. The CP values were displayed in relation to the actual centre of the base of support which was calculated by considering the area of the footplates. A software programme permitted the operator to enter the coordinates of the footplates on the grid etched into the surface supporting the footplates. This feature was designed to enable the footplates to be placed in varying positions determined by the individual foot size characteristics of each subject. The area of the base of support could thus be calculated on an individual basis and its geometric centre determined. Location of CP in the X and Y axes could thus be considered in reference to the centre of the base of support.

The balance testing system (Figure 3.6) was equipped with numerous features designed to ensure subject safety during testing. Frontal and lateral grab bars and an overhead harness were available but not used during this study.

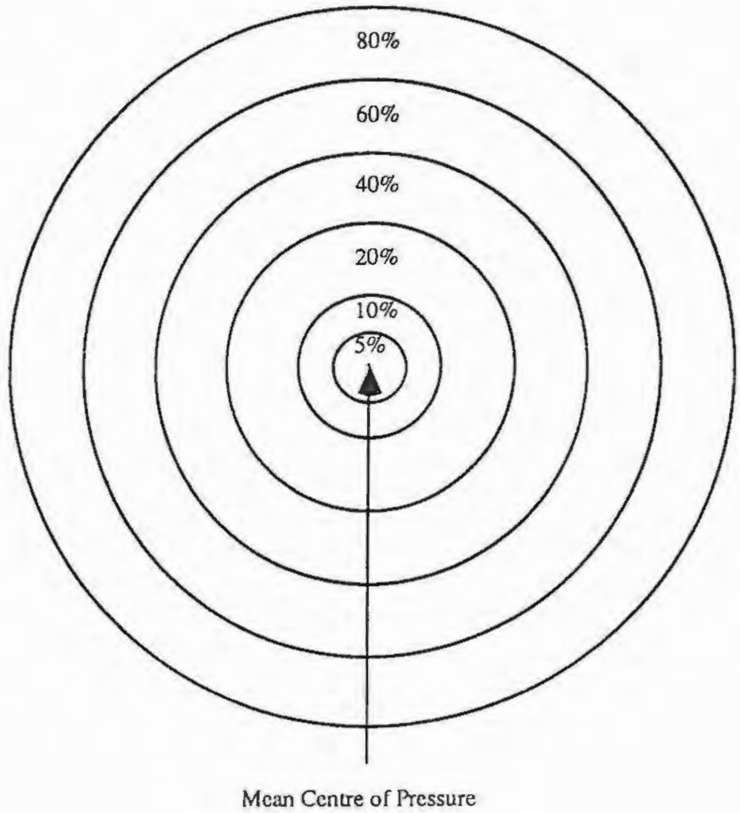

Figure 3.5: Concentric rings which represent varying percentages of body weight displacement away from the mean CP.

The positions of the mean CP in both the X and Y axes were collected during the 10 s testing period for comparative purposes and provided the basis for deriving a measurement of the range over which the subjects were able to safely manipulate their CP during a number of weight shifts. The Disp. and the percentage of time spent in the body weight rings was used to indicate the amount of sway in each of the test positions, thus providing a measure of postural stability in these test positions.

Figure 3.6: The Chattecx BalanceSystem™ with subject standing in the D2 position.

PILOT STUDIES

To test the feasibility of the experimental protocol, two pilot studies were undertaken.

The first pilot tested the experimental protocol using a custom built sway platform to measure the balance parameters. Data were obtained from two subjects. The first subject was a 27 year old male suffering from a left sided hemiplegia sustained as a result of an intracranial hemorrhage 11 years previously. This subject was ambulant and fully functionally independent. The second subject was a normal, 31 year old male. The protocol for the proposed study was adhered to, with gait being measured first on the grid walkway followed by balance using the custom-built sway platform. It was originally the intention to use the sway platform for the study, hence its use in the pilot project

Each subject was required to walk at self-selected "normal", "slower" and "slowest" speeds during which data were collected on an automatic restive grid walkway. This was followed by another "normal" speed walk to reorientate the subjects' perception of normal speed. Data were not collected during this walk. Measurements of the temporal and spatial parameters of gait were then obtained for "fast", "fastest" and a "corrected" walk, the latter trial only being required of the hemiplegic subject by requesting that he walk with equal step lengths and attempt to transfer equal weight through both legs during the gait cycle.

The apparatus used to measure balance performance in this pilot consisted of a custom-built sway platform (SP) interfaced with an Apple II+ computer system. Centre of pressure was measured by means of a pair of strain gauges attached to each of four beams which supported the platform. Signals from these gauges were processed through a control box and input to the computer through the game port. Data, indicating the position of the CP of the subjects on the X and Y axes, were obtained utilizing custom written software. Subjects were tested in a total of five positions. In the first trial, subjects were requested to stand in a quiet manner (quiet standing). In the second trial, subjects were asked to shift their weight as far as possible to the right side and on the third trial they were asked to transfer their weight as far as possible to the left side. Finally, the subjects were asked to displace their weight as far forwards as possible followed by as far backwards as possible. During all trials, the mean position of the CP was calculated on both the "X" and "Y" axes along with the standard deviation from the mean, in both axes, for each during the 30 s data collection period. These data provided a measure of the ability of the subject to shift and sustain this transfer of CP. The length of the line traced by the CP over the data collection period was also determined again to provide a measurement of the magnitude of postural sway in all test positions. Finally, the range of the mean values in both the anteroposterior and lateral positions were calculated and provided an indication of the range over which subjects could voluntarily shift weight in both lateral and anteroposterior directions.

Data were examined in graphic format and the following observations made first for gait and then for balance.

Gait Tests:

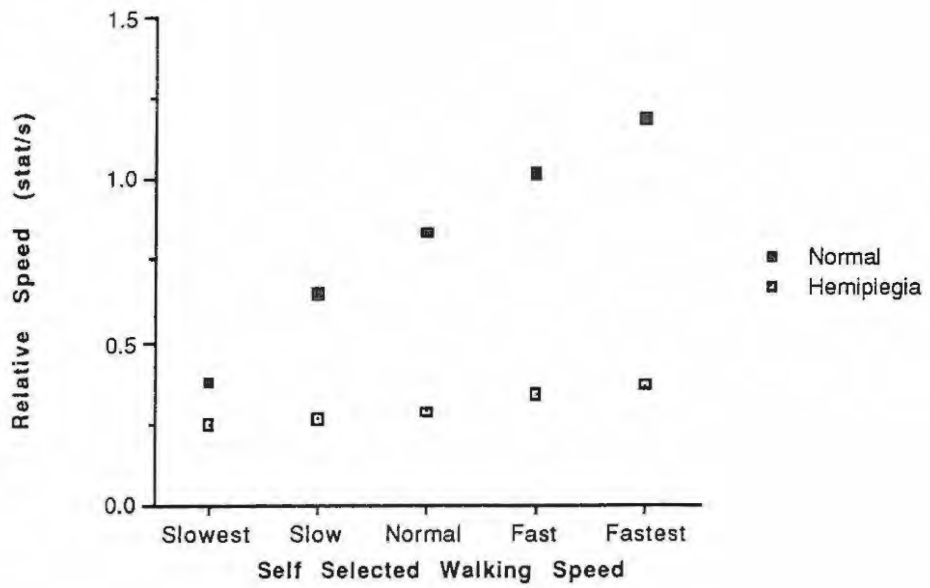

Figure 3.7: Comparison of self-selected walking speeds between hemiplegic and normal subjects at each of the walking conditions.

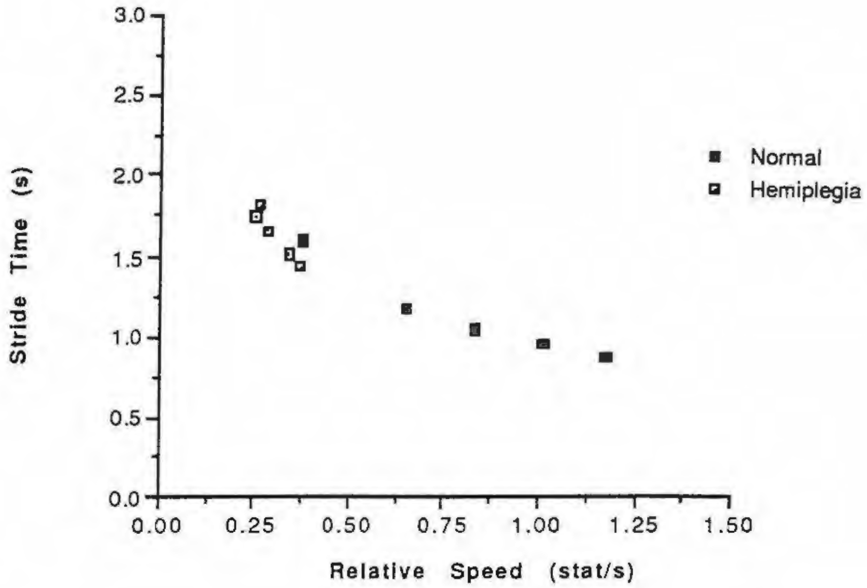

Figure 3.8: Comparison of stride time plotted against relative speed for hemiplegic and normal subjects.

The normal subject demonstrated a range of walking speeds from 0.38 st/s at his "slowest" speed to 1.18 st/s at his "fastest" (Figure 3.7). His walking speed increased in a linear fashion with each condition. The hemiplegic subject walked much slower with speed ranges from 0.25 st/s to 0.37 st/s. His "fastest" walking speed was slower than the "slowest" speed of the normal subject.

The stride time of the normal subject declined in a predictable curvilinear fashion with increasing walking speed from 1.59 s to 0.87 s (Figure 3.8). The values from the hemiplegic subject ranged from 1.81 s at his "slow" speed to 1.44 s at his "fastest". Again the range of the measurements of the hemiplegic subject was small (0.37 s) compared with (0.72 s) for the normal subject.

The stride length of the hemiplegic subject was much shorter than that of the normal subject and varied only from 82 cm to 100.9 cm. (23% increase) compared with 104.6 cm to 179.3 cm. (71% increase) respectively with increasing speed. In terms of relative stride length, the hemiplegic subject, at his "fastest" speed, had a shorter relative stride length than the normal subject at his "slowest" speed (0.54 st/str to 0.6 (st/str) (Figure 3.9).

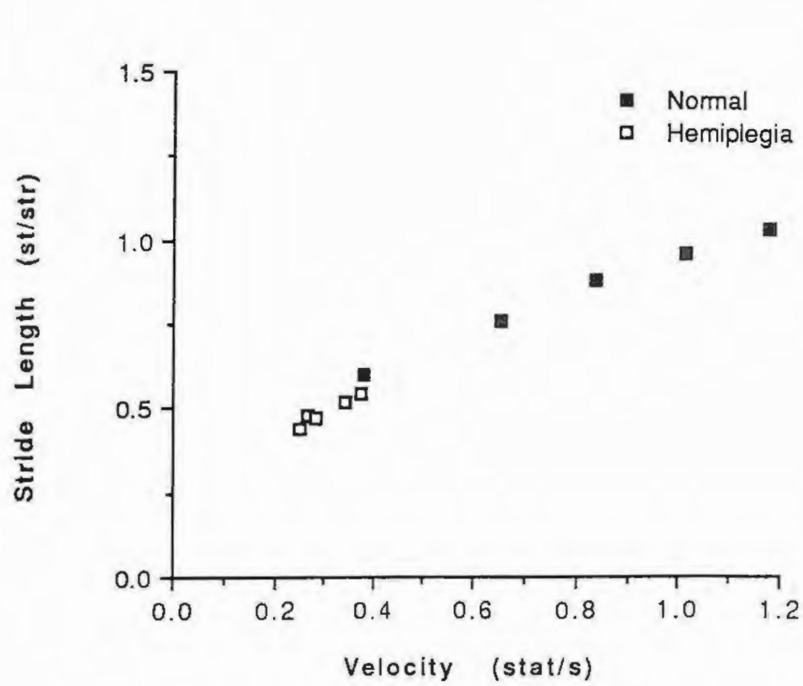

Figure 3.9: Comparison of relative stride length plotted against relative speed between hemiplegic and normal subjects.

Temporal asymmetries were calculated by dividing the percentage of stride of a given phase of the gait cycle derived for the leg with the greater value by that for the leg with the lesser value and subtracting 1 from the product. A score of 0 indicated that no asymmetry existed whereas a positive score signified asymmetry.

The hemiplegic subject was clearly asymmetrical in single support (Figure 3.10) and the braking double support (Figure 3.11) phases of the gait cycle and these asymmetries were greater than those of the healthy subject. There were also clear indications of step length asymmetry in the hemiplegic subject when compared to the normal (Figure 3.12). It appeared that this asymmetry was particularly evident when the hemiplegic walked "very slowly", "fast" and "very fast".

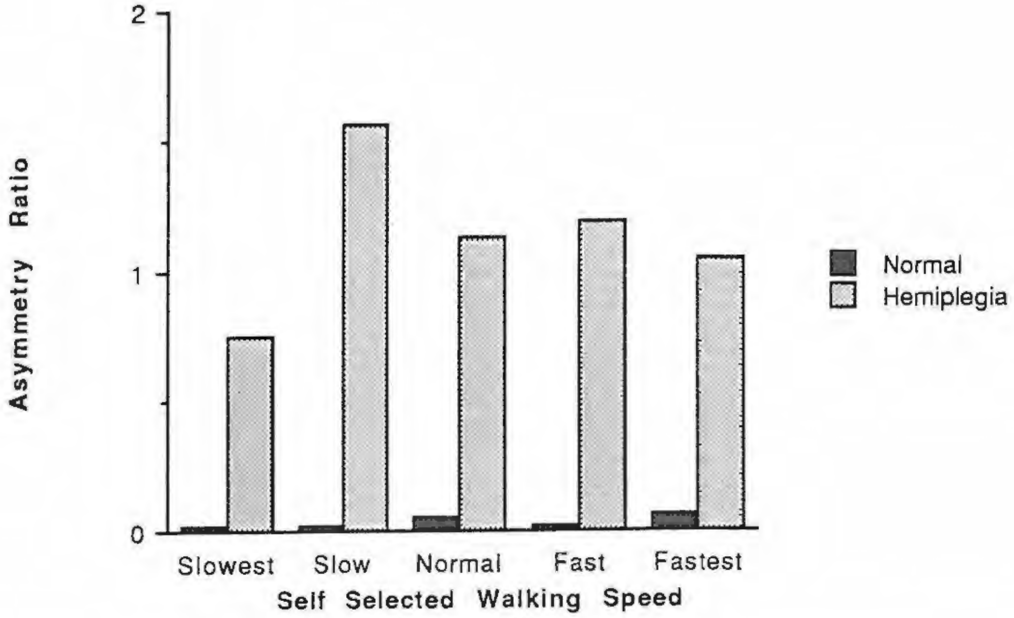

Figure 3.10: Comparison of magnitude of single support asymmetries plotted against self-selected walking speed.

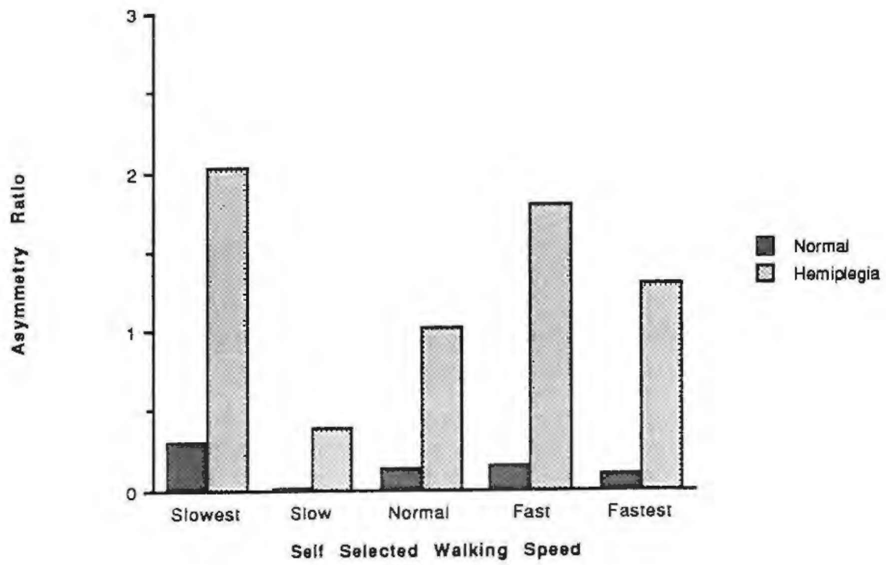

Figure 3.11: Comparison of magnitude of braking double support asymmetries plotted against self-selected walking speed.

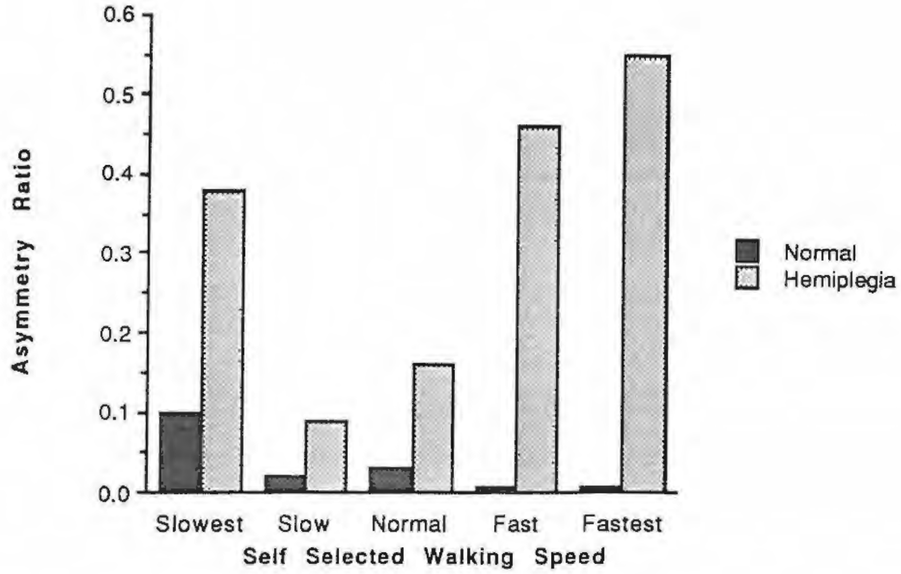

Figure 3.12: Magnitude of step length asymmetries plotted against self-selected walking speed.

Balance Tests:

The length of the line traced by the CP was longer in the hemiplegic subject when compared to the healthy subject (Figure 3.13) except in one trial (displacement of CP anteriorly). This was an artifact of the normal subject almost falling over when challenging his balance. These data suggest that the postural sway of the hemiplegic subject was greater than that of the normal subject implying less stable balance.

Data obtained for the mean position of the CP in the X axis (Figure 3.14) shows that the normal subject was able to displace his CP further to the right and to the left than his hemiplegic counterpart. There is also a suggestion that the hemiplegic patient was able to shift his CP further to the left than the right despite his hemiplegia being left sided. This was an unexpected finding.

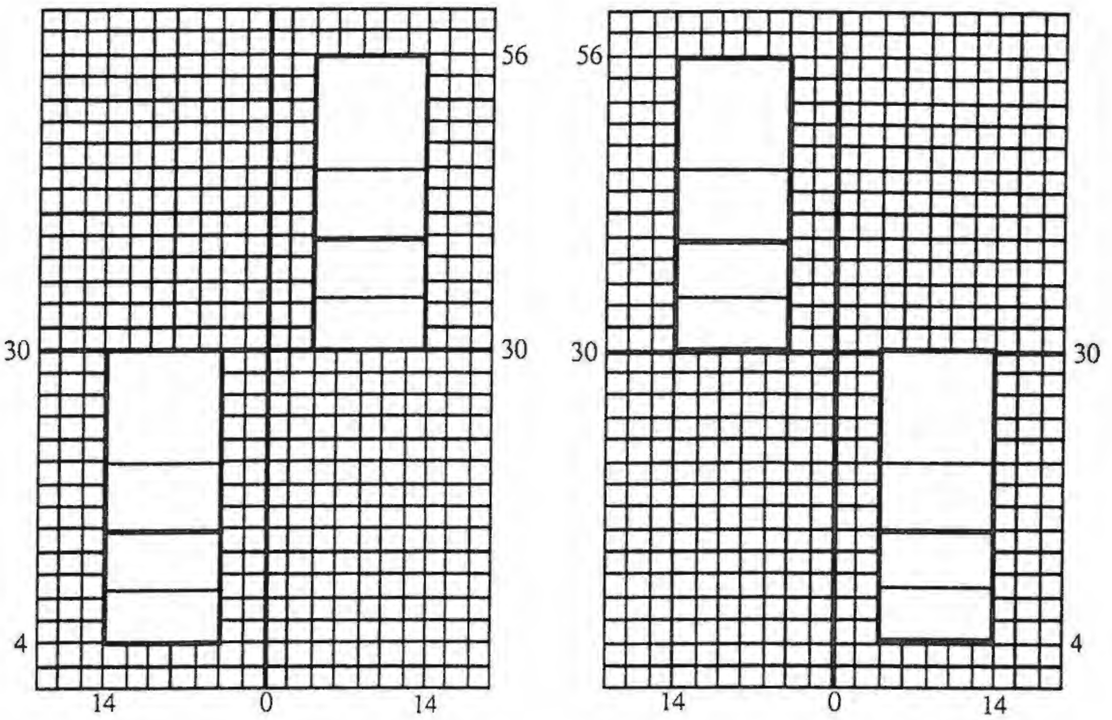

Figure 3.27: Position of the footplates on the supporting surface for the D1 (left) and D2 (right) weight shifts. The coordinates for the anterolateral corner of the anteriorly placed were 56, 14 while that for the posterolateral corner of the posteriorly placed plate was 4, 14.

Data were then collected during the following conditions:

i) The subjects were instructed to place as much weight through the rear foot as possible.

ii) The subjects were then instructed to bear weight symmetrically through both legs.

iii) Subjects were then requested to bear as much weight through the front leg as possible.

The footplates were then moved so that weight shifts could be tested with the feet in the position of the other double support phase of the gait cycle with the left foot placed anteriorly and the right foot posteriorly (Diagonal position D2) (Figure 3.26). The same procedure was followed for measurements in the D1 direction for D2 position. The foot positions were identical to those used for D1 but with the feet in the opposite position. As with the earlier balance tests, data were collected for 10 s at each of these six positions and the same measures were collected.

Two additional tests were included for all subjects which required the subject to transfer weight from the rear foot to the front foot in response to standardised cues from the investigator during the 10s trial in both diagonal balance test conditions. This test was referred to as the "transfer" condition and was dynamic in nature in contrast to the other static balance tests.

In order to control for learning effects caused by repetition of the tests, all trials were randomly ordered for both the study and control groups. This was done by placing scraps of paper on which each test was described and randomly selecting each scrap from a container. In the interests of safety, an operator, a qualified physiotherapist, stood behind the subjects during these movements to ensure the safety of the subject in case of a loss of balance. Subjects were not harnessed in any way and no instructions were provided to the subjects concerning the motor strategies to be used during the test procedures.

During all trials, the mean position of the CP was calculated in both the "X" and "Y" axes along with the Disp. for each during the 10 s data collection period. These data provided a measure of the ability of the subjects to shift and sustain this position of CP and provided an indication of the magnitude of postural sway at each position. The furthermost limits of the mean position of the CP attained laterally and anteroposteriorly were calculated and the area delineating these extremes measured. This measurement was referred to as the range of postural stability. This same procedure was also followed using the mean values obtained for the weight shifts in the directions of D1 and D2 with the range in those directions calculated using the following formula:

$$D = \sqrt{(pX1 - pX2)^2 + (pY1 - pY2)^2}$$

where D = Distance, pX1 = position 1 on the X axis, pX2 = position 2 on the X axis, pY1 = position 1 on the Y axis and pY2 = position 2 on the Y axis.

Thus ranges were obtained for anteroposterior and lateral weight shifts and both diagonal transfers (D1 and D2).

In addition, postural sway was further assessed automatically by the calculation of the percentage of test time spent at various percentages of body weight from the mean position of the CP as represented by a series of concentric circles surrounding the mean position of the CP. Thus, subjects with little postural sway, implying "good" balance, spent a higher percentage of time within the concentric circles closer to the mean position of the CP while those with greater postural sway strayed into the outer concentric rings.

As a result of these testing procedures, the following measures were generated from each of the Chattecx Balance System® conditions applied (Quiet standing, weight shifted anteriorly, weight shifted posteriorly, weight transferred over right limb, weight transferred over left limb, weight shifted posterolaterally to the left (D1), weight shifted anterolaterally to the right (D1), weight shifted posterolaterally to the right (D2), weight shifted anterolaterally to the left (D2) and weight symmetrically distributed between both feet in both D1 and D2 positions):

Mean of CP on "X" axis.

Mean of CP on "Y" axis.

Dispersion Index (Disp.)

Percentage of test time spent in each concentric ring (5%, 10%, 20%, 40%, 60% and 80% of body weight)

Range of weight shift on the "X" axis (lateral cardinal plane shift).

Range of weight shift on the "Y" axis (Anteroposterior cardinal plane shift).

Range of weight shift (D1 and D2 diagonal shift).

Relative positions of the areas of stability and its relationship to the mean position of the CP recorded during quiet standing.

Protocol for Measurement of Gait Parameters: Following the balance measurements, the gait parameters were obtained using the gait measurement system previously described which was housed in the same laboratory as the BalanceSystem™. Self-adhesive aluminium tape was attached to the soles of the subjects' own shoes and each subject was requested to walk at five speeds while temporal and spatial gait data were collected. All subjects received the same instructions prior to each walk in the following manner and sequence:

For Walk I: "Walk down the walkway to the chair placed at the other end at your normal walking speed".

For Walk II: "This time, walk down the walkway slower than normal".

For Walk III: "This time I want you to walk at your slowest speed".

An additional "normal" speed walk, during which data were not collected, was then interposed before the next tested walk. The purpose of this trial was to re-orientate the subject's perception of normal speed which was used as a reference point from which the subject determined variations in velocity.

For Walk IV: "This time, I want you to walk faster than normal"

For Walk V: "Now walk at your fastest speed".

For the hemiplegic subjects only, a sixth walk was included. The instructions provided to the subjects for this walk were as follows:

For Walk VI: "I want you to walk as best you can taking equal steps and putting weight equally on both legs."

All walks of the hemiplegic subjects were videotaped for the purposes of verifying the data and to facilitate interpretation when data were being reduced.

The sequence of walks undertaken by the subjects during gait measurement was not randomized in an attempt to ensure some standardization of walking speed as a reflection of the instructions given.

As a result of these testing procedures, the following measures were generated from each of the walking conditions ("slowest", "slow", "preferred or free", "fast", "fastest" and, for the hemiplegic subjects only, "corrected").

Velocity (m/s)
Relative Velocity (st/s)
Stride Time (s)
Braking Double Support Time (% stride)
Single Support Time (% stride)
Total Support Time (% stride)
Braking Double Support Asymmetry (AR)
Single Support Asymmetry (AR)
Total Support Asymmetry (AR)
Stride Length (cm)
Relative Stride Length (St/str)
Step Length Asymmetry (AR)

STATISTICAL ANALYSIS OF DATA

A number of dependent variables were treated statistically to evaluate the effect of the independent variables of stroke in the stroke sample compared with the control group. Specifically, the research hypotheses were analyzed as follows:

Hypothesis 1

The temporal and spatial parameters of gait of hemiplegic subjects were compared with those of normal age and sex-matched controls utilizing a two factor (subject type and walking condition), repeated measures ANOVA. Repeated measures on one factor (walking condition) were then conducted to identify differences between walking speed conditions for the hemiplegic group and the normal sample. *Post hoc* analyses utilizing the Scheffé test were then used to identify the location of the significant differences. The parameters compared included walking velocity, stride time, stride length, single support asymmetries, double support asymmetries, step length asymmetries, the durations of single support, total support and double support phases of the gait cycle and the range of walking speeds of which the subjects were capable. Unmatched *t* tests were used to test for differences between hemiplegic and normal performances at each of the walking conditions.

Hypothesis 2

The balance performances of the hemiplegic subjects were compared with those of the control sample utilizing unmatched, two tailed *t* tests based upon the position of the CP on the X and Y axes, the variability of the position of CP as measured by the Disp. and the percentage of test time spent in the various body weight rings, the ability of the subjects to voluntarily shift CP in a variety of directions and the range over which CP could be shifted relative to the base of support in both cardinal plane and diagonal directions. Specifically, parameters compared included the mean position of the CP on the X and Y axes (CPX and CPY) and the magnitude of postural sway as indicated by the variance of the mean position of the CP (Disp. and the % of test time within 5%, 10%, 20%, 40%, 60% and 80% of body weight displaced from the mean CP).

Relationships between the gait and balance variables were then explored in the hemiplegic sample by examining correlation coefficients. Pearson product moment correlations were calculated between all balance and gait variables. The research plan anticipated the possibility of making definitive statements about the strength of such relationships which, prior to the study, were expected to exist between gait kinematic factors and balance performance in the hemiplegic subject sample. It was also anticipated that statements would be forthcoming concerning which of the two balance measurement protocols (cardinal or diagonal) were better correlated with gait performance.

STATISTICAL CONFIDENCE

The 0.05 level of probability was used throughout this study as the level of significance. Research in clinical areas which have addressed patient populations commonly use this level of significance (Hogan *et al.*, 1987; Wall and Turnbull, 1987). Although this means that there is still a 5 in 100 chance that a Type I error could occur, in studies relevant to physiotherapy, this has been proposed as an acceptable level of risk (Currier, 1990).

The risk of failing to reject a false hypothesis, a Type II error, is dependent upon the number of subjects tested. Given that a relatively small number of subjects were studied (n = 40) such a possibility is moderately high (Goslin, 1985). However, increasing the numbers, particularly in the hemiplegic group, would have increased the variability of the performance of those studied. Such an action would have compromised the validity of the study. As a result of these considerations, a 0.05 level of significance was felt to be a realistic and pragmatic level of probability.

SELECTION OF SUBJECTS

A total of 40 subjects were selected for this study, 20 of whom had suffered a stroke and 20 who were normal. The stroke subjects were recruited from community based stroke support groups while the normal subjects were identified from senior citizens groups and employees at a local university. Both samples consisted of 12 males and 8 females. Eight subjects in the stroke group exhibited right hemiplegia, indicating left hemispheric brain damage, while the remaining 12 had left sided symptoms resulting from right hemispheric insult. Suitability for qualification for the study was based upon the previously described inclusion/exclusion criteria for both groups and a willingness to participate. Each subject provided written consent, in keeping with the review of the project by the Ethical Review Committee, Faculty of Health Professions, Dalhousie University which approved the study in terms of experimentation using human subjects (Appendix 3).

SUBJECT CHARACTERISTICS

Data concerning age, height, weight and sex for all subjects who participated in this study, were examined. In addition, the side of the hemiplegia and the time elapsed since the stroke were scrutinised. Comparisons were made between subjects in the following manner.

1. All subjects by condition (normal or hemiplegic) (Table 3.I).
2. All subjects by sex (Table 3.II).
3. Normal subject characteristics by sex (Table 3.III).
4. Hemiplegic subject characteristics by sex (Table 3.IV).
5. Hemiplegic subject characteristics by side of hemiplegia (Table 3.V).
6. Male subject characteristics by condition (Table 3.VI).
7. Female subject characteristics by condition (Table 3.VII)

Data were compared using unpaired, two-tailed t tests to determine differences between the hemiplegic and normal subjects studied in this investigation as well as the sub groups within the samples.

No differences were found between the samples in terms of age, height and weight between the groups and the sex composition of both groups was the same (12 males and 8 females).

When all subjects were considered, more males were included (24 to 16 respectively). Expected differences in height and weight between males and females were found with the males being both taller and heavier than the females. These differences were to be expected as a result of gender dimorphism (Goslin, 1985). There was no difference in age between the groups.

When the hemiplegic sample was examined on the basis of sex, no differences were found between males and females in age and time since stroke . Surprisingly, there was no difference in body weight between males and females. Although the mean value for the males appeared to be substantially greater than that of the females (81.93 kg and 71.02 kg respectively), standard deviations and ranges were large in both groups indicating variability within both groups. Differences in height were clearly evident with the males taller than the females.

When only the control group was examined, on the basis of sex, differences were again found in height and weight. Again these differences were to be expected (Goslin, 1985). There was no difference in age between the males and females in the normal sample.

The hemiplegic sample was also scrutinized on the basis of the side of the body affected by the hemiplegia. There were 12 left sided hemiplegics compared with 8 whose stroke had affected the right side of the body. No differences were found between those sub groups regarding age, height, weight and time since stroke.

Subgroups based on sex were also compared between the hemiplegic and normal groups. No differences were detected between the hemiplegic and normal males included in the study in age, height and weight. The same situation was found when the hemiplegic and normal females were compared regarding age, height and weight.

As a result of the tests performed on the data concerning the characteristics of the two samples studied, it can be proposed that the two groups were matched thus permitting comparisons to be made between the hemiplegic sample and the control group.

The age range of the subjects was greater than originally intended. This resulted from the rigorous inclusion/exclusion criteria used to select the hemiplegic sample. As a result of these criteria, it was necessary to recruit younger subjects who were capable of functional ambulation. This meant that the mean age and the range of ages were lower and greater respectively than initially forecasted. The characteristics of the normal group demonstrated similar characteristics as the controls and were recruited with a view to matching the hemiplegic sample in terms of age and sex.

Considerable variability was found in the hemiplegic group regarding the time since the stroke. Despite this, the inclusion/exclusion criteria ensured that the hemiplegic group was relatively homogeneous particularly relating to gait performance, a major factor being considered in this study. The appearance of variability between hemiplegic subjects, even in conditions where careful control was attempted, has been identified in earlier studies (Knutsson and Richards, 1979; Wall and Turnbull, 1987; Richards, 1990) and has been described as an impediment to research in the area of stroke (Bach-y-Rita, 1983). As a result, it was found to be difficult to recruit a sample of subjects without some degree of inter subject variability. However, given that an age and sex-matched control group was used in this study and given the variables being examined it is proposed that the subjects studied were appropriate to answer the research questions posed and permitted the generation of practical suggestions concerning the rehabilitation implications of the findings.

Table 3.I: General subject characteristics.

	Hemiplegic			Normal			
Parameter	Mean	SD	(Range)	Mean	SD	(Range)	p
Number		20			20		
Age (yr.)	57.2	10.7	(30 - 77)	61.5	13.0	(33 - 84)	-
Stature (cm)	170.2	9.8	(151 - 185)	171.6	10.00	(155 - 191)	-
Mass (kg)	77.6	15.7	(52.7 - 114.5)	77.7	15.1	(46.4-101.8)	-
Sex	12 males 8 females			12 males 8 females			n/a
Years Since Stroke	10.7	5.6	(1.4 - 20.0)	Not Applicable			n/a

Table 3.II: General subject characteristics by sex.

	Male			Female			
Parameter	Mean	SD	(Range)	Mean	SD	(Range)	p
Number	24			16			
Age (yr.)	59.0	13.0	(30 - 84)	59.9	10.5	(42 - 79)	-
Stature (cm)	177.4	5.6	(165 - 191)	161.2	5.9	(151 - 171)	< 0.05
Mass (kg)	84.2	12.8	(60.9-114.5)	67.9	13.5	(46.4-96.4)	< 0.05
Sex	12 males 8 females			12 males 8 females			n/a

Table 3.III: Normal subject characteristics by sex

	Males			Females			
Parameter	Mean	SD	(Range)	Mean	SD	(Range)	p
Number	12			8			
Age (yr.)	60.6	14.2	(33 - 84)	62.9	11.6	(45 - 79)	-
Stature (cm)	178.3	5.6	(169 - 191)	161.5	5.2	(155 - 171)	< 0.05
Mass (kg)	86.4	10.8	(60.9-101.8)	64.8	10.7	(46.4-81.8)	< 0.05

Table 3.IV: Hemiplegic subject characteristics by sex

	Males			Females			
Parameter	Mean	SD	(Range)	Mean	SD	(Range)	p
Number	12			8			
Age (yr.)	57.3	12.0	(30 - 77)	57	9.0	(42 - 73)	-
Stature (cm)	176.4	5.5	(165 - 185)	160.9	6.8	(151 -170)	< 0.05
Mass (kg)	81.9	14.6	(62.7-114.6)	71.0	15.9	(52.7-96.4)	-
Years Since Stroke	11.12	5.5	(1.3 - 20.0)	10.0	5.95	(3.0 - 20.0)	-
Hemiplegic Side	Left - 7 Right - 5			Left - 5 Right - 3			n/a

Table 3.V: Hemiplegic subject characteristics by side of hemiplegia.

	Left			Right			
Parameter	Mean	SD	(Range)	Mean	SD	(Range)	p
Number	12			8			
Sex	7 Male	5 Female		5 Male	3 Female		
Age (yr.)	54.4	11.6	(30 - 73)	61.4	7.8	(53 - 77)	-
Stature (cm)	170.4	11.2	(151 - 185)	169.9	8.0	(154 - 178)	-
Mass (kg)	81.5	15.1	(55.5-114.5)	71.6	15.6	(52.7-96.4)	-
Years Since Stroke	11.17	4.6	(4.0 - 20.0)	9.9	7.0	(1.4 - 20)	-

Table 3.VI: Male subject characteristics

	Hemiplegic			Normal			
Parameter	Mean	SD	(Range)	Mean	SD	(Range)	p
Number	12			12			
Age (yr.)	57.3	12.0	(30 - 77)	60.6	14.2	(33 - 84)	-
Stature (cm)	176.4	5.5	(165 - 185)	178.3	5.6	(169 - 191)	-
Mass (kg)	81.9	14.6	(62.7-114.6)	86.4	10.8	(60.9-101.8)	-

Table 3.VII: Female subject characteristics

	Hemiplegic			Normal			
Parameter	Mean	SD	(Range)	Mean	SD	(Range)	p
Number	8			8			
Age (yr.)	57	9.0	(42 - 73)	62 .9	11.6	(45 - 79)	-
Stature (cm)	160.9	6 .8	(151 - 170)	161.5	5.2	(155 - 171)	-
Mass (kg)	71.0	15.9	(52.7-96.4)	64 .8	10.7	(46.4-81.8)	-

CHAPTER 4

RESULTS AND DISCUSSION

THE TEMPORAL AND SPATIAL PARAMETERS OF GAIT

Walking Speed:

Table 4.I shows the velocity data (mean, standard deviation and range) obtained for both hemiplegic and normal groups at the five self-selected walking speeds. The results are presented in both absolute (m/s) and relative (statures/s) terms. These results are shown graphically in Figure 4.1.

The two factor repeated measures ANOVA identified significant differences between type of subject (normal and hemiplegic), the different walking conditions and indicated that a significant interaction existed between the two factors. Repeated measures on one factor for the normal group revealed significant differences between all walking conditions in the normal group (Table 4.II) and significant differences between most of the walking conditions for the hemiplegic group (Table 4.III). Comparison of the same walking conditions between the two groups showed clear differences with the normal group walking significantly faster at all conditions than the hemiplegic sample. Comparison of the "slowest" walking condition for the control group and the "fastest" condition in the hemiplegic group resulted in an insignificant p value (0.0544) indicating that the "fastest" walking speed of the hemiplegic group was not statistically different from the "slowest" walking speed of the normal sample.

Table 4.I Comparison of the data obtained for walking speed (m/s) and relative velocity (stat/s) between hemiplegic and normal subjects for each self-selected walking speed.

	Hemiplegic			Normal			
	Mean	SD	(Range)	Mean	SD	(Range)	p
Slowest (stat/s)	0.21	0.06	(0.11-0.33)	0.38	0.11	(0.19 -0.60)	< 0.05
(m/s)	0.35	0.10	(0.2 - 0.59)	0.66	0.20	(0.30 -1.10)	
Slow (stat/s)	0.26	0.10	(0.13-0.53)	0.54	0.12	(0.38 - 0.83)	< 0.05
(m/s)	0.45	0.17	(0.22-0.82)	0.93	0.21	(0.64 - 1.32)	
Free (stat/s)	0.34	0.13	(0.14-0.69)	0.74	0.09	(0.56 - 0.91)	< 0.05
(m/s)	0.58	0.21	(0.24 - 1.1)	1.26	0.17	(1.00 - 1.56)	
Fast (stat/s)	0.43	0.16	(0.17-0.74)	0.88	0.17	(0.64 - 1.22)	< 0.05
(m/s)	0.73	0.27	(0.29-1.18)	1.51	0.31	(1.06 - 1.95)	
Fastest (stat/s)	0.50	0.23	(0.21-0.94)	1.07	0.20	(0.78 - 1.62)	< 0.05
(m/s)	0.85	0.40	(0.36-1.65)	1.83	0.39	(1.34 - 2.91)	

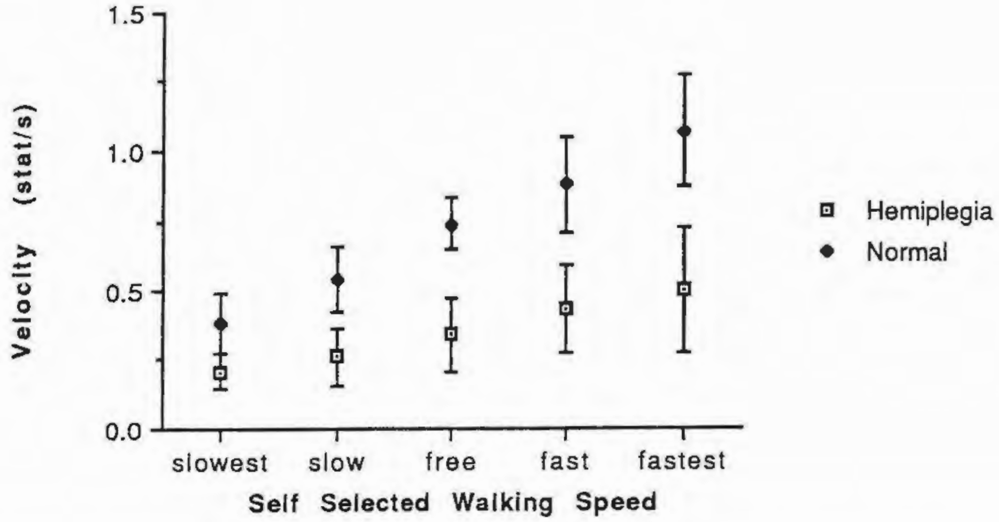

Figure 4.1: Relative speed mean values and standard deviations for both normal and hemiplegic samples for the five self-selected walking speeds.

Table 4.II: Statistically significant differences in walking speeds between conditions in the control group.

	Slow	Free	Fast	Fastest
Slowest	$p < 0.05$	$p < 0.05$	$p < 0.05$	$p < 0.05$
Slow	n/a	$p < 0.05$	$p < 0.05$	$p < 0.05$
Free		n/a	$p < 0.05$	$p < 0.05$
Fast			n/a	$p < 0.05$

Table 4.III: Statistically significant differences in walking speeds between conditions in the hemiplegic group.

	Slow	Free	Fast	Fastest
Slowest	-	$p < 0.05$	$p < 0.05$	$p < 0.05$
Slow	n/a	-	$p < 0.05$	$p < 0.05$
Free		n/a	$p < 0.05$	$p < 0.05$
Fast			n/a	-

With regard to the control group, walking speed increased in a manner to be expected given the nature of the instructions issued to the subjects. Statistical analysis detected differences between all walking conditions. Thus, the normal subjects possessed the capability of consciously varying their walking speed between all conditions requested. Therefore, of the five speeds requested, the normal subjects were able to clearly demonstrate all five.

When the data relating to the hemiplegic sample were considered, no differences in walking speed were found between the "slowest" and "slow", between the "slow" and "free", and between "fast" and "fastest" walking test conditions. In other words only one of the adjacent walking condition pairs, that between the “free” and “fast” speeds, was significantly different. This would tend to suggest that when compared to the control group, there was a limited ability of the hemiplegic group to consciously vary walking velocity.

Variability within the normal sample, as indicated by the standard deviations and the ranges, was least when the subjects walked at their self-selected free speed and increased as the subjects deviated from this preferred speed particularly at the higher velocities. The variability of walking speed in the hemiplegic sample increased as the subjects walked faster. This increasing variability with increasing walking speed appeared to be proportional. This possibly resulted from the variability within the hemiplegic sample where some subjects were able to alter walking speed while others had very limited capability in this regard. Figure 4.2 compares the extreme values of walking velocity (range) for each test condition in both samples studied. In the normal group, the differences between the minimum and maximum values with increasing walking speed are almost parallel with the exception of the "fastest" value whereas, in the hemiplegic sample, the difference between the minima and maxima increased with increasing walking speed.

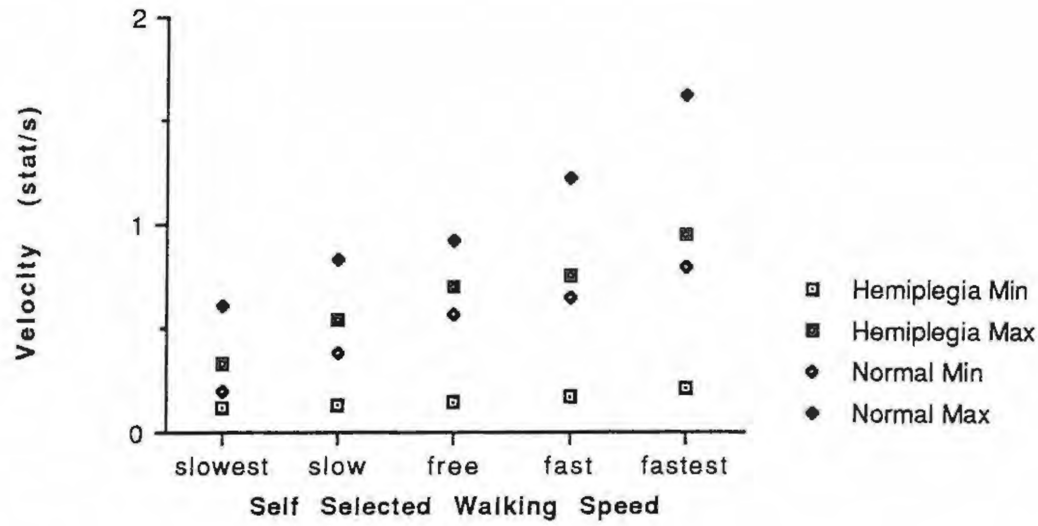

Figure 4.2: The minimum and maximum values obtained for each walking condition for both samples.

In order to examine this phenomenon more closely, the slowest walking speed for each subject was subtracted from the fastest walking speed to determine the ranges of walking speed of which the subjects were capable. The mean range-of-walking-speed for the hemiplegic sample was 0.299 stat/s $\pm$ 0.197 (range = 0.07 - 0.68) while that for the control group was 0.661 stat/s $\pm$ 0.205 (range = 0.44 - 1.13). When subjected to statistical analysis, these differences were found to be significant. Figure 4.3 compares all range-of-walking-speed data between the hemiplegic and control groups. The data were sorted in descending range of walking velocity to facilitate comparison. It can be seen that only six of the stroke subjects had ranges-of-walking-speed in the range of the normal subjects while the remainder were below normal range. Thus, some of the hemiplegic subjects (n = 6), perhaps those with the least amount of residual gait disability, possessed "normal" ranges-of-walking-speed while the majority (n = 14) did not. It is likely that, despite carefully controlling the hemiplegic subjects who qualified for inclusion in this study, inter-subject variability within the hemiplegic sample may have led to the increasing variability found as walking speed increased. However, this finding should be considered in the context that the hemiplegic sample clearly walked slower than the normal group.

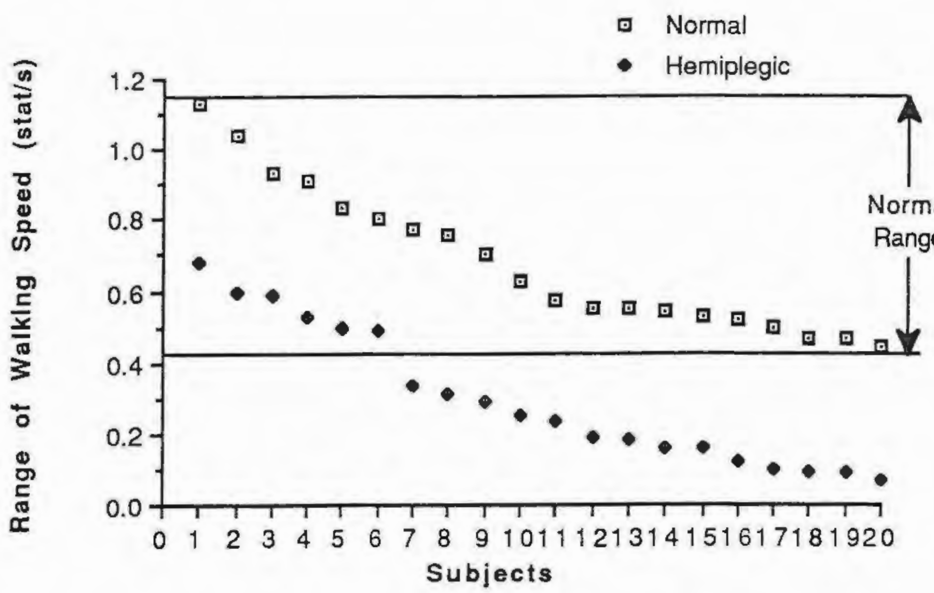

Figure 4.3: Comparison of ranges of walking speed for all subjects. Data were sorted in descending order to facilitate comparison.

Thus, the hemiplegic group walked significantly slower at all self-selected walking speeds than the normal sample. This finding confirms previous work that stroke subjects walk slower than healthy subjects (Knuttson and Richards, 1979; Wall and Turnbull, 1986; Dettman *et al.*, 1987). However, these earlier studies had based that conclusion on information gained at one walking speed which was the preferred speed of these subjects. By walking the subjects at different speeds, as was the case in this study, a greater understanding of the scope of this deficiency in walking speed became apparent. These findings provided stark evidence of the extent of the problem relating to the velocity characteristics of hemiplegic gait. It appeared that the stroke subjects not only walked slower than the controls but only possessed the ability to walk at a velocity which could, in comparison to the normal group, be considered to be very slow even when they attempted to walk as fast as possible. It is likely that such a limitation would compromise the ability of these subjects to cope adequately with their environment, for example, in safely crossing a street in the time provided by a traffic signal.

Although it was useful to examine self-selected walking speeds from slowest to fastest, such comparisons had limitations because the perceptions of "slow", "fast" etc. were specific to each subject. There was no way to control that a given perception of the velocity of gait was consistent between subjects. As such, the limitations of comparing psychometrically derived values were fully recognized. However, in this study, the procedure was designed to yield a range of walking speeds and increments between these extreme values. In this way, it was possible to examine temporal and spatial gait parameters at different speeds of walking. This was useful because it was known that gait velocity influences the other gait parameters, both temporal and spatial (Rosenrot *et al.*, 1980). As a result of the generation of data at different gait velocities, it was possible to compare these other parameters both directly and as a function of gait velocity and compare the performance of the hemiplegic subjects with that of age- and sex-matched controls.

Stride Time:

Table 4.IV shows the mean values, standard deviations and ranges for stride time obtained at each of the self-selected walking speeds for both groups of subjects. These results are shown graphically in Figure 4.4.

Table 4.IV: Data obtained for stride time (s) between hemiplegic and normal subjects for each self selected walking speed.

	Hemiplegic			Normal			
	Mean	SD	(Range)	Mean	SD	(Range)	p
Slowest	1.89	0.36	(1.31-2.78)	1.60	0.28	(1.24-2.28)	< 0.05
Slow	1.70	0.32	(1.26-2.38)	1.30	0.16	(0.94-1.64)	< 0.05
Free	1.53	0.30	(1.12-2.20)	1.08	0.11	(0.87-1.33)	< 0.05
Fast	1.35	0.28	(1.05-2.00)	0.98	0.12	(0.65-1.30)	< 0.05
Fastest	1.27	0.31	(0.85 -1.99)	0.87	0.12	(0.60-1.12)	< 0.05

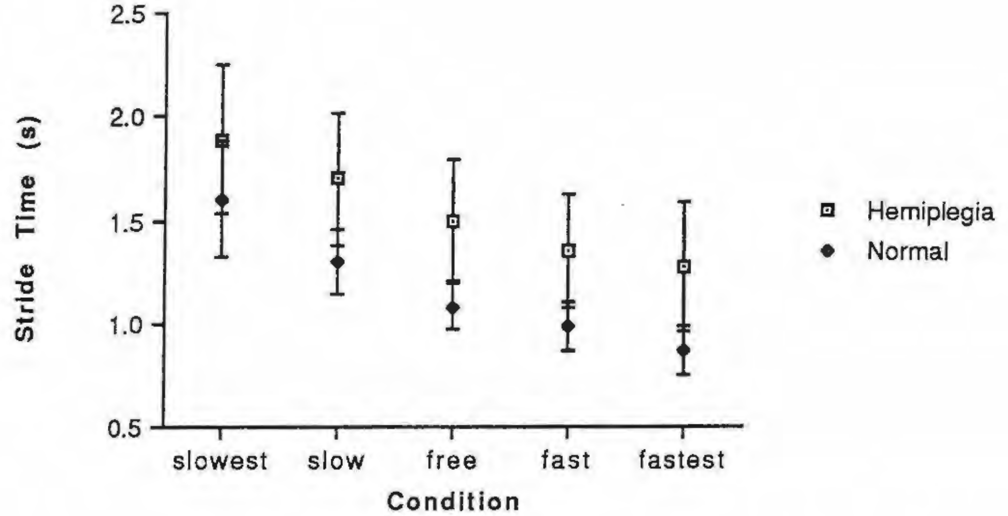

Figure 4.4: Stride time mean values and standard deviations for both normal and hemiplegic samples for the five self-selected walking speeds.

Two factor repeated measures ANOVA revealed that significant differences existed between the two subject types and in the repeated measures of the five walking speeds. However, unlike the findings for walking speed, the interaction between the two factors was not significant indicating that stride time decreased in a similar manner in both groups with increasing walking speed.

In the normal sample repeated measures ANOVA (Table 4.V), revealed significant differences in stride time for all pairwise combinations except for the stride times between the "free" and "fast" and the "fast" and "fastest" conditions for which no significant differences were found. When considering the effect of walking speed on stride time, there were differences between all conditions for relative speed but no differences between "free" and "fast" as well as "fast" and "fastest" for stride time. Therefore, the lack of a significant differences between these trials cannot be explained in terms of changes in walking speed.

Table 4.V: Statistically significant differences in stride time at each self-selected walking speed in the control group.

	Slow	Free	Fast	Fastest
Slowest	$p < 0.05$	$p < 0.05$	$p < 0.05$	$p < 0.05$
Slow	n/a	$p < 0.05$	$p < 0.05$	$p < 0.05$
Free		n/a	-	$p < 0.05$
Fast			n/a	-

Table 4.VI: Statistically significant differences in stride time at each self-selected walking speed in the hemiplegic group.

	Slow	Free	Fast	Fastest
Slowest	$p < 0.05$	$p < 0.05$	$p < 0.05$	$p < 0.05$
Slow	n/a	$p < 0.05$	$p < 0.05$	$p < 0.05$
Free		n/a	$p < 0.05$	$p < 0.05$
Fast			n/a	-

When the stride time values for the hemiplegic sample were scrutinized (Table 4.VI), significant differences were detected between all pairwise combinations except between stride time at the "fast" and "fastest" conditions. This was some also somewhat consistent with the findings for walking speed with only the significant differences detected between the "slowest" and "slow" conditions and the "slow" and "free" conditions being different from those found for walking speed.

When the stride times for each of the walking speeds were compared between the groups, significant differences were detected between the groups for each of the same walking conditions. Further, the stride time for the hemiplegic group at the "fastest" walking trial was almost the same as the stride time for the control group for the "slow" trial. It is likely that these differences between the two groups can be explained by the fact that the hemiplegic subjects walked slower than their normal counterparts. However, the discrepancy between the findings for walking speed and stride time may have been due to the different strategies being used to increase walking speed e.g. increasing stride length rather than reducing stride time. These stride time values were plotted against relative velocity in Figure 4.5 allowing further comparison to be made between the hemiplegic and control groups.

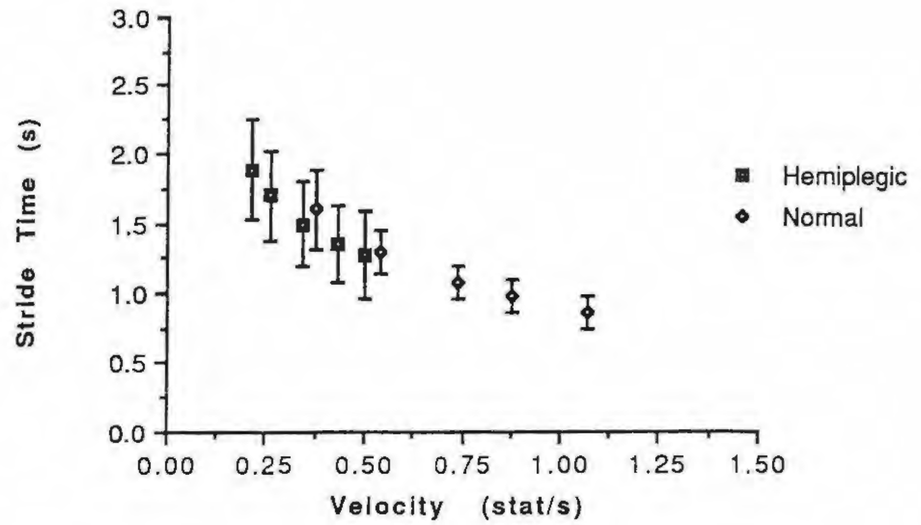

Figure 4.5: Mean values and standard deviations for stride time plotted against relative speed in both normal and hemiplegic samples.

The shape of the graphs for both hemiplegic and control groups are curvilinear and appear somewhat similar although the hemiplegic group demonstrated slower stride times overall probably as a result of the fact that they walked slower than the controls. The absence of an interaction between type of subject and walking speed under the various gait testing conditions would tend to support the contention that the curves generated were similar. The normal group demonstrated greatest variability at the "slowest" speed than any of the other walking speeds for this group. This was probably the result of the subscription to different gait strategies when asked to walk at the "slowest" speed, a velocity which is probably infrequently utilized under normal situations. In the hemiplegic group, variability, as indicated by the standard deviation values, was consistently large and greater than that of the normal group. As suggested earlier, this may have been due to the increased variability within the hemiplegic sample regarding their gait performance compared to the control group.

Stride Length:

Table 4.VII shows the stride length data (mean, standard deviation and range) both in absolute (cm) and relative (stat/str) terms. The results for relative stride length are shown graphically in Figure 4.6.

Table 4.VII: Comparison of the data obtained for stride length between hemiplegic and normal subjects for each self selected walking speed.

	Hemiplegic			Normal			
	Mean	SD	(Range)	Mean	SD	(Range)	p
Slowest (stat/str.)	0.38	0.09	(0.2 - 0.5)	0.58	0.13	(0.31 - 0.77)	$p < 0.05$
(cm)	65.30	14.7	(40.4 - 87.3)	101.90	21.8	(64.3 - 139.2)	
Slow (stat/str.)	0.43	0.12	(0.3 - 0.7)	0.69	0.10	(0.54 - 0.84)	$p < 0.05$
(cm)	72.80	19.40	(44.1-109.0)	119.20	20.5	(89.4 - 151.4)	
Free (stat/str.)	0.49	0.12	(0.3 - 0.8)	0.79	0.08	(0.64 - 0.92)	$p < 0.05$
(cm)	83.70	20.10	(45.8-123.7)	135.8	18.10	(103.7-165.5)	
Fast (stat/str.)	0.54	0.14	(0.3 - 0.8)	0.85	0.12	(0.63 - 1.03)	$p < 0.05$
(cm)	92.50	23.80	(51.7-128.8)	147.10	25.5	(104 - 184.8)	
Fastest (stat/str.)	0.58	0.17	(0.37 - 0.85)	0.91	0.11	(0.78 - 1.11)	$p < 0.05$
(cm)	98.2	28.9	(62.1-146.7)	157.00	25.7	(123.3-199.1)	

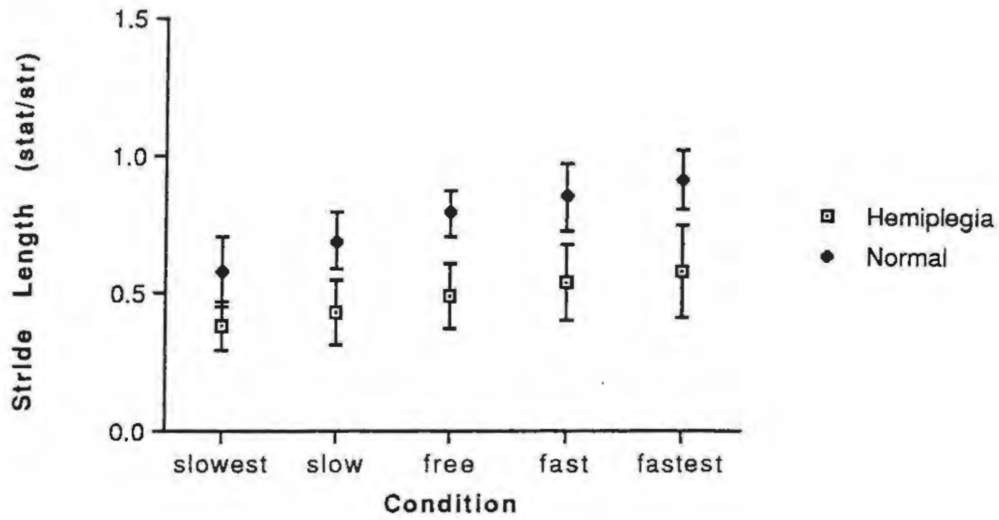

Figure 4.6: Mean values and standard deviations for relative stride length at each self-selected walking speed for both hemiplegic and normal samples.

Two factor, repeated measures ANOVA revealed significant differences between the type of subject, the different walking conditions and a significant interaction between the two factors.

In the normal group, *post hoc* analysis revealed that there were significant differences between all of the walking conditions except between the "fast" and the "fastest" trials (Table 4.VIII). In the hemiplegic sample, no differences were found between the "slowest" and the "slow trials, the "free" and the "fast" and the "fast" and the "fastest". Significant differences were found between all other pairwise combinations (Table 4.IX). Comparison between the two groups showed significant differences at each of the walking trials with the normal groups taking longer strides at all walking trials. The relative stride length for the hemiplegic group at the "fastest" condition was almost identical to that found for the control group at the "slowest" walking trial providing a stark contrast between the relative stride lengths of the two groups. Many of these findings could be explained by the differences between the groups in walking speed and the fact that the normal group walked faster than the hemiplegic sample.

Table 4.VIII: Statistically significant differences in relative stride length at each self-selected walking speed in the control group.

	Slow	Free	Fast	Fastest
Slowest	$p < 0.05$	$p < 0.05$	$p < 0.05$	$p < 0.05$
Slow	n/a	$p < 0.05$	$p < 0.05$	$p < 0.05$
Free		n/a	$p < 0.05$	$p < 0.05$
Fast			n/a	-

Table 4.IX: Statistically significant differences in relative stride length at each self-selected walking speed in the hemiplegic group.

	Slow	Free	Fast	Fastest
Slowest	-	$p < 0.05$	$p < 0.05$	$p < 0.05$
Slow	n/a	$p < 0.05$	$p < 0.05$	$p < 0.05$
Free		n/a	-	-
Fast			n/a	-

It was, therefore, clear that the performance of the hemiplegic subjects was considerably different overall from the normal sample with the same general finding that the stride length of the hemiplegic sample was shorter in comparison to the control group. Thus more evidence was produced to demonstrate the poverty of the adaptability of the hemiplegic gait pattern.

Figure 4.7 shows relative stride length plotted against relative walking velocity which permitted further comparison between the hemiplegic and normal subjects' performance.

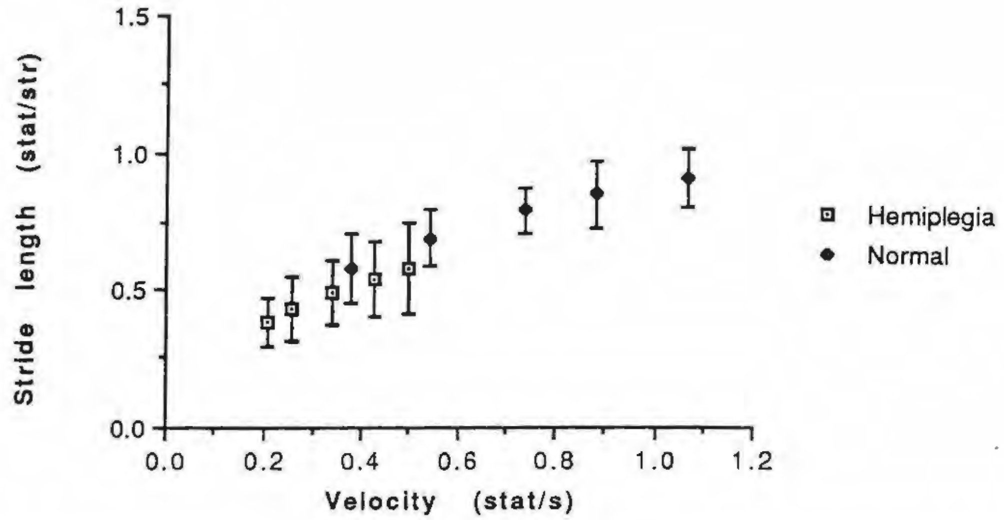

Figure 4.7: Comparison of relative stride length between the hemiplegic and normal sample plotted against relative velocity.

As expected, both groups demonstrated increasing stride length with increasing walking speed. However, like the measures discussed earlier, the hemiplegic stride length was less than that of the controls with only the stride lengths of the two faster speeds generated by the stroke subjects overlapping the stride length of the slowest speed of the normal group.

Stride Time and Stride Length:

It is known that there is a clear relationship between gait velocity and stride time and stride length. As walking speed increases, stride time diminishes while stride length increases (Rosenrot *et al.*, 1980)). Thus, many of the findings for stride time and stride length could be accounted for by the simple fact that walking speed was increasing or decreasing. Figure 4.8 plots both stride time in s and stride length in m against relative speed for both samples with a view to examining the interaction of stride time and stride length with regard to walking speed. It can be seen that the shapes of the curves for the hemiplegic sample was similar to those of the control group except that neither stride time and stride length were of sufficient range given the range of walking speeds available. The normal values intersected at approximately 0.5 stat/s before diversifying again with increasing speed leading to shorter stride times and longer stride lengths. While the same was also true of the hemiplegic sample, the intersection did not occur and stride time and stride length remained restricted apparently by the limitations of the gait velocity to below 0.6 stat/s. Thus, the critical difference appeared to be that subjects of the stroke sample were unable to attain the stride times or the stride lengths of the normal subjects because they were unable to walk as fast. The converse of this statement may also have been true, that is, that the stroke sample was unable to walk as fast as the controls because they were unable to take quick enough or long enough strides. However, when the hemiplegic stride times were compared with those of the controls at the speeds which were common to both groups (the "free", "fast" and "fastest" for the hemiplegic group and the "slowest" and "slow" for the controls), the stride times for the hemiplegic sample were less than that of the controls. The hemiplegic sample also took shorter stride lengths at these speeds (Fig 4.7) Given that gait velocity = stride length/stride time and the velocity was similar, this finding would suggest that the hemiplegic sample took relatively quick and short steps suggesting an urgency to resume the double support phases of the gait cycle which are more stable phases of the gait cycle. Therefore, at the walking speeds which were common to both groups, the hemiplegic sample took quicker and shorter strides which is suggestive of a "mincing" gait pattern, a term used by Wall *et al.*,(1986) to describe the cautious gait of elderly people with balance problems. As a result, this profile lends further support to the contention that the balance of hemiplegic subjects is deficient.

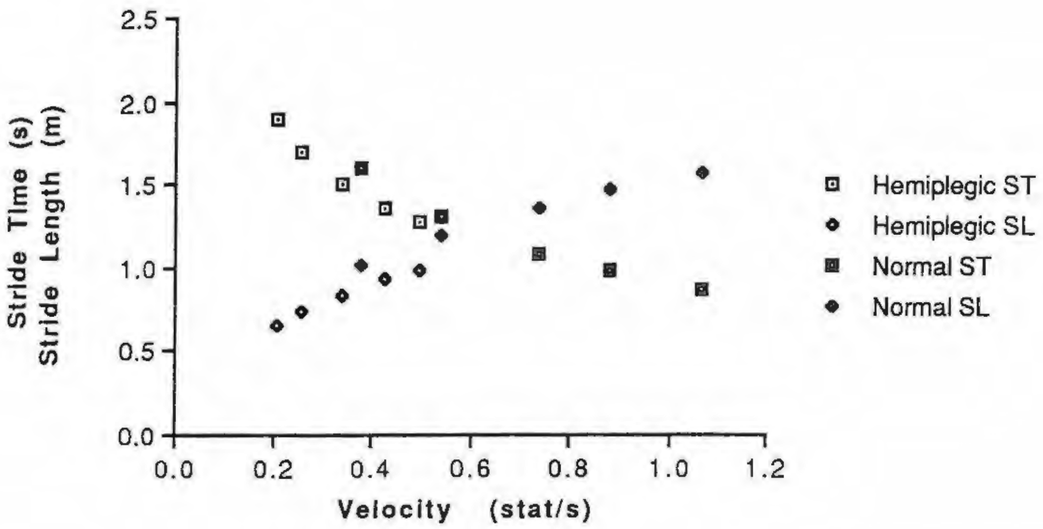

Figure 4.8: Stride time (ST) and stride length (SL) plotted against velocity for both the hemiplegic and normal samples.

Duration of Phases of the Gait Cycle:

Total Support:

Table 4.X shows the data (mean, standard deviation and range) for the duration of total support, expressed as a percentage of stride, spent on both legs during gait analysis at different speeds.

Table 4.X: Comparison of the data obtained for the duration of total support as a percentage of stride between hemiplegic (Un; Aff) and normal subjects (R; L) for each self selected walking speed (Un = Unaffected leg; Aff = Affected leg; L = Left leg; R = Right Leg).

		Hemiplegic			Normal				
		Mean	SD	(Range)		Mean	SD	(Range)	p
Slowest	(Un)	78.5	5.5	(68.8-87.8)	(L)	65.7	4.2	(59.5-76.0)	$p < 0.05$
	(Aff)	64.7	7.3	(53.8-83.5)	(R)	66.3	3.6	(61.3-73.5)	-
Slow	(Un)	76.2	6.2	(66.4-86.7)	(L)	63.0	2.7	(58.5-68.0)	$p < 0.05$
	(Aff)	62.9	6.8	(52.8-82.9)	(R)	63.3	2.9	(58.9-68.4)	-
Free	(Un)	74.1	6.9	(61.9-85.6)	(L)	60.7	1.7	(57.8-63.9)	$p < 0.05$
	(Aff)	61.7	5.8	(53.2-81.1)	(R)	61.2	2.1	(56.3-64.3)	-
Fast	(Un)	72.7	7.6	(60.1-85.6)	(L)	59.5	2.7	(52.0-64.5)	$p < 0.05$
	(Aff)	60.2	5.7	(52.2-79.4)	(R)	60.2	2.3	(55.4-64.2)	-
Fastest	(Un)	71.0	6.9	(57.9-81.3)	(L)	58.1	2.0	(53.4-61.2)	$p < 0.05$
	(Aff)	59.6	6.2	(51.1-77.7)	(R)	57.9	2.8	(50.6-61.9)	-

Figure 4.9 shows these results graphically at each of the self selected walking speeds.

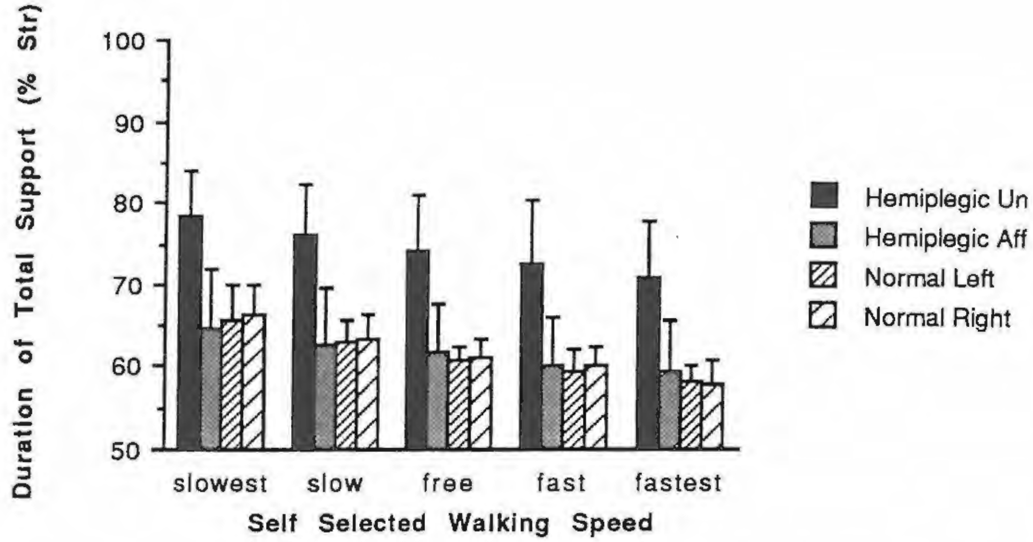

Figure 4.9: Mean and standard deviation for the duration of total support on both legs at each of the self selected walking speeds. The hemiplegic sample are classified by the leg affected (Aff) and unaffected (Un) by the hemiplegia while the normal group are classified by right leg and left leg.

Two factor repeated measures ANOVA revealed significant differences between the type of subject and the walking condition. A significant interaction between those two factors was also revealed.

Post hoc analysis for the normal group revealed significant differences between walking conditions except between the "free" and "fast" trials and the "fast" and "fastest" for the left leg. For the right leg the same results were found with the addition of an insignificant difference between the "slow" and "free" walks. No significant differences were found when the duration of total support for the right and left legs were examined.

In the hemiplegic sample, only two significant differences between walking conditions were found for both legs. These differences were found between the "slowest" walk and the "fast" and "fastest" trials for the unaffected leg. Significant differences were revealed between the duration of total support at all speeds between the unaffected and affected legs.

When the two groups were compared significant differences were found between the duration of total support for the unaffected leg and the data obtained for the normal subjects. However, there were no differences between total support for the affected leg in the hemiplegic group and normal.

Clear asymmetries existed in the hemiplegic group with a greater percentage of time spent in total support on the unaffected leg compared with the affected leg. This finding is consistent with previous work (Brandstater *et al.*, 1985; Wall and Turnbull, 1986). As in other components of this study, considerable variability was found in the hemiplegic sample. The normal subjects, as expected, were much less variable and more symmetrical, however, variability was greatest again when these subjects were walking at their slowest speed. It is likely that this was due to the unfamiliarity of this walking speed as well as the possibility that different motor strategies may have been used at this speed. Similarly, least variability was seen when the normal subjects walked at their "preferred" speed. This may have been due to familiarity as well as this velocity likely being each subject's most energy efficient walking speed. The asymmetry present in the hemiplegic sample is clearly evident. The values obtained from the normal sample and the affected leg of the hemiplegic group were very similar, an observation born out by the statistical analysis.

In comparison, the periods spent in total support on the affected hemiplegic limb appeared consistent with the time spent in total support in the normal group. However, this finding must be viewed in the context of the large amounts of time spent in this phase on the unaffected limb. These findings are again consistent with the findings of previous studies (Wall and Turnbull, 1985).

Figure 4.10 shows total support values for each leg from both samples plotted against relative walking velocity. There is a clear trend of diminishing duration of time spent in total support with increasing walking speed in characteristically linear fashion (Rosenrot *et al.*, 1980). Again this was an expected finding. The most striking finding remains, however, the marked asymmetry present in the gait pattern of the hemiplegic sample with a much greater time spent in total support on the unaffected limb than the affected limb. Conversely, the recordings for the normal group are essentially symmetrical.

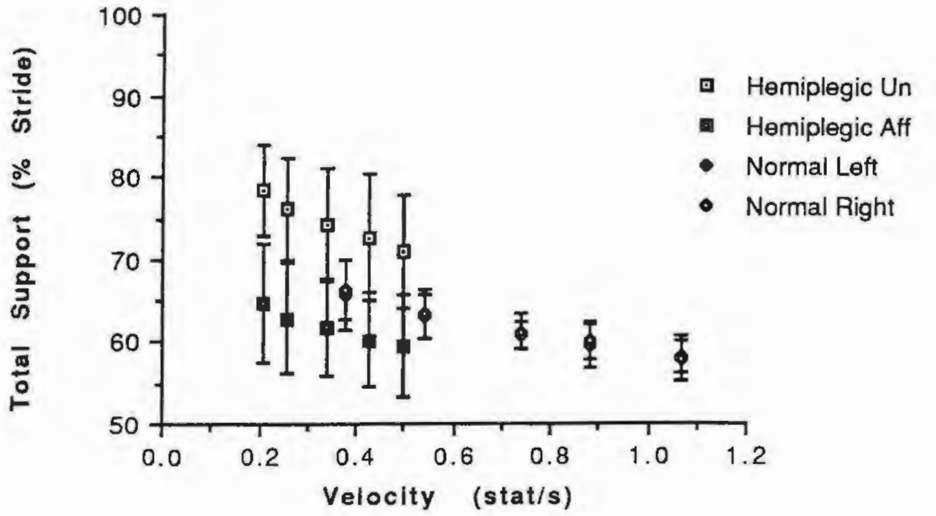

Figure 4.10: The duration of total support as a % of stride plotted against velocity for both lower limbs in both groups.

Single Support

Table 4.XI provides the mean values, standard deviations and ranges for the percentage of stride spent in single support of the gait cycle in both groups. These results are presented graphically in Figure 4.11.

Two factor repeated measures ANOVA on the single support data revealed significant differences for type of subject and between the different walking conditions. A significant interaction between the two factors was also detected. As expected, the normal group demonstrated the same location of significant differences between the duration of stride spent in single support and walking condition as was found for total support. Thus, differences were found between all single support durations except for those between the gait test conditions of "free" and "fast", "fast" and "fastest", both on the left leg and for "slow" and "free", "free" and "fast", and "fast" and "fastest" on the right leg. As expected, there were no differences between the durations of single support between the right and left legs at all walking conditions.

Table 4.XI: Comparison of the data obtained for the duration of single support as a percentage of stride between hemiplegic (Un; Aff) and normal subjects (R;L) for each self selected walking speed (Un = Unaffected leg; Aff = Affected leg; L = Left leg; R = Right Leg).

		Hemiplegic			Normal			
		Mean	SD	(Range)	Mean	SD	(Range)	p
Slowest	(Un)	34.7	7.5	(16.4 - 45.8)	(L) 33.8	3.7	(25.9 - 39.3)	-
	(Aff)	21.6	5.4	(13.6 - 31.0)	(R) 34.1	4.0	(24.0 - 39.9)	$p < 0.05$
Slow	(Un)	36.7	6.6	(17.2 - 46.8)	(L) 36.4	2.8	(30.9 - 40.2)	-
	(Aff)	23.9	6.3	(12.9 - 33.4)	(R) 37.1	2.8	(32.0 - 42.2)	$p < 0.05$
Free	(Un)	38.2	5.7	(18.9 - 46.9)	(L) 38.7	2.1	(35.3 - 43.5)	-
	(Aff)	25.8	6.9	(14.2 - 38.0)	(R) 39.3	1.8	(35.7 - 42.2)	$p < 0.05$
Fast	(Un)	39.4	5.4	(20.1 - 45.3)	(L) 39.8	2.5	(35.0 - 45.9)	-
	(Aff)	27.6	7.2	(14.8 - 38.9)	(R) 40.3	2.4	(35.9 - 47.2)	$p < 0.05$
Fastest	(Un)	40.3	6.1	(22.3 - 48.8)	(L) 41.9	2.6	(38.3 - 48.6)	-
	(Aff)	28.9	7.0	(18.5 - 1.6)	(R) 41.9	2.2	(38.5 - 47.2)	$p < 0.05$

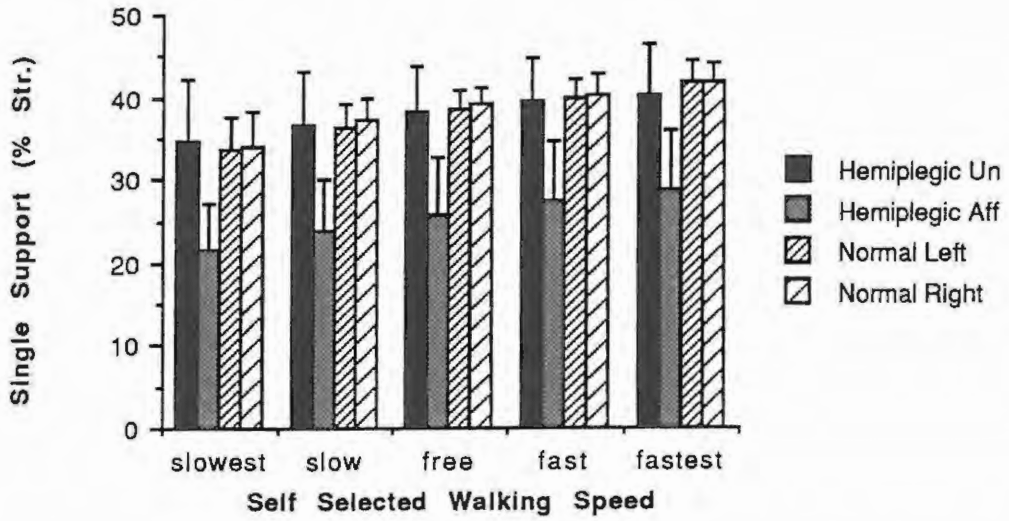

Figure 4.11: Duration of single support for both legs at each of the self selected walking speeds. The hemiplegic sample are classified by the leg affected (Aff) and unaffected (Un) by the hemiplegia while the normal group are classified by right leg and left leg.

In the hemiplegic sample no statistical differences were found except between the single support durations for the "fastest" walking condition and those for "slowest", "slow" and "free" on the unaffected leg and those between "slowest" and "fast" and "fastest" on the affected leg. Significant differences were identified between the single support durations for the unaffected and affected legs at all gait test conditions.

When the two groups were compared significant differences were found between the duration of single support for the unaffected leg and the data obtained for the normal subjects. Again, however, there were no differences between single support for the unaffected leg in the hemiplegic group and normal.

The hemiplegic sample demonstrated a markedly reduced single support duration on the affected limb compared with the control group and compared to the time spent in this phase on the unaffected leg. The expected asymmetry in this parameter was again clearly evident in the hemiplegic sample. It can be proposed that the inability and/or reluctance to bear weight through the hemiplegic leg resulted in the characteristic reduction in single support duration on the hemiplegic leg which is consistent with previous findings (Wall and Ashburn, 1979). In both samples, there was an expected increase in the duration of single support with increasing gait velocity. Variability was much more evident in the hemiplegic sample than the controls the latter of whom again displaying the greatest degree at the slowest walking speed.

Figure 4.12 shows the duration of single support as a percentage of stride plotted against relative speed. All values increased linearly in both samples with increasing walking speed as would be expected. However, the marked asymmetries in the hemiplegic group were clearly identifiable rendering the gait pattern of this group characteristically abnormal.

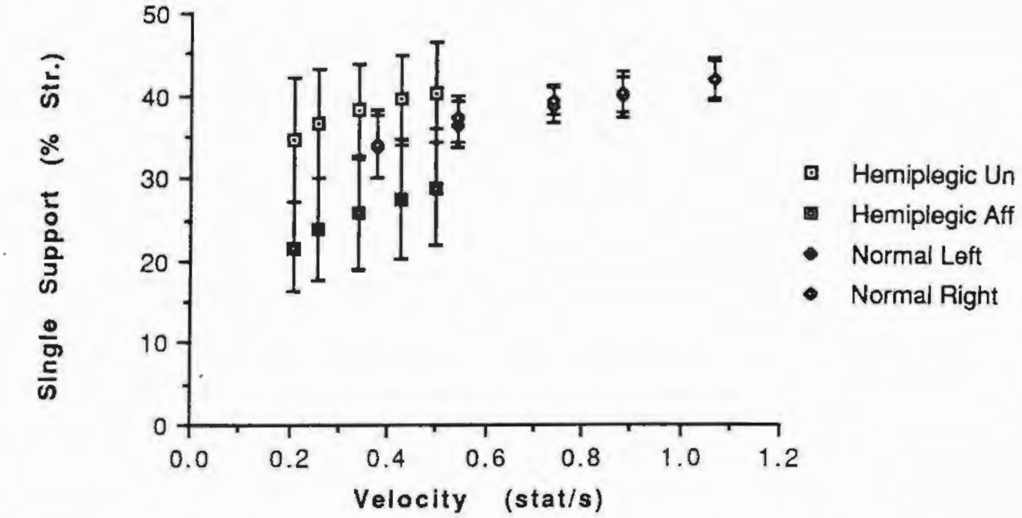

Figure 4.12 The duration of single support as a % of stride plotted against velocity for both lower limbs in both groups.

Braking Double Support:

Table 4.XII provides the mean values, standard deviations and ranges for the percentage of stride spent in double braking support of the gait cycle. These results are presented graphically in Figure 4.13.

Two factor repeated measures ANOVA revealed conducted on the data revealed significant differences both for the type of the subjects and the walking conditions. A significant interaction between these two factors was also detected.

Repeated measures performed on one factor (walking conditions) of the data for the normal group revealed significant differences between the durations of braking double support in seven of the ten possible comparisons for both legs. No differences were found between the data obtained for the "slow" and "free", "free" and "fast", and "fast" and"fastest" gait testing conditions for both legs. No differences were found between the durations of braking double support between the right and left legs at any of the walking conditions.

Table 4.XII: Comparison of the data obtained for the duration of braking double support as a percentage of stride between hemiplegic (Un;Aff) and normal subjects (L;R) for each self selected walking speed (Un = Unaffected leg; Aff = Affected leg; L = Left leg; R = Right Leg).

		Hemiplegic				Normal			
		Mean	SD	(Range)		Mean	SD	(Range)	p
Slowest	(Un)	23.1	7.3	(10.1 - 44.1)	(L)	15.6	4.3	(10.0-26.4)	$p < 0.05$
	(Aff)	19.7	5.4	(11.2 - 37.2)	(R)	16.2	3.7	(9.6 - 24.0)	$p < 0.05$
Slow	(Un)	20.9	6.6	(9.6 - 37.7)	(L)	12.9	2.7	(8.9 - 18.0)	$p < 0.05$
	(Aff)	18.1	6.4	(10.9 - 37.7)	(R)	13.3	2.7	(8.4 - 18.0)	$p < 0.05$
Free	(Un)	19.9	6.1	(10.6 - 37.5)	(L)	10.7	1.8	(7.1 - 14.4)	$p < 0.05$
	(Aff)	15.8	5.9	(9.1 - 36.7)	(R)	11.3	1.8	(7.9 - 15.3)	$p < 0.05$
Fast	(Un)	18.5	5.8	(9.7 - 30.7)	(L)	9.6	2.4	(4.2 - 14.1)	$p < 0.05$
	(Aff)	14.3	6.1	(8.4 - 33.5)	(R)	10.0	2.5	(3.0 - 14.6)	$p < 0.05$
Fastest	(Un)	17.7	6.5	(9.0 - 31.7)	(L)	7.9	2.6	(2.0 - 12.4)	$p < 0.05$
	(Aff)	13.1	5.7	(4.2 - 30.2)	(R)	8.0	2.3	(2.5 - 11.3)	$p < 0.05$

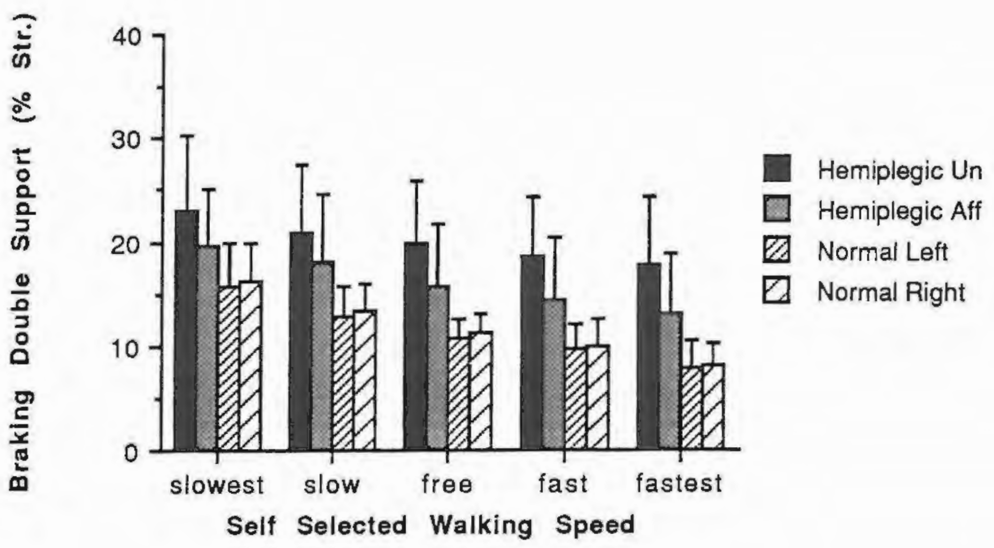

Figure 4.13: Mean and standard deviation for the duration of double braking support on both legs for each group at each of the self-selected walking speeds. The hemiplegic sample is classified by the leg affected (Aff) and unaffected (Un) by the hemiplegia while the normal group is classified by the right and left legs.

In the hemiplegic sample, *post hoc* analysis, revealed no significant differences between braking double support durations on the unaffected leg except for the comparison between the "slowest" and "fastest conditions. For the affected leg, seven of the ten comparisons were not significant. Only the comparisons between "slowest" and "fast", "slowest" and "fastest" and, "slow" and "fastest" walking conditions were significantly different. Comparison of the durations of braking double support between the unaffected and affected legs revealed no differences at any of the walking speeds.

When the data for the durations of braking double support under each of the five walking conditions were compared between the normal and hemiplegic samples, significant differences were found between the unaffected leg for the hemiplegic sample and the left leg of the normal group and the affected leg of the hemiplegic group and the right leg of the controls.

As expected, the durations of braking double support reduced with increasing walking speed in both groups. The hemiplegic sample demonstrated considerable asymmetries at all speeds while the control group remained essentially symmetrical but with most asymmetry at the slowest speed and least at the preferred speed. The asymmetry in the hemiplegic sample was clearly evident while the converse can be said to be true for the control group. Although it was found that there were no significant differences between the unaffected and the affected legs in the hemiplegic group, this was likely a function of the high level of variability in the hemiplegic sample. This probably also accounts for the differences between the samples which were detected for both limbs; a result not found for total or single support. This variability also may have been explained by the strong suspicion that there are probably several different types of hemiplegic gait pattern. Wall and Turnbull (1986) proposed that some hemiplegic subjects walked with a particular type of gait pattern that tended to accentuate a braking double support asymmetry. It may well have been the case that some but not all of the subjects in the present study exhibited this type of pattern. This would have had the effect of increasing the variability.

Figure 4. 14 shows the duration of braking double support for both lower limbs for both groups, plotted against relative velocity. As expected the normal performance is symmetrical and decreases in a curvilinear fashion with increasing walking speed and demonstrates least variability at the "free" walking speed condition. The duration of double braking support for the hemiplegic group was less than normal for the affected leg and greater than normal for the unaffected leg.

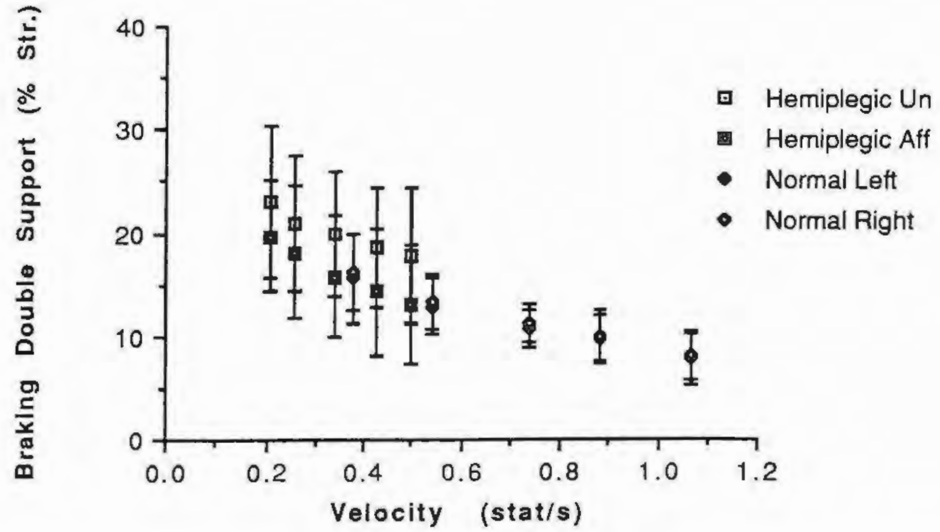

Figure 4.14: Mean and standard deviation for the duration of braking double support on both legs as a percentage of stride plotted against relative velocity. The hemiplegic sample is classified by the leg affected by the hemiplegia while the normal group are classified by right leg and left leg.

Thus, the hemiplegic group was considerably more asymmetrical than the control group when the durations of total support, single support and braking double support were considered. However, in all durations of the gait cycle which were examined, similar curves were obtained for both groups. The hemiplegic group spent greater periods of time in each of the three components of the gait pattern on the unaffected limb than the affected limb, so confirming previous research. Variability in this group was again a major feature probably as a result of the variability within the hemiplegic sample and as a function of the differential effects of the hemiplegia on the gait pattern lending further support to the previous findings by Knuttson and Richards (1979) and Wall and Turnbull (1986) that there exists several different sub groups of the so-called "hemiplegic gait pattern".

Asymmetry

As indicated in the previous section, asymmetry was clearly identified as a major feature of the hemiplegic gait pattern when the durations of total support, single support, braking double support and step length were considered. The effect of walking speed on these asymmetries was studied further by calculating an AR for selected components. The duration of time spent in single support has been the focus of considerable attention in studies examining hemiplegic gait (Wall and Ashburn, 1979; Brandstater et al, 1985; Wall and Turnbull, 1986; Dettman et al, 1987). The typical favouring of the hemiplegic leg, which has been attributed as a major cause of the asymmetrical nature of the hemiplegic gait pattern (Wall and Ashburn, 1979), has been shown to be caused by asymmetries in the single support phase. Similarly, double support asymmetries have been identified as contributing to the asymmetrical nature of the hemiplegic gait pattern (Wall and Turnbull, 1986). A further asymmetry which provides useful clinical information is that of step length. Thus, these parameters were further examined using the AR technique to identify the magnitude of the asymmetries. This study, however, differed from previous work in that the asymmetries at a variety of walking speed were examined thus permitting comments to be made concerning the effect of walking speed on these asymmetries.

Table 4.XIII shows the mean values, standard deviations and ranges for the asymmetry ratios calculated for the single support phase of the gait cycle. These results are presented graphically in Figure 4.15.

Two factor repeated measures ANOVA identified differences in subject type and the effect of the different walking conditions. In addition a statistically significant interaction between the two factors was identified.

Table 4.XIII: Comparison of asymmetry ratio for single support between hemiplegics and normals at each walking condition.

	Hemiplegic			Normal			
	Mean	SD	(Range)	Mean	SD	(Range)	p
Slowest	0.69	0.48	(0.09-1.95)	0.04	0.04	(0-0.16)	$p < 0.05$
Slow	0.62	0.47	(0.15-1.89)	0.05	0.05	(0-0.20)	$p < 0.05$
Free	0.56	0.43	(0.07-1.69)	0.04	0.03	(0-0.10)	$p < 0.05$
Fast	0.51	0.44	(0.06-1.92)	0.03	0.02	(0-0.08)	$p < 0.05$
Fastest	0.46	0.33	(0.04-1.22)	0.03	0.03	(0-0.12)	$p < 0.05$

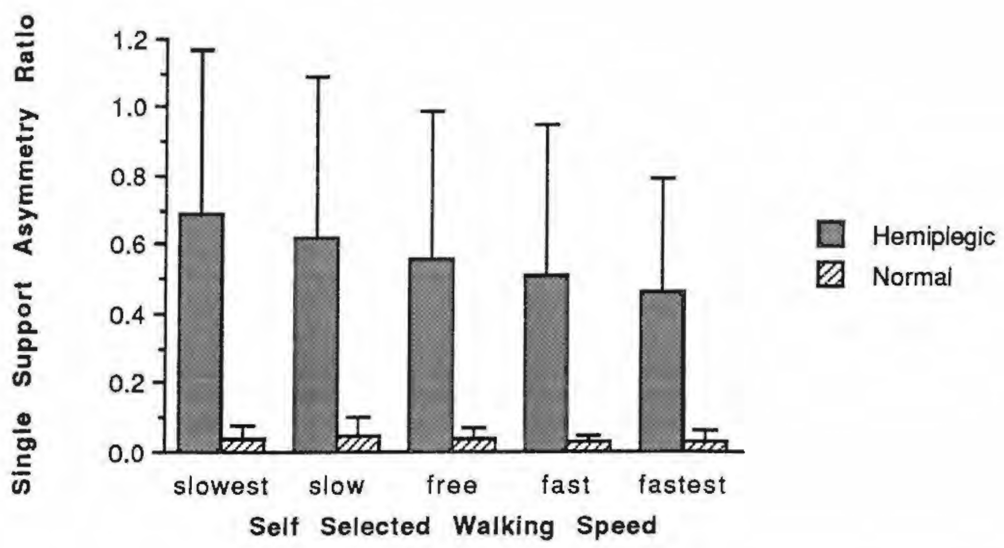

Figure 4.15: Single support asymmetry ratio at each of the self-selected walking speeds.

In the normal group, no significant differences were identified between any of the walking conditions. Symmetry therefore, remained the same at all walking speeds. In the hemiplegic sample significant differences were found in only three of the ten walking condition comparisons. These differences were between the "slowest" test and the "fast" and "fastest" conditions and between the "slow" and "fastest" walks. As the AR diminished with increasing walking speed, this result would tend to suggest that single support asymmetry reduces with increasing walking speed. Comparison between the hemiplegic and normal samples detected significant differences for single support AR for each walking condition with the hemiplegic values consistently higher than those for the controls.

As expected, the normal group were found to be essentially symmetrical in single support. Thus, the consistency of the normal gait pattern in terms of symmetry was confirmed regardless of the self-selected walking speed. Of interest was the extremely small variability in asymmetry at the "fast" walking condition which was even slightly less than that recorded for the "free" speed condition. However, the hemiplegic sample were characteristically asymmetrical. Again, considerable variability within the hemiplegic sample was clearly evident. The asymmetry in some stroke subjects was extremely pronounced (1.95 being the highest value recorded) while some fell within the normal range (0.04) compared to a mean value of 0.03 for the control group. Noticeable variability existed in the hemiplegic group probably resulting from the variability in general walking performance as noted earlier. An unexpected finding, however, was that the single support asymmetries did not deteriorate as the stroke subjects deviated from their preferred walking speed. This was contrary to clinical inclination and the results of the pilot study. Conversely, there appeared to be evidence to support the notion that the magnitude of the asymmetries was inversely proportional to walking speed. This finding deserves to be investigated further because of its rehabilitation implications.

Figure 4.16 shows the single support AR plotted against velocity and shows the declining trend with increasing walking speed as discussed earlier. It is also interesting to note that the variability, otherwise quite marked in the hemiplegic sample, was least at the fastest walking speed.

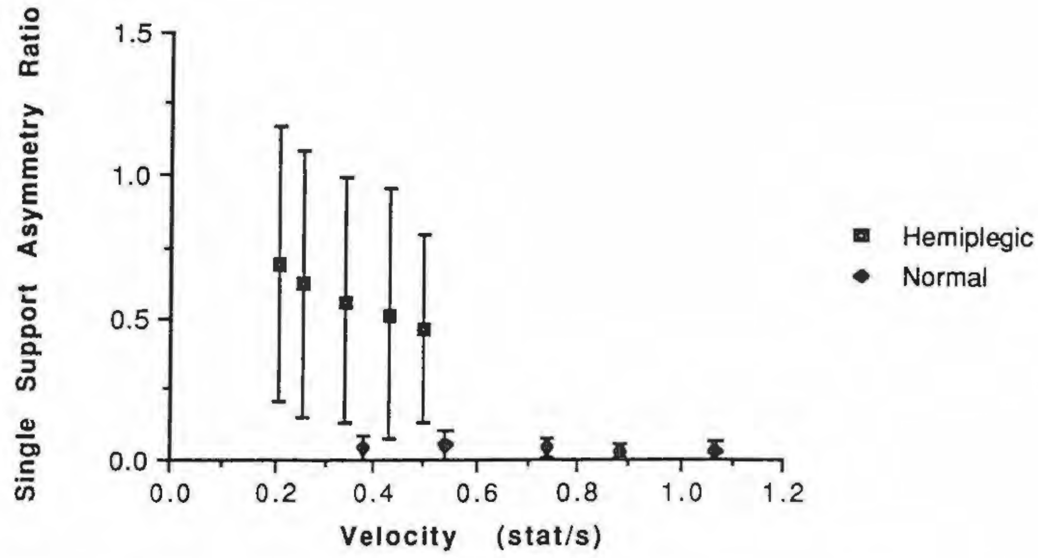

Figure 4.16: Single support asymmetry ratios plotted against relative speed.

Table 4.XIV shows the mean values, standard deviations and ranges for the asymmetry ratios calculated for the braking double support phase of the gait cycle. These results are presented graphically in Figure 4.17.

Two factor repeated measures ANOVA revealed significant differences relating to the type of subjects but no differences were found between the different gait test conditions. In addition, no interaction was found. Significant differences were found between the groups for each gait test condition. The mean values for the normal group were higher than the single support asymmetries. However, this was likely due to the brief duration of the braking double support phase and thus the greater likelihood of a relatively small change resulting in a higher AR value. The mean asymmetries for the hemiplegic group were greater but again with considerable variability within the group. This asymmetry demonstrated no relationship with speed. The mean values for the control group were constant between speeds (0.12 - 0.15) while, in the hemiplegic group, the asymmetry ratios ranged between 0.35 to 0.48 with the slowest speed being the most symmetrical and the fastest speed the least symmetrical and the most variable.

Table 4.XIV: Comparison of the braking double support asymmetries between hemiplegic and normal subjects for each self selected walking speed.

	Hemiplegic			Normal			
	Mean	SD	(Range)	Mean	SD	(Range)	p
Slowest	0.35	0.36	(0.00-1.30)	0.13	0.11	(0.00-0.43)	< 0.05
Slow	0.42	0.38	(0.03-1.34)	0.12	0.10	(0.01-0.34)	< 0.05
Free	0.38	0.39	(0.01-1.48)	0.12	0.11	(0.01-0.45)	< 0.05
Fast	0.41	0.34	(0.10-1.18)	0.14	0.13	(0.00-0.46)	< 0.05
Fastest	0.48	0.42	(0.01-1.42)	0.15	0.10	(0.00-0.33)	< 0.05

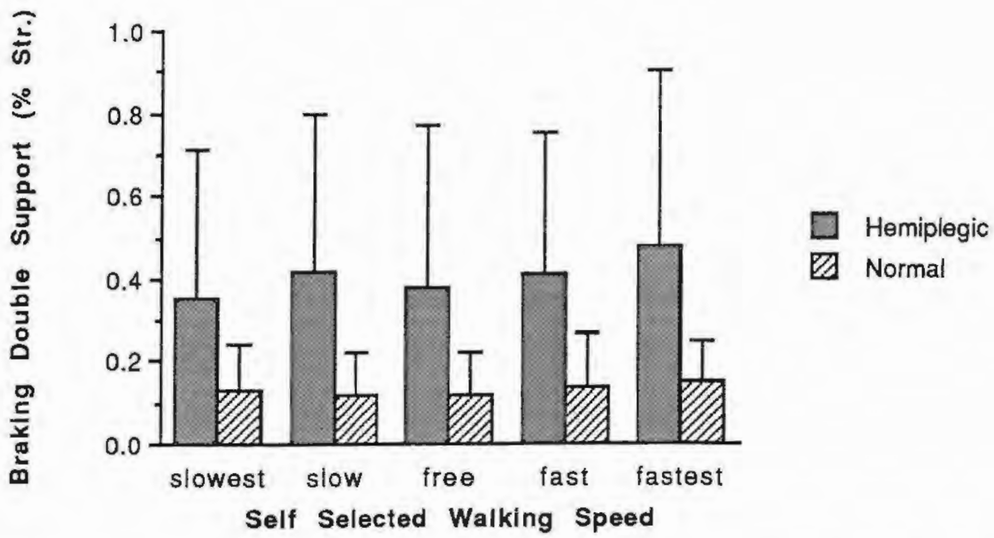

Figure 4.17: The magnitude of braking double support asymmetry ratios at the five self-selected walking speeds in both groups.

Figure 4.18 shows the braking double support asymmetry ratios plotted against relative speed for both groups. It can be seen that variability amongst the hemiplegic sample is striking but that the levels of asymmetry remain somewhat constant in both groups throughout the walking speeds.

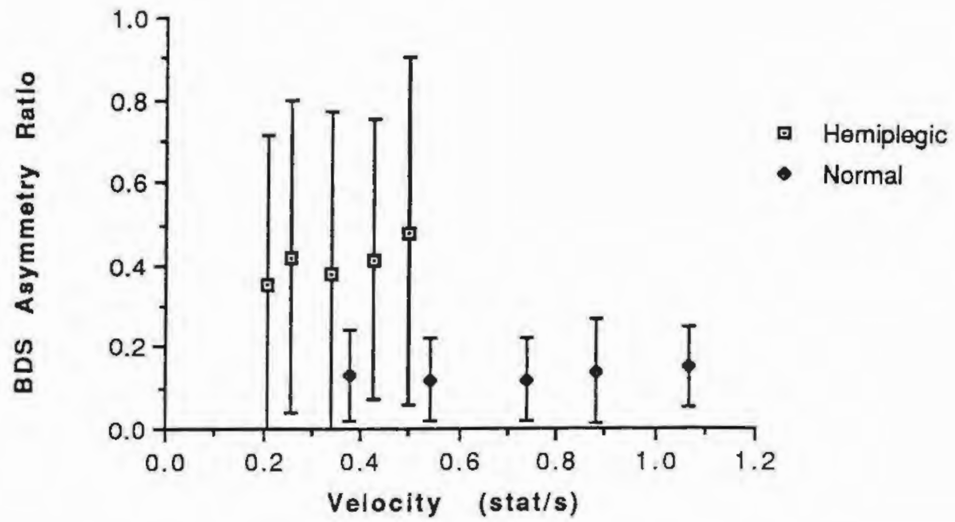

Figure 4.18: Mean and standard deviation for the braking double support asymmetry ratios plotted against relative velocity.

Table 4.XV shows the mean values, standard deviations and ranges for the asymmetry ratios calculated for step length with the right step length being compared with the left in the normal group and those of the affected leg compared with the unaffected leg in the hemiplegic sample. These results are presented graphically in Figure 4.19.

Table 4.XV: Comparison of the step length asymmetries between hemiplegic and normal subjects for each self-selected walking speed.

	Hemiplegic			Normal			
	Mean	SD	(Range)	Mean	SD	(Range)	p
Slowest	0.51	0.58	(0.02-2.17)	0.05	0.05	(0-0.24)	$p < 0.05$
Slow	0.57	0.79	(0.02-3.06)	0.05	0.06	(0-0.26)	$p < 0.05$
Free	0.37	0.50	(0.01-1.82)	0.03	0.02	(0-0.07)	$p < 0.05$
Fast	0.37	0.46	(0.02-1.83)	0.03	0.03	(0-0.09)	$p < 0.05$
Fastest	0.41	0.51	(0.01-1.92)	0.04	0.03	(0-0.14)	$p < 0.05$

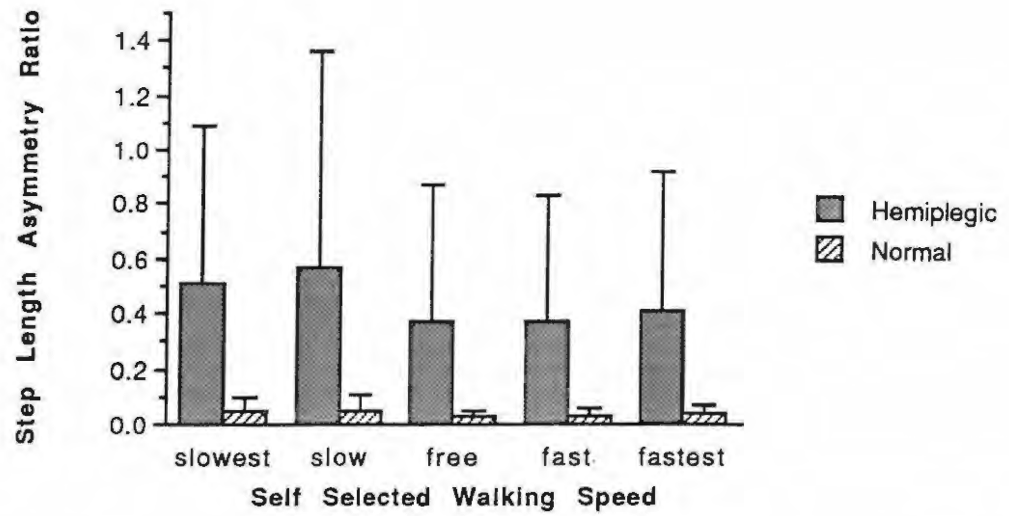

Figure 4.19: Comparison of step length asymmetry between both groups at each self-selected walking speed.

Two factor repeated measures ANOVA detected a significant difference between the types of subject but no difference at the different gait test conditions and no interaction. Significant differences were found between the two groups at each of the walking speeds.

Step length asymmetry in the control group was greatest at the "slow" and "slowest" walking conditions. However, these asymmetries were small overall (0.03-0.05) with minimal variability ($\pm$ 0.02 - 0.06). Similarly, the asymmetry in the hemiplegic sample was greater at the "slowest" and "slow" self-selected speeds but with considerable variability throughout the walking speeds, particularly at the slow speed ($\pm$ 0.79) which was also the walking condition with the greatest standard deviation in the normal group (0.06).

The variability in the normal group was extremely small at the "free speed"; the speed that was most familiar to the normal subjects. In the hemiplegic sample, asymmetry was most pronounced in the "slowest" and "slow" walking conditions and diminished at the "free", "fast" and "fastest" walking speeds. However, those differences were not statistically significant. Thus walking speed did not significantly affect step length asymmetry in either group. This is clearly evident in Figure 4.20 in which step length AR is plotted against relative velocity.

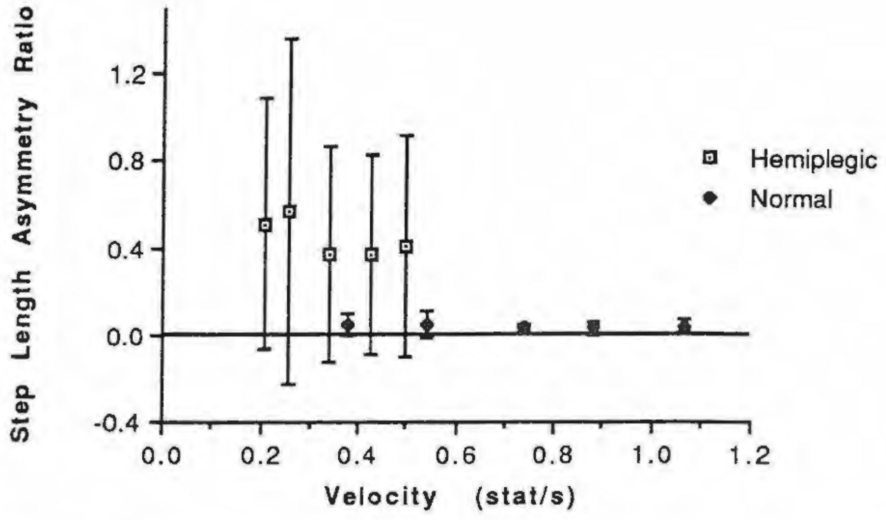

Figure 4.20: Mean and standard deviation for the step length asymmetry ratios plotted against relative velocity in both groups.

The striking asymmetries in the single support, double support and step length in the hemiplegic subjects compared with the controls was consistent with the findings of previous research. However, the effect of a variety of walking speeds on the magnitude of these asymmetries had not been addressed until this time. It appeared that altered walking velocities had little effect on the overall magnitude of the asymmetries with the exception of the single support asymmetry, which appeared to reduce with increasing walking speed. Of some importance was the finding that the asymmetries did not deteriorate when the subjects moved away from their preferred speed of walking. This finding suggests that attempts to improve the range of walking speeds of the hemiplegic subjects, a competency which, according to the findings of this study, is in need of improvement, will not adversely affect the symmetry of the gait pattern.

Corrected Walk

The purpose of requesting that the hemiplegic subjects walk at a "corrected" walking speed was to investigate the extent to which these subjects could modify their gait pattern. The instruction "walk with equal steps and try and spend the same amount of time on each leg" was issued in an effort to have the subjects exert conscious control over their gait pattern thus attempting to produce a more normal gait pattern. The data obtained from this "corrected" walk are shown in Table 4.XVI along with the values obtained for the other walking conditions for comparison. An asterisk (*) is used to denote which parameters are not statistically different from the corrected walk. From this analysis it can be seen that the "corrected" walk and the "free" speed walk are the same in all components with some of the measures, notably relative speed, stride time and relative stride length, identical or almost identical between the two walking conditions. This would suggest that the hemiplegic subjects produce a typical, preferred walking performance when they are attempting to walk as well as possible; an understandable phenomenon. It would also suggest that the habituated gait pattern, which all of these subjects possessed, feels normal to them; a factor which has considerable importance in terms of late-stage rehabilitation. It is likely that, to improve such a situation, the habituated pattern would have to be dismantled before a new, more symmetrical pattern, could be achieved.

As revealed in this study the hemiplegic gait pattern, therefore, can be said to be slow, deficient in range in terms of temporal and spatial performance, asymmetrical and relatively unmodifiable. The walking performance of the group of hemiplegic subjects tested in this study was highly variable despite the application of strict inclusion and exclusion criteria. Some of the stroke subjects demonstrated performances which lay on the fringes of normality while others produced performances indicative of varying degrees of residual disability. However, the stroke subjects in this study could be considered to be elite in that all were able to walk considerable distances without the assistance of a walking cane which, in rehabilitation terms, would be considered to be fully functional. Despite this fact, considerable differences were found between the walking performance of the hemiplegic subjects when compared to the controls. A positive feature of this relative heterogeneity of the stroke sample was that a wide range of performances could be studied, thus enhancing the likelihood of clearly establishing relationships between gait and balance measurements.

Table 4.XVI: Mean and standard deviations (±) for all parameters measured at the five walking speed conditions and the corrected walk (* no difference from "corrected" walk).

	Slowest	Slow	Free	Fast	Fastest	**Corrected**
Speed	0.21	0.26*	0.34*	0.43	0.50	**0.34**
(Stat/s)	±0.06	±0.1	±0.13	±0.16	±0.23	**±0.12**
Stride Time	1.89	1.70*	1.53*	1.35	1.27	**1.56**
(s)	±0.36	±0.32	±0.30	±0.28	±0.31	**±0.30**
Stride Length	0.38	0.43	0.49*	0.54*	0.58	**0.50**
(Stat/Str)	±0.09	±0.12	±0.12	±0.14	±0.17	**±0.12**
TS (Un)	78.5	76.2	74.1*	72.7*	71.0	**73.9**
(% Stride)	±5.5	±6.2	±6.9	±7.6	±6.9	**±5.9**
TS (Aff)	64.7*	62.9*	61.7*	60.2*	59.6	**62.4**
(% Stride)	±7.3	±6.8	±5.8	±5.7	±6.2	**±6.2**
SS (Un)	34.7	36.7*	38.2*	39.4*	40.3	**37.7**
(% Stride)	±7.5	±6.6	±5.7	±5.4	±6.1	**±6.2**
SS (Aff)	21.6	23.9	25.8*	27.6*	28.9	**25.9**
(% Stride)	±5.4	±6.3	±6.9	±7.2	±7.0	**±6.0**
BDS (Un)	23.1*	20.9*	19.9*	18.5*	17.7	**20.6**
(% Stride)	±7.3	±6.6	±6.1	±5.8	±6.5	**±6.8**
BDS (Aff)	19.7	18.1	15.8*	14.3*	13.1*	**15.2**
(% Stride)	±5.4	±6.4	±5.9	±6.1	±5.7	**±5.4**
SS AR	0.69	0.62*	0.56*	0.51*	0.46*	**0.51**
	±0.48	±0.47	±0.43	±0.44	±0.33	**±0.37**
BDS AR	0.35*	0.42*	0.38*	0.41*	0.48*	**0.47**
	±0.36	±0.38	±0.39	±0.34	±0.42	**±0.50**
SLAR	0.51*	0.57*	0.37*	0.37*	0.41*	**0.47**
	±0.58	±0.79	±0.50	±0.46	±0.51	**±0.50**

BALANCE PERFORMANCE

In order to study the balance performance of the groups, the following parameters were compared:

1) The mean position of the centre of pressure (CP) on the X and Y axes in relation to the geometrical centre of the base of support of the subject in the various test positions.

2) The magnitude of the postural sway in the various test positions measured as a function of variance from the mean position in both the X and Y axes, expressed as the Disp. and the percentage of time spent within displacements of percentages of body weight.

3) The magnitude of the ability to shift CP in the anteroposterior, lateral and diagonal positions

4) The characteristics of dynamic weight shifts when standing in positions which simulated both double support phases of the gait pattern.

Cardinal Plane Weight Shifts:

Tables 4.XVII - 4.XIX show the mean positions and standard deviations of the CP of both groups of subjects on the X and Y axes in relation to the geometrical centre of the base of support of the subjects in the test positions of quiet standing and with weight voluntarily shifted and maintained for the test duration of 10 s to the anterior, posterior, left and right. Because these coordinates were real space measures of the location of the CP, the hemiplegic sample was divided into two groups (right and left hemiplegic) so that data normalisation was not necessary to compensate for the side of the hemiplegia. Figures 4.21 - 4.23 show these data graphically comparing the left and right hemiplegic data with normal and the hemiplegic groups respectively.

The normal sample demonstrated the ability to shift weight symmetrically, although there was more variability within the sample on the Y axis; an understandable finding given the predominance of the anteroposterior nature of the postural sway. In contrast to the normal group, both hemiplegic groups demonstrated deficient performance, particularly on the hemiplegic side.

When the results of the left hemiplegic group were compared statistically with those of the normal group, the data for quiet standing showed a significant displacement to the right for the left hemiplegic group on the X axis although there was no difference with the normal group in the Y axis for this test. The anterior displacement was significantly to the right for the left hemiplegic group on the X axis and significantly less anterior than the performance for the normal group in the Y axis. The posterior shift test resulted in the CP of the left hemiplegic group being significantly to the right on the X axis and they were significantly less able to shift weight posteriorly than the controls. With regard to the lateral shifts, there was no difference between the groups in weight shift to the right but the left hemiplegic sample were significantly less able to transfer weight to the left (affected) side. The results of these lateral shifts were predictable and confirmed previous work that found hemiplegic subjects less able to shift weight over the affected lower limb. However, the poor performance of the hemiplegic group to shift weight anteroposteriorly was clearly evident.

Table 4.XVII: Centre of pressure measurements (mm) on the X and Y axes for left hemiplegic and normal samples.

	Left Hemiplegic			Normal			
	Mean	SD	(Range)	Mean	SD	(Range)	p
Quiet (X)	10.23	13.25	(-13.7 - 28.7)	-1.87	7.86	(-15.4 - 12.4)	< 0.05
Quiet (Y)	-4.62	19.86	(-36.7 - 22.3)	-8.37	15.47	(-39.4 - 15.5)	-
Ant. (X)	18.57	15.52	(-14.0 - 38.9)	-0.73	8.90	(-23.6 - 9.7)	< 0.05
Ant. (Y)	32.71	29.18	(-1.4 - 88.7)	56.69	26.31	(0.4 - 94.0)	< 0.05
Post. (X)	14.54	16.98	(21.0 - 34.2)	-2.68	8.33	(-15.4 - 14.2)	< 0.05
Post. (Y)	-21.82	22.60	(-52.6 - 18.7)	-53.67	21.50	(-85.9 - -7.2)	< 0.05
Right (X)	58.41	13.42	(29.0 - 75.5)	63.29	17.30	(25.7 - 83.5)	-
Right (Y)	10.94	13.00	(-17.9 - 33.5)	-1.15	16.47	(-34.8 - 33.1)	< 0.05
Left (X)	-31.02	23.41	(-73.4 - 5.5)	-66.45	15.37	(-84.38 - -21.8)	< 0.05
Left (Y)	-2.13	24.47	(-51.3 - 23.9)	-5.33	20.95	(-39.5 - 46.8)	-

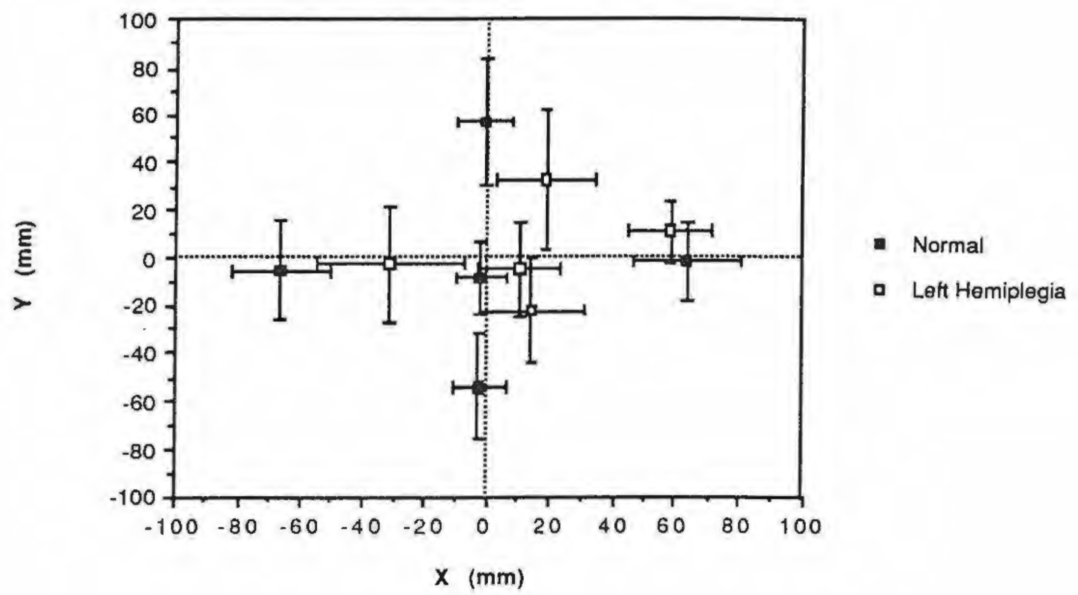

Figure 4.21: Mean position and standard deviations on both the X and Y axes for the normal and left hemiplegic subjects in each of the five test positions.

When the right hemiplegic group were compared to the normal sample statistically, the quiet standing condition was significantly displaced to the left on the X axis and anteriorly on the Y axis to the performance of the normal sample. The anterior shift for the right hemiplegic sample was significantly to the left of the normal sample in the X axis but there was no difference in the magnitude of the anterior shift. Posteriorly, again there were significant differences in the tendency to shift CP backwards to the left and there was a significant difference in the Y axis demonstrating less ability to transfer weight backwards compared to the normal group. The lateral shifts were also significantly diminished in the right hemiplegic group both to the right and the left, however, no differences were found on the Y axis during the weight transfer to the right while, in the transfer to the left, the hemiplegic sample transferred weight significantly to the anterior compared to the normal group.

Table 4.XVIII: Centre of pressure measurements (mm) on the X and Y axes for right hemiplegic and normal samples.

	Right Hemiplegic			Normal			
	Mean	SD	(Range)	Mean	SD	(Range)	p
Quiet (X)	-16.85	11.88	(-40.2 - 0.4)	-1.87	7.86	(-15.4 - 12.4)	< 0.05
Quiet (Y)	8.82	15.51	(-21.4 - 24.3)	-8.37	15.47	(-39.4 - 15.5)	< 0.05
Ant. (X)	-16.88	14.38	(-4.0 - 9.3)	-0.73	8.94	(-23.6 - 9.7)	< 0.05
Ant. (Y)	41.64	13.03	(19.4 - 57.4)	56.69	26.31	(0.4 - 94.0)	-
Post. (X)	-25.04	15.63	(-57.5 - -7.0)	-2.68	8.33	(-15.4 - 14.2)	< 0.05
Post. (Y)	-9.15	27.04	(-38.2 - 33.1)	-53.67	21.50	(-85.9 - -7.2)	< 0.05
Right (X)	30.32	12.75	(7.8 - 47.2)	63.29	17.30	(25.7 - 83.5)	< 0.05
Right (Y)	4.79	18.50	(-18.46 - 34.6)	-1.15	16.47	(-34.8 - 33.1)	-
Left (X)	-47.18	20.21	(-82.0 - -20.8)	-66.45	15.37	(-84.38 - -21.8)	< 0.05
Left (Y)	16.24	21.43	(-22.0 - 39.51)	-5.33	20.95	(-39.5 - 46.8)	< 0.05

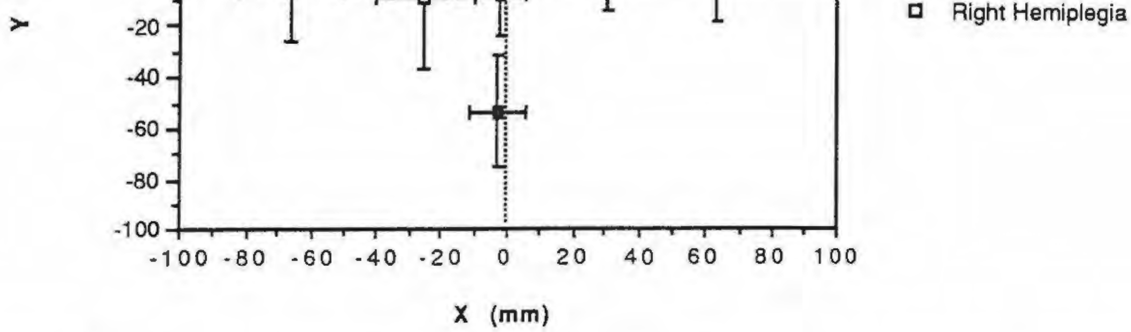

Figure 4.22: Mean position and standard deviations on both the X and Y axes for the normal and right hemiplegic subjects in each of the five test positions.

When the two groups of hemiplegics were compared statistically, quiet standing, anterior shift and posterior shift resulted in significant differences on the X axis with the left hemiplegics bearing weight to the right and the right hemiplegics bearing weight to the left. Laterally, the left hemiplegics were able to shift weight further to the right than the right hemiplegic group, which was to be expected. However, there were no differences in the shift to the left between the two groups.

Table 4.XIX: Centre of pressure measurements (mm) on the X and Y axes for left and right hemiplegic subjects.

	Left Hemiplegic			Right Hemiplegic			
	Mean	SD	(Range)	Mean	SD	(Range)	p
Quiet (X)	10.23	13.25	(-13.7 - 28.66)	-16.85	11.88	(-40.2 - 0.4)	< 0.05
Quiet (Y)	-4.62	19.86	(-36.7 - 22.3)	8.82	15.51	(-21.4 - 24.3)	-
Ant. (X)	18.57	15.52	(-14.0 - 38.9)	-16.88	14.38	(-4.0 - 9.3)	< 0.05
Ant. (Y)	32.71	29.18	(-1.4 - 88.7)	41.64	13.03	(19.4 - 57.4)	-
Post. (X)	14.54	16.98	(21.0 - 34.2)	-25.04	15.63	(-57.5 - -7.0)	< 0.05
Post. (Y)	-21.82	22.6	(-52.6 - 18.7)	-9.15	27.04	(-38.2 - 33.1)	-
Right (X)	58.41	13.42	(29.0 - 75.5)	30.32	12.75	(7.8 - 47.2)	< 0.05
Right (Y)	10.94	13.00	(-17.9 - 33.5)	4.79	18.50	(-18.5 - 34.6)	-
Left (X)	-31.02	23.41	(-73.4 - 5.5)	-47.18	20.21	(-82.0 - -20.8)	-
Left (Y)	-2.13	24.47	(-51.3 - 23.9)	16.24	21.43	(-22.0 - 39.51)	-

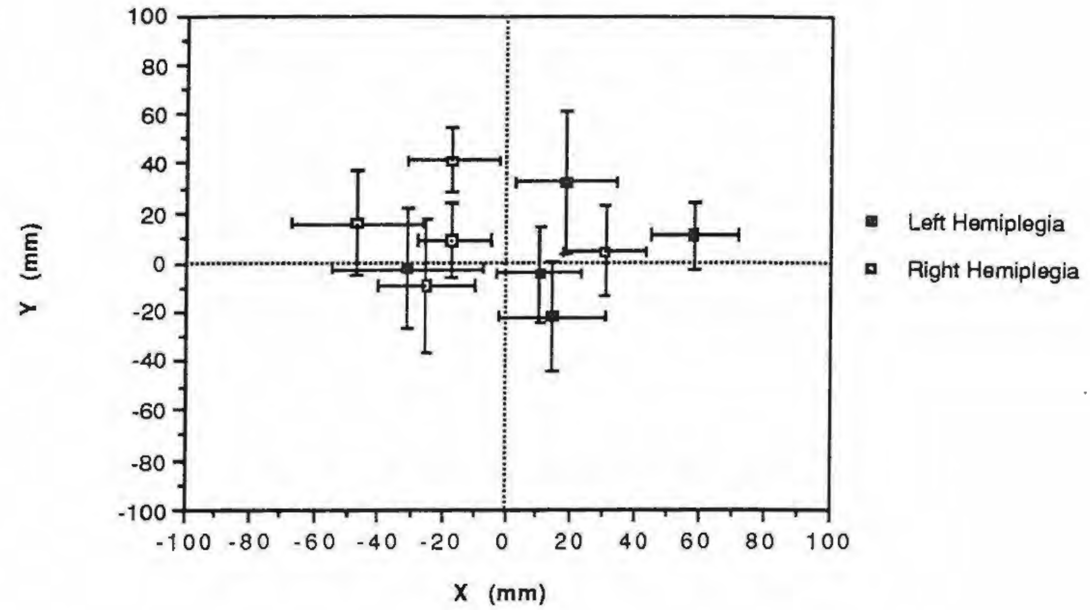

Figure 4.23: Means and standard deviations on the X and Y axes for both left and right hemiplegic subjects at all five positions (quiet, anterior, posterior, right and left).

In an attempt to summarise the findings of this component of the study, Figure 4.24 shows the mean data of the control group compared with the left hemiplegic group, with the right hemiplegic group and between the two hemiplegic groups respectively. The mean values obtained for the anterior, posterior, left and right were joined thus highlighting the skewed nature of the hemiplegic data towards the sound side in the hemiplegic groups. This figure also clearly demonstrates the poverty of the range of weight shifting ability in both lateral and anteroposterior directions resulting in a much smaller area over which the hemiplegic subjects could control their CP compared with normal. These findings were consistent with those of previous studies (Murray et al, 1975; Dettmann et al, 1987). However, the poverty of the ability of both hemiplegic samples to shift weight anteroposteriorly particularly the considerable difficulties experienced in the posterior shift is a finding which may have considerable ramifications in the rehabilitation environment.

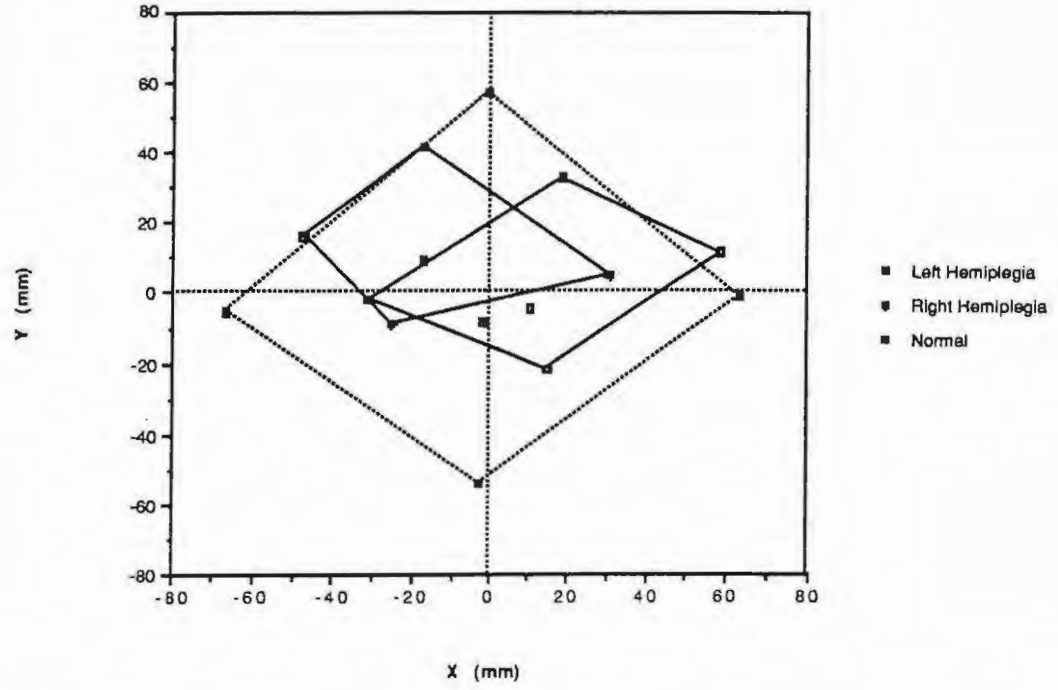

Figure 4.24: Comparison of the mean area of displacement between the normal sample and the right and left hemiplegic groups when voluntarily shifting weight anteriorly, posteriorly, right and left.

Diagonal Weight Shifts

In the D1 position, subjects were required to stand with the left foot back and the right forward. They were then measured after being instructed to shift weight back over the posteriorly placed foot, then symmetrically and finally, with weight over the anteriorly positioned foot. Tables 4.XX - 4.XXII show the results of these diagonal weight transfer tests. Data are presented for both the X and Y axes and include the normal subjects, left hemiplegic subjects and right hemiplegic subjects. Figures 4.25 - 4.27 display the means and standard deviations for both axes, graphically, for the shifts in the D1 direction.

In these test positions, the normal subjects were relatively symmetrical. However, they were able to shift further left and posteriorly (D1B) than to the right and anteriorly (D1F). The mean symmetrical position (D1S) for the normal sample was slightly posterior and to the left of the centre of the base of support (0 X, 0 Y). Variability within the normal group was greater in the Y axis, again probably resulting from the anteroposterior direction of normal postural sway.

The left hemiplegic group, when compared to the normals in the D1 tests, demonstrated considerably less ability to transfer weight through the rear placed leg in both the X and Y axes both of which differences were statistically significant when compared to the performance of the normal group. This was not unexpected in that the rear placed foot in this group was the hemiplegic lower limb. A significant difference between the two groups was also detected in the position of the left hemiplegic symmetrical condition (D1S) on the X axis with the hemiplegic position to the right of the equivalent value for the normal group. Again, this finding was not surprising given the left sided hemiplegia. No statistically significant differences were found between the right, anterior positions (D1F) between the two groups. The variability in the left hemiplegic group was considerable both on the X and Y axes when weight was posterior and to the left (D1B). However, this variability diminished considerably when the weight was anterior and to the right (D1F) and, in this position, was less than that of the control group.

Table 4.XX: Centre of pressure measurements on the X and Y axes (mm) for left hemiplegic and normal samples during the D1 diagonal weight shift.

		Left Hemiplegic			Normal			
		Mean	SD	(Range)	Mean	SD	(Range)	p
D1B	(X)	-22.87	29.76	(-71.6 - 29.9)	-66.60	8.20	(-81.6 - -55.3)	< 0.05
D1B	(Y)	-4.74	27.57	(-46.5 - 53.2)	-31.63	17.99	(-69.2 - 0.5)	< 0.05
D1S	(X)	5.61	14.08	(-20.5 - 33.5)	-9.96	8.18	(-27.4 - 5.1)	< 0.05
D1S	(Y)	4.95	17.62	(-21.8 - 35.9)	-4.18	13.10	(-31.2 - 19.9)	-
D1F	(X)	31.93	15.95	(8.6 - 61.6)	39.53	16.7	(11.9 - 69.6)	-
D1F	(Y)	25.45	15.93	(5.2 - 56.6)	23.49	22.44	(-13.3 - 62.7)	-

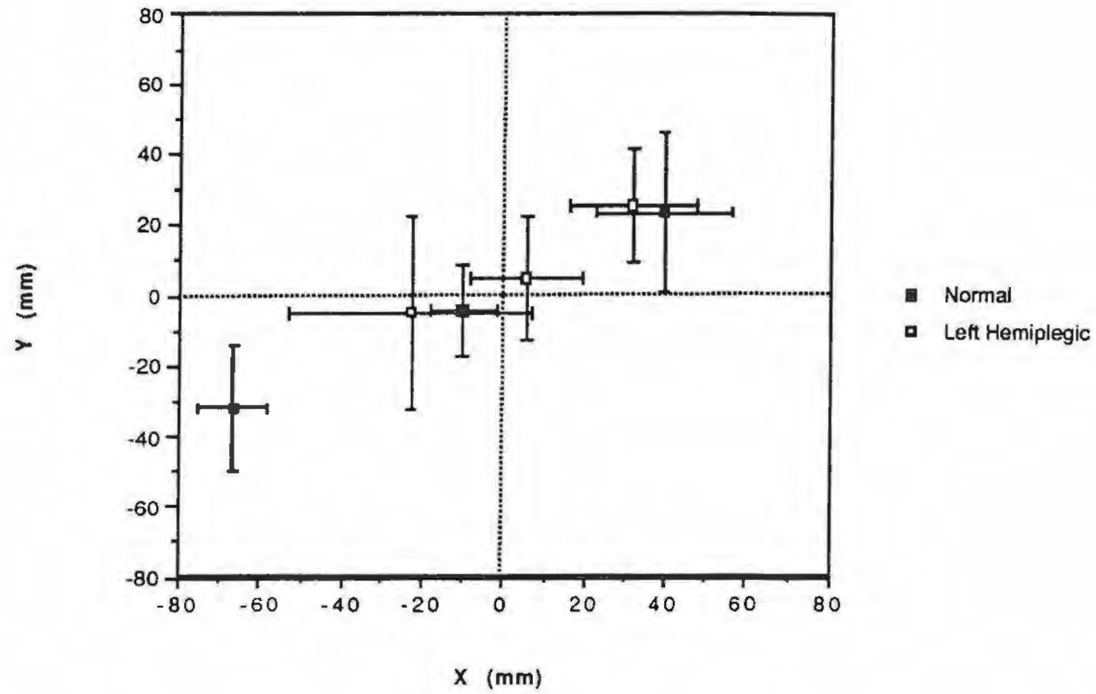

Figure 4.25: Means and standard deviations of CP on X and Y axes of normal and left hemiplegic subjects during left to right diagonal weight shift (D1).

It can be seen from Table 4.XXI and Figure 4.26 that the performance of the right hemiplegic group was also deficient compared to the normal group. Measures for both groups in the D1B condition were significantly different in both the X and Y axes with the right hemiplegic group less able to shift weight laterally to the left and posteriorly the latter of which was striking compared to normal. Although the right hemiplegic group was able to shift weight to some degree to the left, the measures remained statistically significant when compared to the normals. In the D1S condition, a significant difference was detected on the X axis, with the mean position of the right hemiplegic group displaced to the left, when compared with the normal group. This was expected as the right lower limb was the affected leg in this hemiplegic subgroup. There were no differences on the Y axis in this testing condition. Comparison of the D1F condition showed a statistically significant difference on the X axis between the right hemiplegic group and the controls. Thus, the ability of the right hemiplegic group to shift weight to the left was severely compromised to the extent that the mean value in this transfer was to the left of the midline demonstrating a clear reluctance to weight bear through the affected limb. There was no difference between the two groups, in this condition, on the Y axis. Variability in the right hemiplegic sample was particularly noticeable on the X axis at the D1F position. This would be due to the variability within the sample at the position which probably would have caused most trouble to this group, that is a shifting and sustaining of weight on the affected hemiplegic leg.

Table 4.XXI: Centre of pressure measurements on the X and Y axes for right hemiplegic and normal samples during the D1 diagonal weight shift.

		Right Hemiplegic			Normal			
		Mean	SD	(Range)	Mean	SD	(Range)	p
D1B	(X)	-41.46	18.12	(-57.4 - -6.7)	-66.60	8.20	(-81.6 - -55.3)	< 0.05
D1B	(Y)	-2.26	23.92	(-35.4 - 23.3)	-31.63	17.99	(-69.2 - 0.5)	< 0.05
D1S	(X)	-24.72	18.10	(-48.0 - 4.8)	-9.96	8.18	(-27.4 - 5.1)	< 0.05
D1S	(Y)	2.70	16.59	(-33.9 - 22.8)	-4.18	13.10	(-31.2 - 19.9)	-
D1F	(X)	-6.18	33.00	(-37.1 - 53.3)	39.53	16.70	(11.9 - 69.6)	< 0.05
D1F	(Y)	16.49	17.79	(-21.8 - 35.7)	23.49	22.44	(-13.3 - 62.7)	-

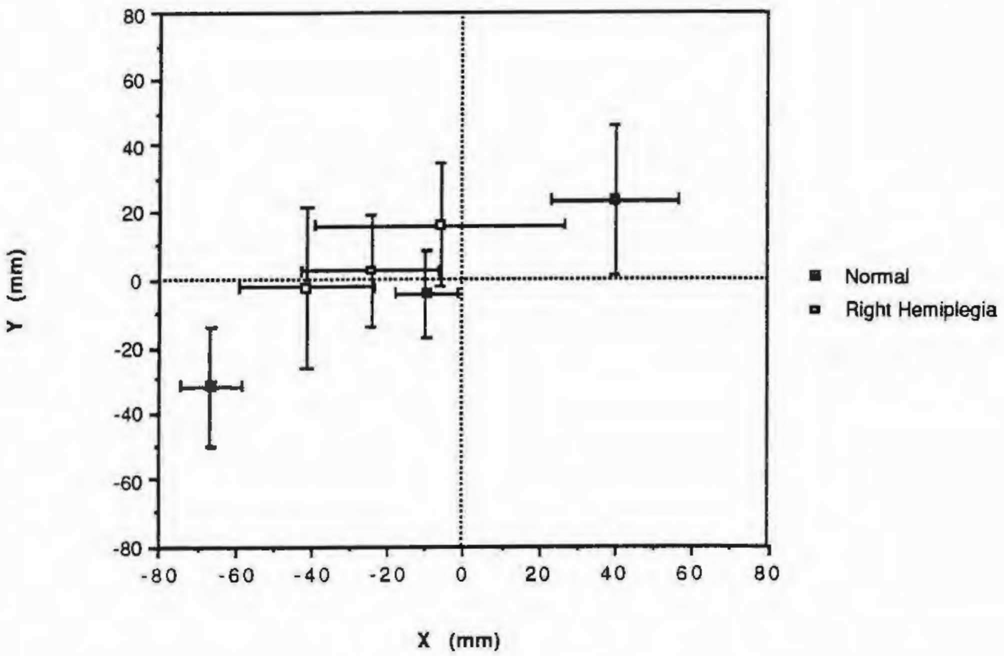

Figure 4.26: Means and standard deviations of CP on X and Y axes of normal and right hemiplegic subjects during left to right diagonal weight shift (D1).

When the left and right hemiplegic subjects were compared, both groups appeared deficient in a manner consistent with the side of the hemiplegia (Table 4.XXII and Figure 4.27). In the D1S condition, a significant difference was found between the X axis measurement between the two groups with the CP of the left hemiplegic subjects to the right of the midline and the right hemiplegic group to the left. This finding is entirely consistent with clinical observation and the findings of previous studies in that the stroke person is reluctant to bear weight on the affected leg (Lane, 1978; Murray *et al.*, 1975; Dettman *et al.*, 1987). Similarly, a significant difference was found in the D1F condition on the X axis in that the left hemiplegic subjects were able to transfer and maintain weight much further to the right than their right sided counterparts. Again, for the reasons stated earlier, this finding was also expected. No differences were found on the Y axis for either the D1S and the D1F conditions between the two groups. No differences were found between the two hemiplegic subgroups in the D1B position either on the X or Y axes. Although the mean value for the right sided group was noticeably further to the left on the X axis than the left sided group, the large standard deviations probably combined with the relatively small numbers in each group precluded the detection of differences between the groups.

Table 4.XXII: Centre of pressure measurements on the X and Y axes for left and right hemiplegic subjects during the D1 diagonal weight shifts.

		Left Hemiplegic			Right Hemiplegic			
		Mean	SD	(Range)	Mean	SD	(Range)	p
D1B	(X)	-22.87	29.76	(-71.6 - 29.9)	-41.46	18.12	(-57.4 - -6.7)	-
D1B	(Y)	-4.74	27.57	(-46.5 - 53.2)	-2.26	23.92	(-35.4 - 23.3)	-
D1S	(X)	5.61	14.08	(-20.5 - 33.5)	-24.72	18.10	(-48.0 - 4.8)	< 0.05
D1S	(Y)	4.95	17.62	(-21.8 - 35.9)	2.70	16.59	(-33.9 - 22.8)	-
D1F	(X)	31.93	15.95	(8.6 - 61.6)	-6.18	33.00	(-37.1 - 53.3)	< 0.05
D1F	(Y)	25.45	15.93	(5.2 - 56.6)	16.49	17.79	(-21.8 - 35.7)	-

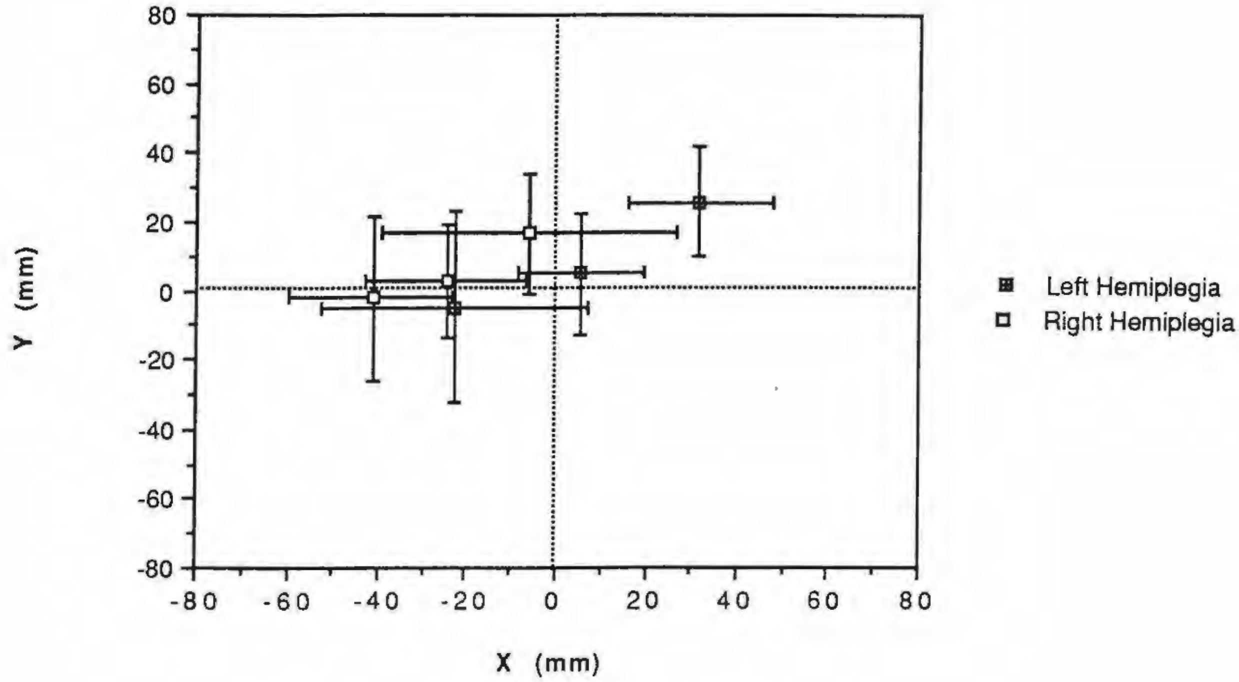

Figure 4.27: Means and standard deviations of CP on X and Y axes of left and right hemiplegic subjects during left to right diagonal weight shift (D1).

The D2 diagonal required that subjects stood with the right foot back and the left forward while measured in the same way as for the D1 conditions with the only difference being the position of the feet of the subjects. Tables 4.XXIII - 4.XXV compare the values obtained during the D2 weight shift tests for the normal and left hemiplegic groups, the normal and right hemiplegic groups and the left and right hemiplegic samples respectively. These results are presented graphically in Figures 4.28 - 4.30.

Again, as shown in Figure 4.28 and Table 4.XXIII, the normal subjects were relatively symmetrical. As in D1, they were able to shift further laterally and posteriorly (D2B), despite the reciprocal position of the feet, than to the left and anteriorly (D2F). The mean symmetrical position (D2S) was fractionally anterior to the zero point on the X axis and this time slightly right of the central point of the base of support on the Y axis. Variability within the normal group was greater in both axes than that obtained for the D1 measures.

In comparison to the normal group, the left hemiplegic subgroup reflected a smaller excursion on the X axis and an inability to shift weight posteriorly this time to the right (Figure 4.28). Significant differences were found between the values for the normal and the left hemiplegic group on both the X and Y axes in the D2B condition. Therefore, the left hemiplegic group was unable to shift as far to the right and as far posteriorly as the controls. There were no differences between the two groups on both the X and Y axes for the D2S condition with the mean points very close together particularly on the Y axis but with the left hemiplegic mean displaced slightly to the right of both the centre of the base of support and the mean for the normal sample. A significant difference between the normal sample and the hemiplegic group was found between positions attained during the D2F test on the X axis with the left hemiplegic group deficient in shifting weight to the left an understandable finding given the left sided nature of the hemiplegia.

Table 4.XXIII: Centre of pressure measurements on the X and Y axes (mm) for left hemiplegic and normal samples during the D2 diagonal weight shifts.

		Left Hemiplegic			Normal			
		Mean	SD	(Range)	Mean	SD	(Range)	p
D2B	(X)	49.38	14.89	(23.9 - 69.0)	65.53	13.91	(24.6 - 85.7)	< 0.05
D2B	(Y)	-7.14	21.65	(-52.2 - 23.5)	-26.94	22.83	(-75.0 - 12.7)	< 0.05
D2S	(X)	8.94	20.17	(-32.6 - 42.8)	3.08	16.07	(-20.3 - 41.8)	-
D2S	(Y)	0.91	13.65	(-25.9 - 20.1)	1.09	15.05	(-28.5 - 34.2)	-
D2F	(X)	-12.56	25.90	(-50.6 - 41.9)	-44.11	18.27	(-74.4 - -8.3)	< 0.05
D2F	(Y)	21.93	20.51	(-18.2 - 47.5)	27.53	26.96	(-13.5 - 77.4)	-

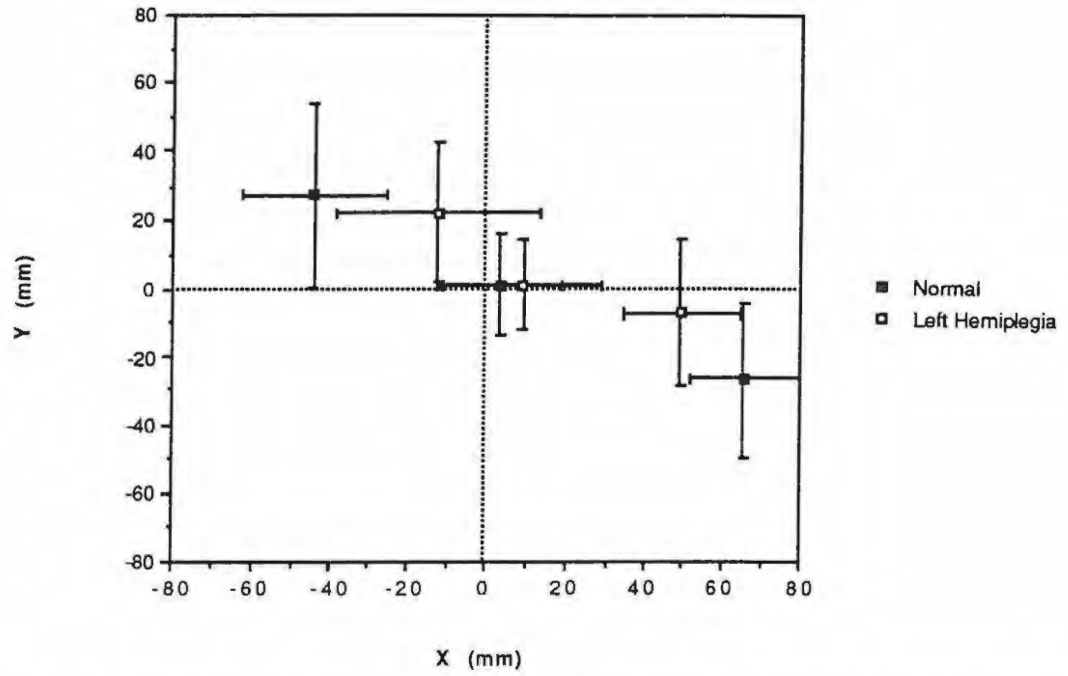

Figure 4.28: Means and standard deviations of CP on X and Y axes of normal and left hemiplegic subjects during right to left diagonal weight shift (D2).

When the performance of the right hemiplegic subjects was compared with the normals in the D2 positions, significant differences were detected on both the X and Y axes for the D2B condition (Table 4.XXIV and Figure 4.29). The right hemiplegic subjects not only faired poorly in comparison to the control group but, also, they were unable to shift weight to the right of the midpoint of the base of support on the X axis. On the Y axis, the mean value for the right hemiplegic group was slightly anterior to the centre of the base of support (4.6 mm). There was a significant difference on the Y axis between the two groups at the D2S position with the CP of the right hemiplegic group significantly further anteriorly than the controls. There was no difference on the X axis between the two groups in this position although the mean value for the right hemiplegic group was to the left in comparison to the controls.

No statistical differences were detected between the normal and right hemiplegic groups in either the X or Y axes at the D2F condition although the mean position of the normals was farther to the left and anterior to those of the right hemiplegic group. However, considerable variability in the right hemiplegic group rendered these differences statistically insignificant. Again, these findings were to be expected given the right sided nature of the hemiplegia which resulted in a reluctance to shift weight through the hemiplegic leg with a subsequent compensation using the sound lower limb.

Table 4.XXIV: Centre of pressure measurements on the X and Y axes (mm) for right hemiplegic and normal samples during the D2 diagonal weight shifts.

		Right Hemiplegic			Normal			
		Mean	SD	(Range)	Mean	SD	(Range)	p
D2B	(X)	-0.35	32.20	(-49.2 - 39.1)	65.53	13.91	(24.6 - 85.7)	< 0.05
D2B	(Y)	4.55	22.56	(-23.1 - 37.5)	-26.94	22.83	(-75.0 - 12.7)	< 0.05
D2S	(X)	-9.67	24.86	(-41.6 - 25.1)	3.08	16.07	(-20.3 - 41.8)	-
D2S	(Y)	17.50	16.70	(-12.7 - 35.6)	1.09	15.05	(-28.5 - 34.2)	< 0.05
D2F	(X)	-27.75	31.62	(-61.7 - 29.5)	-44.11	18.27	(-74.4 - -8.3)	-
D2F	(Y)	22.79	27.47	(-21.1 - 59.2)	27.53	26.96	(-13.5 - 77.4)	-

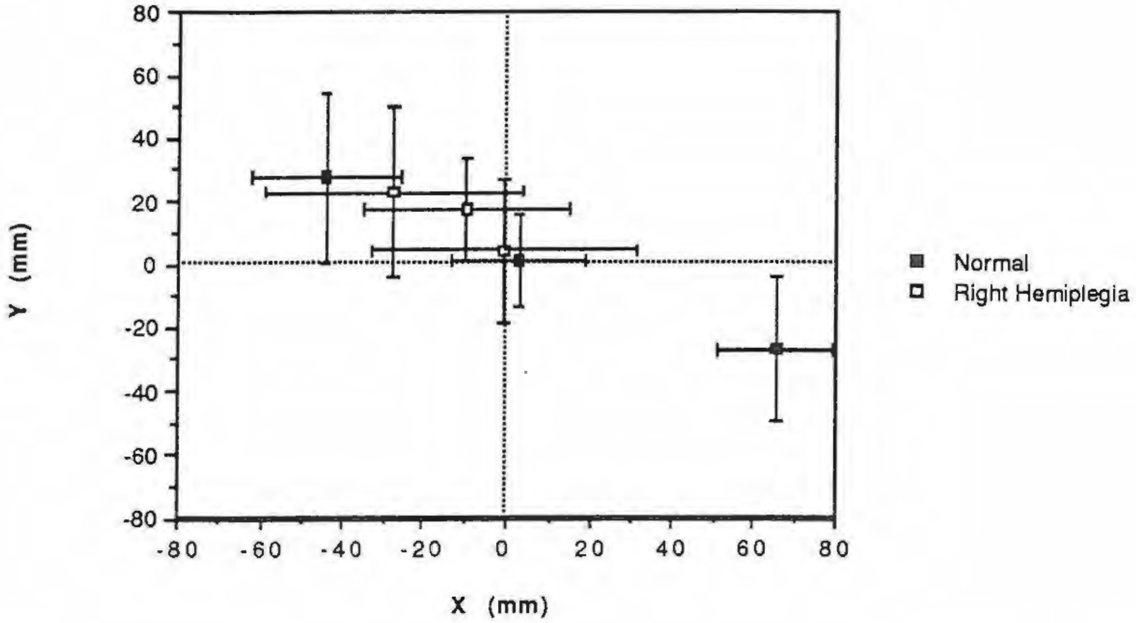

Figure 4.29: Means and standard deviations of CP on X and Y axes of normal and right hemiplegic subjects during right to left diagonal weight shift (D2).

When the two hemiplegic groups were compared (Table 4.XXV and Figure 4.30), the left hemiplegics bore weight predominantly through the right leg and the right hemiplegic group bore weight through the left as would be expected. However, the overall range of values was relatively small and, as a result, only two of the six comparisons were statistically significant. There was a difference between the two groups on the X axis in the D2B condition where the left hemiplegic sample were able to shift weight significantly farther to the right. In addition, there was a significant difference between the two groups in the D2S condition on the Y axis with the CP of the right hemiplegic group anterior to that of the left hemiplegic group. In this test condition, the mean for the left hemiplegic group on the Y axis was situated centrally. In the D2F condition, the mean value for the right hemiplegic group was situated further to the left than that of the left hemiplegic sample, but, as mentioned earlier, the variability, particularly in the right hemiplegic group rendered this apparent difference statistically insignificant.

The large variability within the right hemiplegic group during the tests in the D2 diagonal was notable and was probably due to the variability found amongst the right hemiplegic group and the small number in that sample.

Table 4.XXV: Centre of pressure measurements on the X and Y axes (mm) for left and right hemiplegic samples during the D2 diagonal weight shifts (D1 and D2).

		Left Hemiplegic			Right Hemiplegic			
		Mean	SD	(Range)	Mean	SD	(Range)	p
D2B	(X)	49.38	14.89	(23.9 - 69.0)	-0.35	32.20	(-49.2 - 39.1)	< 0.05
D2B	(Y)	-7.14	21.65	(-52.2 - 23.5)	4.55	22.56	(-23.1 - 37.5)	-
D2S	(X)	8.94	20.17	(-32.6 - 42.8)	-9.67	24.86	(-41.6 - 25.1)	-
D2S	(Y)	0.91	13.65	(-25.9 - 20.1)	17.5	16.70	(-12.7 - 35.6)	< 0.05
D2F	(X)	-12.56	25.90	(-50.6 - 41.9)	-27.75	31.62	(-61.7 - 29.5)	-
D2F	(Y)	21.93	20.51	(-18.2 - 47.5)	22.79	27.47	(-21.1 - 59.2)	-

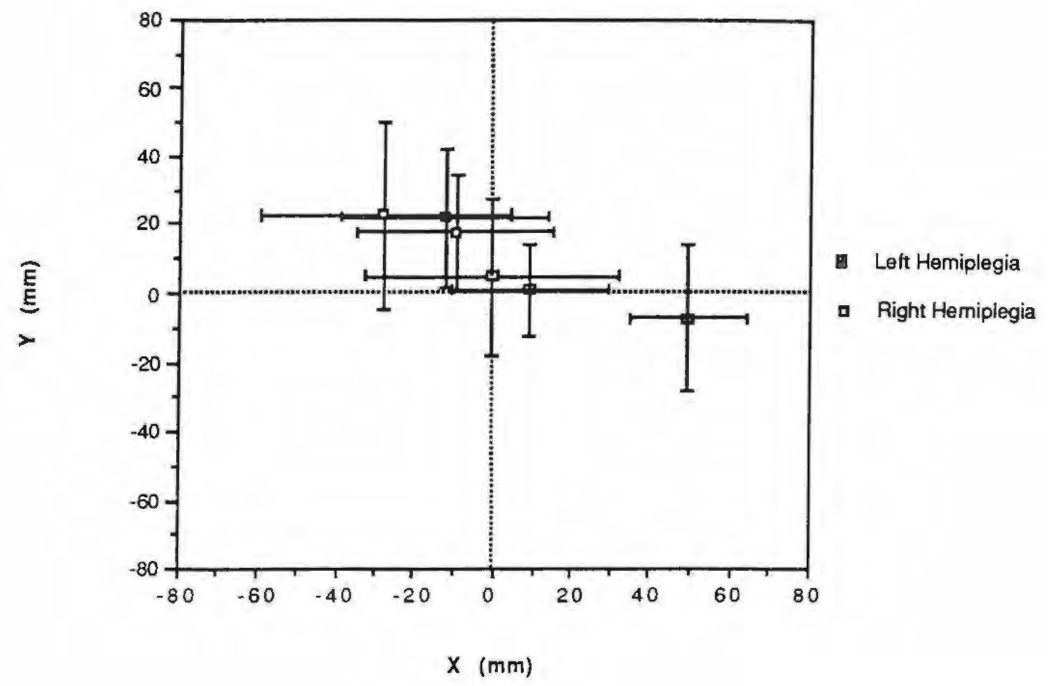

Figure 4.30: Means and standard deviations of CP on X and Y axes of left and right hemiplegic subjects during right to left diagonal weight shift (D2).

The results of the diagonal weight shift measurements confirmed that the cardinal plane weight shifts through the lower limbs (as studied in previous research) can be extrapolated to diagonal weight shifts in very general terms in that the hemiplegic subjects tended to shift further to the sound side compared to the hemiplegic side. However, additional findings were generated which were common for both hemiplegic groups. Both left and right hemiplegic subjects, when compared with the performance of the normal group, were significantly different from the controls on both the X and Y axis for the tests which required them to stand with weight through the posteriorly situated hemiplegic limb. It would, therefore, appear that this was a significant problem for the hemiplegic subjects. Similarly, when the hemiplegic subjects were required to place the hemiplegic leg anteriorly and then transfer weight through that limb, their performance was significantly different from that of the normal subjects on the X axis but not on the Y axis. Thus, the anterior weight shift component of the hemiplegic subjects was the same as the controls in an anteroposterior direction but was lacking in the magnitude of the lateral component of the displacement. It is possible that these findings may have substantial rehabilitation ramifications.

As a way of summarising the results of the diagonal weight shift tests, Figure 4.31 compares the areas, as generated by joining the mean points for D1B, D2F, D1F and D2B in the three groups. It can be clearly seen that both hemiplegic groups have much smaller areas compared with the normals particularly posteriorly and to the hemiplegic side. Further, the area of the right hemiplegic subgroup is the least of the three. This figure illustrates the findings described above.

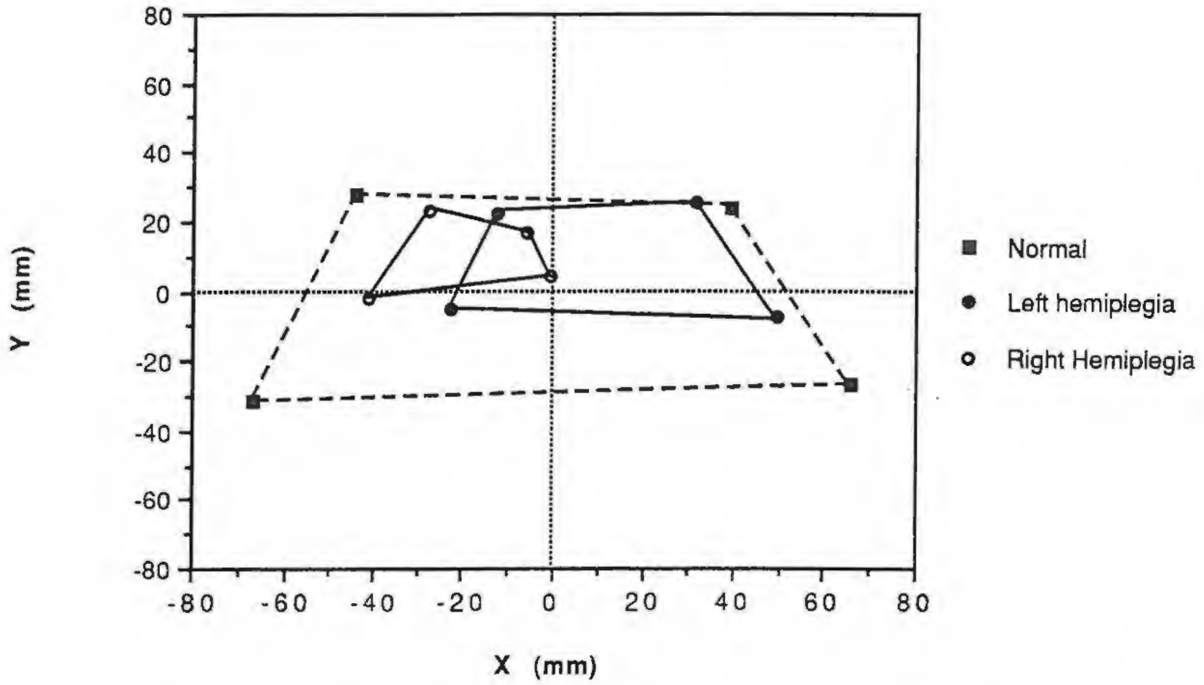

Figure 4.31: The area of postural control of the two hemiplegic groups (right and left sided) compared with the controls as determined by the mean CP positions in the diagonal weight shift positions.

Figure 4.32 shows the areas derived by joining all mean points of the test positions, both cardinal plane and diagonal, with the exception of those for quiet standing, D1S and D2S. From this, an appreciation of the difference between the hemiplegic groups and the normal sample is clearly evident. The normal sample area is symmetrical laterally but with more displacement anterior than posterior, a fact which is understandable given the biomechanics of the ankle joint and the foot. Similarly the posteriorly placed diagonal shifts in the normal sample are more laterally situated than the anterior diagonal shifts. This may be because the motor strategy of standing with the weight situated posteriorly is a more common motor act and may be more energy efficient than the anteriorly placed diagonal shift. It was also noted during the data acquisition phase of the study that different motor strategies were used by different subjects when undertaking the anterior diagonal shifts. This may have led to the diminished performance anteriorly and is worthy of future study. The mean area of balance for the left hemiplegic group is shaped similarly to the normal group but with the predictable tendency to be displaced laterally to the non-affected side. Similarly, the inability to shift weight posteriorly in any direction is striking. The pattern of the mean area of balance for the right hemiplegic group is erratic and unlike the pattern for the normal group and the left hemiplegic sample. All weight shifts are deficient particularly those to the right side and posteriorly.

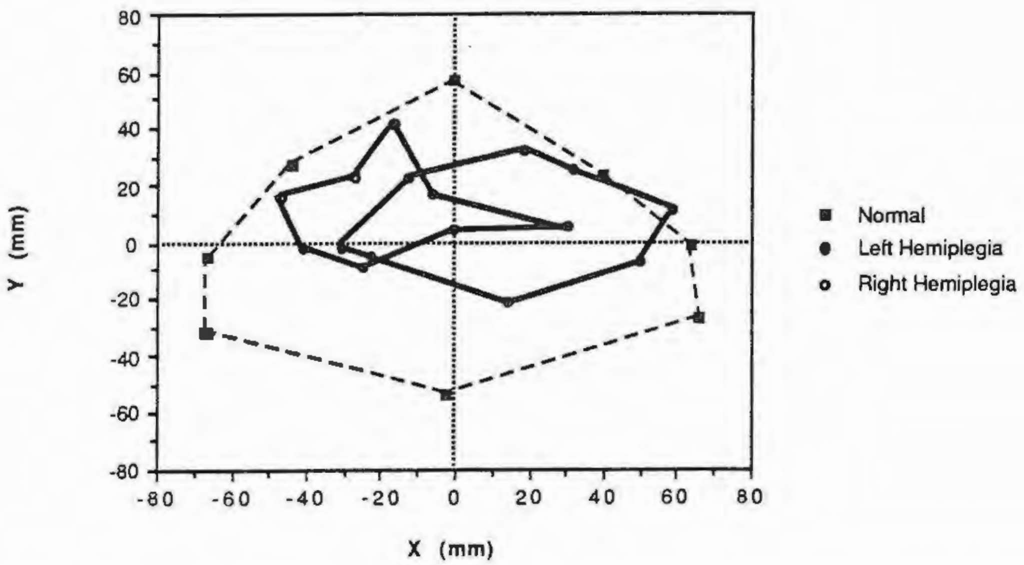

Figure 4.32: The area of postural control of the two hemiplegic groups (right and left sided) compared with the controls as determined by the mean CP positions for both cardinal plane and diagonal weight shift positions.

Clear deficiencies are, thus, present in the balance performance and weight shifting ability of the hemiplegic groups compared to normal as indicated by the range over which CP can be moved voluntarily in the frontal and sagittal planes and in diagonal directions. While the former has been studied before by Murray et al (1975) and Dettmann et al. (1987), the diagonal weight shifts have not until this time and, because the control of CP during walking is diagonal and the test positions mimicked the position of the feet during the two double support phases of the gait cycle, it may be that these findings will bear more relevance to the gait pattern of the hemiplegic group.

Postural Sway.

Two separate methods to assess postural sway were used in the study. The first method was to calculate the Disp. which is a composite, single figure which is derived from the standard deviations from the mean values in both the X and Y axes. The Disp., therefore, indicates the amount of variability about the mean point during the testing period in both axes. The higher the value of the Disp., therefore, the greater the postural sway. The second method concerned the percentage of time during the duration of the test which fell into a series of concentric circles, the centre point of which is the mean value of the sway on the X and Y axes. Each concentric circle represents a percentage of the subject's body weight with the innermost ring representing 5%, the next 10% and so on. If the sway displaces the body weight of the subject by less than 5% during the test period, this will be recorded as a 100% period of time within the 5% ring. Thus, the greater the postural sway, the greater percentage of time will be spent in the rings which represent the displacement of greater percentages of body weight.

For these tests, the hemiplegic group were considered as a single group because it was appropriate to normalise the data regarding the side of the hemiplegia. Thus, lateral shifts were considered over the "affected" leg and the "unaffected" leg. In the case of the diagonal shifts, the same adjustment was made so that the hemiplegic leg was either situated posteriorly, represented by the code BAff (Back Affected) or anteriorly (FAff or Forward Affected) irrespective of the side of the hemiplegia. As a result, it was possible to include the measures for all hemiplegic subjects as one group.

Table 4.XXVI shows the Disp. means, standard deviations and ranges for both the hemiplegic and normal samples for each of the test positions. Figures 4.33 and 4.34 show these results graphically.

Table 4.XXVI: Means, standard deviations and ranges for the Disp. in each of the test positions for both groups (R: Right; L:Left: Aff: Affected Leg; Un: Unaffected leg; D: Diagonal; B: Back; S: Symmetrical; F: Forward).

	Hemiplegic				Normal			
	Mean	SD	(Range)		Mean	SD	(Range)	p
Quiet	5.05	2.02	(1.8 - 8.7)		4.46	2.38	(2.6 - 12.2)	-
Anterior.	6.74	3.11	(2.7 - 14.7)		4.62	1.93	(2.5 - 8.2)	< 0.05
Posterior.	7.38	2.33	(3.7 - 11.9)		6.73	1.98	(3.4 - 10.4)	-
Affected.	8.23	3.88	(3.6 - 17.9)	(R)	5.86	2.29	(2.9 - 10.6)	< 0.05
Unaffected.	7.93	3.04	(3.5 - 17.3)	(L)	5.77	2.17	(2.4 - 11.8)	< 0.05
BAff B	10.06	2.66	(5.8 - 15.3)	(D1B)	7.89	2.37	(4.2 - 12.7)	< 0.05
BAff S	10.78	3.79	(5.2 - 20.9)	(D1S)	6.95	2.28	(3.5 - 11.7)	< 0.05
BAff F	11.07	3.91	(3.0 - 18.0)	(D1F)	6.36	3.13	(2.8 - 15.2)	< 0.05
FAff B	11.15	5.06	(4.4 -23.7)	(D2B)	7.26	3.72	(2.9 - 18.7)	< 0.05
FAff S	9.87	3.81	(4.1 - 19.1)	(D2S)	7.82	3.13	(3.3 - 14.2)	-
FAff F	11.56	4.99	(3.9 - 25.7)	(D2F)	6.83	3.01	(3.1 - 13.4)	< 0.05

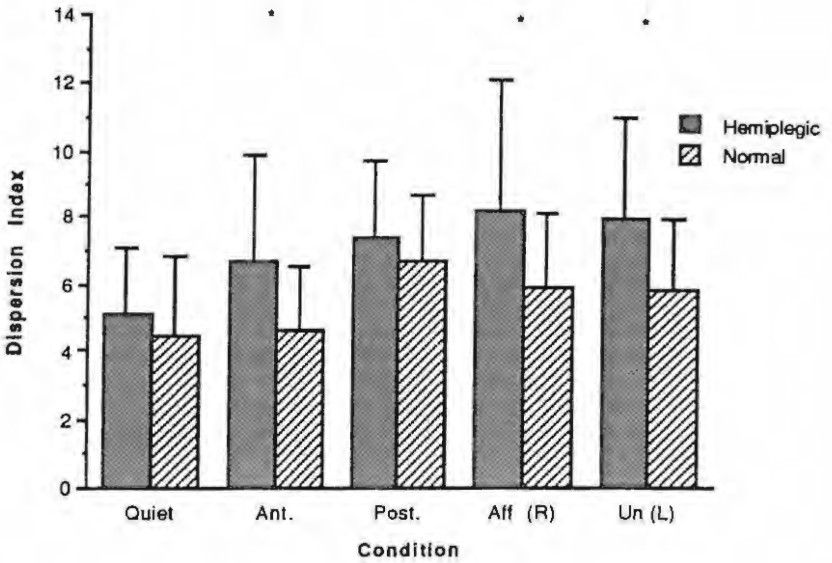

Figure 4.33: Means and standard deviations for Disp. in the Quiet standing, anteroposterior shifts and lateral shifts in both groups (Ant.: Anterior; Post: Posterior: R: Right Leg normal group; L: Left leg normal group; Aff: Affected leg hemiplegic group; Un: Unaffected leg hemiplegic group). (* significant at the $p < 0.05$ level)

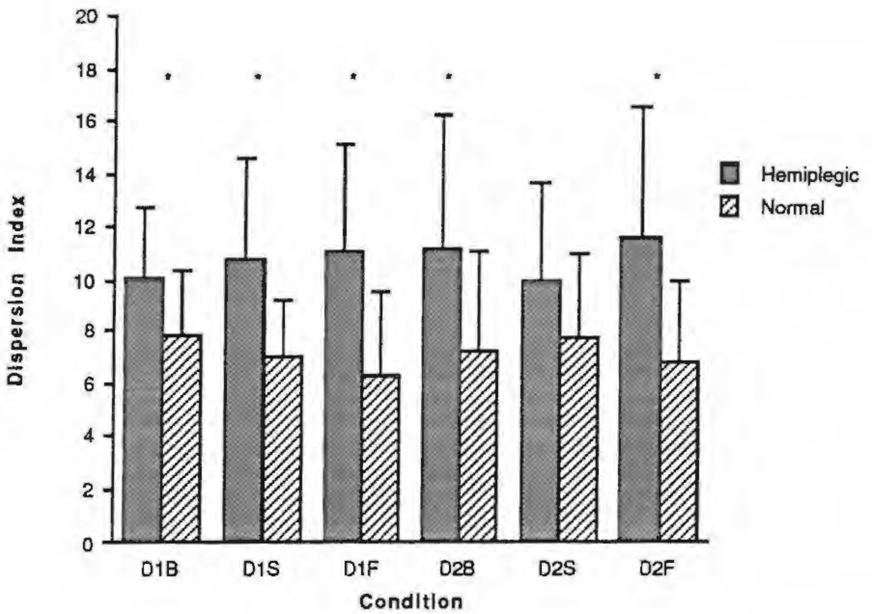

Figure 4.34: Means and standard deviations for dispersion index in the diagonal shifts in both groups (D1 and D2 labels represent DBAff and DFAff respectively for the hemiplegic sample). (* significant at the $p < 0.05$ level)

The performance of the normal group over the right and left legs was assumed to be equal based upon the research conducted by Caldwell *et al.* (1986) and Dettman *et al.* (1987). Thus, the data obtained for the shift to the right in the hemiplegic group was compared with that for the affected leg in the hemiplegic sample and the shift to the left in the normals was compared with the shift to the unaffected side in the hemiplegic group. Overall, the Disp. were less for the cardinal plane shift tests than those for the diagonal shifts. Similarly, they were greater in the hemiplegic group for all test conditions when compared to the same test condition in the normal group. All but three of those comparisons between the hemiplegic and normal group were found to be statistically significant. The only test conditions in which there were no significant differences were in the quiet standing, posterior shift and the diagonal shift FAFFS (D2S). In the latter shift, the hemiplegic subjects stood with the affected leg anteriorly and the test in this position was the "symmetrical" condition. Thus, the postural sway, as measured by the Disp., was greater in the hemiplegic group than the normal sample in all but three instances. This finding was consistent with the findings of previous research. In the three situations where there were no differences, the finding for quiet standing probably resulted from the fact that this would be the most comfortable, least taxing of all the tests. The postural sway, therefore, for both groups would be minimal. In the "posterior" shift condition, the normal group demonstrated the greatest degree of postural sway of all the cardinal plane test conditions and, as a result, no differences were detected. This finding should be seen, however, in the proper context in that the normal group displaced the CP much further posteriorly than the hemiplegic group whose performance was severely limited. It could be proposed that the normal subjects swayed more because they displaced their CP to the limits of their balance capability whereas the hemiplegic subjects did not. This may have accounted for the lack of difference between the two groups in this test condition.

In the third and final condition in which there were no differences between the two groups, the diagonal shift FAffS (D2S), it was unclear why this was found. Examination of the mean data showed that the hemiplegic group had its lowest Disp. reading of all the diagonal tests while the normal group had its highest. As a result, the measurements under this condition were not found to be different. It must be recognized, however, that the two measures are not necessarily equivalent. For the treatment of these data, the condition for the hemiplegic subjects was normalised so that the hemiplegic leg was placed anteriorly. Thus, the left sided hemiplegics would be standing in the same position as the controls but the right hemiplegic subjects would be standing in the equivalent of the D1S position. As a result, it is probably unwise to propose reasons for the insignificant differences found in this testing condition. However, this was the condition during which the hemiplegic sample demonstrated least sway during the diagonal weight shifts.

This test of postural sway, therefore, showed that the hemiplegic group swayed more than the controls in all but three of the eleven test positions. These findings are suggestive of less stable balance in the hemiplegic group compared to the controls. In terms of the performance of the two types of test, cardinal plane and diagonal, the Disp. for the diagonal shifts was greater in both groups than the cardinal plane shifts reflecting the less stable nature of the diagonal tests.

Percentage of Time Spent in Concentric Circles.

Table 4.XXVII shows the percentage of test time spent in the 5%, 10%, 20% and 40% of body weight rings for both groups during the cardinal plane shifts. Figures 4.35 - 4.38 show these results graphically.

Time spent in the 5% body weight ring was greater in the normal group compared to the hemiplegic sample. These differences were found to be statistically significant between the two groups for the anterior and both lateral shifts. For the 10% ring, the hemiplegic group spent a greater amount of time than the normal group in every test and the statistical significant differences were located in the same place but this time with the hemiplegic group having the greater values.

Table 4.XXVII: Percentage of test time spent in the 5%, 10%, 20% and 40% of body weight rings during cardinal plane shifts.

	Hemiplegic			Normal			
	Mean	SD	(Range)	Mean	SD	(Range)	p
Quiet							
(5%)	77.9	22.7	(27.9 - 100.0)	84.6	22.1	(25.0-100.0)	-
(10%)	20.3	20.7	(0.0 - 68.1)	13.1	16.7	(0.0 - 50.7)	-
(20%)	1.9	4.1	(0.0 - 17.6)	2.4	7.3	(0.0 - 31.8)	-
(40%)	0.0	0.0	(0.0 - 0.0)	0.0	0.0	(0.0 - 0.0)	-
Anterior							
(5%)	66.8	25.3	(11.6 - 99.8)	84.0	15.9	(54.3 - 100.0)	< 0.05
(10%)	24.9	16.0	(0.2 - 64.6)	14.2	14.2	(0.0 - 45.5)	< 0.05
(20%)	8.0	12.0	(0 - 41.7)	1.9	3.6	(0.0 - 12.9)	< 0.05
(40%)	0.3	1.4	(0 - 6.2)	0.0	0.0	(0.0 - 0.0)	-
Posterior							
(5%)	56.9	20.4	(11.5 - 91.6)	64.4	20.2	(30.3 - 97.1)	-
(10%)	35.6	14.3	(8.4 - 71.9)	23.0	15.8	(2.9 - 55.5)	-
(20%)	7.2	8.5	(0.0 - 26.1)	5.6	6.4	(0.0 - 21.0)	-
(40%)	0.4	1.4	(0.0 - 6.0)	0.0	0.0	(0.0 - 0.1)	-
Affected (Right)							
(5%)	52.6	27.5	(12.5 - 97.0)	73.2	21.5	(27.5 - 98.4)	< 0.05
(10%)	34.7	19.8	(3.0 - 79.2)	22.6	17.5	(0.8 - 64.6)	< 0.05
(20%)	11.7	17.9	(0.0 - 65.5)	4.1	7.5	(0.0 - 31.0)	-
(40%)	1.1	3.2	(0.0 - 13.9)	0.1	0.6	(0.0 - 2.5)	-
Unaffected (Left)							
(5%)	54.6	23.2	(5.8 - 94.7)	73.4	19.6	(22.9 - 97.8)	< 0.05
(10%)	33.5	15.4	(5.3 - 53.3)	23.2	14.4	(2.2 - 49.9)	< 0.05
(20%)	11.7	17.0	(0.0 - 74.2)	3.3	7.1	(0.0 - 25.3)	< 0.05
(40%)	0.2	0.8	(0.0- 3.5)	0.1	0.4	(0.0 - 1.9)	-

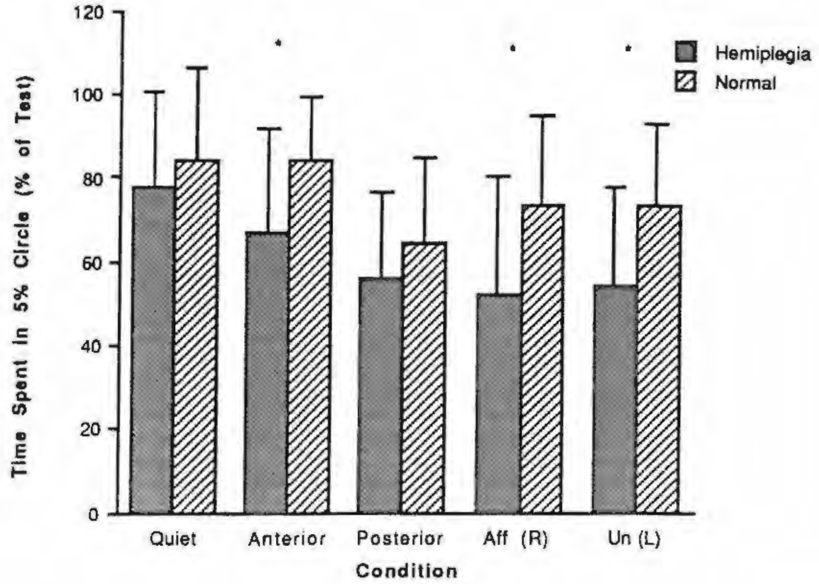

Figure 4.35: Comparison of the percentage of test duration (10 s) spent within 5% of body weight between hemiplegic and normal subjects for the cardinal plane tests (* p = < 0.05).

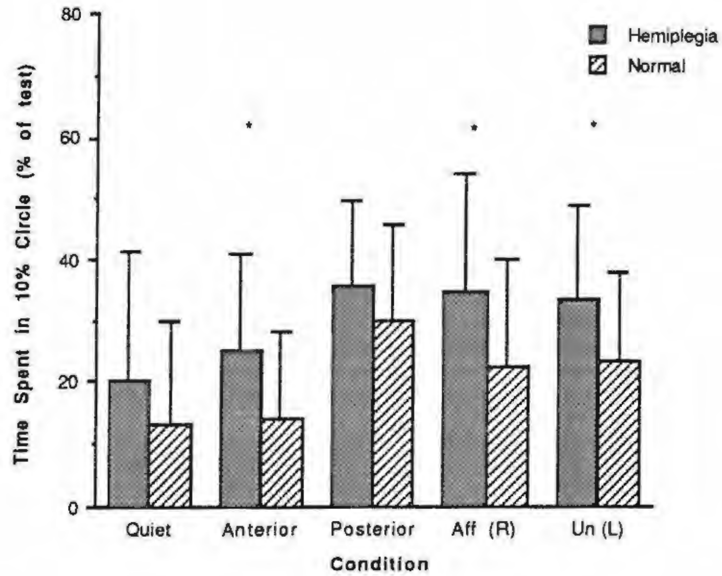

Figure 4.36: Comparison of the percentage of test duration (10 s) spent within 10% of body weight between hemiplegic and normal subjects during cardinal plane shifts (* p = < 0.05).

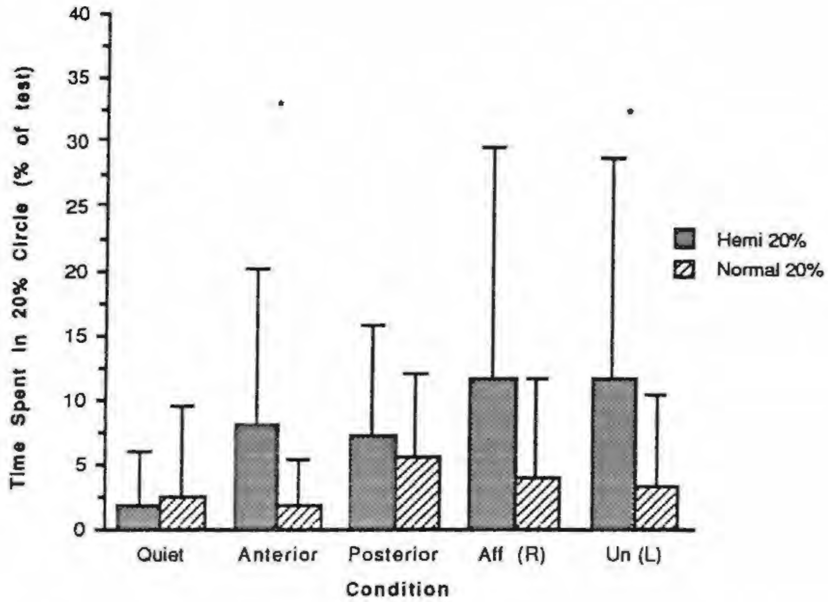

Figure 4.37: Comparison of the percentage of test duration (10 s) spent within 20% of body weight between hemiplegic and normal subjects during cardinal plane shifts (* p = 0.05).

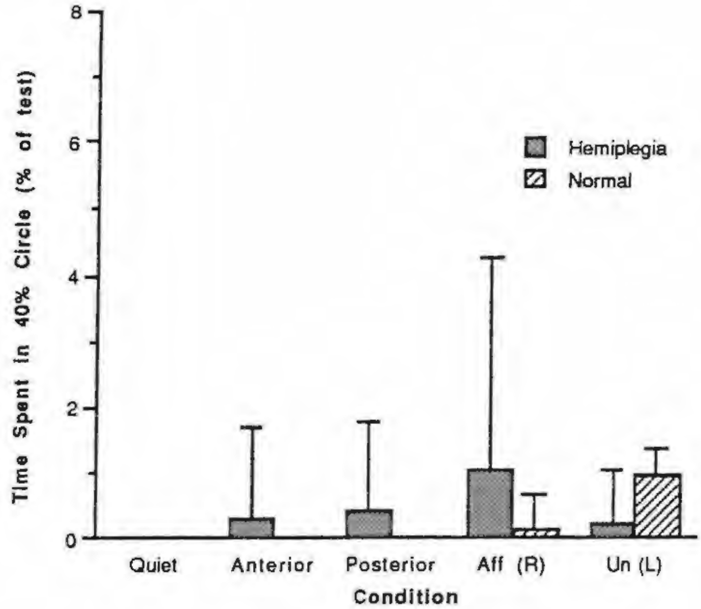

Figure 4.38: Comparison of the percentage of test duration (10 s) spent within 40% of body weight between hemiplegic and normal subjects during cardinal plane shifts.

The statistical significance between the groups for the affected (right) shift was lost when the data concerning the time spent in the 20% of body weight circle were compared. However, the statistical significance was retained for the anterior and unaffected (left) shifts. This probably resulted from the large variability detected particularly in the hemiplegic group for the affected (right) test. No differences were detected for the time spent in the 40% circle as a result of the small values obtained for this test. Thus, the normal group spent more time within the 5% ring and less in the 10% and 20% rings. This indicated that the sway of the hemiplegic group was greater than that of the controls. These findings were consistent with those found for the Disp. and indicated a tendency of the hemiplegic to sway more than the normal group. An interesting finding was that significant differences were detected for both lateral shifts between the groups in the 5% and 10% rings. If postural sway can be considered to be an indicator of balance performance then the performance of the hemiplegic sample to sustain a weight shift over the affected leg and the unaffected leg were both different from normal. Again this assumes that the performance of the normal group is symmetrical which appears to be a sound principle given examination of the data (73.22% $\pm$ 21.5 to the right and 73.43% $\pm$ 19.55 to the left in the 5% ring). It was expected that the sway for the weight shift sustained over the sound limb would not be different from normal. This was not found to be the case in the 5% and 10% rings and in the 20% ring a difference was detected between the unaffected side shift in the hemiplegic group and the left shift in the normal but not between the affected shift in the stroke group and the right shift in the normals. These findings may suggest that either the unaffected limb is affected to some degree or that balance on the sound leg is influenced by the effect of the hemiplegic leg.

Table 4.XXVIII shows the reduced data for the percentage of time spent in the 5%, 10%, 20% and 40% of body weight rings for both the hemiplegic and normal sample for the diagonal weight shifts. Figures 4.39 - 4.42 show these results graphically.

Table 4.XXVIII: Percentage of test time spent in the 5%, 10%, 20% and 40% of body weight rings during diagonal weight shifts.

	Hemiplegic			Normal			
	Mean	SD	(Range)	Mean	SD	(Range)	p
BAff B (D1B)							
(5%)	43.5	17.6	(10.8 - 77.0)	58.8	17.2	(29.1 - 90.3)	< 0.05
(10%)	36.5	7.8	(17.4 - 48.8)	29.8	11.0	(9.7 - 59.7)	< 0.05
(20%)	18.4	11.6	(0.1 - 38.4)	10.6	9.6	(0.0 - 34.9)	< 0.05
(40%)	1.6	2.1	(0.0 - 7.7)	0.7	1.3	(0.0 - 4.5)	-
BAff S (D1S)							
(5%)	43 2.	49.0	(4.2 - 85.6)	68.1	18.3	(30.6 - 92.1)	< 0.05
(10%)	36.3	10.4	(11.4 - 52.8)	24.2	10.9	(7.9 - 48.0)	< 0.05
(20%)	18.8	15.1	(0.1 - 38.4)	7.6	9.8	(0.0 - 38.3)	< 0.05
(40%)	1.9	2.9	(0.0 - 10.5)	0.1	0.4	(0.0 - 1.7)	< 0.05
BAff F (D1F)							
(5%)	42.6	22.3	(18.4 - 98.3)	71.3	21.3	(13.1 - 98.4)	< 0.05
(10%)	32.0	13.4	(1.7 - 52.2)	21.5	9.9	(1.6 - 34.8)	< 0.05
(20%)	22.4	15.1	(0.0 - 55.6)	6.8	13.6	(0.0 - 47.1)	< 0.05
(40%)	3.1	4.1	(0.0 - 12.6)	0.4	1.6	(0.0 - 7.1)	< 0.05
FAff B (D2B)							
(5%)	46.6	20.3	(12.1 - 88.0)	64.8	23.9	(9.5 - 96.7)	< 0.05
(10%)	31.4	10.3	(12.0 - 50.2)	24.5	13.1	(3.3 - 48.0)	-
(20%)	19.2	12.1	(0.0 - 44.6)	9.7	13.2	(0.0 - 47.7)	< 0.05
(40%)	3.2	5.1	(0.0 - 15.4)	1.0	3.6	(0. - 16.2)	-
FAff S (D2S)							
(5%)	47.5	20.6	(17.1 - 89.0)	60.3	20.9	(27.1 - 95.6)	-
(10%)	31.6	10. 7	(11.0 - 55.2)	27.5	11.8	(4.4 - 47.1)	-
(20%)	18.0	12.3	(0.0 - 38.4)	11.5	12.3	(0.0 - 43.1)	-
(40%)	2.1	5.2	(0.0 - 17.5)	0.8	2.4	(0.0 - 8.4)	-
FAff F (D2F)							
(5%)	42.1	22.3	(7.9 - 91.3)	67.7	21.2	(29.6 - 99.2)	< 0.05
(10%)	33.7	11.6	(8.7 - 56.1)	23.5	13.1	(0.8 - 47.3)	< 0.05
(20%)	21.1	15.4	(0.0 - 44.2)	8.4	12.3	(0.0 - 41.9)	< 0.05
(40%)	2.8	4.2	(0.0 - 15.8)	0.5	1.2	(0.0 - 4.4)	< 0.05

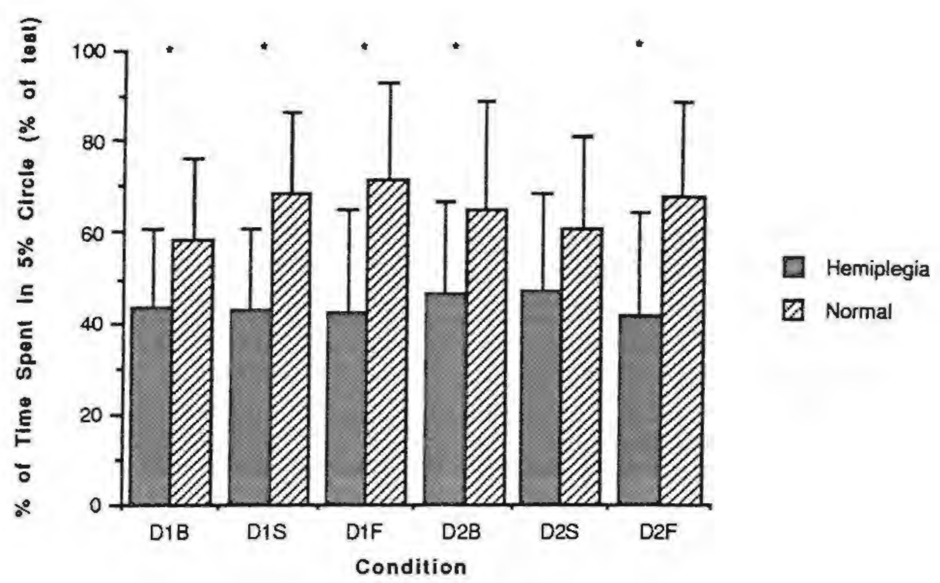

Figure 4.39: Comparison of time spent in 5% of body weight ring between hemiplegic and normal subjects. (D1 and D2 labels represent DBAff and DFAff respectively for the hemiplegic sample). (* significant at the p < 0.05 level).

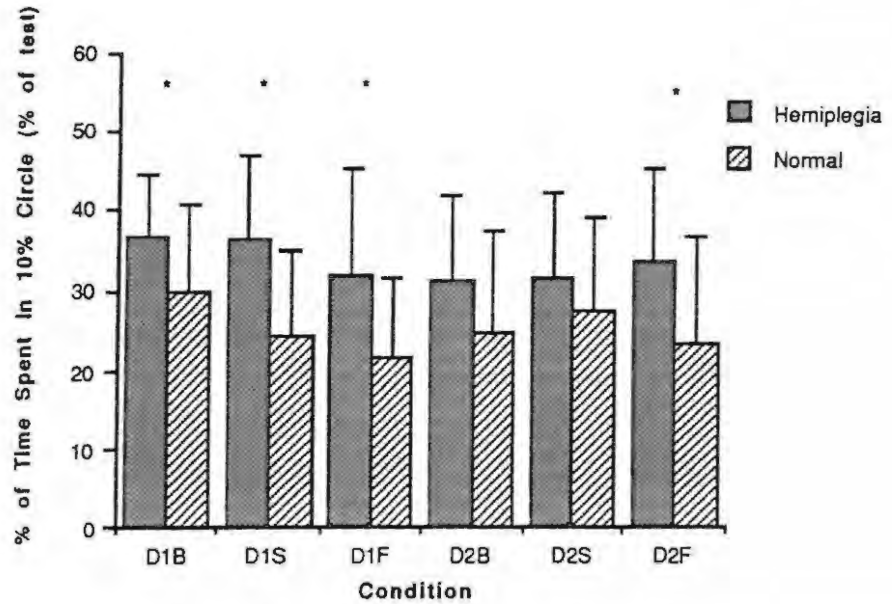

Figure 4.40: Comparison of time spent in 10% of body weight ring between hemiplegic and normal subjects. (D1 and D2 labels represent DBAff and DFAff respectively for the hemiplegic sample). (* significant at the p < 0.05 level).

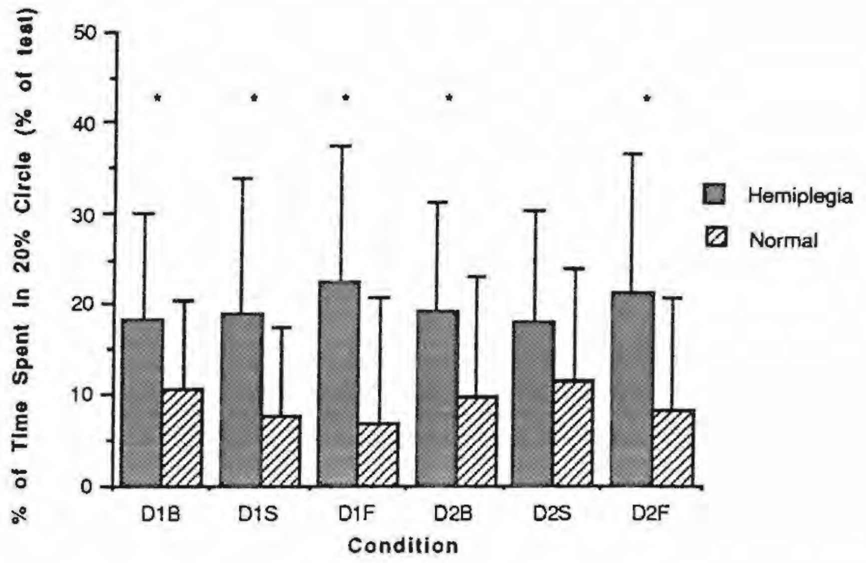

Figure 4.41: Comparison of time spent in 20% of body weight ring between hemiplegic and normal subjects. (D1 and D2 labels represent DBAff and DFAff respectively for the hemiplegic sample). (* significant at the $p < 0.05$ level).

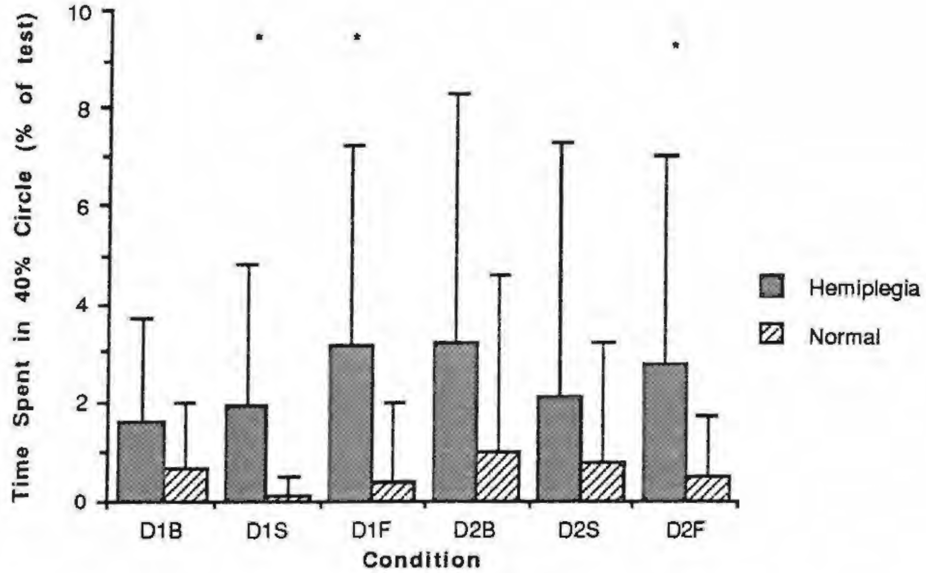

Figure 4.42: Comparison of time spent in 40% of body weight ring between hemiplegic and normal subjects. (D1 and D2 labels represent DBAff and DFAff respectively for the hemiplegic sample). (* significant at the $p < 0.05$ level).

In all test conditions, the normal subjects spent longer in the 5% ring than the hemiplegic subjects. All but one of the comparisons was statistically significant. No difference was found between the D2S (FAffS) condition between the two groups. A similar picture emerged when the data for the 10% ring was examined, however, the D2B (FAffB) was also not significantly different. This latter finding can be explained by the return of the significant difference in the D2B (FAffB) condition when the 20% ring data was analysed. This would suggest that the sway under this condition was greater in the hemiplegic group because more time was spent in the 20% ring indicating a greater postural sway. Significant differences were found between three of the six comparisons made when the data for the 40% ring were considered. However, these values were very low and the variability in both groups considerable lending some question to the value of this measure. Like the cardinal plane shifts, these findings are consistent with those for the Disp. and, again, provide evidence for less stable balance in the hemiplegic sample. However, the lack of difference between the samples at the D2S (FAffS) condition deserves comment. In this position, the hemiplegic subjects stood with the hemiplegic leg forward. In this position, the hip joint on the hemiplegic side is flexed which is a position more likely to be comfortably tolerated than with the affected hip in extension. This is due to the fact that the primitive movement pattern seen in many hemiplegic subjects favours flexion at the hip as opposed to an extended position. With the hip joint in extension, the rectus femoris muscle is elongated; a position likely to stimulate the overactive stretch reflex a characteristic sign of spasticity (Nathan, 1975). Thus, having the hip joint in flexion as opposed to the more difficult to maintain position of extension is more likely to be associated with greater stability. Further, stroke patients in this position can place the knee joint in extension and the foot can be maintained in plantarflexion, all of which are consistent with the primitive synergistic pattern associated with hemiplegia. This may result in an apparent improvement in stability while standing "symmetrically". However, this contrived posture was not dynamic and may only be relatively stable in this one position. It should also be pointed out that the subjects were not provided with any instructions as to the motor strategy to be used in any of the weight shifts. As a result, the strategy which came naturally to the subject was the one adopted and could have been "abnormal" in the hemiplegic sample.

Out of a total of 24 comparisons made between the hemiplegic and normal samples, 17 were found to be statistically significant (70.83%) during the diagonal weight shifts. This compared with 8 significant differences out of 20 comparisons made during the cardinal plane shifts (40%). The diagonal weight shift tests, therefore, appeared to highlight more stark differences in the balance performance between the hemiplegic sample and the controls.

Range of Shift of Centre of Pressure:

Table 4.XXIX shows the reduced data for the range of weight shift undertaken by both groups of subjects in the lateral direction (X axis), the anteroposterior direction (Y axis), the D1 diagonal direction (Left foot back and right forward) and the D2 direction (right leg back and left forward). In the hemiplegic sample, the side of the hemiplegia was again normalised so that D1 entailed placing the affected leg back and the normal leg forward (BAff) while D2 placed the sound limb posteriorly and the affected leg anteriorly (FAff). Figure 4.43 shows these results graphically.

Table 4.XXIX: Comparison of the ranges of weight shift (mm) in both cardinal plane and diagonal directions between hemiplegic and normal subjects.

	Hemiplegic			Normal			
	Mean	SD	(Range)	Mean	SD	(Range)	p
Lateral	84.3	27.9	(40.5 - 140.6)	130.1	29.2	(65.4 - 167.6)	<0.05
Anteropost.	54.4	31.8	(9.1 - 110.0)	110.4	45.2	(8.0 - 162.4)	<0.05
D1 (BAff)	63.1	24.8	(14.1 - 100.4)	123.4	25.0	(79.5 - 168.2)	<0.05
D2 (FAff)	60.3	35.2	(1.5 - 123.3)	127.6	29.1	(80.6 - 179.7)	<0.05

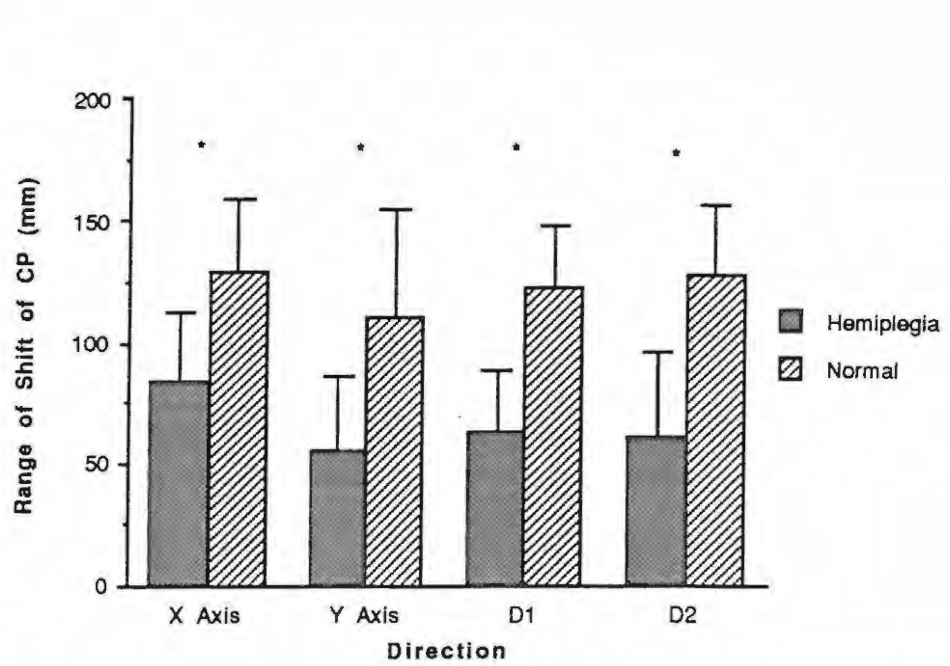

Figure 4.43: Comparison of the range of weight shift in both cardinal plane and diagonal directions between the hemiplegic and normal samples (D1 and D2 labels represent DBAff and DFAff respectively for the hemiplegic sample). (* significant at the $p < 0.05$ level).

Statistical differences were detected in the shifts for all four directions compared, with the hemiplegic performance significantly inferior to the performance of the control group. In relative terms, the hemiplegic group performed best in the lateral shifts, the range achieved being 64.83% of that found for the normal group. They performed worst in the diagonal shift when standing with the hemiplegic leg situated anteriorly (47.24%). In the anteroposterior cardinal plane shift, the performance of the hemiplegic sample was 49.32% of that achieved by the normal sample while in the D1 (BAff) diagonal shift, the hemiplegic performance was 51.1% of that of the normal group. Clear differences, therefore, existed in the range over which the subjects could shift and sustain CP with the performance of the hemiplegic subjects clearly deficient compared with the controls. The cardinal plane anteroposterior and the two diagonal shifts were very similar in terms of comparison with the normals (approximately half). The lateral weight shift, which, relatively speaking was that performed best by the hemiplegic subjects, was probably attributable to the distance the feet were apart during the testing procedure. Mechanically, it was probably easier for the stroke subjects to shift weight laterally, particularly over the unaffected leg. However, a flaw in this proposal is that the same could be said for the normal subjects. A further possibility was that all the hemiplegic subjects in this study had received rehabilitation, a major component of which would have been the teaching and practice of lateral weight shifts. Thus, these subjects may have learned to shift weight laterally during their rehabilitation programme an acquired competency which may have accounted for their relative success in the lateral shifts.

The anteroposterior shift was the least of the four in both groups of subjects, a factor probably attributable to the anatomical arrangement of the foot.

Dynamic Weight Shift

Except for the tests conducted in quiet standing during the cardinal weight shift protocol all tests of balance described to this point in this study considered situations where subjects were required to sustain a balance displacement. Given that walking is a dynamic event and that the type of balance required during gait is dynamic, tests of dynamic balance were included although this was not part of a recognized balance testing protocol for this testing equipment. Subjects were required to shift weight from the posteriorly placed limb to the anteriorly placed leg during the test period. The procedure was conducted for both diagonal directions (D1 and D2). During these tests, Disp. was measured as well as the percentage of time spent in each of the body weight rings. Normally, these parameters are used to indicate postural sway but do this by examining the variability of the position of the CP during the testing period. In the dynamic weight shift tests, an indication of the magnitude of weight shift was gleaned by examining those parameters. Thus a limited performance would be associated with a low Disp. and greater percentages of test time being spent in the more central concentric rings (5%, 10% and 20%) whereas a more extensive weight transfer would be associated with a greater Disp. and a greater percentage of test time spent in the outer concentric rings (20%, 40%, 60% and 80%).

Table 4.XXX compares the reduced values obtained for Disp. and the time spent in the

concentric rings between the hemiplegic group and the normal sample. Figures 4.44 - 4.46 present this data in graphic format.

Table 4.XXX: Disp. and the percentage of time spent in the 5%, 10%, 20%, 40%, 60% and 80% of body weight rings during the dynamic weight shifts for both hemiplegic and normal samples.

	Hemiplegic			Normal			
	Mean	SD	(Range)	Mean	SD	(Range)	p
Disp. D1 (BAff)	30.4	13.1	(12.7 - 63.1)	56.7	12.8	(31.4 - 74.7)	<0.05
Disp. D2 (FAff)	33.0	14.5	(15.9 - 62.3)	59.5	12.1	(36.1 - 79.6)	<0.05
D1 (BAff) 5%	7.6	12.4	(0.0 - 44.4)	0.2	0.4	(0.0 - 1.0)	<0.05
D1 (BAff) 10%	12.8	12.7	(0.0 - 35.9)	2.5	8.1	(0.0 - 36.5)	<0.05
D1 (BAff) 20%	38.0	18.2	(0.8 - 65.8)	14.5	19.3	(0.0 - 64.9)	<0.05
D1 (BAff) 40%	29.7	18.8	(2.6 - 74.3)	34.9	22.5	(3.8 - 81.1)	-
D1 (BAff) 60%	9.9	12.3	(0.0 - 34.9)	30.4	23.7	(0.9 - 68.9)	<0.05
D1 (BAff) 80%	2.1	6.8	(0.0 - 29.3)	17.6	13.5	(0.0 - 47.2)	<0.05
D2 (FAff) 5%	7.6	12.4	(0.0 - 44.4)	0.6	1.0	(0.0 - 2.9)	<0.05
D2 (FAff) 10%	13.1	12.6	(0.0 - 37.9)	1.1	1.5	(0.0 - 5.6)	<0.05
D2 (FAff) 20%	36.5	22.1	(2.5 - 68.4)	9.3	16.9	(0.0 - 78.2)	<0.05
D2 (FAff) 40%	33.4	19.2	(8.7 - 67.3)	37.2	23.9	(3.6 - 72.0)	-
D2 (FAff) 60%	9.9	11.2	(0.0 - 31.1)	31.5	19.3	(0.8 - 58.0)	<0.05
D2 (FAff) 80%	4.1	9.7	(0.0 - 30)	20.2	12.7	(0.0 - 37.1)	<0.05

Comparison of the Disp. data revealed statistically significant differences between the hemiplegic performance and that of the control group with the Disp. for the normal group greater than that for the hemiplegic sample in both diagonal directions. This strongly suggests that the control group were able to transfer weight over a greater distance than the hemiplegic sample. The Disp. for both directions was similar in both groups (30.36 and 33.02 for the hemiplegic group and 56.7 and 59.48 for the controls). These findings were reinforced when the data concerning the percentage of time spent in the concentric rings was considered. Statistically significant differences were detected between all comparisons between the two groups except for the 40% ring. In the 5%, 10% and 20% ring, the hemiplegic times were greater than those for the control group. However, this situation reversed for the 60% and 80% rings with the normal group spending the greater percentage of test time in these rings indicating a greater displacement of body weight.

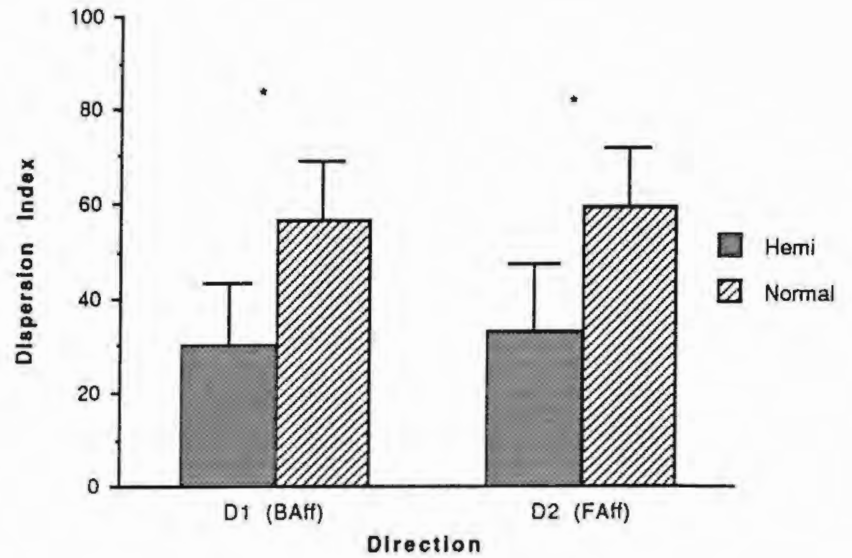

Figure 4.44: Comparison of Dispersion index between the hemiplegic and normal samples for the dynamic weight shift (* Significant difference between groups at $p < 0.05$).

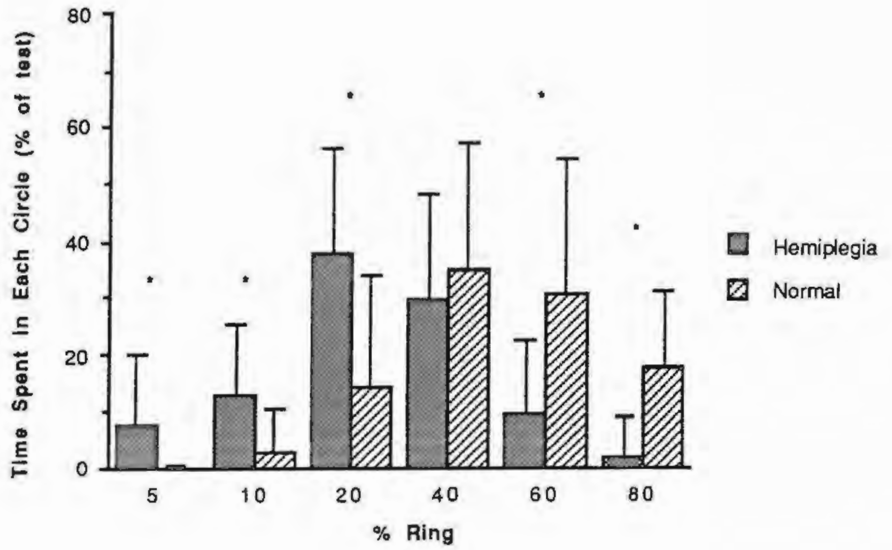

Figure 4.45: Comparison of the time spent in each of the % body weight rings during the dynamic weight shift in the D1 (BAff) direction between the hemiplegic and normal samples (* Significant difference between groups at $p < 0.05$).

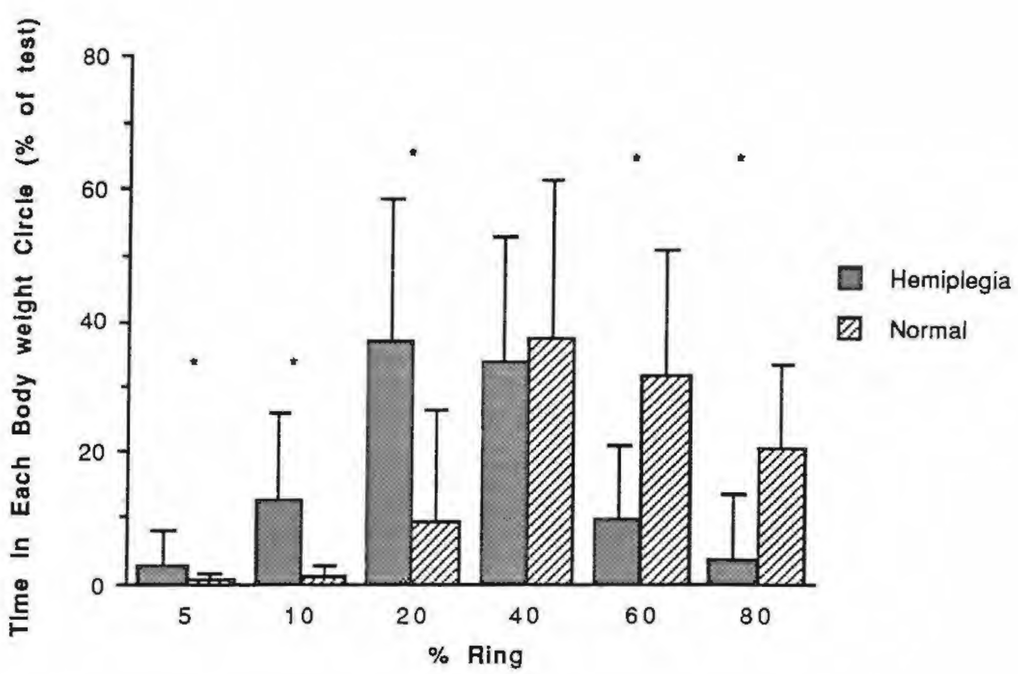

Figure 4.46: Comparison of the time spent in each of the % body weight rings during the dynamic weight shift in the D2(FAff) direction between the hemiplegic and normal samples (* Significant difference between groups at $p < 0.05$).

Figures 4.47 and 4.48 show the same data but in a different format. It can be seen that the curves for both groups are similar but with the hemiplegic group spending more time in the innermost rings and the normal sample spending more time in the outermost rings. The reason that no difference was found at the 40% ring was that this is close to the bisection point of the two curves. This indicates that the normal group shifted weight over a much greater range than the hemiplegic sample. Thus, considerable evidence was found to support the contention that the ability of the hemiplegic group to shift weight dynamically is significantly less when compared to normal. It seems reasonable to speculate that this limitation in performance is likely to affect the gait pattern of the hemiplegic group.

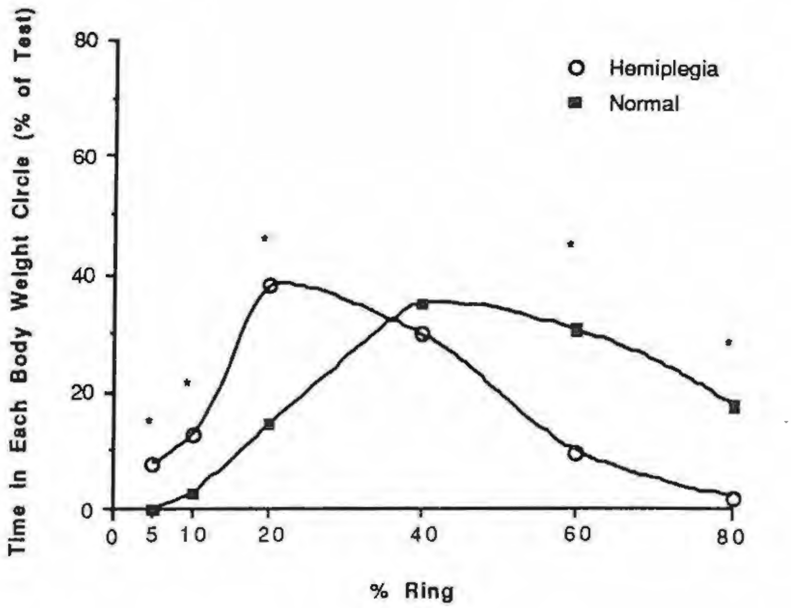

Figure 4:47: Comparison of the characteristics of the percentage of test time spent in each of the body weight rings between the two study groups during the D1 (BAff) dynamic weight shift (* Significant difference between groups at $p < 0.05$).

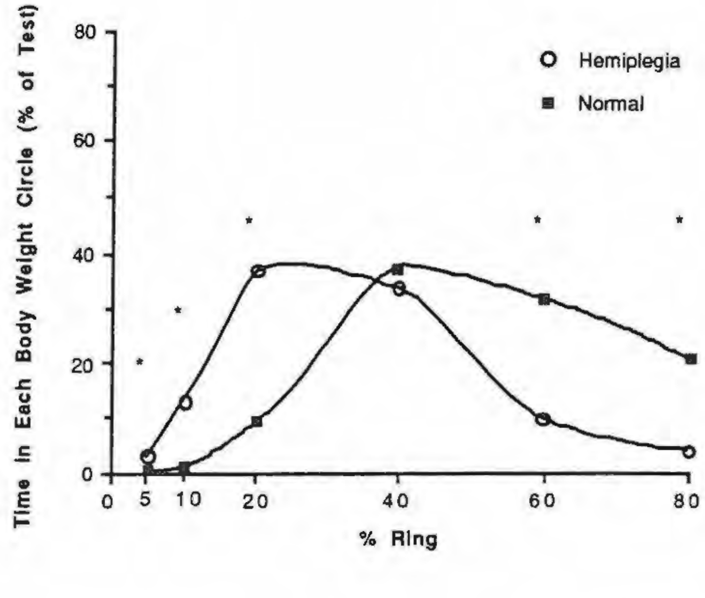

Figure 4:48: Comparison of the characteristics of the percentage of test time spent in each of the body weight rings between the two study groups during the D2 (FAff) dynamic weight shift (* Significant difference between groups at $p < 0.05$).

THE RELATIONSHIP BETWEEN BALANCE MEASURES AND GAIT

The purpose of this analysis was threefold and included:

i) the identification of relationships between the gait and balance measurements in the hemiplegic sample.

ii) the exploration of the strength of these relationships.

iii) the provision of a basis for discussion concerning the relative values of the cardinal plane balance tests compared with the diagonal tests regarding their relationship with gait.

Correlational matrices were constructed and Pearson product moment correlations calculated between all gait and balance parameters for the hemiplegic subjects. In tests where all subjects were included ($df = 18$), critical r values of 0.4438 or greater were representative of a 0.05 level of probability that the correlation was different from 0 (Friedman, 1972). In tests which included only the left hemiplegic group ($df = 10$) the critical r value was set at 0.576 while in the right hemiplegic group ($df = 6$) the critical value for r was 0.7067 (Friedman, 1972).

Given the voluminous data generated in this statistical analysis (in excess of 11,000 relationships were examined), only those relationships with a probability of < 0.05 are reported. These correlations are presented in Appendix 6 (pages 248 - 258). An important factor taken into account in commenting on the detected relationships pertained to the patterns of correlation. For example, if a relationship was found between a balance measure and one of the gait parameters at all five walking speed test conditions then this was viewed as a consistent correlation between the gait variable and the balance measure. Conversely, an isolated correlation between a balance measure and a gait parameter at only one of the walking test conditions was regarded as being correlated less consistently. Further, while recognising the flaws and imprecision of using descriptive terms to describe the strength of correlations due to variability in context, the proposals of Munro *et al.*, (1986) were used to assist with the discussion. Thus, correlations of r values, between the critical r value of

0.4438 and 0.49, were considered to be "low correlations" while r values of 0.5 - 0.69 were regarded as being indicative of a "moderate correlation". Values of 0.70 - 0.89 were viewed as being "highly correlated" and a "very high correlation" was considered to exist if the r values were between 0.9 - 1.00. However, it is important that such a system of reporting correlations be clearly placed in a realistic context. The coefficient of determination (r^2) expresses the "explained" variance. Therefore, a "high" correlation of 0.75 really only explains 0.75^2 or 56% of the variance. The coefficient of non-determination ($1 - r^2$) expresses the "unexplained" variance. Thus, when $r = 0.75$, $r^2 = 0.56$, and $1 - r^2 = 0.44$. This indicates that 44% of a relationship, with an r value of 0.75, remains unaccounted for. The correlations reported in this study must, therefore, be viewed with a degree of caution. This is in contrast to some instances in the rehabilitation literature which report "significant" correlations at a predetermined level such as < 0.05 (Dettman *et al.,* 1987; Bohannon, 1991). This type of reporting has a tendency to lead the reader to a definitive conclusion which may be inappropriate.

Position of CP on the X and Y axes:

For these analysis procedures, the hemiplegic sample was divided into two groups: those with left sided hemiplegias and; those whose hemiplegia was right sided. This was done because the values obtained during these tests were dependent on the side of the hemiplegia as previously shown during the analysis of the balance parameters in this study and in previous related studies (Murray *et al.,*1976; Dettman *et al.,* 1987).

Appendix 6 (page 249, 250) show the significant correlations between the gait parameters and balance tests on the X axis for the left and right hemiplegic samples respectively. Surprisingly, the pattern of correlations did not mirror each other with the pattern for the right hemiplegics appearing to differ from that of the left hemiplegic group. This may, in part, have been due to the smaller number of subjects in the right hemiplegic group which necessitated the attainment of a critical r value of 0.7067 before a < 0.05 level of significance that the correlation was different from 0 could be accepted.

For the left hemiplegic group, on the X axis, the tests which were correlated with the most gait parameters were the "left" weight shift (28), the "TD1" condition (16) and the "D1B" condition (13). The "left" weight shift was negatively correlated with relative speed at all five walking speed conditions The "left" weight shift condition also correlated negatively with four of the five walking speed conditions for relative stride length and the % of stride spent in single support on the affected leg. Positive correlations for this balance test were found in four of the five walking conditions for stride time, single support AR and the % of stride spent in total support on the unaffected side. In terms of the strength of these relationships, the values for the five relative speed conditions (-0.718 - -0.786), could be considered to be high (Munro *et al.*,1986). The correlation coefficients for stride time were also high (0.634 - 0.783) particularly at the three fastest speeds although only four out of the five gait testing conditions were significant.

Relative walking speeds at all five gait test conditions were found to be negatively correlated with the "D1B" balance condition at moderate to high level (-0.614 - -0.722). In addition, this balance test was positively correlated with stride time values at all walking test conditions (0.649 - 0.791) again at the moderate to high level.

The dynamic weight shift condition in the "TD1" test was positively correlated with all five stride time values (0.634 - 0.777) and all five step length AR measures (0.577 - 0.764). Both of these sets of correlations could be considered to be moderate to high in strength.

Of the other tests, the weight shift to the right was negatively correlated with all five step length AR measures at moderate to high strength (-0.576 - -0.723) while the "D2F" balance test was positively and moderately correlated with four of the five stride time measures (0.642 - 0.652).

Most relationships, therefore, were detected in the X axis for the left hemiplegics in the cardinal plane shift to the hemiplegic side. The next balance test which correlated with most gait variables was the dynamic weight transfer, "TD1" followed by the static test, "D1B".

Some of these relationships were strong but were less pronounced than the cardinal plane static shift to the hemiplegic side. In summary of the significance of these relationships, the greater the ability of the left hemiplegic subjects to shift weight into the negative component of the X axis (to the left and thus, over the affected leg), the better was the gait parameter. For example, the greater the negative value for the balance test to the "left", the faster was relative speed. This was a consistent finding amongst tests confirming previous studies and rehabilitation theory.

In the right sided hemiplegic sample, all correlations on the X axis were strong due to the high critical *r* value of 0.7067. The dynamic balance test "TD1" correlated with 15 gait measures while the static tests in "quiet" standing correlated with 11 gait variables and the "posterior" weight shift with 10. In terms of patterns, the "TD1" test correlated positively with all five measures for % of stride spent in total support on the affected side (0.755 - 0.87) and negatively with single support on the unaffected side (-0.761 - -0.868). For the same balance test, there was a negative correlation with single support AR in four of the five gait test conditions (-0.729 - -0.88). The static "quiet" standing balance test was correlated positively with four of the five gait test conditions for the % of stride spent in total support on the affected side (0.744 - 0.765) and negatively at the same test conditions for the % of stride spent in single support on the affected side (-0.731 - -0.763). For the "posterior" static weight shift, a negative correlation was found at three of the five gait test conditions for single support AR. The "D1F" balance test was found to positively correlate with all five gait conditions for total support on the affected side (0.742 - 0.817).

Unlike the left hemiplegic sample the right hemiplegic group did not demonstrate the same correlation between the gait measures and the cardinal plane weight shift to the hemiplegic side where only one significant relationship was detected. Similarly, 10 correlations were detected for the "posterior" weight shift compared to none for the left hemiplegic group. A similarity was found in the number of correlations detected for the "TD1" dynamic shift. Intuitively, given the side of the hemiplegia, it may have been thought that the "TD2" transfer would have yielded some correlations in the right hemiplegic group, however, this was not the case. More understandable was the relatively high number of strong relationships found in the "D1F" position where the test required weight shift through the right, hemiplegic limb. From the findings of this part of the study, little difference was found between the cardinal plane and static diagonal shifts although more relationships were evident in the former group of tests. The dynamic transfer condition "TD1", however, demonstrated the greatest number and the strongest relationships with the gait variables. All of these relationships indicated that X axis CP positions which were displaced toward the negative (left), indicating an inability to load the hemiplegic leg, were associated with more abnormal gait performance. For example, in the "quiet standing" test position, the more the CP was displaced to the left, the greater was the % of stride spent in single support on the unaffected side and, in the "TD1" balance test, the more the CP was displaced to the left, the greater was the single support AR.

The lack of a consistent pattern between the two hemiplegic groups was also observed in the position of the CP in the Y axis (Appendix 6, page 251 and 252). A total of 65 correlations were detected between the gait and balance of the left sided hemiplegic group compared with only 16 for the right sided group. Again, the high critical r values, brought about by the small number of right sided hemiplegic subjects, which were required to be set to ensure that a probability of < 0.05 that the correlation was significantly different from 0, probably accounted for this discrepancy.

In the left hemiplegic sample, the "posterior" cardinal plane shift was correlated with 29 gait parameters, the "D1B" balance test with 12 and the "D2B" test with 8. The "posterior" weight shift was negatively correlated with all five relative speed measures at a moderate to high level (-0.589 - -0.787) and all five relative stride length values (-0.765 - -0.843) which could be considered to be high. This same balance test was positively correlated with all five gait test conditions at a moderate to high level for the % of stride spent in braking double support on the unaffected side (0.599 - 0.84). In the "D1B" balance test, the 12 correlations were somewhat scattered except for the moderate, positive relationship with four of the five measures for step length AR (0.623 - 0.714). The balance test of "D2B" was correlated positively for 4 of the five gait test conditions for % of stride spent in total support on the affected leg (0.637 - 0.708) and, correlated negatively for the same walking conditions for % of stride spent in single support on the unaffected side(-0.657 - -0.745) both of which could be considered to be moderate to high. Moderate positive correlations were detected between the "D1F" balance test and four of the five relative stride length values (0.617 - 0.683).

The strength of the correlations suggest that the "posterior" static balance test demonstrated the strongest correlations with gait compared with the diagonal tests. However, the number of correlations detected in the cardinal tests other than the "posterior" shift was negligible whereas the diagonal shifts particularly in the D1 direction showed a greater yield. It was interesting to note that the two dynamic diagonal weight shift tests were correlated with a total of one gait variable at one speed. Therefore, in general terms, the greater the ability of the left hemiplegic subjects to displace the CP in the Y axis from 0 during the balance tests, the better was their walking ability. For example, a greater excursion of the CP into the negative Y coordinates was associated with a greater relative speed in the "posterior" shift test and, in the "D1F" test positions, a greater ability to transfer the CP forward was associated with an increase in relative stride length.

Only 16 correlations between the gait parameters and the position of the CP on the Y axis during the balance tests were detected for the right hemiplegic group with 8 being found at the "D2S" condition. This test position was strongly negatively correlated with four of the five gait conditions for single support AR (-0.718 - -0.942). The correlation between "D2S" and the single support AR at the "fast" walking condition demonstrated an r value of -0 .942. Unlike the left hemiplegic sample, only one correlation was found between the CP on the Y axis and the gait parameters for the cardinal plane shifts, the remainder being detected for the diagonal weight shifts. Again, the limited sample size of the right hemiplegic group probably accounted for this finding. The small number of correlations detected prevented any definitive conclusions to be proposed other than the fact that in the "D2S" balance test position, the more posterior the location of the CP in the Y axis, the greater was the single support AR. This is an interesting finding because in the "D2S" position, the right sided hemiplegic sample stood with the affected leg in a posteriorly situated position. Thus, it would be thought that a posteriorly situated CP in the Y axis would be indicative of an ability to load the affected limb. However, this was not found. It is possible that subjects with less stable balance did not possess the ability to shift their CP anteriorly which would require placing the hemiplegic leg in an extended position at the hip joint and rotating the tibia rotated forwards on the fixed foot. This would result in a primitive propping of body weight through the affected leg with reduced ability to move the CP forwards towards the midline as might be expected in this test. This phenomenon may be worthy of further study.

Therefore, the tests conducted on both groups of hemiplegic subjects showed that, in general terms, the ability to shift weight was reflected in the gait pattern with the parameters influenced by the type of balance test conducted. Relationships were present when the posture required by the balance test favoured the unaffected leg and when the particular test placed demands on the affected leg. This correlation analysis appeared to show that difficulties in placing the CP through the affected leg and posteriorly had a negative effect upon the gait pattern of the subjects

There was little to choose in terms of correlation superiority between the static cardinal plane tests and the static diagonal tests with the cardinal plane tests yielding more correlations in terms of the total relationships detected when adjustments were made for the disparity in the number of tests conducted for each procedure (5 cardinal plane versus 6 diagonal tests). The strength of the correlations appeared similar for both static cardinal and diagonal tests. The dynamic diagonal test showed some potential but only in the X axis.

Dispersion Index (Disp.):

Because the Disp. was a measure of postural sway and was not influenced by the side of the hemiplegia, the hemiplegic sample were considered as one group. However, it was necessary to modify the the terms describing the tests, as discussed earlier, to standardise for the side of the hemiplegia. Because the hemiplegic sample were considered as one group, the critical r value was reduced to 0.4438. The correlations detected between the Disp. obtained for the balance measures and the gait variables are shown in Appendix 6 (page 253).

In terms of the static cardinal plane balance tests, a total of 12 relationships were detected, 5 of which were found for the balance test to the "unaffected" side. These correlations were all positively related to the gait measure of single support AR and could only be considered moderate in strength at best (0.487 - 0.641). Only four scattered correlations were found between Disp. and the gait values when the static diagonal weight shifts were analysed. However, 45 relationships were detected between Disp. and the gait measures in the dynamic weight transfer tests of "TBAFF" and "TFAFF". This was not a surprising finding in that the Disp. measure is strongly influenced by variability of position of the CP. By definition, the diagonal weight transfer tests were designed to transfer the position of the CP from the rear placed foot to the foot positioned anteriorly. Therefore, it was not surprising that correlations between the Disp. of the weight transfer tests and gait were found. The "TFAFF" test yielded 20 significant relationships while the "TFAFF" test produced 25 significant correlations.

The "TBAFF" test correlated positively with all five relative speed conditions, all five relative stride length tests and four of the five for the % of time spent in single support on the affected leg . Four of the five gait values for the % of time spent in total support on the unaffected side correlated negatively. The strength of the correlations with relative speed could be described as moderate (0.502 - 0.711) while those for the relative stride length measures ranged from low to moderate (0.466 - 0.698) as did those for the % of stride spent in single support on the affected side (0.455 - 0.538) and % of stride spent in total support on the unaffected side (-0.46 - -0.554).

For the "TFAFF", balance test the Disp. values correlated positively with the all gait test conditions for relative stride length, four of the five for relative speed and four of the five for the % of stride time spent in single support on the affected side. Negative correlations were detected between the Disp. values for this test and all five gait measures for the % of stride spent in total support on the unaffected side and four of the five for single support AR. The strength of the correlations ranged from moderate for relative stride length (0.497 - 0.68), the % of stride spent in total support on the unaffected side (-0.521 - -0.628) and the % of stride time spent in single support on the affected side (0.539 - 0.605). The relationships between Disp. in this test and relative speed (0.47 - 0.645) and single support AR (-0.463 - -0.498) were weaker and could be considered to be relatively low.

With regard to the relationship between Disp. and the gait measures, few correlations existed in the static weight shift balance tests with the cardinal plane shifts being correlated with a few more variables than the diagonal shifts although the low numbers in both groups preclude definitive comparison. It could be concluded that the relationship between the balance measure of Disp. and gait is weak as far as the static tests are concerned. However, Disp. values for the diagonal weight shifts in both directions were correlated with a number of gait measures. As such, the Disp. values for these diagonal tests appeared to possess some potential as shown by their relationship with several gait factors. This probably resulted from the dynamic nature of the tests.

In summary, in the static tests, a higher Disp. was associated with a greater single support asymmetry ratio which is a characteristic feature of the hemiplegic gait pattern. This would confirm previous research and clinical wisdom that poorer postural stability, as indicated in this study by the Disp., is associated with a more asymmetrical gait pattern. However, in the dynamic diagonal tests, higher Disp. values were indicative of greater ability to shift weight which, in turn were associated with more normal gait measures. For example, the positive correlation between the Disp. and relative speed in the "TBAFF" condition indicates that higher Disp. values were associated with faster walking speeds. Conversely, the negative correlation between the Disp. and single support AR in the "TFAFF" condition is indicative of increasingly abnormal AR in single support, a key hemiplegic gait indicator, with decreasing Disp. in this testing condition. The promising relationship between gait and the dynamic diagonal weight transfer tests was, again, clearly demonstrated.

Percentage of test time spent in concentric rings representative of % of body weight:
These balance measures were also indicators of the magnitude of postural sway. However, detecting correlations using this balance parameter was complicated by the high number of 0 values encountered particularly in the rings representing 20% - 80% of body weight. To compensate for reducing the number of data points analysed, the critical *r* values had to be adjusted individually for each test to prevent misleading correlations from being identified.

With regard to the relationship between the gait parameters and the percentage of time spent within the 5% ring, the critical *r* value of 0.4438 or greater was used as representative of a 0.05 level of probability that the correlation was different from 0 except for the two dynamic tests "TBAFF" and "TFAFF" In those tests five and seven zero values respectively were present. As a result, the critical *r* values were adjusted for the "TBAFF" to 0.5139 (df = 13) and for the "TFAFF" tests to 0.5529 (df = 11).

The significant *r* values for the correlations between the percentage of test time spent in the 5% of body weight ring for all balance tests and the gait parameters are shown in Appendix 6 (page 254).

A total of correlated with 28 significant correlations were detected, 17 of which were accounted for by the static cardinal plane weight shifts and only 1 for the static diagonal weight shifts The dynamic diagonal tests correlating with 10 gait variables. A total of 11 correlations were found for the cardinal plane shift to the "unaffected" side. Five of these correlations were for the five gait values regarding single support AR where the relationships were negative and moderate in strength (-0.511 - -0.647). These correlations indicated that a higher percentage of test time spent in the 5% body weight ring (implying greater postural stability) was associated with a reduced single support AR.

For the dynamic, diagonal tests, three correlations were detected for the test, "TBAFF",while the other diagonal balance test, "TFAFF", was correlated with 6 gait variables, five of which were for step length AR. These latter relationships were positive and strong (0.687 - 0.813) and indicated that the lesser the amount of time spent in the 5% body weight ring the lesser the step length AR. This was an understandable finding indicating that a greater ability to shift CP was associated with more symmetrical step lengths.

A similar picture emerged for the relationships between gait and the % of time spent within the 10% body weight ring (Appendix 6, page 255). A total of 39 correlations were detected of which 19 were accounted for by the cardinal plane shifts compared with 8 for the diagonal weight shifts. The remaining 11 correlations were between the gait measures and the dynamic diagonal shifts. The "unaffected" weight shift accounted for 10 relationships, 5 of which were positively correlated with all five gait measures for % of stride spent in single support on the unaffected leg indicating that an increasing amount of time spent in the 10% body weight ring was associated with an increase in the % of stride spent in single support on the unaffected side. However, the strength of these positive relationships could be described as low to moderate (0.461 - 0.66).

Nine positive correlations were detected in the dynamic weight shift of "TFAFF" while two were detected for the "TBAFF" condition. For the "TFAFF" test, five of the correlations were with all gait conditions relating to step length AR. The strength of the correlations with step length AR was moderate (0.489 - 0.689) and indicated that the greater the time spent in the 10% body weight ring, the greater was the step length asymmetry. As this was a dynamic test, a higher percentage of time spent in the inner body weight rings was indicative of a decreased ability to shift CP. Thus, for this correlation a weaker balance performance was associated with an accentuation of the hemiplegic gait pattern, in this instance, step length AR.

When the percentage of test time spent in the 20% body weight ring was correlated with the gait variables, the frequency of occurrence of significant relationships was slightly greater in the static cardinal plane shifts (4) than the diagonal (2) but the dynamic weight shifts produced the greatest yield with "TBAFF" producing twenty four and "TFAFF" yielding three (Appendix 6, page 256). Of the static balance tests, the "unaffected" weight shift again produced the most correlations (4), all of which were with four of the five gait values for single support AR. These relationships were positive, were on the low end of moderate in strength and were indicative of the fact that as the percentage of time within the 20% ring increased, the single support asymmetry increased. In the dynamic balance tests, twenty five correlations were found between the gait values and the "TBAFF" test. Moderate negative correlations were found for five of the five relative stride length measures (-0.571 - -0.650), four of the five relative speed values (-0.5 - 0.61) and four of the five gait measures for % of stride time spent in single support on the affected side (-0.546 - -0.644). Moderate positive correlations were found between the five gait values for % of stride spent in total support on the unaffected side and this balance variable (0.539 - 0.661). These correlations all indicated that smaller percentages of test time spent in the 20% body weight ring, which is indicative of less postural sway and implies greater postural stability, were associated with a more normal gait pattern. For example decreasing % of time spent in this ring were associated with a longer relative stride length, an increased relative speed and an increase in the % of stride spent in single support on the affected leg.

None of the static tests for the 40%, 60% and 80 % of body weight rings were analysed because of the high number of registered zero percentages of time spent in each of these tests. This was done because any correlations detected would have been derived from only a few subjects thus biasing the findings in an unacceptable manner and nullifying the ability to legitimately generalise the results. This problem did not arise with the dynamic weight shift tests at the 40% body weight ring parameter. In the dynamic diagonal shifts for this ring, thirteen correlations were detected for the "TBAFF" condition and one for the "TFAFF" test (Appendix 6, page 257). The "TBAFF" correlations were generally low to moderate in strength and included positive relationships for four out of the five gait measures for relative speed (0.472 - 0.517) and negative relationships with four out of the five gait values for stride time (-0.497 - -0.564). Given that a more normal gait pattern is associated with an increased relative speed and a decreased stride time compared with the more characteristic gait pattern, these results indicate that a higher percentage of test time spent in the 40% body

weight ring in the dynamic test is associated with an improving gait pattern. This was an understandable finding in that those hemiplegic subjects with a better ability to transfer weight would have spent a greater percentage of time in the 40% body weight ring during this test. These findings, however, were restricted to the "TBAFF" test in which the hemiplegic subjects stood with the hemiplegic limb back and the non hemiplegic leg forward, transferring weight from the hemiplegic leg to the normal leg.

No significant relationships were detected between the gait parameters and the time spent in the 60% or 80% of body weight rings during either of the dynamic diagonal weight shifts.

Overall, 114 correlations were detected between the gait measures and the percentage of time spent in the various % of body weight rings measured in this study. The distribution of these relationships is shown in Table 4.XXXI. Of these, 62 were accounted for by the dynamic diagonal weight shift with the "TBAFF" test accounting for more than twice that for the "TFAFF". Of the remaining 52 correlations, 40 were attributed to the cardinal plane shifts, 25 of which were found in the weight shift to the "unaffected" side, and only 12 were detected in the static diagonal shifts. This would tend to suggest that, for this measure of balance performance, the static cardinal plane shifts are more likely to be correlated with gait factors than the static diagonal tests. For these particular tests, the dynamic diagonal tests appeared to be useful particularly the "TBAFF" condition. However, it should also be noted that the "TFAFF" position accounted for a substantial number of the correlations for these tests which also underscores the potential of both dynamic, diagonal tests in this testing context.

Table 4.XXXI: Frequency of occurrence of correlations at each % of body weight ring and the totals of correlations for each balance condition.

	5%	10%	20%	40%	60%	80%	Totals
Quiet	0	0	0	0	0	0	0
Anterior	6	6	0	0	0	0	12
Posterior	0	3	0	0	0	0	3
Unaffected	11	10	4	0	0	0	25
Affected	0	0	0	0	0	0	0
BBAFF	0	3	0	0	0	0	3
SBAFF	1	4	2	1	0	0	8
FBAFF	0	0	0	0	0	0	0
BFAFF	0	1	0	0	0	0	1
SFAFF	0	0	0	0	0	0	0
FFAFF	0	0	0	0	0	0	0
TBAFF	3	2	24	13	0	0	42
TFAFF	7	9	3	1	0	0	20
Totals	28	38	33	15	0	0	114

Range of Weight Shift:

The correlations for the range of weight shift tests, in the X and Y axes as well as the two static diagonal weight shifts, are shown in Appendix 6 (page 258).

Range of weight shift in the X axis and for the "BFAFF" condition were responsible for all but one of the 61 correlations with the gait parameters. The "Range X" accounted for 34 of the correlations detected while the "Range DBAFF" correlated with 26 of the gait measures. The "Range Y" accounted for the other correlation. The "Range X" correlated positively with all five gait values for relative speed at a moderate to high level (0.51 - 0.743) as well as % of stride time spent in single support on the affected leg again at the moderate level (0.545 - 0.639). Further positive correlations were detected between four of the five measures for relative stride length at moderate strength (0.619 - 0.644). The "Range X" balance measure was negatively and moderately correlated with all five gait measures for the

% of stride spent in total support on the unaffected side (-0.562 - -0.655) and all five gait measures of % of time spent in braking double support on the unaffected side (0.47 - - 0.657). Additional negative correlations were detected for four of the five gait values for stride time, all of which were moderate in strength (-0.544 - -0.626), and four of the five gait measures of % of stride spent in braking double support on the affected side the latter of which could be regarded as low to moderate in strength (-0.45 - -0.526).

The "Range BBAFF" (hemiplegic leg back, weight back) was correlated with four gait parameters at all five of the gait testing conditions. All four could be regarded as moderate in strength. Positive correlations were detected for relative stride length (0.556 - 0.69) and % of stride spent in single support on the affected side (0.512 - 0.586) while negative correlations were found for single support AR (-0.51 - -0.558) and % of stride time spent in total support on the unaffected leg (-0.533 - -0.586). Positive correlations were found for four of the five gait relative speed measures in this balance testing condition all of which were moderate in strength (0.501 - 0.613). Thus the lateral static cardinal plane shift appeared to be correlated with more gait values than the diagonal range. The range on the X axis was negatively correlated with all five gait values for both stride time and the % of stride spent in double braking support neither of which were correlated with the "DBAFF diagonal range of shift. However, the "DBAFF" test performed reasonably well, negatively correlating with one gait parameter which the "Range X" test did not, namely single support AR. There was little difference between the two tests in terms of the strength of the correlations, although the correlations with the "Range X" procedure appeared to be slightly stronger. In both tests, a greater range was associated with a more normal walking performance. For example, increasing range of weight shift in the X axis was associated with an increase in relative walking speed and in the "DBAFF" condition, a greater range was associated with a smaller single support AR.

Range of Walking Speed with Balance Variables:

Because of the interesting gait finding in this study that the range of walking speeds was significantly diminished in hemiplegic subjects, it was decided that it may be useful to search for correlations between range of walking speeds and the balance variables.

A high negative correlation was found between the range of walking speed and the position of CP in the left hemiplegic sample (-0.766) indicating that the greater the ability to shift into the negative range on the X axis (the left) the greater the range of walking speed. A moderate negative correlation was also found between the same two variables in the balance testing position of D1B in which the left hemiplegic sample stood with weight back over the hemiplegic leg. Again a greater shift to the left was associated with a greater range of walking speed. No significant correlations between the position of CP in the X axis for the right hemiplegic group were detected, as was the case for the Y axis for the left hemiplegic sample. A strong positive correlation was found between the position of the CP on the Y axis during the "posterior" cardinal plane displacement and range of walking speed (0.776). This indicated that the more anterior the CP in this test, the greater the range of walking speed. This was a surprising finding and is similar to the earlier finding for the same hemiplegic group that a relationship existed between the CP on the Y axis in the "D2S" position and single support AR. The same reasons may apply in this instance. Further possible explanations for this finding are that these stroke subjects had suffered their stroke some time previously. As a result, it was possible that functional adaptations had permitted them to become adept at producing an abnormal but effective gait pattern. An addition, it is possible that although the range of walking speed was high in relative terms, the gait pattern may have been in the slow range. A final interpretation of this finding and its inconsistency with most other findings in this study is that the relationship may have been spurious.

Two low correlations were found between range of relative walking speed and the Disp. measures for the two dynamic weight shifts "TBAFF" (0.468) and "TFAFF" (0.458) both of which were indicative of a greater Disp. being associated with a greater the range of walking speed, an expected finding given the dynamic nature of these tests.

When correlation coefficients were calculated between the range of relative speed and the range of weight shift balance test conditions, two significant relationships were detected.

The range calculated in the cardinal plane, "Range X", correlated positively and at a moderate level with range of relative speed (0.641) while the "Range DBAFF" correlated less strongly (0.493).

One objective of this study was to identify relationships between gait measures and balance parameters in the hemiplegic subjects studied in this research project. Numerous correlations and the strengths of those associations were identified and defined between balance and gait which demonstrated the relationship between those two factors. Although cause and effect cannot be established from studies of correlation, this study has provided the platform for further research into the relationships between balance and gait. However, this author is aware that the "strongest" of correlations here revealed (see Appendix 6, pages 248 - 258) still leave a great deal of unexplained variance. However, the nature of hemiplegia, insofar as it affects gait and postural balance is such that meaningful rehabilitative interventions can be planned around the correlations revealed in this study.

From a rehabilitation perspective, ample evidence was produced to support the notion that improving balance would result in a more normal gait pattern. Conversely, but less plausible from an intuitive point of view, it could be proposed that by improving gait, balance would be enhanced. If the improvement of gait was the objective of rehabilitation, it would seem desirable to improve those balance competencies which seem to impact on gait. This study has identified many avenues of promise in this regard.

From a testing viewpoint, little difference was found between the cardinal plane weight shifts and the diagonal shifts. Thus, it can be concluded that the diagonal weight shifts, although seemingly more relevant to gait, were no better correlated to gait than the cardinal plane shifts. However, before definitive statements are made, several factors have to be considered and, perhaps, studied further. The motor strategies used by subjects during the diagonal weight shift tests were extremely variable. This variability may have affected the results obtained. Clearly, further study of this important variable is required. Another factor which may have influenced subject performance during the diagonal weight shifts was the position of the footplates. More revealing data may have been forthcoming if the footplate locations had been placed further apart. Again this factor should be studied further. Finally, the data obtained from the dynamic weight shift may have yielded more information had the timing of the dynamic shift been controlled in some way. Given the potential of the dynamic tests and their possible rehabilitation consequences, this is an important area of further investigation.

CHAPTER 5

SUMMARY, CONCLUSIONS AND RECOMMENDATIONS

Studies of the gait patterns of people who have suffered a stroke have tended to examine only one walking speed ("free", "preferred" or "comfortable") and, as a result, little is known about the range of walking speeds of which hemiplegic subjects are capable and how walking speed affects the other temporal and spatial gait parameters as well as the characteristic gait asymmetries. Deficient balance in the hemiplegic leg, which is thought to preclude effective limb loading particularly during single support, has been proposed to be a contributing factor to the abnormal gait pattern of this clinical group. A common objective during stroke rehabilitation, therefore, has been to retrain the ability to transfer weight through the affected lower limb in an attempt to re-educate a more normal gait pattern. Although a few studies have examined this contention, little is known about the precise relationship between balance in the hemiplegic leg and the gait disorder particularly when balance performance is tested with the lower limbs of subjects placed in positions which appear more relevant to the gait pattern than the traditional balance testing protocols.

HYPOTHESES

It was hypothesised that:

1) The temporal and spatial parameters of gait of fully ambulant hemiplegic subjects significantly differ from those of normal age- and sex-matched controls at a variety of psychometrically derived walking speeds designed to identify the extremes of range of gait velocity.

2) The balance performance of fully ambulant hemiplegic subjects differs significantly from that of normal subjects.

3) Significant relationships exist between gait kinematic factors and balance performance and that these relationships may be enhanced by measuring balance performance with the feet of the subjects in positions which are more relevant to the double support components of the gait cycle.

SUMMARY OF PROCEDURES

The purpose of this study was to examine the spatiotemporal gait kinematics and various elements of balance performance of a group of hemiplegic subjects and compare their performance with a sample of age- and sex-matched controls. Before the project was conducted, two pilot studies were undertaken to test the protocol of the proposed study. Gait and balance data were collected from two subjects, a hemiplegic person and a normal, for both pilots. The hemiplegic subject was a 27 year old, male, stabilised stroke survivor who was fully ambulant and demonstrated clear subjective evidence of an asymmetrical "hemiplegic" gait pattern. His data were compared with those of the normal, 31 year old male.

In the first pilot, the gait analysis system performed without incident collecting, storing and analyzing temporal and spatial kinematic data. This was expected in that the gait system had been thoroughly validated by Crouse *et al.*, (1987) and had been used to collect gait data in a number of published articles in refereed journals (O'Brien *et al.*, 1983; Turnbull and Wall, 1985; Wall and Turnbull, 1986; Wall and Turnbull, 1987; Bagley *et al.*, 1990; Gaudet *et al.*,1990).

Obtaining valid and reliable balance measures, however, was more complex. The measurement system tested in the first pilot consisted of a custom-built sway platform (SP) interfaced with an Apple II+ computer system. Centre of pressure (CP) was measured by means of a pair of strain gauges attached to each of four beams which supported the platform. Subjects were tested in a total of five positions. In the first trial, subjects were requested to stand in a quiet standing position. In the second trial, subjects were asked to shift their weight as far as they could to the right side and on the third trial they were asked to transfer their weight as far as possible to the left side. The subjects were then required to displace their weight as far forward as they could followed by as far backward as possible. During these trials, the mean position of the CP was calculated on both the "X" and "Y" axes along with the standard deviation from the mean, on both axes, for each during the 30 s data collection period. These data provided a measure of the ability of the subject to shift and sustain these transfers of CP. The length of the line traced by the CP over the data collection period was also determined to provide a measurement of the magnitude of postural sway in all test positions. Finally, the mean values in both the anteroposterior and lateral positions were used to calculate and provide an indication of the range over which subjects voluntarily shifted weight in both lateral and anteroposterior directions.

Problems, were experienced, in relation to the performance of the SP. While collecting data, the healthy individual consistently moved his CP beyond the limits of the equipment. In order to overcome this, it was necessary to have the normal subject stand with his feet closer together. However, this solution would have required that both subjects stand with their feet together during balance testing so that valid comparisons could be made. This scenario would have destabilized the already fragile balance of the hemiplegic subject so creating additional difficulties. An additional problem was that the position of the feet on the SP had to be traced on to a template, situated on the surface of the platform, so that the geometric centre of the base of support could be calculated. With this system, this was the only way to identify the position of the CP of the subject relative to the centre of his base of support. This was time consuming and reduced the consistency of the testing procedure creating problems if the subject had to step off the platform for any reason. Another issue which became obvious in this pilot study was that the weight shifts requested appeared unrelated to gait function and did not resemble any component of the walking pattern, thus raising the question of the relevance of such a testing protocol. The weight shifts requested, although typical of those required during rehabilitation and the same as those used by previous researchers (Murray *et al.*,1975; Dettman *et al.*,1987), were cardinal plane shifts, and appeared unlike those associated with the gait cycle. Thus, it appeared, it might be appropriate to consider weight shifts with one foot in front of the other such as those which occur during the double support phases of the gait cycle. Attempts to measure diagonal shifts on the SP were unsuccessful mainly as a result of the limited size of the supporting surface. The testing period of 30 s also appeared to be excessive in that fatigue appeared to become a factor towards the end of the test thus biasing the validity of the balance data.

A second pilot procedure tested a revised protocol for measuring the balance parameters. Balance data were obtained from the same two subjects but gait was not measured in this pilot. To test the balance parameters, the commercially available Chattecx BalanceSystem™ (Chattanooga, Tennessee) was used. This system measured similar parameters to the sway platform but the method of obtaining this information was different. Subjects stood on two footplates both of which had anterior and posterior components. As a result, the footplates could be adjusted to accommodate the foot lengths of each subject. The positions of the footplates, which represented the base of support of the subject, were input to the computer which calculated the geometric centre of the base of support. Thus, the position of the CP of the subject could be studied in relation to the base of support of the subject. Subjects were tested while performing the same weight shift movements as were tested in the previous pilot. In addition, a series of diagonal tests were included. For these, the subjects were requested to stand first with the left foot back and the right forward (D1) in a manner which resembled the foot placements during the double support phase of the gait cycle. Subjects were asked, in random order, to weight bear on the posteriorly positioned foot (B), then symmetrically with weight through both legs (S), and then through the anteriorly placed foot (F). The procedure was repeated but with the feet placed in opposite positions in a manner similar to that assumed during the other double support phase of gait (D2). Data were collected for 10 s for each weight shift condition.

This second pilot study confirmed that the Chattecx BalanceSystem™ resolved all of the shortcomings of the SP used earlier, particularly in relation to calculating the position of the geometric centre of the base of support and so dealing with the issue of the location of the centre of the base of support of each subject. The problem of the normal subject exceeding the bounds of the equipment was also resolved. Similarly, because the footplates could be positioned independently of each other, the use of this system permitted the measurement of diagonal weight shifts which may be more strongly correlated to gait performance than the cardinal plane protocols which have been typically used in previous studies. As a result of the pilot studies, it was decided to proceed with the resistive grid walkway to measure the spatiotemporal parameters of gait and the Chattecx BalanceSystem™ to measure balance performance.

Twenty, stabilised, fully-ambulant hemiplegic subjects (12 men and 8 women) with a mean age of 57.2 year (±10.7) were compared with 20 age- and sex-matched controls. The temporal and spatial kinematics of gait were measured and subjects received a series of instructions which yielded five different gait velocities ("very slow", "slow", "free", "fast" and "very fast"). A number of balance parameters were then measured while the subjects maintained a variety of weight shift postures both with the feet parallel to each other and then in diagonal positions similar to those assumed during the double support phases of the gait cycle. The location and variability of the centre of pressure (CP) were measured and the ranges over which subjects could shift CP were calculated for both the cardinal plane and diagonal shifts.

Statistical analysis of these data were performed as follows: The temporal and spatial parameters of gait of hemiplegic subjects were compared with those of the normal age and sex-matched controls utilizing two factor (subject type and walking condition), repeated measures ANOVA. Repeated measures on one factor (walking condition) were then conducted to identify differences between walking speed conditions for the hemiplegic group and the normal sample. *Post hoc* analyses utilizing the Scheffé test were then used to identify the location of the significant differences. The parameters compared included walking velocity, stride time, stride length, single support asymmetries, double support asymmetries, step length asymmetries, the durations of single support, total support and double support phases of the gait cycle and the range of walking speeds of which the subjects were capable. Unmatched *t* tests were used to test for differences between hemiplegic and normal performances at each of the walking conditions.

The balance performances of the hemiplegic subjects were compared with those of the control sample utilizing unmatched, two tailed *t*-tests based upon: the position of the CP on the X and Y axes; the variability of the position of CP as measured by the Disp. and the percentage of test time spent in the various body weight rings; the ability of the subjects to voluntarily shift CP in a variety of directions and; the range over which CP could be shifted relative to the base of support in both cardinal plane and diagonal directions. Specifically, parameters compared included the mean position of the CP on the X and Y axes (CPX and CPY) and the magnitude of postural sway as indicated by the variance of the mean position of the CP (Disp. and the % of test time spent within 5%, 10%, 20%, 40%, 60% and 80% of body weight displaced from the mean CP).

Relationships between the gait and balance variables were then explored in the hemiplegic sample by examining correlation coefficients. Pearson product moment correlations were calculated between all balance and gait variables.

The 0.05 level of probability was used throughout this study as the level of significance.

SUMMARY OF RESULTS

The gait patterns of the hemiplegic subjects were significantly slower than that of the controls at all 5 walking conditions. This deficiency in gait velocity was clearly highlighted by the finding that the "fastest" speed of the hemiplegic sample was no different, statistically, from the "slowest" walking condition for the control group. In the control group, there were significant differences between each adjacent walking speed, however, in the hemiplegic group, there were no differences in walking speed between the "slowest" and "slow", between the "slow" and "free", and between "fast" and "fastest" walking test conditions in the hemiplegic sample. Only one of the adjacent walking condition pairs, that between the "free" and "fast" speeds, was significantly different. This would suggest that, when compared to the control group, there was a limited ability of the hemiplegic group to consciously vary walking velocity. Range of walking speeds was also found to be significantly different between the two groups with the hemiplegic sample possessing less than half the range of the controls.

Significant differences were found between the groups for both stride time and stride length with the stride times quicker and the stride lengths longer in the normal sample. These findings could be attributed to some extent to the differences in walking speed. However, at the walking speeds which were common to both groups, the hemiplegic sample took quicker and shorter strides than the controls suggesting a cautious gait pattern which is characteristic of the gait pattern of people with a balance disorder. As a result, this profile lends further support to the contention that the balance of hemiplegic subjects is compromised.

When the various phases of the gait pattern were compared between the two groups, the hemiplegic sample spent significantly longer durations in total support on the unaffected leg, significantly shorter durations in single support on the affected leg and significantly longer durations in braking double support on both legs than the controls. A central feature of the durations spent in these phases of the gait pattern was the marked asymmetries present in the hemiplegic sample. This finding was confirmed when the AR for the durations of total support, single support and braking double support and the AR for step length were compared. The hemiplegic sample were significantly more asymmetrical than the controls. In three of the four parameters, the magnitude of the asymmetries in the hemiplegic sample remained the same irrespective of walking speed. However, single support AR diminished with increasing walking speed.

A sixth walking condition, undertaken only by the hemiplegic sample, required that the subjects walk in a manner where they attempted to exert conscious control over their step length and support phase asymmetries in an effort to produce a more normal, symmetrical gait pattern. The result of this trial was the production of an identical pattern to that produced at the "free" walking speed condition. This would suggest that during preferred walking speed, hemiplegic subjects have the perception that they are walking optimally.

The walking performance of the group of hemiplegic subjects tested in this study when compared with normals, therefore, was slow, deficient in range in terms of temporal and spatial performance and asymmetrical. In addition, the hemiplegic gait pattern was relatively unmodifiable and variability, within each parameter, was high despite the application of strict inclusion and exclusion criteria.

When the positions of CP in both the X and Y axes were considered, the CP positions of the hemiplegic subjects were displaced towards the unaffected side and, in the cardinal plane weight shifts, their ability to shift CP posteriorly towards the hemiplegic side was significantly less than that of the controls. In the diagonal shifts, the inability of the hemiplegic sample to shift weight posterolaterally over the affected leg was clearly demonstrated. In addition, the diagonal shifts showed that, in the anterior diagonal weight shift over the affected leg, the location of the CP in the Y axis was not different from the controls but was significantly different in the X axis indicating that the hemiplegics were able to shift CP anteriorly but not laterally. When the mean values for all mean CP positions were compared, the diminished ability of the hemiplegic subjects to control CP particularly to the hemiplegic side and posteriorly was clearly evident.

Both methods of measuring postural sway (Disp. and the % of test time spent in the % of body weight circles) showed that, in the majority of test positions, the hemiplegic sample had a significantly greater postural sway than the controls thus implying less stable balance. For this balance testing procedure, the diagonal balance tests revealed more differences than the cardinal plane shifts.

Comparison of the ranges over which CP was shifted, laterally and anteroposteriorly in the cardinal plane shifts and posterolaterally to anterolaterally in both diagonal directions, revealed significant differences between the two groups with the hemiplegic sample less able than the controls to shift weight in all four directions. Laterally, the hemiplegic sample were able to transfer weight 64.8% of that attained by the normal sample while, anteroposteriorly, the hemiplegic performance was 49.3% of normal. The relative performance of the hemiplegic group was 51.1% and 47.3% of the normals for the D1 (DBAff) and D2 (DFAff) transfers, respectively.

The dynamic diagonal weight shift measurements confirmed the range of weight shift data as indicated by the Disp. measurements and the percentages of time spent in the body weight circles that the control sample shifted CP over significantly greater distances than the hemiplegic group. This result was consistent with the static balance tests. This finding provides evidence to support the contention that the ability of the hemiplegic group to shift weight dynamically is significantly less when compared to normal. It seems reasonable to speculate that this limitation in performance is likely to affect the gait pattern of the hemiplegic group.

When correlation coefficients were calculated between the gait and balance measures for the hemiplegic sample, many significant relationships were detected. However, these correlations must be viewed with caution. A significant correlation indicated that the correlation was different from 0 (no correlation) at the $p < 0.05$ level. As a result, an r value of 0.4438 was significant when all hemiplegic subjects were considered. It must be noted that such an r value is less than impressive in accounting for the variance of the data. It is most important to temper these findings with the fact that most of the "strongest" correlations described (see Appendices6.1 - 6.10)still leave a great deal of unexplained variance. However, the nature of hemiplegia, insofar as it affects gait and balance performance, is such that meaningful rehabilitative interventions can be planned around the correlations revealed in this study.

The hemiplegic sample was divided into two groups (left and right) for the correlations between the gait parameters and CPX and CPY because the values obtained for CP were dependent on the side of the hemiplegia. Surprisingly, the pattern of correlations between the right and left hemiplegics did not mirror each other. This may, in part, have been due to the smaller number of subjects in the right hemiplegic group which necessitated the attainment of a critical r value of 0.7067 before a < 0.05 level of significance that the correlation was different from 0 could be accepted. The tests conducted on both groups of hemiplegic subjects showed that, in general terms, the ability to shift weight, as measured by the location of CPX and CPY, was positively related to the normality of the gait pattern with the more sensitive parameters influenced by the type of balance test conducted. Relationships were present when the posture required by the balance test stressed the unaffected leg and when the particular test placed demands on the affected leg. This correlation analysis appeared to show that difficulties in placing the CP through the affected leg and posteriorly had a negative effect upon the gait pattern of the subjects.

There was little to choose in terms of superiority between the static cardinal plane tests and the static diagonal tests in detecting correlations with the cardinal plane tests yielding more correlations in terms of the total relationships detected when adjustments were made for the disparity in the number of tests conducted for each procedure (5 cardinal plane versus 6 diagonal tests). The strength of the correlations appeared similar for both static cardinal and diagonal tests. The dynamic diagonal test performed well as a useful testing procedure, being correlated with several gait parameters, but only in the X axis.

With regard to the relationship between Disp. and the gait measures, the static tests showed that a higher Disp. was associated with a greater single support asymmetry ratio which is a characteristic feature of the hemiplegic gait pattern. This would confirm previous research and clinical wisdom that poorer postural stability, as indicated by the Disp., is associated with a more asymmetrical gait pattern. However, in the dynamic diagonal tests, higher Disp. values were indicative of greater ability to shift weight which, in turn were associated with more normal gait measures. For example, the positive correlation between the Disp. and relative speed in the "TBAFF" condition indicates that higher Disp. values were associated with faster walking speeds. Conversely, the negative correlation between the

Disp. and single support AR in the "TFAFF" condition was indicative of increasingly abnormal AR in single support, a key hemiplegic gait indicator, with decreasing Disp. in this testing condition. The promising relationship between gait and the dynamic diagonal weight transfer tests was, again, clearly demonstrated. In comparing the cardinal plane and diagonal testing protocols, few correlations existed in the static weight shift balance tests with the cardinal plane shifts being correlated with a few more variables than the diagonal shifts although the low numbers in both groups precluded definitive comparison. It could be concluded that the relationship between the balance measure of Disp. and gait was weak as far as the static tests were concerned. However, Disp. values for the diagonal weight shifts in both directions were correlated with a number of gait measures. The Disp. correlations for these diagonal tests appeared to possess some potential as shown by their relationship with several gait factors. This probably resulted from the dynamic nature of the tests.

The findings for the Disp. correlations were similar to those found for the relationship between the % of test time spent in the body weight rings because both balance parameters are measures of postural sway. Overall, 114 correlations were detected between the gait measures and the percentage of time spent in the various % of body weight rings measured in this study. Of these, 62 were accounted for by the dynamic diagonal weight shift with the "TBAFF" test accounting for more than twice that for the "TFAFF". Of the remaining 52 correlations, 40 were attributed to the cardinal plane shifts, 25 of which were found in the weight shift to the "unaffected" side, and only 12 were detected in the static diagonal shifts. This would tend to suggest that, for this measure of balance performance, the static cardinal plane shifts are more likely to be correlated with gait factors than the static diagonal tests. For these particular tests, the dynamic diagonal tests appeared to be useful particularly the "TBAFF" condition. However, it should also be noted that the "TFAFF" position accounted for a substantial number of the correlations for these tests which also underscores the potential of both dynamic, diagonal tests in this testing context.

Range of weight shift in the X axis and for the "BFAFF" condition were responsible for all but one of the 61 correlations between the gait parameters and the range of CP shift. The "Range X" accounted for 34 of the correlations detected while the "Range DBAFF" correlated with 26 of the gait measures. The "Range Y" accounted for the other correlation. In the tests which detected correlations, a greater range was associated with a more normal walking performance. For example, increasing range of weight shift in the X axis was associated with an increase in relative walking speed and, a greater range was associated with a smaller single support AR in the "DBAFF" condition.

When correlation coefficients were calculated between range-of-walking-speed and the balance values, a high negative correlation was found between the range-of-walking-speed and the position of CP in the left hemiplegic sample indicating that the greater the ability to shift to the left, the greater the range of walking speed. A moderate negative correlation was also found between the same two variables in the balance testing position of D1B in which the left hemiplegic sample stood with weight back over the hemiplegic leg. Again a greater shift to the left was associated with a greater range of walking speed. No significant correlations between the position of CP in the X axis for the right hemiplegic group were detected as was the case for the Y axis for the left hemiplegic sample. A strong positive correlation was found between the position of the CP on the Y axis during the "posterior" cardinal plane displacement and range of walking speed. This indicated that the more anterior the CP in this test, the greater the range of walking speed. The reason for this finding is unclear and the detected relationship may have been spurious. Two low correlations were found between range of relative walking speed and the Disp. measures for the two dynamic weight shifts "TBAFF" and "TFAFF" both of which were indicative of a greater Disp. being associated with greater range-of-walking-speed, an expected finding given the dynamic nature of these tests.

When correlation coefficients were calculated between the range of relative speed and the range of weight shift balance test conditions, two significant relationships were detected.

The range calculated on the X axis correlated positively and at a moderate level with range of relative speed while the "Range DBAFF" also correlated significantly but less strongly. Both findings implied that a greater range of relative walking velocity was associated with a greater ability to weight shift

A sufficient number of significant correlations, therefore, were detected between the balance measures and gait to support the contention that improving balance should result in a more normal gait pattern. If the objective of a rehabilitation programme was the improvement of gait, it would seem desirable to improve those balance competencies which seem related to gait. The correlation section of this study has identified many avenues of promise in this regard. From a testing viewpoint, little difference was found between the cardinal plane weight shifts and the diagonal shifts. Thus, it can be concluded that the diagonal weight shifts as used in this study, although seemingly more relevant to gait, were no better correlated to gait than the cardinal plane shifts.

CONCLUSIONS

Based on the findings of this study, the following conclusions were drawn:

1. The first hypothesis was retained. The temporal and spatial gait kinematics, as measured in this study, were significantly different between the hemiplegic subjects and the age- and sex-matched controls. The hemiplegic subjects walked slower, were more asymmetrical and possessed a reduced range-of-walking-speeds. They were less able to consciously vary walking speed when compared to the controls and demonstrated that attempts to modify their gait pattern resulted in a gait performance identical to "preferred" walking speed. Many of the differences detected could be explained by the differences in walking speed between the two groups. However, when the gait patterns where the samples walked at similar velocities were examined, the hemiplegic group took quicker and shorter strides. The implication of this finding is that the hemiplegic group walked in a cautious manner.

2. The second hypothesis was retained. The balance performance of the hemiplegic sample was significantly different from that of the controls. The stroke subjects were less able to shift CP, particularly over the affected leg and posteriorly, had an increased postural sway at most test positions, suggesting less stable balance, and could shift CP over significantly smaller ranges than the controls both in the cardinal plane and the diagonal directions.

3. A number of significant correlations between balance and gait were revealed in the hemiplegic sample thus confirming the relationship between gait and stationary balance. However, many of these relationships left high percentages of the variance unexplained. The cardinal plane shifts detected more correlations than the diagonal tests although the diagonal tests revealed useful information from a rehabilitation perspective. The dynamic diagonal weight shift test produced a promising yield of correlations with gait and may be more relevant to gait than static balance tests.

4. The procedure of walking subjects at a variety of psychometrically derived walking speeds, clearly demonstrated the extent of the gait velocity decrement in stroke. This finding would suggest that more attention has to be paid to this competency during rehabilitation. In addition, this technique made it possible to study the effects of different walking speeds on the other gait parameters. As a result this procedure proved effective in revealing more detailed information concerning hemiplegic gait.

5. An important finding was that single support asymmetry diminished with speed. This has important rehabilitation implications. By retraining stroke subjects to walk faster, single support asymmetry may be reduced thus leading to a more symmetrical gait pattern.

6. The use of the Chattecx BalanceSystem™ to measure the balance parameters was most successful. In addition to collecting information on a variety of static balance tests including the diagonal shifts, the results of the dynamic test appear to show promise and may be more relevant to gait because of its dynamic characteristic.

RECOMMENDATIONS

The following recommendations for future study merit consideration:

1. The deficiency of gait speed clearly needs to be addressed during the rehabilitation process. This would require that those re-educating gait find an effective method of instructing the patient how to walk faster. Given that increases of walking velocity are associated with shorter stride times, longer stride lengths or both, instructions designed to bring about these changes might be effective. Alternatively, gait velocity may improve if the subject is simply asked to walk faster although the findings of this present study have demonstrated the limitations of such a method. There is a need to examine this question in future research.

2. The distance between the footplates during the diagonal weight shift procedures was used because of the limitations of the size of the supporting surface on which the footplates rested. Had the footplates been placed further apart or closer together, the data generated from the diagonal tests might have been different. In the normal sample, the feet were closer together than they would be in a typical step. Similarly, footplates for the hemiplegic sample were set symmetrically which, for this group, may have been abnormal. There is a need to investigate the effect on the balance data, derived from diagonal tests, brought about by alterations in the spacing of the footplates.

3. The timing of the dynamic weight shift procedure was standardised in a somewhat variable fashion. When data collection was initiated, the subjects were asked to shift weight from their rear placed foot to the anteriorly placed foot at a self determined speed. Some subjects performed this motor pattern quickly while others took a longer period of time. There is a need to conduct these tests in a manner where the velocity of the weight shift is standardised between subjects thus reducing a potential source of variability.

4. While maintaining the extreme postures in both the cardinal plane and the diagonal tests, subjects were not instructed as to the motor patterns to be used to achieve the desired objective. As a result, different motor strategies were used particularly during the diagonal weight shifts. Further study of these motor strategies is required to differentiate between "normal" responses and those resulting from compensatory maneuvers.

5. It is clear from the present study that hemiplegic subjects have great difficulty in loading the affected limb particularly if the limb is positioned posterolaterally. In the study conducted by Wall and Turnbull (1987), they proposed that their disappointing results may have been due to the inability of the hemiplegic subjects to load the affected limb and so derive benefit from these exercises. This study would tend to support that contention in that, despite being asked to weight bear in a manner which would have caused loading of the affected limb, this was not achieved. These results do not support the findings of Bohannon and Tinti-Wald (1991) that hemiplegics were capable of voluntarily shifting weight over the affected leg. It would, therefore, appear that hemiplegic subjects should be provided with some form of augmented feedback so that they are made aware of the precise motor patterns required to load the hemiplegic leg. Perhaps this technique would enhance future gait re-education. There is a need to test this hypothesis in a controlled clinical study.

6. The finding that the hemiplegic gait may contain elements of a "cautious" pattern is worthy of further investigation. This could be done by comparing the spatiotemporal gait patterns of hemiplegic subjects with normal controls at walking speeds common to both groups perhaps controlled by a treadmill.

7. The inability of the hemiplegic subjects to control balance posterolaterally probably results from abnormal motor control. Patterns of hypertonicity may play a role in this deficiency. To investigate this contention, it would be most useful to examine the electromyographic activity of selected muscles while these weight shift postures were being maintained. This would provide useful information as to the cause of this weight shift discrepancy and provide opportunities to devise appropriate remedial strategies.

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APPENDIX 1

SAMPLE OF COMPUTER PRINTOUT OF SPATIOTEMPORAL GAIT MEASURES

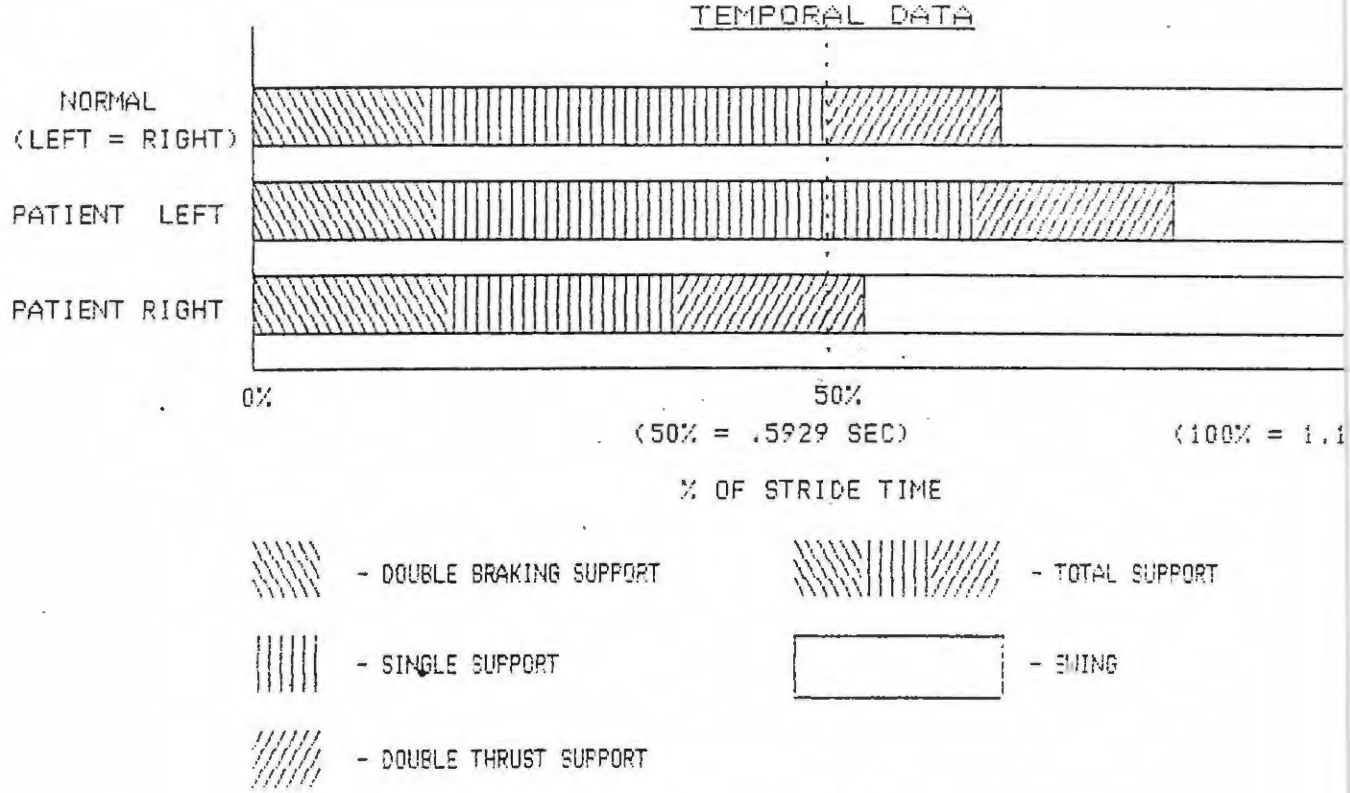
NAME :

DATA FILE NAME : [illegible]
REVISION DATE : [illegible]

AGE : 61
SEX : M
HEIGHT (CM) : 168
DATE OF INJURY : 00/00/82

DATE : 12/04/91
WALK NUMBER : 1
COMMENT : FREE SPEED

TEMPORAL DATA



DURATION OF PHASES (% OF STRIDE TIME)

	DOUBLE BRAKING SUPPORT	SINGLE SUPPORT	DOUBLE THRUST SUPPORT	TOTAL SUPPORT	SWING TIME
LEFT	15.8	46.9	16.5	79.6	20.1
RIGHT	16.5	20.1	15.8	53.2	46.9
SYMMETRY RATIO (L/R)	-0.041	+1.333	+0.041	+0.496	-1.333

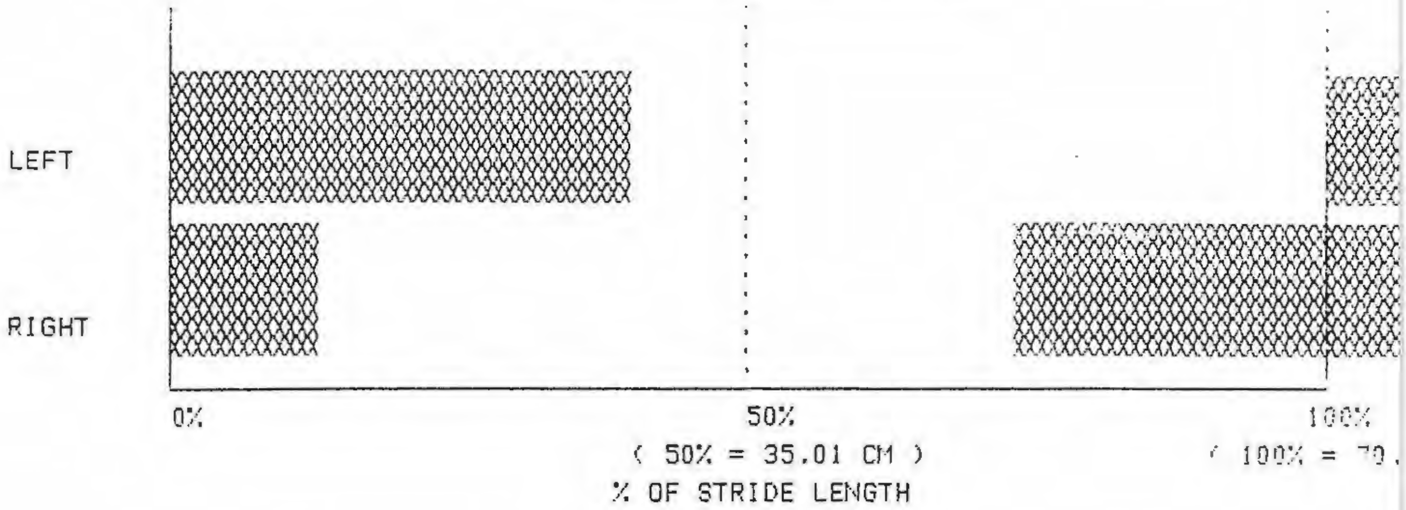
ACTUAL & PREDICTED STEP TIME

	ACTUAL AVERAGE STEP TIME (SEC)	PREDICTED STEP TIME (SEC)
LEFT	.437	.319
RIGHT	.748	.862

RELATIVE SPEED (STAT/SEC) : 0.351
AVERAGE STRIDE TIME (SEC) : 1.18
AVERAGE VELOCITY (M/SEC) : .590

NAME :

DISTANCE INFORMATION



RELATIVE SPEED : 0.351
RELATIVE STRIDE LENGTH : 0.416

STRIDE LENGTH (CM)

PATIENT	70.02
NORMAL	109.7

STEP LENGTH DATA

	LENGTH(CM)	% OF STRIDE LENGTH	SYMMETRY RATIO
LEFT	18.87	26.9	-1.699
RIGHT	50.94	72.7	
NORMAL	54.87	50	0

NUMBER OF FOOTLENGTHS PER STRIDE : 2.43

NAME :

DATA FILE NAME : [illegible]
REVISION DATE : [illegible]

*** STRIDE DATA (R 1 12/04/91 W 1 D1 FL31) ***

STRIDE #	LEFT LENGTH (CM)	LEFT STRIDE TIME (SEC)	LEFT SWING TIME (SEC)	**	RIGHT LENGTH (CM)	RIGHT STRIDE TIME (SEC)	RIGHT SWING TIME (SEC)
1	73.65	1.258	.256	**	71.81	1.291	.604
2	70.89	1.234	.217	**	69.97	1.24	.658
3	67.21	1.153	.22	**	69.05	1.138	.541
4	76.41	1.204	.274	**	69.97	1.182	.558
5	69.05	1.144	.241	**	76.41	1.177	.576
6	63.52	1.131	.228	**	62.60	1.072	.471
7	69.05	1.171	.261	**	74.57	1.228	.562
8	70.89	1.168	.207	**	68.13	1.141	.472
9	68.13	1.189	.244	**	69.05	1.225	.571
AVG	69.87	1.183	.2386	**	70.17	1.188	.557
S.D.	3.724	.04212	.02240	**	3.954	.06576	.05902

*** STEP DATA (R 1 12/04/91 W 1 D1 FL31) ***

STEP #	LEFT LENGTH (CM)	LEFT DOUBLE SUPPT. (SEC)	LEFT STEP TIME (SEC)	**	RIGHT LENGTH (CM)	RIGHT DOUBLE SUPPT. (SEC)	RIGHT STEP TIME (SEC)
1	18.41	.231	.456	**	53.40	.167	.835
2	20.25	.159	.423	**	49.71	.2	.817
3	21.17	.18	.417	**	47.87	.212	.721
4	19.33	.192	.432	**	50.63	.18	.75
5	[illegible]	[illegible]	[illegible]	**	50.63	.18	.723
6	[illegible]	[illegible]	[illegible]	**	[illegible]	.252	.651
7	19.33	.186	.48	**	55.24	.162	.748
8	13.81	.246	.423	**	54.32	.243	.718
9	16.57	.204	.45	**	52.47	.17	.775
10	15.65	.159	.414				
AVG	18.87	.1884	.437	**	50.94	.1962	.7486
S.D.	3.283	.03158	.02178	**	3.444	.03316	.05559

APPENDIX 2

SAMPLE OF COMPUTER PRINTOUT OF BALANCE MEASURES

BALANCE TEST RESULT

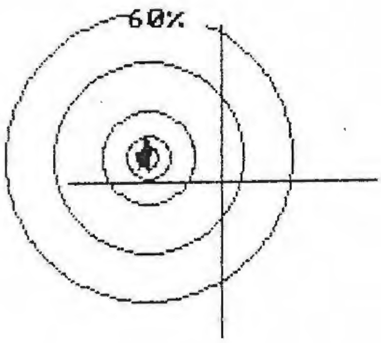
Department of Physiotherapy
Dalhousie University Halifax NS

Patient:
Birth Date: / /
Physician: Aldon
Group:
Comments:

Date: 03/28/91
Sex: Male
Diagnosis: R Hemiplegia
Sub Group:

EYES OPENED
Stable

60%	:	0.00
40%	:	0.00
20%	:	0.00
10%	:	8.90
5%	:	91.10
Disp	:	4.26
COB X	:	-40.17
COB Y	:	14.22



BALANCE TEST RESULT

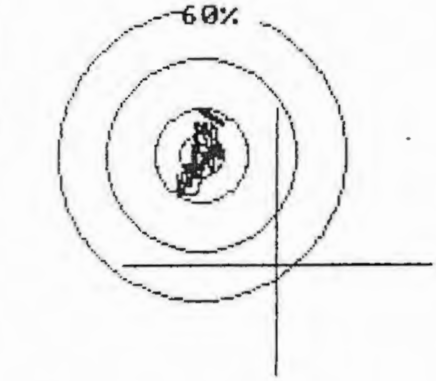
Department of Physiotherapy
Dalhousie University Halifax NS

Patient:
Birth Date: / /
Physician: Aldon
Group:
Comments:

Date: 03/28/91
Sex: Male
Diagnosis: R Hemiplegia
Sub Group:

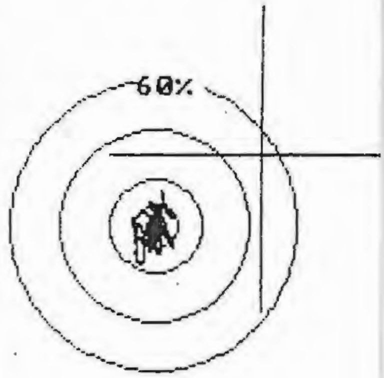
EYES OPENED
Stable

60% :	0.00
40% :	0.00
20% :	37.10
10% :	33.40
5% :	29.50
Disp :	12.75
COB X:	-40.36
COB Y:	57.40



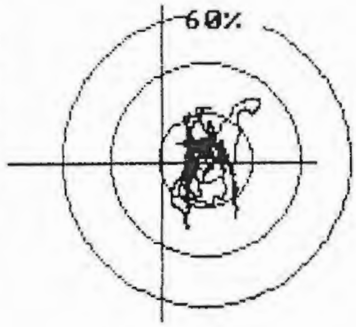
EYES OPENED
Stable

60% :	0.00
40% :	0.00
20% :	10.90
10% :	41.70
5% :	47.40
Disp :	8.66
COB X:	-57.47
COB Y:	-38.16



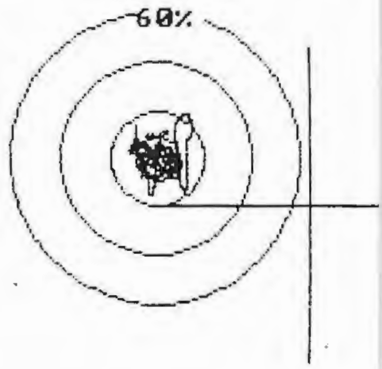
EYES OPENED
Stable

60% :	0.00
40% :	13.90
20% :	36.70
10% :	33.90
5% :	15.50
Disp :	17.90
COB X:	23.65
COB Y:	1.82



EYES OPENED
Stable

60% :	0.00
40% :	.80
20% :	24.70
10% :	50.70
5% :	23.80
Disp :	11.57
COB X:	-81.95
COB Y:	25.41



APPENDIX 3

ETHICAL REQUIREMENTS

DALHOUSIE UNIVERSITY
HALIFAX, N.S.
CANADA B3H 3J5

FACULTY OF HEALTH PROFESSIONS
ETHICS COMMITTEE

FACULTY OF HEALTH PROFESSIONS

ETHICS REVIEW FINAL APPROVAL FORM

Title of Research Proposal: A Study of Automatic Balance Reactions and Gait Following Stroke

Principal Investigator(s): George Turnbull

On behalf of the Faculty of Health Professions Ethics Committee for Human Research, I hereby acknowledge that the above proposed research is ethically acceptable with respect to the use of human subjects.

Signed on behalf of the Committee: (Elizabeth Sarakin-Doyle Ph.D.)

Date: June 1, 1996

Revised: July, 1987

CONSENT FORM

The rehabilitation of walking in patients who have had a stroke is of great importance. It is vital that research is undertaken so that such rehabilitation can be made more effective. This study, which is being undertaken by Professor George Turnbull of the School of Physiotherapy at Dalhousie University, is designed to study the relationship between balance and walking in people who have suffered a stroke. A clearer understanding of how balance affects walking will permit better rehabilitation techniques to be developed. In addition, part of the study will study the performance of healthy elderly people. This will provide information about the abilities of normal, healthy elderly people so that valid comparisons can be made with the stroke group.

To ensure that you are suitable to take part in this study, you will be assessed by a licensed physiotherapist. You will be asked to visit the School of Physiotherapy at Dalhousie University. There, you will be asked to stand on a platform which will measure your balance. Measurements will be taken from the platform while you stand in eleven different positions which will be explained to you at the time. Following this, you will be asked to walk a short distance (10 yards) on a walkway while measurements are taken. For the walking tests, a piece of metal tape will be attached to the soles of your own shoes. You will be asked to walk at different walking speeds. At the same time, you will be videotaped to assist with the analysis of the results.

No measurements will be collected that you would consider offensive in any way. You will not be identified by name but by number only. All information from the study will be kept completely confidential and will only be used by the researcher in reporting the results of the project. The data collection will take approximately one hour of your time. None of the procedures which you will undertake could be considered in any way dangerous or unpleasant except for the extremely small risk of you losing your balance while on the platform. To avoid any danger to you from this risk, an experienced physiotherapist will stand immediately beside you during data collection at all times.

Permission to participate as a subject in this study is entirely voluntary and you are free to refuse to take part if you so wish. You are free to withdraw from the study at any time even during the testing. If you do withdraw you will not be penalised in any way. Questions dealing with any part of this study are very welcome. You are urged to ask about any concerns you may have and you have been given the telephone numbers of the researcher (Home 462 7081 (evenings) and Business 424 2633 (during the day) should you wish to contact Professor Turnbull at any time.

SUBJECT CONSENT FORM

I am satisfied with the explanation I have been given about my part in the study. I have had a full opportunity to ask questions. I understand that my taking part in this study is entirely voluntary and that I may withdraw from the study at any time even after the study has started.

I also understand that individual results will remain completely confidential.

Date:__________________ Signed:X______________________________

Witnessed by:______________________________

I have given a full and complete explanation of all procedures involved in this study. No form of deception is involved and the anonymity of the participant will be fully respected.

Date:__________________ Signed:X______________________________

Witnessed by:______________________________

APPENDIX 4

SCREENING CHECK LIST USED FOR HEMIPLEGIC SUBJECT SAMPLE

HEMIPLEGIC SUBJECT SCREENING EVALUATION

Personal Details

Name:_________________________________ Age:_____ Sex:_____

Address:____

____

Phone:___________ Physician:________________(phone)______________

Time Since Stroke:______________________ Hemiplegic Side:____R______L_____

INCLUSION CRITERIA

Age . . . .
Time Since Stroke . . . .
Rehabilitation Undertaken . . . .
Length of Rehabilitation . . . .
Can Walk Without Cane . . . .
Presence of Asymmetrical Gait Pattern . . . .

EXCLUSION CRITERIA

Serious or Unstable Medical Conditions . . . .
Unilateral Neglect . . . .
Major Perceptual Dysfunction . . . .
Significant Peripheral Sensory Loss . . . .
Severe Homonymous Hemianopsia . . . .
Cognitive Disturbance including Memory Loss . . . .
Severe Intractable Pain . . . .
Receptive Aphasia . . . .
Incontinence of Bowel or Bladder . . . .
Presence of Gait Disorder Not Due to Stroke . . . .

DECISION: INCLUDE EXCLUDE

APPENDIX 5

SCREENING CHECK LIST USED FOR NORMAL SUBJECT SAMPLE

NORMAL SUBJECT SCREENING EVALUATION

Personal Details

Name:______________________________ Age:_____ Sex:_____

Address:____

____

Phone:__________ Physician:__________________(phone)__________

INCLUSION CRITERIA

Age . . . .

EXCLUSION CRITERIA

Neurological Disorder . . . .
Vestibular/Inner Ear Disorder . . . .
Severe Visual Disability . . . .
Significant Peripheral Sensory Loss . . . .
Severe Degenerative Osteoarthritis . . . .
Cognitive Disturbance including Memory Loss . . . .
Marked Skeletal Deformity . . . .
Postural Hypotension . . . .
Chronic Alcohol Abuse . . . .
Cervical Myelopathy . . . .
Normal Pressure Hydrocephalus . . . .
Multiple Sensory disorder . . . .
History of Gait Disorder . . . .
Observable Gait Disorder . . . .

DECISION: INCLUDE EXCLUDE

APPENDIX 6

CORRELATION COEFFICIENTS BETWEEN BALANCE PARAMETERS WITH GAIT.

APPENDIX 6 - 1

Significant correlation coefficients (r) between gait parameters and position of CP on the X axis for left hemiplegic subjects (critical r value: 0.576).

LHCPX	Q	A	P	R	L	D1B	D1S	D1F	D2B	D2S	D2F	TD1	TD2
RS Slowest					-0.718	-0.632					-0.678	-0.672	
RS Slow					-0.786	-0.722					-0.638	-0.664	
RS Free	-0.579				-0.781	-0.681							
RS Fast	-0.642				-0.767	-0.614							
RS Fastest					-0.779	-0.633					-0.602	-0.596	
St T Slowest				-0.620		0.649	0.630	0.625				0.742	
St T Slow					0.634	0.748				0.584	0.642	0.777	
St T Free					0.726	0.791				0.611	0.651	0.753	
St T Fast	0.614	0.596			0.758	0.718		0.588		0.585	0.652	0.680	
St T Fastest	0.578	0.577			0.783	0.663					0.645	0.634	
RSL Slowest													
RSL Slow					-0.674								
RSL Free					-0.655								
RSL Fast					-0.604								
RSL Fastest					-0.638								
SSAR Slowest					0.626								
SSAR Slow					0.603								
SSAR Free					0.645								
SSAR Fast													
SSAR Fastest					0.593								
BDSAR Slowest								0.676					
BDSAR Slow													
BDSAR Free													
BDSAR Fast													
BDSAR Fastest													
SLAR Slowest				-0.576		0.709	0.712	0.702				0.764	
SLAR Slow				-0.669		0.597	0.667	0.604				0.694	
SLAR Free				-0.720		0.619	0.683	0.577				0.703	
SLAR Fast				-0.658								0.577	
SLAR Fastest				-0.723								0.623	
TSUN Slowest					0.633								
TSUN Slow					0.680								
TSUN Free					0.699								
TSUN Fast													
TSUN Fastest					0.681								
TSAFF Slowest													
TSAFF Slow												0.623	
TSAFF Free				-0.582									
TSAFF Fast													
TSAFF Fastest										-0.584			
SSUN Slowest													
SSUN Slow												-0.633	
SSUN Free				0.599									
SSUN Fast													
SSUN Fastest													
SSAFF Slowest					-0.625								
SSAFF Slow					-0.677								
SSAFF Free					-0.689								
SSAFF Fast													
SSAFF Fastest					-0.657								
BDSUN Slowest													
BDSUN Slow													
BDSUN Free				-0.642									
BDSUN Fast									-0.647				
BDSUN Fastest									-0.663				
BDSAFF Slowest													
BDSAFF Slow					0.577							0.653	
BDSAFF Free					0.581								
BDSAFF Fast													
BDSAFF Fastest					0.581								

Significant correlation coefficients (r) between gait parameters and position of CP on the X axis for right hemiplegic subjects (critical r value: 0.7067).

RHCPX	Q	A	P	R	L	D1B	D1S	D1F	D2B	D2S	D2F	TD1	TD2
RS Slowest													
RS Slow													
RS Free													
RS Fast													
RS Fastest													
St T Slowest			0.716		0.719								
St T Slow			0.739										
St T Free													
St T Fast													
St T Fastest													
RSL Slowest				0.769									
RSL Slow													
RSL Free													
RSL Fast													
RSL Fastest													
SSAR Slowest	-0.809		-0.742									-0.880	
SSAR Slow	-0.712		-0.764									-0.740	
SSAR Free			-0.763									-0.731	
SSAR Fast												-0.729	
SSAR Fastest													
BDSAR Slowest													
BDSAR Slow													
BDSAR Free						0.815					0.758		
BDSAR Fast						0.744							
BDSAR Fastest						0.755							
SLAR Slowest							-0.731						
SLAR Slow													
SLAR Free			-0.824										
SLAR Fast			-0.719										
SLAR Fastest													
TSUN Slowest													
TSUN Slow													
TSUN Free													
TSUN Fast													
TSUN Fastest													
TSAFF Slowest	0.753		0.708					0.742				0.823	
TSAFF Slow	0.765						0.730	0.804				0.870	
TSAFF Free	0.744		0.730		0.709			0.817				0.861	
TSAFF Fast	0.744						0.738	0.817				0.755	
TSAFF Fastest								0.747				0.769	
SSUN Slowest	-0.731											-0.761	
SSUN Slow	-0.760											-0.852	
SSUN Free	-0.763		-0.749		-0.735			-0.819				-0.868	
SSUN Fast	-0.747							-0.810				-0.767	
SSUN Fastest								-0.738				-0.764	
SSAFF Slowest													
SSAFF Slow													
SSAFF Free													
SSAFF Fast													
SSAFF Fastest													
BDSUN Slowest													
BDSUN Slow							0.761						
BDSUN Free	0.773				0.732							0.715	
BDSUN Fast													
BDSUN Fastest													
BDSAFF Slowest													
BDSAFF Slow													
BDSAFF Free													
BDSAFF Fast													
BDSAFF Fastest													

APPENDIX 6 - 3

Significant correlation coefficients (r) between gait parameters and position of CP on the Y axis for left hemiplegic subjects (critical r value: 0.576).

LHCPY	Q	A	P	R	L	D1B	D1S	D1F	D2B	D2S	D2F	TD1	TD2
RS Slowest			-0.787										
RS Slow			-0.683					0.591					
RS Free			-0.623					0.609					
RS Fast			-0.603					0.593					
RS Fastest			-0.589										
St T Slowest													
St T Slow													
St T Free													
St T Fast													
St T Fastest													
RSL Slowest			-0.765					0.617					
RSL Slow			-0.843					0.683					
RSL Free			-0.779					0.655					
RSL Fast			-0.796					0.605					
RSL Fastest			-0.814										
SSAR Slowest													
SSAR Slow													
SSAR Free													
SSAR Fast													
SSAR Fastest													
BDSAR Slowest			0.639										
BDSAR Slow													
BDSAR Free					0.649	0.692							
BDSAR Fast													
BDSAR Fastest												0.664	
SLAR Slowest			0.588										
SLAR Slow						0.623							
SLAR Free						0.653							
SLAR Fast						0.679							
SLAR Fastest			0.576			0.714	0.609						
TSUN Slowest													
TSUN Slow			0.626										
TSUN Free			0.623										
TSUN Fast													
TSUN Fastest													
TSAFF Slowest			0.587						0.672				
TSAFF Slow			0.725						0.708				
TSAFF Free			0.578			0.617							
TSAFF Fast				0.671					0.656				
TSAFF Fastest						0.653	0.652		0.637				
SSUN Slowest			-0.603						-0.694				
SSUN Slow			-0.714						-0.699				
SSUN Free			-0.613			-0.632							
SSUN Fast				-0.718					-0.745				
SSUN Fastest						-0.625	-0.636		-0.657				
SSAFF Slowest													
SSAFF Slow			-0.613										
SSAFF Free			-0.620										
SSAFF Fast													
SSAFF Fastest			-0.579										
BDSUN Slowest			0.763			0.632							
BDSUN Slow			0.840										
BDSUN Free			0.742			0.703							
BDSUN Fast			0.744			0.620	0.709						
BDSUN Fastest			0.599				0.692						
BDSAFF Slowest													
BDSAFF Slow													
BDSAFF Free													
BDSAFF Fast													
BDSAFF Fastest													

APPENDIX 6 - 4

Significant correlation coefficients (r) between gait parameters and position of CP on the Y axis for right hemiplegic subjects (critical r value: 0.7067).

RHCPY	Q	A	P	R	L	D1B	D1S	D1F	D2B	D2S	D2F	TD1	TD2
RS Slowest													
RS Slow													
RS Free													
RS Fast													
RS Fastest													
St T Slowest												-0.730	
St T Slow									0.804				
St T Free												-0.873	
St T Fast													
St T Fastest													
RSL Slowest													
RSL Slow													
RSL Free													
RSL Fast													
RSL Fastest													
SSAR Slowest													
SSAR Slow										-0.754			
SSAR Free										-0.780			
SSAR Fast										-0.942			
SSAR Fastest										-0.718			
BDSAR Slowest							-0.744						
BDSAR Slow													
BDSAR Free													
BDSAR Fast													
BDSAR Fastest													
SLAR Slowest										-0.780			
SLAR Slow													
SLAR Free													-0.877
SLAR Fast			-0.726							-0.786	-0.752		
SLAR Fastest										-0.757			
TSUN Slowest													
TSUN Slow													
TSUN Free													
TSUN Fast										-0.723			
TSUN Fastest													
TSAFF Slowest													
TSAFF Slow													
TSAFF Free													
TSAFF Fast													
TSAFF Fastest													
SSUN Slowest													
SSUN Slow													
SSUN Free													
SSUN Fast												0.707	
SSUN Fastest													
SSAFF Slowest													
SSAFF Slow													
SSAFF Free													
SSAFF Fast													
SSAFF Fastest													
BDSUN Slowest													
BDSUN Slow													
BDSUN Free													
BDSUN Fast													
BDSUN Fastest													
BDSAFF Slowest													
BDSAFF Slow													
BDSAFF Free													
BDSAFF Fast													
BDSAFF Fastest													

Significant correlation coefficients (r) between gait parameters and Dispersion Index for all hemiplegic subjects (critical r value: 0.4438).

DISPERSION	Q	A	P	UN	AFF	BBAFF	SBAFF	FBAFF	BFAFF	SFAFF	FFAFF	TBAFF	TFAFF
RS Slowest												0.502	
RS Slow												0.711	0.645
RS Free												0.609	0.470
RS Fast												0.639	0.529
RS Fastest												0.516	0.488
St T Slowest													
St T Slow											0.444		
St T Free													
St T Fast												-0.499	
St T Fastest													
RSL Slowest												0.529	0.579
RSL Slow												0.698	0.680
RSL Free												0.524	0.497
RSL Fast												0.539	0.544
RSL Fastest												0.466	0.514
SSAR Slowest				0.556									-0.463
SSAR Slow				0.595									-0.498
SSAR Free				0.596									-0.490
SSAR Fast				0.487							0.456		-0.463
SSAR Fastest			0.445	0.641									
BDSAR Slowest													
BDSAR Slow							0.603						
BDSAR Free													
BDSAR Fast													
BDSAR Fastest													
SLAR Slowest												-0.558	-0.556
SLAR Slow							0.447						
SLAR Free					0.528								
SLAR Fast													
SLAR Fastest													
TSUN Slowest												-0.522	-0.586
TSUN Slow												-0.554	-0.628
TSUN Free												-0.460	-0.554
TSUN Fast													-0.521
TSUN Fastest												-0.471	-0.574
TSAFF Slowest		-0.462	-0.464										
TSAFF Slow													
TSAFF Free													
TSAFF Fast													
TSAFF Fastest		-0.474											
SSUN Slowest													
SSUN Slow													
SSUN Free													
SSUN Fast			0.454										
SSUN Fastest		0.486											
SSAFF Slowest												0.538	0.605
SSAFF Slow												0.520	0.592
SSAFF Free												0.459	0.544
SSAFF Fast													0.539
SSAFF Fastest												0.455	
BDSUN Slowest													
BDSUN Slow													
BDSUN Free													
BDSUN Fast													-0.466
BDSUN Fastest													-0.484
BDSAFF Slowest													
BDSAFF Slow													
BDSAFF Free													
BDSAFF Fast													
BDSAFF Fastest													

APPENDIX 6 - 6

Significant correlation coefficients (r) between gait parameters and time spent in 5% of body weight ring for all hemiplegic subjects (critical r value: 0.4438).

5% Body Weight	Q	A	P	UN	AFF	BBAFF	SBAFF	FBAFF	BFAFF	SFAFF	FFAFF	TBAFF	TFAFF
RS Slowest													
RS Slow													
RS Free													
RS Fast													
RS Fastest													
St T Slowest												0.565	
St T Slow												0.495	
St T Free												0.544	
St T Fast												0.513	
St T Fastest													
RSL Slowest													
RSL Slow													
RSL Free													
RSL Fast													
RSL Fastest													
SSAR Slowest		-0.444		-0.603									
SSAR Slow				-0.612									
SSAR Free				-0.647									
SSAR Fast				-0.511									
SSAR Fastest				-0.642									
BDSAR Slowest												0.559	0.466
BDSAR Slow													
BDSAR Free													0.559
BDSAR Fast													
BDSAR Fastest							-0.456						
SLAR Slowest												0.458	0.712
SLAR Slow													0.725
SLAR Free													0.794
SLAR Fast													0.813
SLAR Fastest													0.687
TSUN Slowest													
TSUN Slow													
TSUN Free													
TSUN Fast													
TSUN Fastest													
TSAFF Slowest		0.514		0.481									
TSAFF Slow				0.464									
TSAFF Free				0.523									
TSAFF Fast													
TSAFF Fastest		0.496											
SSUN Slowest		-0.494		-0.510									
SSUN Slow				-0.482									
SSUN Free				-0.521									
SSUN Fast		-0.451											
SSUN Fastest		-0.509											
SSAFF Slowest													
SSAFF Slow													
SSAFF Free													
SSAFF Fast													
SSAFF Fastest													
BDSUN Slowest													0.537
BDSUN Slow													
BDSUN Free													0.599
BDSUN Fast													
BDSUN Fastest													
BDSAFF Slowest													
BDSAFF Slow													
BDSAFF Free													
BDSAFF Fast													
BDSAFF Fastest													

APPENDIX 6 - 7

Significant correlation coefficients (r) between gait parameters and time spent in 10% of body weight ring for all hemiplegic subjects (critical r value: 0.4438).

10% Body Weigh	Q	A	P	UN	AFF	BBAFF	SBAFF	FBAFF	BFAFF	SFAFF	FFAFF	TBAFF	TFAFF
RS Slowest													
RS Slow													
RS Free													
RS Fast							-0.446						
RS Fastest													
St T Slowest						-0.525							
St T Slow						-0.456						0.479	
St T Free						-0.512						0.446	
St T Fast							0.557						
St T Fastest							0.533						
RSL Slowest													
RSL Slow													
RSL Free													
RSL Fast													
RSL Fastest													
SSAR Slowest		0.461											0.517
SSAR Slow													0.494
SSAR Free													0.485
SSAR Fast													0.446
SSAR Fastest													0.455
BDSAR Slowest													
BDSAR Slow													
BDSAR Free													
BDSAR Fast													
BDSAR Fastest													
SLAR Slowest												0.655	0.689
SLAR Slow												0.566	0.581
SLAR Free													0.489
SLAR Fast													0.566
SLAR Fastest													0.583
							0.447						
TSUN Slowest													
TSUN Slow													0.444
TSUN Free													
TSUN Fast													0.478
TSUN Fastest													0.445
TSAFF Slowest		-0.517		-0.625									
TSAFF Slow				-0.557									
TSAFF Free			-0.446	-0.585									
TSAFF Fast													
TSAFF Fastest		-0.461											
SSUN Slowest		0.508		0.660					-0.465				
SSUN Slow				0.567									
SSUN Free			0.457	0.578									
SSUN Fast		0.463	0.451	0.461									
SSUN Fastest		0.475		0.470									
SSAFF Slowest													
SSAFF Slow													
SSAFF Free													
SSAFF Fast													
SSAFF Fastest													
BDSUN Slowest													
BDSUN Slow													
BDSUN Free													
BDSUN Fast													
BDSUN Fastest													
BDSAFF Slowest				-0.592									
BDSAFF Slow				-0.471									
BDSAFF Free													
BDSAFF Fast													
BDSAFF Fastest													

APPENDIX 6 - 8

Significant correlation coefficients (r) between gait parameters and time spent in 20% of body weight ring for all hemiplegic subjects (critical r value: 0.4438).

20% Body Weight	Q	A	P	UN	AFF	BBAFF	SBAFF	FBAFF	BFAFF	SFAFF	FFAFF	TBAFF	TFAFF
RS Slowest													
RS Slow												-0.577	-0.516
RS Free												-0.500	
RS Fast												-0.534	
RS Fastest												-0.611	
St T Slowest													
St T Slow													
St T Free													
St T Fast													
St T Fastest													
RSL Slowest												-0.586	-0.524
RSL Slow												-0.650	-0.522
RSL Free												-0.574	
RSL Fast												-0.571	
RSL Fastest												-0.620	
SSAR Slowest				0.489									
SSAR Slow				0.587									
SSAR Free				0.567									
SSAR Fast				0.498									
SSAR Fastest				0.600									
BDSAR Slowest							0.463						
BDSAR Slow													
BDSAR Free													
BDSAR Fast		-0.551											
BDSAR Fastest		-0.516											
SLAR Slowest													
SLAR Slow													
SLAR Free													
SLAR Fast													
SLAR Fastest													
TSUN Slowest												0.539	
TSUN Slow												0.576	
TSUN Free												0.583	
TSUN Fast												0.544	
TSUN Fastest												0.661	
TSAFF Slowest													
TSAFF Slow													
TSAFF Free													
TSAFF Fast													
TSAFF Fastest													
SSUN Slowest													
SSUN Slow													
SSUN Free													
SSUN Fast													
SSUN Fastest												-0.444	
SSAFF Slowest												-0.546	
SSAFF Slow												-0.582	
SSAFF Free													
SSAFF Fast												-0.560	
SSAFF Fastest												-0.644	
BDSUN Slowest													
BDSUN Slow													
BDSUN Free													
BDSUN Fast												0.532	
BDSUN Fastest												0.604	
BDSAFF Slowest							-0.524						
BDSAFF Slow													
BDSAFF Free												0.549	
BDSAFF Fast												0.519	
BDSAFF Fastest												0.612	

APPENDIX 6 - 9

Significant correlation coefficients (r) between gait parameters and time spent in 40% of body weight ring for all hemiplegic subjects (critical r value: 0.4438).

40% Body Weigh	Q	A	P	UN	AFF	BBAFF	SBAFF	FBAFF	BFAFF	SFAFF	FFAFF	TBAFF	TFAFF
RS Slowest												0.472	
RS Slow												0.517	
RS Free													
RS Fast												0.475	
RS Fastest												0.488	
St T Slowest													
St T Slow												-0.497	
St T Free												-0.530	
St T Fast												-0.551	
St T Fastest												-0.564	
RSL Slowest													
RSL Slow													
RSL Free													
RSL Fast													
RSL Fastest													
SSAR Slowest													
SSAR Slow													
SSAR Free													
SSAR Fast													
SSAR Fastest													
BDSAR Slowest													
BDSAR Slow													
BDSAR Free													
BDSAR Fast													
BDSAR Fastest													
SLAR Slowest												-0.541	-0.479
SLAR Slow													
SLAR Free													
SLAR Fast													
SLAR Fastest													
TSUN Slowest												-0.482	
TSUN Slow												-0.478	
TSUN Free													
TSUN Fast													
TSUN Fastest													
TSAFF Slowest													
TSAFF Slow													
TSAFF Free													
TSAFF Fast													
TSAFF Fastest													
SSUN Slowest													
SSUN Slow													
SSUN Free													
SSUN Fast													
SSUN Fastest													
SSAFF Slowest												0.490	
SSAFF Slow												0.452	
SSAFF Free													
SSAFF Fast													
SSAFF Fastest													
BDSUN Slowest													
BDSUN Slow													
BDSUN Free													
BDSUN Fast													
BDSUN Fastest													
BDSAFF Slowest													
BDSAFF Slow													
BDSAFF Free													
BDSAFF Fast													
BDSAFF Fastest													

Significant correlation coefficients (r) between gait parameters and range of weight shift for all hemiplegic subjects (critical r value: 0.4438).

Range of Shift	X	Y	BFAFF	FFAFF
RS Slowest	0.510			
RS Slow	0.743		0.613	
RS Free	0.716		0.501	
RS Fast	0.721		0.574	
RS Fastest	0.686		0.538	
St T Slowest				
St T Slow	-0.544			
St T Free	-0.578			
St T Fast	-0.626			
St T Fastest	-0.613			
RSL Slowest			0.612	
RSL Slow	0.629		0.690	
RSL Free	0.619		0.556	
RSL Fast	0.621		0.641	
RSL Fastest	0.644		0.621	
SSAR Slowest			-0.552	
SSAR Slow			-0.510	
SSAR Free			-0.533	
SSAR Fast			-0.539	
SSAR Fastest			-0.558	
BDSAR Slowest				
BDSAR Slow				
BDSAR Free				
BDSAR Fast				
BDSAR Fastest				
SLAR Slowest	-0.485		-0.610	
SLAR Slow			-0.470	
SLAR Free				
SLAR Fast				
SLAR Fastest				
TSUN Slowest	-0.589		-0.586	
TSUN Slow	-0.644		-0.564	
TSUN Free	-0.629		-0.533	
TSUN Fast	-0.562		-0.555	
TSUN Fastest	-0.655		-0.565	
TSAFF Slowest				
TSAFF Slow				
TSAFF Free				
TSAFF Fast				
TSAFF Fastest				
SSUN Slowest				
SSUN Slow		0.484		
SSUN Free	0.465			
SSUN Fast				
SSUN Fastest				
SSAFF Slowest	0.584		0.586	
SSAFF Slow	0.639		0.512	
SSAFF Free	0.622		0.523	
SSAFF Fast	0.545		0.516	
SSAFF Fastest	0.633		0.548	
BDSUN Slowest	-0.470			
BDSUN Slow	-0.534			
BDSUN Free	-0.657			
BDSUN Fast	-0.502			
BDSUN Fastest	-0.509			
BDSAFF Slowest				
BDSAFF Slow	-0.526			
BDSAFF Free	-0.471			
BDSAFF Fast	-0.450			
BDSAFF Fastest	-0.508			