LIMITS OF BALANCE RECOVERY IN HEMIPARETIC STROKE INDIVIDUALS DURING CORRECTIVE INPLACE AND STEPPING RESPONSES

by
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A Thesis/Practicum submitted to the Faculty of Graduate Studies of The University of Manitoba in partial fulfillment of the requirements of the degree

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ABSTRACT

Introduction: Stroke is a leading cause of motor impairment leading to disability and restricted mobility. Individuals suffered as stroke have difficulty walking, low gait speeds and are unable to meet different demands like walking on all support surfaces, uneven terrain, ramps and stairs. Researchers have suggested a role of motor cortex in regulation of posture and balance, however none of the physiological studies done as yet have been able to prove a cortical influence in regulation of corrective balance reactions. Limited information is available regarding corrective (feed-back) balance responses in strokes during dynamic tasks such as walking, stepping or gait initiation. A number of studies have shown correlation between impaired standing balance and ioss of walking function in the stroke subjects. The purpose of this study was to evaluate the limits of corrective balance reactions in chronic stroke individuals and healthy controls to unexpected disturbances of varying magnitudes during a leg-lifting task. This task which places the subject in single limb support and reflects the single limb support phase during the task of walking. The single leg-lifting task allows one to quantify corrective inplace as well as stepping responses.

Methods: Ten chronic hemiparetic stroke subjects with moderate residual motor deficits and nine healthy controls were subjected to sudden forward (FT) and backward (BT) support surface translations over three magnitudes (acceleration/velocities). Corrective balance reactions were elicited during lifting of the paretic and non-paretic legs. Linear and angular kinematics were obtained bilaterally using the peak 2D motion analysis system. Responses from each subject were classified into an inplace, touchdown or stepping strategy based on linear displacement of lateral malleolus marker. Performance levels based on strategy selection and centre of mass displacements (CM-S and CM-F) were graded as successful-good, successful-fair, unsuccessful-stumble and unsuccessful-falls. Multi-link normative movement patterns were identified in the controls and stroke subjects. Deviations from normative movement patterns in magnitude, direction and/or timing of the angular displacements were classified into abnormal patterns. Associations of normative movement patterns with successful performance and of abnormal movement patterns with unsuccessful performance levels were determined. EMG recordings from bilateral Hamstrings (HA), Quadriceps (QU), hip adductors (AD) and abductors (AB) were analyzed for onset latencies to identify any significant delays. Chedoke McMasters impairment and disability scores and gait speed were recorded for each subject. For purpose of data analysis and statistical analysis subjects were divided into three groups controls (CON), non-paretic side stance (NST) and paretic side stance (PST). A chi-square analysis was done to compare difference in strategy selection and performance levels between groups. A Spearman-Rho rank order correlation analysis was done to correlate performance levels of stroke subjects on this balance test with gait speed and Walking Index scores from the Chedoke Disability Inventory.

Results: The control group mainly exhibited a successful inplace strategy with the exception of single steps taken at highest level of disturbance during FT. In contrast a stepping corrective strategy was observed in over one half of the trials for stroke subjects both during paretic side and non-paretic side stance which was equally distributed over the three different rates of platform translations. There was a significantly greater success rate in the controls compared to the stroke subjects (100% in controls for BT and FT, 60% in stroke subjects during BT and 50% during FT). There was no significant difference in strategy selection or performance levels between NST and PST. The associations of movement patterns with performance levels were in the range of 70-90% except for the associations of normative pattern to successful performance success for PST and abnormal pattern to unsuccessful performance for NST during BT. The timing of the corrective muscle response in the stroke subjects was delayed compared to the controls. The correlation of performance levels with gait speed and walking index was moderate during FT.

Conclusion: Limitations in corrective balance reactions seen in the stroke subjects signified cortical influence in regulating feed-back balance responses during a single-limb support task. In a number of trials the stroke subjects achieved success by adopting a strategy switch from inplace to stepping right from the lowest level of disturbance. The results show that both the non-paretic and paretic side contributed and are required during corrective balance responses. One should therefore not comment on balance or postural deficits by assessing and training only the paretic side of the body. Rehabilitation training to improve walking function in stroke population should thus involve functional bilateral tasks, and concentrate to improve both stance control and swing limb abilities.

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1. INTRODUCTION

Balance is a complex function of the nervous system. It involves two main aspects: one maintaining stability at rest or during movement (predictable control) and second restoring body stability in response to sudden unexpected disturbances or loss of balance.

A number of researchers have studied balance reactions to unexpected platform translations and rotations during standing and the task of walking. On basis of the onset latencies of these reactions to perturbations, they suggest that supraspinal structures are necessary in the control of balance (Nashner et al, 1980; Eng and Winter, 1994; Tang et al, 1996). These balance reactions are impaired or abnormal in terms of timing of the muscle activity and strategy selection in subjects having cortical lesions such as stroke. A need therefore exists to further explore the underlying patho-physiological mechanisms of balance impairment during functional tasks in cortically lesioned patient population. Residual balance impairment is a common consequence after a stroke (Hamrin et al, 1982; Keenan et al., 1984; Soderback et al., 1991; Wade et al., 1992; Berg et al., 1992; Sackley, 1990). Four aspects of balance control in stroke subjects have been examined by researchers: 1) weight bearing at rest 2) sway patterns at rest 3) voluntary swaying and 4) corrective responses to unexpected disturbances. The results indicate that the stroke subjects experience excessive sway, decreased weight bearing on the affected limb and decreased weight shifting capability as compared to non-stroke subjects during simple balance tasks. Corrective responses examined during standing in response to sudden support surface translations and rotations report differences in EMG responses of the paretic leg as compared to the non-paretic leg. The question that now arises is how do these differences impact on the ability to perform function? Research studies have indicated that balance impairment correlates with level of ambulation and with motor recovery of the lower extremity (Hamrin et al, 1982; Dettman et al, 1987; Sackley, 1990). A number of studies have also reported on gait abnormalities post stroke (Carlsoo et al, 1974; Wall and Turnbull, 1986; Bohannon, 1987; Lehamann et al, 1987; Olney et al, 1994,1998). Substantial deficiencies and difficulties in the control of body stability during all phases of gait were observed after stroke. These deficits have a direct impact on what one can do with the non-paretic limb for progression and the ability to effectively transfer weight from non-paretic limb to paretic limb and vice versa. Weight bearing asymmetries in stroke individuals have been linked with abnormal walking patterns.

Feedback balance responses in stroke patients during dynamic tasks such as walking, stepping or gait initiation have not been studied. The only ones who examined such tasks (Roger et al 1993, Pai et al 1994, Brunt et al 1995) have looked at preparatory (feed-forward) balance control.

Evaluation of balance control during tasks that have similar dynamics and stability requirements to that of walking for example, is necessary if specific causes of balance impairment are to be identified and effective management and treatment regimes are to be developed. Before examining corrective balance responses of stroke subjects during the task of walking, the present study first evaluated the task of single leg lifting with single leg support as a prelude to gait initiation. This is a task that is in progression from standing to walking. This task also allows us to compare responses from the paretic and non-paretic limb.

This thesis evaluates corrective balance responses to unexpected disturbances during the task of single –leg lifting in individuals who have suffered a stroke.

2. REVIEW OF LITERATURE

The human body is a multi-link system. The stability of each link or body segment is achieved by maintaining a balance of external (gravity dependent and motion dependent) forces and internal (muscular, bony and ligamentous) forces acting about each joint. Balance is a functional term that is most commonly defined as the ability to maintain and control position and motion of total body center of mass (COM) relative to the base of support (BOS) (Winter, Patla, and Frank, 1990; Mackinnon and Winter, 1993; Maki and McIlroy, 1996; Pai and Patton, 1997).

Two mechanisms of balance control have been recognized:

1. **Feed forward mechanism** – in which the CNS is involved in the planning of anticipatory or preparatory postural responses preceding or accompanying voluntary tasks/movements. In preparatory balance control, the nervous system has adequate time to predict the upcoming balance requirements and makes necessary adjustments to prevent any disturbance. This process is believed to involve cortical and sub-cortical structures of the CNS for planning and execution of the response. These preparatory adjustments demonstrate the predictive role of the nervous system in controlling stability (Massion, 1994; Aruin and Latash, 1995).
2. **Feed back mechanism** – in which the CNS and PNS deals with postural responses to sudden unexpected disturbances. These are usually referred to as automatic postural reactions and exclude volitional responses. Automatic balance control involves making rapid adjustments of the body in response to the unexpected disturbance in order to maintain COM within the BOS. This thesis restricted to studying automatic balance responses. The exact neural structures and mechanisms required for automatic balance reactions have not as yet been identified.

There are two main areas of research, to automatic balance reactions one examining spinal reflexes and their contribution to the generation of automatic corrective responses (Dietz et al., 1984; Frossberg, 1979, Zehr and Stein, 1997). Others study the role of supraspinal structures to the generation of adequate corrective responses (Horak and Macpherson, 1996; Eng and Winter, 1994; Nashner, 1979,1980).

A number of studies in animals and humans have used the moving platform paradigm to suddenly disturb standing balance by rotating or translating the support surface resulting in a sudden change in the position of the COM relative to the BOS. This is one way in which researchers have been able to elicit and study automatic corrective responses. The EMG recordings from all the different studies showed that the earliest muscle responses occurred within a range of 60-100 ms following the onset of the balance disturbance. The onset latencies of these "balance" responses were longer that the stretch reflex latencies (40 - 50 ms), but shorter than voluntary reaction times (150 ms). The exact neural structures and mechanisms underlying required for automatic balance reactions have not as yet been identified.

Another feature of these corrective responses was that different parameters of platform motion (acceleration/velocity) resulted in different movement synergies; small magnitude (displacement) and slow movement disturbances produced a single link movement synergy called the "ankle strategy" (Horak and Nashner, 1986; Diener, Horak and Nashner, 1988), large magnitude or fast disturbances produced multi-segmental movement synergies involving movement about the trunk, hip, knees and ankle (Szturm and Fallang, 1998) and was called by some researchers as the "hip strategy" (Nardone et al, 1990). Still larger and faster perturbations resulted in stepping responses called the "stepping strategy" (McIlroy and Maki, 1996).

The first strategy to be observed was the ankle strategy, where muscle activity commences about in the ankle muscles and the body behaves as a single rigid link swaying anteriorly or posteriorly about the ankle joint. Backward translations (BT) resulted in activation of dorsal muscles of the body (gastrocnemius, hamstrings and paraspinals) and forward translations (FT) resulted in the anterior muscles being activated. The 'hip strategy' resulted in a proximal-to-distal muscle activation sequence or pattern. BT resulted in activation of abdominals followed by quadriceps and FT resulted in activation of paraspinals followed by hamstrings. The hip strategy and the ankle strategy are examples of fixed-support or "inplace" strategies, which serve to control the displacement of the center of mass, without altering the base of support. The stepping strategy comes into play to maintain balance when the disturbances are of sufficiently large magnitude or fast and serves to relocate the base of support under the projected position of the COM. The step is initiated within a range of 165-365 ms after onset of platform motion and lasts for a moderate duration. The stepping response is an automatic one. The step occurs rapidly and is not associated with a preparatory phase usually observed in voluntary leg lifting or gait initiation tasks.

The task of regulating balance during functional tasks such as stepping and walking requires much more complex CNS control. During steady state walking, the center of mass is outside the base of support for about 75-80% of the stride time. One of the reasons why it is more difficult to maintain stability during single support phase is due to the movement of the COM forward and along the medial border of the foot, compared to the double support phase where the COM is between the two stance limbs.

A number of animal and human studies have examined balance responses during walking. In the initial work done by Frossberg natural light touch and electrical stimuli were applied to cats during the swing and stance phases of walking. EMG activity was recorded from muscles of the leg and thigh and limb position was detected by light emitting infrared markers placed on the joints. Consistent short latency (30-50 ms) responses were observed, which were phase dependent with reflex reversals found in the EMG response during stance and swing with the same stimuli. There was no difference in the responses between electrical and the natural stimuli. Forssberg (1979) coined the term stumbling corrective response. Much of this work has been done to examine the role of individual sensory inputs (cutaneous, proprioception) on regulating muscle activity or kinematic patterns during the gait cycle. Different paradigms have been used by these researchers such as tactile or electrical stimulation, actual physical disturbances like obstacles or holding stimuli applied to the limb, and support surface translations involving acceleration or deceleration of a moving treadmill belt or sudden movement of a stationary platform. Forssberg (1979), Wand et al (1980), Drew and Rossignol (1987), Yang and Stein (1990), Zehr et al (1997,98) and others studied short latency EMG responses to sensory/cutaneous stimulation during walking. These have been referred to as "Stumbling Responses" (Forssberg, 1979. For review see Zehr et al, 1999).

Studies similar to Frossberg's have been done on humans by Duysens et al. (1990,1992), Zehr et al (1997) and Van Wezel (1997). Zehr et al (1997), have studied EMG and kinematic responses by stimulating different nerves during treadmill walking.

Superficial peroneal (SP) nerve stimulation during early swing gave rise to knee flexion and ankle plantar flexion that electrical with inhibition of the tibialis anterior (TA) muscle and facilitation biceps femoris muscle. Tibial nerve stimulation also resulted in increased knee flexion and plantar-flexion when stimulated at the stance to swing transition and produced knee extension and dorsiflexion when stimulated at late stance. Kinematic changes in the joint trajectories were a consequence of the muscle activity produced by the stimulation and not due to the initial nerve stimulation.

The question arises if there was any true balance disturbance posed by the stimuli in all the above studies. For most of these studies described above, it is questionable if any of the stimuli used by the researchers restricted the trajectory of the movement or lead to an actual disturbance. The responses seen may be more appropriately called tactile /obstacle avoidance rather than stumbling reactions. Tactile or cutaneous electrical stimulation may be considered a very low level of disturbance, which may have very minimal effect on the individual body segment or the balance of the body as a whole. Such a disturbance could be corrected by automatic mechanisms operating at the spinal level both in animals and humans.

Wand, Prochazka and Sontag (1980) compared the early responses to mechanical and electrical stimulation in cats during treadmill locomotion. Their results were different than the results of Frossberg. They found a difference in response between electrical and mechanical stimulation. The response after electrical stimulation was of shorter onset (9 ms compared to 17 ms for the knee flexor) and brief duration as compared to the mechanical stimuli. It was concluded that these responses were evoked in response to cutaneous afferents rather than muscle afferents as anesthesia of the dorsum abolished these responses. Buford and Smith (1993) also found substantial differences in early responses between electrical pulses and obstructive mechanical taps. They showed substantial difference in response to electrical stimuli as compared to taps that actually tripped the animals. Taps and electrical stimuli were applied to the dorsal or ventral aspect of the hind paw of four cats during swing and stance phases of forward and backward walking. They showed direction dependent responses to obstructive stimuli (taps) during forwards and backwards walking. Obstruction of forward swing resulted in a response that drew the limb backwards away from the obstacle, and obstruction of backward swing resulted in a response the drew the limb forward away from the obstacle. However muscle responses to non-obstructive (electrical cutaneous stimulation) stimuli were the same regardless whether the cat walked forwards or backwards. Thus the above two studies show that subjects react with substantial different corrective responses when there is an actual balance disturbance compared to electrical/tactile stimulation.

Using a different paradigm, Hibert et al. (1994) provided evidence to show that supraspinal structures were required for recovery from an unexpected balance disturbance. Intact and chronic spinal cats were made to walk on a treadmill. The chronic spinal cats had recovered their ability to step with their hind limbs. Their disturbance consisted of a trap door built into the treadmill belt and triggered to open just before hind limb contact. The trap door consisted of a force sensor to detect the forepaw loading. Decrease in the force signal at the end of the foreleg stance led to opening of the trap door. EMG and kinematic recording of joint angles were obtained from the hind limb of intact and spinal cats. The results indicated that corrective flexor responses were present in both spinal and intact cats however the latency of the flexors burst to remove the foot from the hole was much longer in spinal cats (130-150 ms) compared to intact cats (70-150 ms). Secondly, in the intact cat, the flexion movement was stronger enabling the cat to remove the paw clearly from the hole and regain support on the treadmill. In the spinal cats, the flexion movement was not sufficient to enable the cat to remove the paw from the hole. Thus the authors strongly suggest that spinal system is facilitated by supraspinal influences for corrective balance responses in intact cats. However, which supraspinal structures are involved and the mechanisms through which they influence the spinal system still remains to be established.

HUMAN STUDIES:

Few studies on healthy human subjects have looked at actual physical disturbances such as, restriction of the swing limb trajectory or support surface translations during walking. These types of disturbances cause a change in relationship between the center of mass and the base of support and challenge the balance requirements of an individual.

Dietz et al. (1986) investigated bilateral ankle muscle responses following an obstruction of the forward swing leg while walking on treadmill. The disturbance consisted of a holding impulse applied through a cord to the swing leg, and was generated by a torque motor. The impulse was applied at the beginning and the end of the swing phase. The EMG responses recorded from the ankle muscles differed in sign (excitation of inhibition) and latency depending on the period of the gait cycle when the holding stimulus was applied that is early swing or late swing. Thus these results indicate that different patterns of muscle response were used depending on the phase of the gait cycle when the disturbance was applied. Ghori et al. (1989) in a similar study applied momentary resistance to one leg at three selected points in the gait cycle. They recorded EMG from both the lower limbs. They also found phase dependent responses evoked bilaterally with onset latencies between 76 to 88 ms.

The study by Eng and Winter (1994) examined responses to a sudden perturbation that simulated a tripping response. The purpose of their study was to examine recovery of human subjects from perturbation obstructing the limb during early swing and late swing. The subjects were asked to walk on a walkway in which two force plates were embedded. EMG recording was done from sixteen muscles, which included eight from each stance and swing limbs. The force platform provided the temporal gait parameters. Temporal and kinematic parameters analyzed included step length and duration, maximum-swing hip flexion, maximum swing knee flexion, maximum swing ankle dorsiflexion and maximum trunk angle. The onset latencies and patterns of the EMG response were dependent on the period of the gait cycle when the obstacle occurred. An elevating strategy (60-130 ms) was seen in response early swing perturbation. This consisted of increased flexion of the swing limb over the obstacle, with increased flexion angles of the ankle, knee and hip following the perturbation. Where as lowering strategy (70-100 ms) consisting of a rapid lowering of the limb to the ground and shortening of the step-length was seen following a late swing perturbation. Thus the elevating strategy helps to propel the body center of mass upwards and away from the obstacle. Similarly, the lowering strategy seemed the most appropriate way of preventing a fall to the late swing perturbation. In such a case lowering the foot in front of the obstacle resulted in a temporary double support position from which further recovery responses could take place. Thus it seems that the nervous system could efficiently deal with the type of disturbance by generating appropriate synergies to stabilize the body at that point. The results of a similar study done by Schillings et al. (1996), were in accordance with those of Eng and Winter (1994). Both of these studies demonstrate that tripping perturbations imposed at different periods of the gait cycle result in different stability requirements and different corrective adjustments; with the subjects being able to compensate rapidly, prevent falling and maintain progression.

Nashner (1980) used the moving platform paradigm to examine the responses to unexpected platform movements during walking. He studied forward and backward platform translations. The perturbations occurred randomly during heel strike, beginning of single support phase, mid-stance and beginning of double support phase. He studied EMGs from bilateral ankle muscles and recorded the movements and forces associated with them during each perturbation. Early corrective muscle responses were observed in two ankle muscles. The onset latency of muscles was in the range of 95 to 110 ms and lasting for about 100-400 ms. The Author concludes that the changes in the EMG patterns were rapid, large in magnitude and movement specific (different for FT vs. BT). In response to backward translation initial dorsiflexion was seen (passive component), which changed to compensatory plantar flexion after onset of EMG activity in the posterior ankle muscles. The authors propose the conceptual model of "locomotor-balance control", in which the same movement generator can be used simultaneously to organize both, the stepping movements and the balance adjustments.

Dietz et al. (1984) found similar rapid reactions with early onset latencies. They studied bilateral EMG and kinematic responses from the ankle during acceleration and deceleration of the treadmill belt during constant speed walking. The pattern of the bilateral EMG responses was dependent on the direction of the perturbation and on the phase of the gait cycle in which the perturbation occurred. The onset latencies of responses from both the sides were between 65-70 ms and lasted for about 150 ms. Treadmill acceleration evoked an ipsilateral gastrocnemius and contralateral tibialis anterior response, where as treadmill deceleration evoked a bilateral tibialis anterior response. According to the authors, the reflex response generated served to support the body, compensate for imbalance and prevent falling.

Tang et al. (1998), in a similar study to Nashner (1980), examined corrective balance responses from the trunk and proximal and distal limb muscles. The responses had onset latencies between 90-140 ms, magnitude ranging between four to nine times that during normal walking and lasting relatively long (70-200 ms). Thus like Nashner's their results also exhibited early onset and large magnitude muscle responses.

Riediger (1999) went beyond examining phase-dependent EMG responses and focused mainly on analyses of kinematics and ground reaction forces. The purpose of this study was to examine which parameters of the stepping cycle (i.e. pattern, phase and magnitude) are modifiable during disturbances changing the relationship between the center of body mass (COM) and the base of support (BOS). Backward and forward translations were presented just after right heel-off and after right mid-swing. Eight muscles were recorded bilaterally from the lower limbs. Task performance during the perturbed trials compared to the non-perturbed trials was evaluated based on the analysis of displacement and velocity of COM relative to space (CM-S). Magnitude, duration and timing of divergence in trajectories of a) knee, hip and ankle joint angular displacements, b) trunk segment rotation, and c) COM displacement relative to the foot (CM-F) were determined for forward translation, backward translation and no perturbation trials. The results showed that corrective muscle responses were observed in all the muscles with onset latencies between 80-180 ms, with some muscles exhibiting excitatory responses and some exhibiting inhibitory responses. The magnitude and velocity of the COM displacement did not change significantly between FT, BT and no platform translations (NPT). There was a phase shift in the angular displacement trajectories, the COM relative to foot, and the trunk segment rotation, which was evident during perturbations at heel-off and mid swing. There was no significant difference in the magnitude or pattern of the response for the BT and FT compared to the NPT. There was a phase lead in the angular trajectories of the knee, hip and ankle for the BT and a phase lag in the trajectories for FT compared to NPT. The pattern of the corrective responses did not differ depending on where in the gait cycle the disturbance occurred, that is heel strike or mid swing. Balance adjustments in the locomotor patterns were exhibited by a phase-shifting response. The author proposed that the centers for balance and locomotion were integrated similar to Nashner's conceptual model.

The above studies (Nashner, 1980; Schillings et al,1996; Tang et al, 1998; Eng and Winter 1994; Dietz et al, 1984; Ghori et al, 1989; Hibert et al, 1994, Wand, Prochazka and Sontag, 1980; Buford and Smith, 1993) have characterized automatic balance responses and suggested the possible neural mechanisms involved generation of these. The spinal cord reflexes alone can be accounted to correct for very minor disturbances, disturbances that truly do not destabilize the individuals balance either by changing relative segment positions or the relationship between the center of mass and the base of support. Most of the researchers now strongly believe that supra-spinal structures (brain stem and cortex) are required for the control of balance during functional tasks such as walking or stepping. Hibert et al. (1994) showed that spinal cats were not capable of generating an effective corrective response and continue the interrupted locomotion. Eng and Winter (1994) showed that the nervous system actively controls the corrective response by selecting the appropriate synergy depending on the stability requirements of the body at a particular instance in the gait cycle. It is highly unlikely that these type of corrective responses solely be produced by the poly-synaptic spinal reflexes, and therefore must require some control from supra-spinal structures. Ghori et al. (1989), and Nashner (1979,80), also suggest that supra spinal structures may be required for the control of balance. However, none of the studies examining corrective responses during locomotion have reached a definite conclusion as to which higher sub-cortical and cortical centers are responsible balance control.

2.1 NEUROPHYSIOLOGICAL MECHANISMS OF BALANCE CONTROL

A number of anatomical and electrophysiological studies have been conducted in animals to identify the supra spinal structures responsible for balance and locomotor control. One of the main structures identified for posture and balance control is the ponto-medullary reticular formation (PMRF) and its descending pathways. Few researchers have suggested the role of cortex in balance control.

Anatomical, electrophysiological and lesion studies suggest that the PMRF within the brain stem is involved in control of posture, balance and locomotion (Matsuyama et al, 1988,1997; Drew and Rossinol, 1990; Mori S, 1987,1992; Lawrence and Kuypers, 1968; Peterson 1975,1979). Electrophysiological studies have shown that the PMRF consisting of nucleus reticulo ponto-caudalis (NRPc), nucleus reticulo giganto-cellularis (NRGc), nucleus reticulo ponto-oralis (NRPo), the dorsal part of the tegmental field (DTF) and the ventral part of the tegmental field (VTF) in the caudal pons is involved in regulating balance control. Stimulation of the VTF in the pons along the midline in decerebrate cats results in an increase in the level of hind limb muscle activity that persisted for some time after cessation of stimulation (Mori et al., 19780,1982,1987). Stimulation of the VTF resulted in increased weight bearing by the animals as indicated by increased force recorded on the force transducers underneath the limbs and decreased body weight support offered by the rubber hammock. On stimulation of the VTF, the cat showed changes from no weight bearing to partial weight bearing and eventually to full weight bearing where the cat maintained a quiet standing posture without the aid of the of a rubber hammock. Stimulation of the DTF resulted on opposite changes, such that the cat changed its posture from full weight bearing to partial weight bearing to almost no weight bearing in sequence, with continuation of DTF stimulation.

The NRPo is situated in the pons, the DTF is situated caudally to it and the NRGc is situated most caudally in the medulla oblongata. Stimulation of the NRPo activates fibers passing through the DTF and VTF which in-turn activates the NRGc. The motor neurons innervating the axial, hind limb and forelimb muscles are under the control of descending reticulospinal system pathways either through direct connections or via interneurons (Kiezer and Kuypers, 1984; Matsuyama et al, 1988, 1993). A signal from the DTF would lead to activation of descending fibers from the NRGc terminating on inhibitory interneurons in the gray matter, leading to suppression of postural muscle activity or muscle tone. Similarly a signal from the VTF, would activate descending fibers terminating on excitatory inter-neurons leading to an increase in the postural muscle activity (Oka T et al., 1993; Takakusaki K et al., 1989). Biochemical stimulation of these same structures, by carbachol or serotonin within the medial pontine and medullary reticular formation in both decerebrate and intact cats showed similar effects on postural muscle activity as the electrical stimulation studies, thereby strengthening the evidence of the role of PMRF in regulating balance control (Takausaki et al 1993a, 1993b, Katayama Y et al, 1984).

Stimulation of the mesencephalic locomotor region (MLR) in decerebrate cats is known to produce rhythmical alternating locomotion (Mori et al 1978,82). The locomotor activity induced by the MLR is influenced by stimulation of the DTF and the VTF. The stimulation of the DTF resulted in a suppression of the ongoing locomotor activity induced by MLR stimulation, where as stimulation of the VTF converted hind limb stepping into a four-legged locomotion. Responses to electrical stimulation of the MLR, VTF and the DTF in intact freely moving cats were similar to those in decerebrate cats (Mori S, 1987). Eidelberg et al. (1981) studied hind limb stepping and locomotor activity in macaque monkeys after lesions affecting the pyramidal tracts but sparing ventromedial system containing the reticulospinal (RST) and vestibulo spinal tracts (VST). They found that these monkeys could continue locomotion and use hind limbs for support even after the lesion. They suggest role of descending brain stem pathways (RST and VST) in regulation posture and locomotion. These studies have enabled us to identify the areas of brain stem essential for weight bearing and balance control and have shown that these areas interact with the area responsible for locomotion.

Lesion studies interrupting the PMRF at the level of the brain stem also provide evidence for the role of these structures in regulation of balance. Kuypers et al. (1968) studied the effect of lesions of the pyramidal and brain stem pathways in monkeys. After bilateral pyramidal tract lesions the animals could immediately right from either side and sit up in their cage with body upright and within 24 hours they could stand and climb a few steps. They could perform reaching movements with their arms but could not use their hands independently to pick up pellets from food wells, for clinging to the cage or climbing. The recovery period ranged from five weeks to five months, however at the end of full recovery also individual finger movements did not return. The authors then lesioned the brainstem pathways in the same monkeys after they had recovered. They grouped the descending brain stem pathways into two systems, the ventromedial system the reticulospinal and vestibulospinal tracts and the lateral system comprising of the rubrospinal tract. After interruption of the ventromedial system the animal showed unsteadiness in sitting, difficulty to maintain walking with a slight movement or sound resulting in a fall, and difficulty in avoiding obstacles. The animal also showed an abnormality of posture, adapting a flexion posture of the body when sitting or held supine, delayed righting to sitting position, and severe deficit in axial and proximal limb movements. However the distal extremity movements were considerably less impaired.

There was very little improvement in these deficits over time. The authors also studied the effects of lesions on the brain-stem pathways without prior pyramidotomy. The results showed similar but less severe changes in balance control and mobility. The animals righted and sat immediately after operation, but still maintained a flexion posture. In summary, the animals were left with relatively high control of balance after cortico-spinal lesions alone, however substantial deficits appeared after lesion of the ventromedian system and were maximum with simultaneous lesions of the cortico-spinal and ventromedian system.

From the above studies (Mori et al 1978,82; Eidelberg et al., 1981; Kuypers et al., 1968), we can conclude that the ventromedian brain stem pathways are important in the control of balance during standing, stepping and locomotor tasks.

Role Of Motor Cortex And Descending Cortical Pathways In Controlling Postural Activity

A number of animal studies tracing anatomical pathways by injection of florescent tracers indicate presence of abundant connections between the motor cortex and the PMRF (Kuypers, 1958; Keizer and Kuypers, 1984; Matsuyama and Drew, 1997). Neural traces injected in the medial reticular formation of the lower brain stem labeled corticobulbar neurons situated mainly in the pericrutiate area and the SMA (areas 4 and 6) and a few in area 3a (Keizer and Kuypers, 1984). Distribution of corticospinal and corticobulbar neurons overlap and the corticobulbar fibers from the sensori motor cortex are distributed bilaterally.

Electrophysiological studies stimulating the cortex and recording from the brain stem and vice versa (Magni and Willis, 1964; Peterson, 1974; Tower, 1944; Woolsey et al., 1972) have confirmed the presence of projections from the sensori-motor cortex to the PMRF. These studies indicate the presence of a fast di-synaptic descending pathway from the sensori-motor cortex to the spinal motor neurons comprising of the fast cortico-bulbar fibers, and of collaterals from fast conducting corticospinal fibers.

A few physiological studies have been done to examine effects of cortical influence in control of balance. To provide direct evidence for or against the hypothesis that the cortex plays a role in the control of balance and movement, Kably and Drew

(1998) studied the projection patterns of motor cortical neurons during a preparatory posture control task. Cats were trained to step over obstacles attached to a moving treadmill belt. The cats could see the obstacle 2-3 seconds before stepping over it. Action potentials were recorded from cortical neurons during periods of locomotion. Stimulating electrodes were placed in the PMRF and the PT and recording was done from layer five neurons of the pericrutiate cortex which were grouped in three categories, the pyramidal tract neurons (PTNs), the corticoreticular tract neurons (CTNs) and thirdly the both the PT and the CTNs depending on their discharge patterns. EMG recording was done from a number of forelimb and hind limb muscles. Most of the projection from the motor cortex to the PMRF was almost from collaterals of PT axons whereas most of the SMA projection to PMRF was direct. A large percent of CRNs projected bilaterally compared to the other two groups. All the three groups of neurons discharged before onset of swing during the obstacle avoidance. The CRNs discharged slightly earlier before the onset of stance in the gait cycle compared to the PTNs and the PTN/CRNs. The authors have not been able to link or time lock the onset of cortical neuronal activity with the preparatory EMG activity associated with the obstacle avoidance task. Despite this, the earlier discharge of the PTNs, PTN/CRNs and more so of the CRNs recorded during the stance phase of the limb indicates anticipatory activity of these respective structures.

2.2 BALANCE AND STROKE

Researchers have attributed restrictions in walking function to a number of factors, one of them being deficits in balance control. A number of researchers have examined balance impairment post stroke during different tasks like standing still, maintaining single stance, weight shifting from one limb to another, voluntary swaying and support surface perturbations while standing.

Shumway-cook et al. (1988) analyzed the pattern of weight shifting in stroke subjects during standing tasks. Subjects were made to stand on a force platform and the anterior posterior and medio lateral center of foot pressure or the sway was calculated. The results indicated that the stroke subjects showed a greater area of sway as compared to normals during quiet standing and had restricted sway during target reaching compared to the normals, with the path of displacement of the sway being away form the paretic leg. This shows that the stroke subjects have much less stability during standing still as well as during reaching tasks. Dettman et al. (1987) examined body sway during standing still and voluntary weight shifting task. They recorded the displacement of the center of foot pressure (CFP) beneath each limb. They also found similar results wherein during standing still the CFP excursion was greater than normal controls but had much less ability to voluntarily weight shift forwards, backwards or laterally. DiFabio, Badke and Duncan (1990) measured voluntary sway to targets placed at 50% of subjects' maximum sway. They calculated the path of sway as well as frequency of sway. Their results indicated that the stroke subjects could sway to the targets with restricted movement in the lateral direction and excess movement in the forward direction compared to healthy subjects.

A number of studies have examined corrective balance responses to sudden support surface translations in stroke subjects (Badke et al., 1983,1987; Di Fabio et al., 1986). Badke et al. (1983) perturbed stroke subjects in the anterior-posterior (A-P) direction at a very low velocity (5 cm/s). This elicited the ankle synergy i.e., gastrocnemius (GA)-hamstrings (HA) to BT and tibialis anterior - quadriceps (Q) in response to FT. Subjects were divided into two motor impairment groups, mild and moderate, based on the Fugyl-Meyer assessment lower extremity scores. Those subjects with low maximum scores (moderate impairment) showed a more variable response and longer onset latencies than those with higher scores (mild impairment). The mean onset latencies of controls to forward translations in were in the range of 95-115 ms. The mild impairment group had onset latencies within a range of 120-220 ms and moderate impairment group had onset latencies in the range of 200-475 ms. For backward translations the onset latencies for the control, mildly impaired and moderately impaired groups were 95-118 ms, 140-215 ms and 200-350 ms. The level of significance for the onset latencies between controls and the strokes has not been given. Also, responses with onset latencies in the range of 380-475 ms are classified as voluntary responses and should not be classified under automatic corrective reactions. Although the strokes showed activation in appropriate muscles for a small percentage of trials (GA-HA for BT and TA-Q for FT), they activated multiple muscles for most of the trials. More than two muscles were activated for 60% of the trials during BT and 87% of trials during FT. The percentage of body weight on the right lower limb 100 ms after the platform movement was calculated using limb-load monitor measurements. The healthy controls and the hemiplegic subjects from group 2 had 48-54 percent weight bearing on the right lower extremity whereas the group 1 hemiplegic subjects exhibited only 38-42 percent of body weight on the right lower limb. According to the authors, the variable response seen in the hemiplegics in term of onset latencies and pattern may be due to alteration in the weight bearing capability of the stroke subjects.

In a further experiment Badke et al. (1987) presented data on onset latencies from four groups of muscles for FT and BT. However mean onset latency of all the four muscle groups was calculated and compared for the control and hemiplegic group using the student's t test. Thus we do not know the onset latencies of the individual muscle groups. The results show no differences in mean and SD between the controls right and left leg (147-161 ms) and the non-paretic leg (144-146 ms) of the hemiplegic subjects to the unexpected support surface translations. The hemiplegic subjects exhibited longer (169-197 ms) and more variable muscle onset latencies in the paretic leg, compared to the controls. Again here we do not know if the differences in onset latencies were significantly different or not (level of significance not given).

In contrast to the studies by Badke et al. (1983) and Badke et al. (1987), Difabio and Duncan (1986), in spite of using an identical experimental set-up to theirs' found early onset latencies in a few muscles in the paretic limb compared to the controls. The findings indicated that the paretic limb the GA and TA had significantly longer onset latencies than controls, however the proximal muscles (Q and HA) had shorter onset latencies than the controls (significance not given). The authors report that a proximal to distal sequence of activation was seen on the paretic of the stroke subjects compared to a distal to proximal one seen on the non-paretic side and in the controls.

Dietz et al. (1984) examined balance responses in 12 spastic hemiparetic (7 strokes) subjects and 12 healthy controls. They used a different form of disturbance. Here, the subjects stood on separate see saw platforms on top of two parallel force platforms. EMG signals were recorded form the TA and the GA from each leg and the onset latencies and the rectified and averaged amplitudes were compared between both the groups, and between both the legs. The authors state that during the experiment it was made sure that the body weight of the subjects was equally distributed between the two legs. The balance response was elicited by tilting one of the seesaws downwards either by tibial nerve stimulation of mechanically. In the healthy controls, there was no difference in the response between the two sides in terms of onset latencies and amplitude of the muscles. In the hemiplegic group, irrespective of which leg was displaced, the onset latency of the tibialis anterior for the non-paretic side was 57 ms ±13. This was similar to the onset latencies of the controls (55 ms ± 12). The paretic leg of the stroke subjects had a mean onset of 80 ms when it was displaced and 78 ms when the non-paretic leg was displaced. Thus there was a significant delay of about 22 ms ($p<0.01$) in the onset of TA activity on the paretic side of the subjects. Onset latency of the gastrocnemius was not given. The authors attribute the delayed response (20-30 ms) from the TA of the paretic leg due to impaired supraspinal control.

Most of the above studies have quantified automatic balance reactions by analyzing EMG responses, mainly from distal muscles of the limbs. Important kinematic information about angular displacements of the leg and trunk segments and displacement of the total body center of mass, which enable us to detect the presence or absence of corrective balance synergies have not been presented. It must be noted that in all of the above studies none of the stroke subjects actually lost their balance.

Brunham (1996) examined limitations in balance recovery and the types of movement synergies elicited during sudden FT and BT in 18 hemiplegic stroke subjects and 9 age-matched controls. The magnitude of balance disturbance was varied using three levels of platform acceleration/velocity. Centre of mass displacement and angular displacement about the ankle, knee and hip/trunk (HT) were calculated. EMG recordings from bilateral tibialis anterior, gastrocnemius, Hamstrings and quadriceps were rectified and analyzed for onset latencies. Each individual's performance for every translation was graded as good, fair, poor and fall depending on the onset and pattern of the COM displacement. Overall they found that performance levels of stroke subjects were substantially reduced as compared to healthy age-matched controls. The essential components of successful, normal motor synergies were present in the majority of stroke subjects (16 of 18) and evident on the paretic side at the lower levels of platform motion (accelerations/velocities). These were multi-segmental movement patterns involving corrections about the ankle, knee and hip. Significant limitations in balance recovery and correspondingly, emergence of abnormal movement synergies became evident as the task demands (level of acceleration/velocity) increased and for different types of tasks (FT vs BT). The most abnormal pattern during both FT and BT was failure to produce adequate hip/trunk motion. FT was the most difficult task for all groups, and was associated with high number of falls especially at high platform velocities. Performance grades were significantly better for subjects with right cardio-vascular accident (RCVA) than those with left cardio-vascular accident (LCVA) for both FT and BT. There was no difference in the onset latencies between the control group and the non-paretic side of the strokes. During FT, the onset latencies of the TA, GA and HA were significantly delayed on the paretic side compared to the non-paretic side. For BT only one muscle showed significantly different onset latencies on the paretic side, the HA being activated significantly later in the RCVA group and the QU being activated significantly earlier in the LCVA group.

Zehr et al. (1998) studied short latency responses to superficial peroneal nerve stimulation from 8 stroke subjects and eleven healthy subjects while walking on a treadmill. As discussed in the previous section this type of electrical stimulation does not pose a real threat to balance and the response obtained can be more appropriately classified as the tactile avoidance task. The EMG and kinematic responses from the thigh and leg were studied from the affected side of the strokes both during stance and the swing phases. There was no significant difference in the reflex muscle responses in the swing limb or during swing between the healthy and the stroke group. However the kinematic changes were much smaller in the strokes than the controls and a significant correlation between the EMG and angular displacements occurred only at one point in the gait cycle. During the stance phase the stroke subjects showed reflex suppression in all the extensors examined, i.e. soleus, medial gastrocnemius and vastus lateralis. The authors conclude that although part of the corrective response is present in the stroke subjects, they exhibit more suppressive responses than the healthy controls.

From all the above studies we can conclude that there is a greater variability in terms of onset latencies and pattern of muscle activation in the stroke subjects compared to healthy individuals. There is a controversy about the stroke subjects exhibiting longer onset latencies. Badke et al. (1983) show longer onset latencies in all muscles whereas Difabio et al. (1986) show longer onset latencies only in distal muscles, with the proximal muscles of the paretic side having shorter onset latencies compared to normals.

Brunham (1996) showed that five of the eight muscles tested showed no difference in onset latencies during FT both in the right and left hemiplegics. Three muscles had significantly delayed onset latencies during FT. They found the QU to have early onset latency in right hemiplegics compared to controls during BT.

Most of the studies have quantified corrective responses on the bases of onset latencies and pattern of muscle activation. None-the-less, studies have shown that automatic balance reactions are present in stroke subjects on both non-paretic and paretic limbs. Brunham (1996) observed that balance performance was dependant on the magnitude of the disturbance. Successful and normative corrective reactions were observed at the lower magnitudes of balance disturbance. Limits in balance recovery became evident at higher rates of disturbances. This study did not quantify the relative contributions of the paretic and non-paretic limb to recovery of balance. It is thus important to know how and to what extent both limbs participate in recovery of whole body balance.

Analysis of posture and balance function during quiet standing may not be best suited for more dynamic activities involving changes in the base of support and acceleration of the center of mass. Investigators have examined a few components of balance control during single leg flexion and gait initiation tasks while standing on a stationary surface (Pai et al., 1993; Rogers et al., 1994; Brunt et al., 1995). The leg-lifting task resembles many components of gait initiation and walking. Like walking, in the leg-lifting task we need to voluntarily lift one of the lower limbs from the support surface and transfer the body's center of mass over to the stance limb. This also includes a period of single limb weight support. In hemiplegic subjects it allows us to compare the weight bearing abilities of the paretic and the non-paretic sides. Similar to that of gait initiation a preparatory phase is evident during single leg flexion before the onset of unloading. This includes an increase in the resultant horizontal component of the ground reaction force in the frontal plane, which leads to the linear displacement of the COM towards the single-stance limb (Roger et al, 1990).

Further studies by Roger et al., (1993) examined resultant ground reaction forces acting on the body in the frontal plane during single leg-flexion movements in post acute hemiparetic subjects. The subjects stood on separate force platforms and performed single leg flexion movements with the paretic (PL) and non-paretic (NL) limbs. The vertical (FZ) and the horizontal medio-lateral (FY) forces were recorded. The initial onset latency of the FY component was recorded from the relative onset of limb unloading recorded by FZ. All the hemiparetic subjects like healthy individuals commenced lateral weight transfer in advance of the onset of unloading of either the paretic or non-paretic flexing limb however half of the subjects showed delayed onset of changes in the FY under the PL compared to the NL. For the NL movements, 86% of the resultant FY propulsive force was contributed from under the upcoming non-involved swing (flexing) limb. Where as in the PL limb movements, 70% of the resultant FY propulsive force came from beneath the non-involved stance limb. Thus there was an overall reduction in the proportion of contribution from the paretic side to the resultant FY. This horizontal force component is required for the initial propulsion of the center of mass laterally towards the stance-limb.

Pai and Rogers (1994) examined the ground reaction forces and position of the COM in stroke patients and healthy controls during the same task. Motion analysis was used to determine the displacements of the COM. Subjects were scored on the Fugyl-Meyer assessment as a clinical test. The performance was spilt into completed transfer and uncompleted transfer. The completed transfer was divided into a) successful performance in which the subjects could transfer the COM to the opposite limb and could maintain position for five seconds and b) failure to hold in which the COM was successfully transferred to the stance leg but was not held for five seconds. An incomplete transfer consisted of an undershoot, where the COM did not shift over to the stance limb. The results indicated a successful performance for 48% of trials on the non paretic side and 20% of trials on the paretic side; a failure to maintain single stance for 26% of all the trials on the non- paretic side and 63% of the trials to the paretic side. There was insufficient displacement of the COM to the paretic side (non-paretic side in swing) for 26% of trials and for 63% of trials on the non-paretic side (paretic side in swing). Thus unsuccessful trials when the non-paretic side was flexing were equally due to insufficient transfer of the COM and a failure to hold the final position of the COM. Unsuccessful attempts while flexing the paretic side were mainly due to failure to maintain the COM within the paretic limb BOS. Successful performance was highly to moderately associated with the Fugyl-Meyer motor function and balance scores.

The above studies have concentrated on examining preparatory postural adjustments in stroke subjects, during a self-paced leg-lifting task. To further extend the work done by Rogers and Pai (1993, 94), Brunt et al., (1995) investigated ground reaction forces during the task of gait initiation in stroke subjects. The purpose of their study was, to explore relationship between limb loading asymmetry in stroke subjects and their ability to produce appropriate lateral and forward propulsive forces for gait initiation. Thirteen stroke subjects participated in this study. They were divided into two groups, namely, the SLL (symmetrical limb loading) group and the ALL (asymmetrical limb loading) group depending on the weight bearing during quiet standing. The SLL group had a mean weight bearing of 44% of total BW on the involved limb during stance compared to the ALL group who bore 25% of total BW on the involved limb. The patients performed three trials of gait initiation with the involved limb on the force plate and three trials with the non-involved limb on the force plate and always initiating walking with the involved leg. The peak ground reaction forces FZ (vertical ground reaction force), FY (medial/lateral ground reaction force) and FX (fore/after ground reaction force) were measured. Subjects from both the groups loaded involved swing limb before toe-off, however the amount of loading was correlated with the initial loading of the swing limb before gait initiation. In the SLL group the swing limb was loaded more than 50% of body weight (BW) before toe-off, where as in the ALL group although the limb loading was present it was less than 50% of the BW. With increase in the swing limb loading there was an unloading of the stance limb observed in both the groups, however the unloading was significantly less in the ALL group compared to the SLL group. Prior to foot off a backward and lateral-directed ground reaction force (negative FX a negative FY) was observed in the SLL group. This is what is typically observed in healthy subjects. A negative FX and a negative FY were also observed in the ALL group but were significantly reduced compared to the SLL group. The tibialis anterior and gastrocnemius activity was recorded both during swing and stance of the gait initiation.

In the SLL group there was a bilateral TA activity before movement onset, which contributed to the forward progression of the center of mass along with some co-contraction of the G. In the ALL group four of the patients showed only G activity in the absence of TA activity. The EMG pattern for the noninvolved stance limb was similar in both the groups.

2.3. SUMMARY

Walking is one of the most important functional task that is affected post-stroke. Restrictions in walking can be due to inadequate voluntary control, balance control, speed and endurance. Most of the stroke subjects strive hard to achieve independent walking. Research has shown that 60% of the stroke population regains independent walking function (Skillbeck et al., 1983; Keenan, Perry and Jordan, 1984; Wade et al., 1987; Friedman, 1991; Jorgensen et al., 1995). However, it must be noted that these people are mainly indoor walkers, have a very low gait speed, and rely heavily upon their assistant devices. These individuals have considerable difficulty in walking outdoors and walking on all terrain independently.

A moderate correlation has been found between impaired standing balance after stroke and loss of walking function and other standing activities of daily living (Hamrin et al., 1982; Dettman et al., 1987; Sackley, 1990). One sees significant deficiencies and difficulties in the control of body stability during all phases of gait in particular single leg support. These deficiencies have a direct impact on what one can do with the non-paretic limb for progression and the ability to effectively transfer weight from non-paretic limb to paretic limb and visa versa. Weight bearing asymmetries have also been linked with abnormal walking patterns in hemi-paretic individuals (Carlsoo et al., 1974; Bogardh and Richards, 1981; Wall and Turnbull, 1986). Asymmetrical weight bearing has been associated with exaggerated body sway in the frontal plane (Dickstein et al., 1984, Shumway-Cook et al., 1988). This frontal plane instability has been considered a major cause of falls towards the affected side in stroke subjects (Diller and Winberg, 1981).

It is important to know how a person with cortical damage due to stroke would react to sudden disturbances while performing tasks of daily living. As identifying the deficient components of balance control and the underlying neural structures responsible for these impairments/deficits could lead to development of better treatment strategies and provide information regarding which specific components of balance to focus the rehab treatment. Brunham (1996) has shown that abnormal components in movement synergies appeared only at considerable high levels of disturbance, and maximum falls occurred with the highest velocity of platform motion. To achieve independent outdoor walking, training stroke subjects with tasks demanding high balance requirements might be more beneficial than training them for simple tasks such as standing, weight shifting and reaching.

A few investigators have examined a few components of feed forward balance control during a self-paced single leg-flexion task while standing on a stationary surface (Pai et al. 1993; Rogers et al., 1994; Brunt et al., 1995). However no studies have been done to examinc feedback control during dynamic tasks such as walking, gait initiation or leg lifting as yet.

Secondly it is important to know the proportion of contribution in recovery of whole body balance from the paretic side and non-paretic side. It is well known in literature that stroke subjects have asymmetrical weight bearing and have difficulty in bearing weight through the paretic side (Bohannen and Larkin, 1985; Dettmann et al., 1987). This weight bearing on the paretic limb is very important to achieve single support phase during gait (Brunnstorm, 1965; Carr and Shepherd, 1983; Duncan and Badke, 1987; Lane RE, 1978). Most of the subjects even after rehabilitation therapy do not re-learn to equally bear weight through the paretic limb, as they learn to walk with help of an assistant device, and transmit most of their body weight through it. According to Weinstein et al. (1985), improvement in gait and balance of stroke subjects undergoing intensive rehabilitation seems to come from increasing reliance on the non-affected leg rather than functional use of the affected leg.

Before examining walking, we chose to examine a preliminary task such as single leg lifting. Disturbing the subjects' balance unexpectedly during a leg-lifting task would give one an exact indication of the limits of the subjects balance control system during a task that is in progression from standing to walking. It is very important to look at balance recovery during single leg support as it is very similar to the gait initiation task (Brunt et al., 1995) and, requires dynamic balance adjustments to maintain stability while the COM is being accelerated over the stance limb.

This study was thus directed to examine the feedback control of balance in individuals suffered from a stroke during the functional task of single leg lifting.

2.4. STUDY PURPOSE, OBJECTIVES AND HYPOTHESIS

The purpose was to evaluate and compare corrective balance responses (and performance levels) to anterior-posterior balance disturbances during a single leg-lifting task in healthy individuals and individuals suffered a stroke.

2.4.1 Study Relevance

The findings of this research would:

1. Provide a better understanding of corrective balance responses required during unexpected balance disturbances post stroke during functionally related tasks.
2. Provide information regarding the relative contribution from the paretic and non-paretic sides during automatic balance reactions.
3. Provide a base for future studies on platform translations in the stroke population during tasks of gait initiation and walking.
4. Provide an understanding of the underlying patho-physiological mechanisms to balance impairment and how they relate to locomotor dysfunction.
5. Act as a laboratory test for examining dynamic balance dysfunction in the stroke population.

2.4.2 Objectives

The main objectives of this study were:

1. To identify the limits of stability of stroke subjects during a single leg support situation, which was achieved by performing the task of single leg lifting.
2. To identify the means by which the corrective responses were achieved.
3. To identify if there was a strategy switch or scaling of the response present to increasing degrees of difficulties, in both the strokes and the controls.
4. To correlate balance performance on this laboratory test to other clinical tests : 1) Walking Index (WI) section from the disability inventory of the Chedoke-McMaster Stroke Assessment Scale and 2) Gait speed obtained from a 25 m and 3 m walking test.

2.4.3 Hypothesis Related To The Objectives

1. The overall performance levels of corrective balance responses to sudden forward and backward platform perturbations between the stroke group and healthy controls were determined and compared. The subjects were asked to perform the task with both the right and left limbs. This allowed for examination of responses from the paretic limb of the strokes both when it was in stance (sole weight bearing condition) and when it was in swing (ability to step). The corrective responses of each subject were classified into one of the following strategies:

    a) An inplace strategy described by Horak (1986) and Szturm and Fallang (1998) where the subjects restored balance without any change in the BOS.

    b) A (neutral step) touchdown - in which the subjects could not maintain an inplace and immediately brought the swing leg to the ground.

    c) A single stepping strategy as described by Maki (19996) and Luchi (1996), where the subject took one rapid forwards or backwards step (depending on direction of platform motion) within 100 ms with the swing leg.

    Based on operational definitions, the performance was quantified into four levels, good, fair, stumble and falls.

    i) Good performance - was one in which the subjects exhibited an inplace strategy.

    ii) Fair performance- was one when a single step was taken in the correct direction, for example backward step for forward translations or a neutral step was taken (touchdown).

    iii) Stumble (multiple steps)- was one in which subjects took more than one step on the swing or stance side to control their balance and prevent a fall.

iv) Fall - was one in which the subjects were not able to control their balance and would have fallen if not caught by the investigator standing besides them.

The hypothesis is that the performance levels would be significantly reduced in the stroke subjects compared to the controls, both while paretic side in stance (PST) and non-paretic side in stance (NST).

2. To identify the means by which the corrective responses were achieved EMG responses from selected muscles and selected kinematic parameters were recorded and analyzed. The following dependent variables, which represent key temporal and spatial features of the corrective balance responses, were quantified for all individuals:

   a) The onset latency, and frequency of occurrence of the initial corrective activity in bilateral hip abductors (gluteus medius), hip adductors (adductor magnus), hamstrings (HA) and rectus femoris (RF) to see if the balance reactions were delayed form an EMG perspective. The hypothesis is that in general the onset latency or the frequency of response would not be significantly different between the controls and the stroke groups.

   b) Displacement of the Center of Mass (CM-S) relative to space and centre of mass relative to the foot (CM-F) in the sagittal plane to identify how the center of mass to base of support relationship was altered during the corrective responses.

   c) The timing, direction and magnitude of trajectories of angular displacement of the knee and hip and trajectories of trunk segment rotation in the sagittal and frontal plane. Here the normative movement synergies present in healthy subjects for restoration of balance were identified, and then evaluated for which component(s) of the normative synergy was/were absent or excess in the stroke population.

Further the association between normative/abnormal movement synergies and performance levels successful- (good + fair), unsuccessful (poor + fall) was quantified. **It is hypothesized that a high association would be found between normal movement synergies and good performance and abnormal movement synergies and poor or unsuccessful performance.**

3. To identify if there was a strategy switch or scaling of the response present to increasing degrees of difficulties, in both the strokes and the controls. Increasing the platform translation acceleration/ velocity puts increasing demands on the balance system and thus increases complexity of the task. (Brunham, 1996). Three increasing rates of platform motion (V1, V2, V3) were chosen to alter the magnitude of the disturbance.

   It is hypothesized that a switch in strategy from an inplace strategy to a stepping strategy would be observed in the control group between V1 and V3, which would probably occur at the highest velocity (V3). Further it was hypothesized that in the stroke group an inplace strategy while standing on the non-paretic side would be seen at least at the lowest velocity. An inplace strategy will not be seen while standing on the paretic side, instead a stepping strategy will be seen.

   In cases where the same movement synergy was observed across all the platform acceleration/velocities, presence of scaling of the corrective response from V1 to V3 was examined. For an inplace response a similar approach to Brunham (1996) was used i.e. to examine effect of platform acceleration/velocity on selected kinematic variables from corrective angular displacements. **The hypothesis is that in the stroke group, if scaling of corrective response was present, it would correlate with a successful-good performance and if no scaling was present, it would correlate with successful-poor or an unsuccessful performance. This would enable to identify the reason for the limits of balance recovery in the stroke population.**

3. METHODOLOGY

Motor control and Neurological Dysfunction Laboratory in the School of Medical Rehabilitation provided the research environment to undertake this research task. The University of Manitoba, Faculty of Medicine, Ethics Committee approval was granted prior to recruiting the subjects. Informed consent was obtained from all subjects.

3.2. SUBJECTS

The intent of this study was to examine residual balance deficits in stroke survivors who are community dwellers but unable to function independently outdoors. This included subjects who were at least 6 months post–stroke and who had completed an inpatient-rehabilitation program. Healthy adults served as "normative" controls.

Ten individuals with hemiplegia from a first stroke, and nine healthy subjects were recruited. Stroke subjects were recruited in two ways:

1. The Stroke Association of Manitoba was contacted and informed of the study, its purpose, and what was expected of the participants. This information was presented at one or more of their general meetings and people interested to participate in the study were recruited.
2. From the Physical Therapy Department, Rehabilitation Hospital, Health Sciences Center. Potential outpatient clients were asked if they would be interested in volunteering to participate in this study and after their consent they were recruited in the study.

INCLUSION CRITERIA FOR STROKE SUBJECTS:

- At least six months post-stroke
- Cortical stroke in the fronto-pareital region.
- Between 40 and 70 years of age
- Stand independently and walk, with or without aids, for at least 20 meters.
- Having a score between 2-5 on the Chedoke-McMaster Stroke Assessment Motor Impairment Inventory.

EXCLUSION CRITERIA:

- Individuals with receptive aphasia, cognitive or perceptual deficits who could not satisfy the investigators that they understood what was expected, follow directions or give consent.
- Individuals with history of a previous stroke
- History of any other musculo-skeletal or orthopedic diseases.

The healthy control subjects were recruited from family, friends and colleagues. Each participant was required to sign a consent form that fully described the study and what was expected of the individual. The consent form also stated that the participant was free to withdraw from the study at any time. Each subject was asked to wear running shoes, black shorts with cutouts for placement of reflective markers, and T-shirts. The control subjects were weighed before the experiment. The stroke subjects were weighed by placing the left and right limbs two separate weighing scales. This gave an indication of the weight bearing asymmetry in the stroke subjects.

3.2. MOVABLE PLATFORM APPARATUS

The task of single leg-lifting was performed on a movable platform with an AMTI biomechanical force plate firmly mounted into the standing surface. The movable platform was constructed to provide forward and backward support surface translations. Platform translations were controlled by an electrical DC motors/linear actuator (model H105B, Industrial Devices Corporation, 35 Pamaron Way, Novato, CA 94949) with a factory installed linear potentiometer mounted inside the cylinder of the motor. After calibration, the linear potentiometer signals were used to determine linear displacement of the platform during the translations. The displacement data was low-pass filtered at 20 Hz using a fourth order Butterworth type zero phase lag digital filter and then differentiated with respect to time to obtain velocity. An Amiga 2000 computer equipped with a digital to analog (D/A) converter is interfaced to the control units of the DC motors.

The AMTI force plate was used to record outputs of the ground reaction forces in the antero-posterio (sagittal) plane (Fy), mediolateral (frontal) plane (Fx) and the horizontal plane (Fz). The force plate also provided the moments for each force Mx, My, and Mz.

The moment (My) about the anterior-posterior axis was used to trigger the Amiga 2000 computer. According to the pilot data there was a positive or negative change in My from a flat base-line in standing as the swing leg unloads in preparation of leg-lifting. Thus the platform was triggered after the subjects shifted their body weight to the stance limb.

An IBM compatible computer equipped with a 16 channel 12-bit analog to digital (A/D) converter (RC Electronics Inc., 6464 Hollister Ave, Goleta CA, USA) was used to collect EMG signals, force plate signals, linear potentiometer signals and the Amiga trigger pulse signal (analog signals). At onset of the motor cylinder movement, one output of the control unit generates a square pulse. This pulse activated two light emitting diodes (LEDs) placed on either side of the walkway and in view of three video cameras. This signal was also collected on the IBM computer, and used to synchronize video motion analyses with the analog signals described above.

3.3. EXPERIMENTAL PROTOCOL

The movable platform apparatus was used to suddenly and unexpectedly displace the support surface in order to disturb the subjects' balance in a controlled and safe manner. In order to assure that the subjects would not fall, they were fitted in a safety harness and assisted by a physiotherapist standing beside them through out the experiment. The safety harness was fitted around the pelvis and trunk. It was secured above to a rigid over- head support system, which is mounted to the main joints of the floor above.

Subjects were assisted to stand on the moveable platform. The subjects were instructed to stand straight with feet parallel and bent the arms at elbows. Bending arms ensured ASIS and hip markers would not be obscured during the test. Each subject was instructed to stand still initially to ensure proper weight bearing on both the feet. Subjects performed 24 self-paced leg lifts where they lifted their foot off the ground for 2 blocks of 12 trials. In one block of 12 trials, the subjects were instructed to lift their right foot and in the other block their left foot, i.e. non-paretic and paretic limbs. As the foot was raised the support surface suddenly moved either forwards or backwards. The following platform motion parameters were used:

- Displacement of translation was 10 cm

- Three target velocities of translation; 20 cm/s, 25 cm/s and 30 cm/s, which were all reached in 100 ms.

The platform moved in only one-half of the trials in each block (6 Of 12 trials, 3 forward and 3 backward). The subjects were given adequate rest as required between trials. Between blocks the subjects were seated during the rest period.

3.4. DATA RECORDING

FORCE PLATE:

The (OR-6, AMTI Inc.) force plate was used to record the resultant ground reaction forces and moments during the platform movements and balance reactions. It measured the ground reaction forces in antero-posterior (sagittal) plane (Fy), mediolateral (frontal) plane (Fx), and vertical plane (Fz) and the three moments Mx, My, and Mz. The force platform data was collected on the A/D converter and stored for analysis and used to look at the preparatory phase of the task to ensure that the task was performed consistently within and between groups.

VIDEO BASED MOTION ANALYSIS:

The subjects were filmed using three synchronized video cameras positioned on left (Sony SLV-R5UC) and right (Panasonic AG-450) sides of sagittal plane and the frontal plane (Sony CCD-V801). One camera (Sony CCD-V801), which was connected to a video cassette recorder (VCR) (Sony SLV-R5UC), was placed in the left side of sagittal view. The shutter speeds chosen for the left, right and frontal cameras were 1/250. All cameras were placed on a stationary box at a fixed distance from the platform. In order to highlight reflective markers, spotlights were focused on left, right and front of the subjects' body. These spotlights were placed adjacent to the cameras. Prior to each experiment, a calibration rod was placed in the middle of the force plate and filmed by each of the camera, separately. The calibration rod provided a reference to scale the coordinate data and was used as a scaling factor to relate pixel units to real units when digitizing the markers.

The circular light reflective markers were placed on anatomical landmarks defining end points of body segments joint axes of rotation. The markers were placed on the following:

- The vertex of the skull - head
- The zygomatic angle of the mandible - chin-neck angle
- The acromium process - shoulder in frontal plane
- The mid-point of the lateral aspect of humeral head - shoulder in sagittal plane
- The lateral epicondyle of humerus - elbow
- The anterior superior iliac spine (ASIS) - pelvic crest/L4
- The antero-superior margin of greater trochanter - hip
- The lateral condyle of the femur - knee
- The lateral malleolus - ankle

A reflective marker was used as fixed earth reference. The fixed earth reference marker was placed on a fixed rod attached to the ceiling and was behind the subject. The x and y coordinates of common earth fixed reference marker were subtracted from coordinate data of each body marker to obtain a common coordinate system for right and left sagittal plane video data. The peak 2D video motion analysis system (Peak Performance Technologies Inc., 7385 S. Revere Parkway, Suite 601, Englewood, Colorado) was used to digitize x and y coordinates of the centroid of each marker, relative to the earth fixed marker. A sampling rate of the video system of 60 Hz (60 images per second) was used. For each trial, a total of two hundred images were digitized, twenty before the onset of platform motion, and one hundred and eighty after the onset. This raw coordinate data was low-pass filtered at 5 Hz, using a fourth order Butterworth zero phase lag digital filter. Figure 1 illustrates the experimental set-up and recording, with a stick figure showing placement of light reflective markers and surface EMG electrodes.

A customized software package was used to calculate kinematic data from the raw coordinate data in the right and left sagittal and frontal planes. The floodlights placed next to each camera, reflected off the surface of the platform, and merged with the calcaneus and 5th metatarsal markers and obscured them. Therefore it was not possible to digitize these markers, and analyze angular displacements about the ankle. The following parameters were calculated:

i) Joint angular displacement and velocity of the knee and hip joint

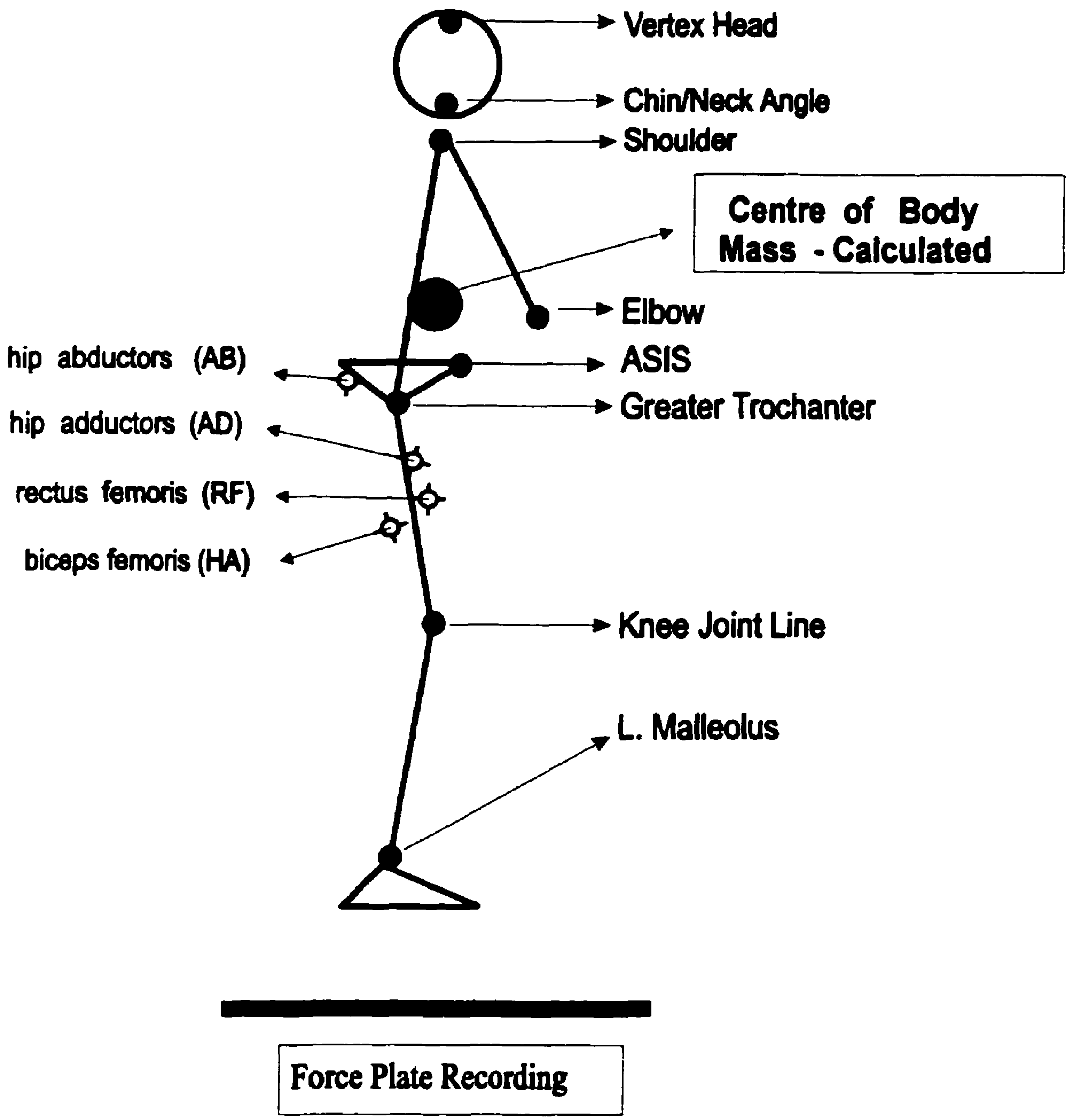


Figure 1: Illustration showing experimental set-up. Shaded small circles show locations of bilateral light reflective markers placed at segment endpoints and axes of rotations. Open circles mark locations of EMG electrodes, placed bilaterally. Centre of foot pressure (CFP) is derived from force plate recordings.

ii) Angular displacement and velocity of trunk segment rotation

iii) COM displacement relative to space and displacement of centre of mass relative to foot in the sagittal plane

iv) Linear displacement and velocity of the lateral malleolus marker x and y-coordinate to obtain foot placement data in the sagittal plane

The individual joint angles were defined as follows:

i) Knee angle: line from ankle-knee and the line from knee-hip

ii) Hip angle: line from knee-hip and the line from hip-pelvic crest

iii) Trunk segment: line from pelvic crest to shoulder

iv) Shank segment: line from the ankle to knee

The TBCM in the sagittal plane was calculated using the method described by Winter's (1980) from

i) The end-point coordinate data

ii) Anthropometric data obtained from Chandler 1975

iii) The subject's body mass.

The displacement of the total body centre of mass was defined as the centre of mass relative to space (CM-S). The displacement of the x coordinate of the lateral malleolus from the stance side was subtracted from the x coordinate of the CM-S to obtain the relationship of center of mass relative to the foot (CM-F). The CM-F indicates the position of the total body centre of mass in relation to the base of support. Positive values during BT indicated, that the foot was moving away from the CM-S and negative values indicated vice-versa. For FT the directions were opposite to those of BT.

EMG:

EMG surface electrodes were placed over the muscle bellies of eight muscle groups which will included gluteus medius (hip abductor, AB), adductor magnus (hip adductor, AD), rectus femoris and hamstrings bilaterally. Care was taken to ensure that the skin surface is clean and also that there was the same bilateral placement of the electrodes. The placement of the electrodes was done according to the locations described by Winter (1991). Pre-packaged disposable surface electrodes (Medicotrace) were used to record the EMG signals. Three surface electrodes were placed on each muscle belly, two of which served as the active electrodes and the third as the earth. The electrodes were connected to miniaturized lightweight pre-amplifiers (30 grams) with the help button shaped leads and wires. The pre-amplifiers amplified the raw EMG signal 100 times. The electrodes and pre-amplifiers were taped to the skin to achieve safe and consistent placement throughout the experiment. The pre-amplifiers were connected to the differential amplifiers by cables that were hooked at the back of subject. The cables did not interfere with the subjects during the task.

Two 4-channel EMG amplifiers (Biosys.) were used to amplify and process the differential EMG signal obtained. The signals were there after band-pass filtered (10 Hz to 1 kHz, –3db), rectified and low-pass filtered at 50 Hz. The EMG signals were recorded on the IBM compatible A/D converter. For each platform trigger trial a 4 second sweep of data was collected at a sampling rate of 333 Hz. The data file for each subject was saved on the computer and further analysis.

FUNCTIONAL ASSESSMENTS:

Chedoke-McMaster Stroke Assessment (Impairment and Disability Inventory) and gait speed were examined and recorded. These are standard clinical assessment scales of motor impairment or disability used in the Physiotherapy departments at the Health Sciences Center and St. Boniface Hospital.

The Chedoke-McMaster Stroke Assessment is a standardized outcome measure that consists of the Impairment Inventory and Disability Inventory. It has undergone extensive study of reliability, validity and responsiveness (Gowland et al 1993a, 1993b, 1995, Barclay-Goddard 1994). The Impairment Inventory determines the presence and severity of physical impairments of, the arm including shoulder, the hand, the leg, the foot, and postural control. It is scored using a modified seven point Brunnstrom scale, determined by the quality and pattern of voluntary movements present. To assess outdoor walking competency the Walking Index section of the disability inventory of the Chedoke-McMaster Stroke Assessment Scale was used. It consists of the following items, walking indoor 25m, walking outdoors over rough ground, ramps and curbs for 140 m, walking up and down stairs. Scaling of the different walking activities is based on the 7-point FIMM scale. In addition walking distance for a 2-minute walk is determined and scored according to age. See APPENDIX 1, for description of Chedoke-McMaster Stroke Assessment Scale.

The subjects were asked to walk a distance of ten meters on a flat surface. After a steady state gait velocity was reached, the time taken to walk a distance of 3 m was noted. This was repeated once again. The steady state velocity over three meters was then calculated. A similar procedure was repeated to calculate gait speed over 25 m.

3.5. DATA ANALYSIS

The subjects were divided into three groups for purpose of data analyses and interpretation: 1) CON – controls with right side in stance, 2) NST - stroke group with non paretic side in stance (lifting paretic limb) and 3) PST- stroke group with paretic side in stance (lifting non- paretic limb)

FORCE PLATE RECORDINGS:

The antero-posterior (AY) and medio-lateral (AX) center of foot pressure was calculated from the calibrated force plate data using a custom software program. The peak magnitudes of AY and AX displacement during the leg-lifting task before onset of platform motion were obtained. The group means and standard deviations were calculated. A T-test was done to identify any group differences in the displacement of the center of foot pressure. This was done to check that the task was performed in a similar fashion between the three groups being evaluated, in particular that there was no systematic difference in either AP or ML centre of foot pressure displacement during the task.

STRATEGIES AND PERFORMANCE LEVELS FOR CORRECTIVE RESPONSES:

Corrective balance responses were grouped into three strategies 1) inplace 2) neutral step (touchdown) and 3) single step. The strategies were determined by analyzing displacements of the x and y coordinates of the lateral malleolus (ankle) marker from the swing side within the first 300 ms after onset of platform motion.

For an inplace strategy, the horizontal displacement (x) was in the direction of the platform motion. The vertical displacement reached its peaks and plateau signifying that the foot remained off the ground.

For a touchdown, the foot was lifted off the ground during the task, but shortly after onset of platform motion, it returned to the ground within 1-3 cm of its original place. The x displacement was in the direction of platform motion similar to that seen in an inplace strategy and there was no indication of a forward or backward step.

For a Stepping strategy the horizontal displacement reverses direction within 100ms of the platform onset signifying a step taken in a direction opposite to the platform motion. The vertical displacement in this case, similar to the touchdown reached its peaks, but immediately reversed direction after onset of platform and returned to the base-line, signifying that the foot had touched the ground.

Performance was divided in to four levels and a grade given for each level. The following four grades were assigned to the performance levels:

Grade 1 - good: was given to subjects who exhibited an inplace strategy

Grade 2 - fair: was given to subjects who exhibited either a touchdown strategy or a primary stepping strategy

Grade 3 - stumble: was given to subjects who took multiple steps and required assistance to maintain balance and prevent a fall.

Grade 4 - fall: fall is made operational for this study to signify when subject immediately lost their balance and needed assistance from the physiotherapist standing besides them to prevent a fall.

It should be noted that the subjects were allowed to react to their limits of stability before assisting them to regain balance. Since our sample size was small for purpose of statistical analysis and associations we collapsed the good and fair into successful performance and the stumbles and falls into unsuccessful performance

<u>CENTRE OF MASS:</u>

To identify how the total body centre of mass was regulated during the corrective response we analyzed trajectories of COM displacement relative to space (CM-S) and centre of mass relative to the foot (CM-F) in sagittal planes during the response.

Analysis of CM-S and CM-F trajectories was also used to confirm the presence of different corrective strategies and performance levels. The direction and magnitude of the CM-S displacement after platform onset and the direction and peak-to-peak magnitude of the CM-F were used for this purpose.

KINEMATIC ANALYSIS:

In order to identify and group the observed movement synergies (kinematic patterns) used to restore balance, angular displacements at the knee and hip joint and trunk segment rotation in the sagittal plane were evaluated. The following kinematic parameters were quantified:

i) The direction and peak magnitude of the initial corrective joint angular displacements (extension/flexion) and trunk segment rotation from the stance leg in a time period of about 300-400 ms after onset of platform translation.

ii) The timing of the corrective response, wherever possible was determined by looking at the onset of change in direction of angular displacements on the stance side after onset of platform translation. The mean onset timings obtained from the control group were used to interpret a delayed movement response in the stroke groups.

iii) The direction and peak magnitude of the hip angular displacement and trunk segment rotation in the frontal plane after onset of platform motion were analyzed to look for an excessive response in this plane. An excessive response was one when the magnitude of displacement was double of that present in the trials with no platform translation (TO).

The step onset could not be determined from the x and y linear displacements of the lateral malleoulus marker as during the task the foot was already off the ground and moving. However a neutral step (touchdown) a single step and a multiple step could be easily distinguished from each other from the x and y displacements of the lateral malleolus marker as described above.

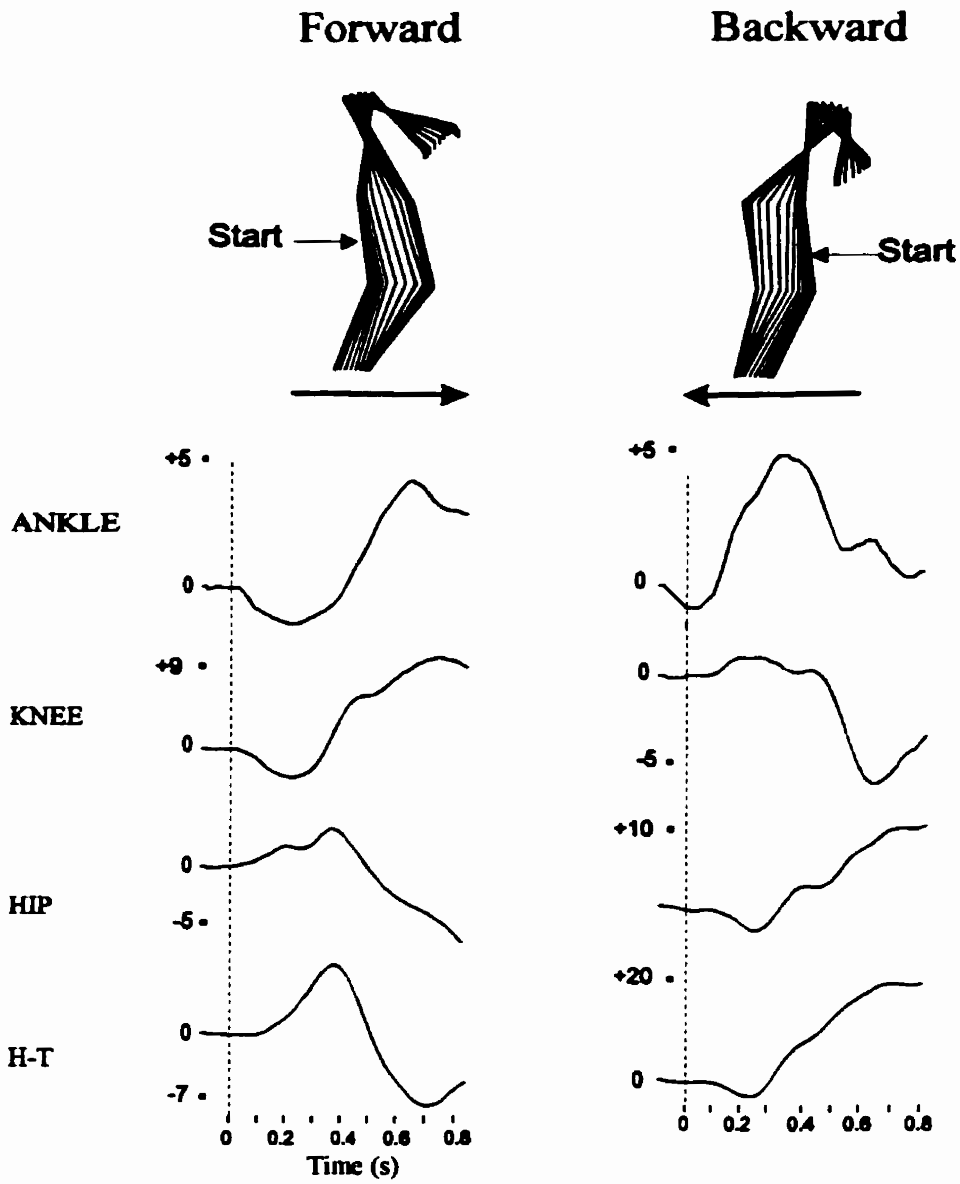
From the knee, hip and trunk angular displacements, for each trial the pattern on the stance leg was determined. Brunham (1996) has identified different inplace corrective movement synergies for BT and FT with both feet on the ground, illustrated in Figure 2. The normal pattern for backward translations is early onset of hip flexion, and ankle planter flexion to move the body centre of mass backwards over the feet. The magnitude of angular knee displacement is typically small 1-5 degrees, and the direction is variable, slight extension, or slight flexion.

The normal pattern for forward translations is active knee flexion and ankle dorsiflexion followed by active hip extension, thrusting the pelvis forward in order to move the body centre of mass over the feet. It is expected that these same inplace movement patterns would be observed in most of the controls in the present study. The direction and peak magnitude of the hip and trunk displacements for the corrective response to BT and FT were quantified. Wherever possible the onset of the corrective response was also quantified. Deviations from the normal pattern seen in controls were quantified for the NST and PST depending on the onset, direction and magnitude of the hip and knee displacements trunk segment rotation. For example if hip flexion was seen instead of hip extension or knee flexion was delayed in onset during FT then it was classified into an abnormal pattern. If the magnitudes of angular displacements were outside the range obtained from controls, the pattern was classified as abnormal. The different abnormal patterns seen in stroke groups are described in detail in the results section.

Further each trial was classified as having a normative pattern or an abnormal pattern. The frequency of the trials with successful performance having a normative pattern was associated with the total number of successful trials. Similarly, the frequency of unsuccessful trials having an abnormal pattern was associated with the total number of unsuccessful trials. This was done for BT and FT separately.

EMG ANALYSIS:

The onset latency of the early muscle activity during the corrective response will be determined by employing methods described by Eng et al. (1994) and Tang et al. (1998). Each subject's averaged EMG activity for all the leg-lifting trials with no perturbations (NPT) was calculated for each muscle separately. EMG records of each perturbed trial (FT and BT) were subtracted from the averaged control EMG waveform to



(+) is flexion / dorsiflexion
(-) is extension / plantar flexion

Figure. 2: Stick figures and plots of angular displacement illustrating a normative in-place pattern for FT (left panel) and BT (right panel) at V3 with both feet on the ground. Stick figures illustrate the direction and pattern of the movement strategy used to restore balance in response to FT and BT. Each stick figure line represents the coordinate data from every third digitized image (33.3 ms interval). Bold lines indicate overlapping images prior to platform movement and the stable equilibrium position reached at 500-600 ms. Plots of angular displacement indicate pattern of the corrective response. Vertical line at time 0 is the start of platform motion. The Y axis is degrees. Positive (+) direction indicates dorsiflexion, knee, hip and HT flexion. Negative (-) direction indicates plantarflexion and knee, hip, and HT extension.

produce a difference waveform. The following was quantified from the difference waveform obtained:

a) **Onset latency**: Onset latency was defined as the time interval between the start of the platform motion and the beginning of the first detectable change in difference waveform exceeding the mean pre-movement difference waveform level (first 700 ms of the signal) by at least 2 standard deviations. We looked for onsets within a time period of 300ms from the initiation of platform motion, as we were interested in the early muscle responses, which represent automatic corrective balance reactions. Onset of EMG signals was quantified for each of the eight muscle groups.

b) **Occurrence frequencies**: The frequency of presence (excitatory or inhibitory response) or absence of EMG corrective response was analyzed. For the response to be marked as present, it had to occur within a window of 300ms and be greater than two standard deviations of the pre-movement level expressed as percentages.

__FUNCTIONAL ASSESSMENTS:__

Each subject received a lower limb impairment score, a posture control score and a disability score from the Chedoke impairment and disability assessment. The gait velocity scores for each subject over both the 3m and 25 m distance were calculated. The average 3 m gait speed and the walking index score from the disability inventory were tabulated for a rank order correlation with performance levels.

3.6. STATISTICAL ANALYSIS

Statistical analysis of the data related to the objectives were done to determine if the hypothesis stated were acceptable or not. Comparisons were made between the independent and dependent variables. The independent variables were:

1. GROUPS: consisting of three levels, the controls, non-paretic stance and paretic stance.
2. Rate of platform translation consisting of three levels, V1, V2, and V3

The following analysis was performed:

i) The Chi-square statistical procedure was used to determine if there was a significant difference in performance levels between groups CON, NPT, and PST. For each level of platform acceleration/velocity the number of trials with a performance grade of good and fair were totaled and put into one group called successful performance. Similarly the number of trials with performance grade of stumble or fall were totaled and put into a group called unsuccessful. The values for successful and unsuccessful performance for all the three groups were tabulated and entered in a 3 x 2 Chi-squared analysis. The analysis was done separately for BT and FT. If a significant difference was found between groups, then individual 2 x 2 Chi-square analysis was done between pairs of groups. A significance level of 0.05 was used.

ii) A similar analysis was done to look for a difference in strategy selection between groups for BT and FT separately. All the trials with a touchdown or an inplace were collapsed into a single inplace strategy group. The values for inplace strategy and primary stepping strategy were entered in a 3 x 2 Chi-Square analysis. If a significant difference was found post-hoc analysis with 2 x 2 Chi-square tables was done.

iii) To examine differences in timing of the corrective balance responses between groups a one-way ANOVA by group (CON, NPT, and PST) was performed on EMG onset latencies. Thus this analysis was only possible for a limited number of muscles i.e. when the majority subjects (at least 75%) in each group exhibited a response. When appropriate, a post-hoc analysis using Bonferroni's correction was used to determine pair-wise group differences.

iv) A repeated measures ANOVA was used to look at the within group difference (scaling effect due to rate of platform translation) and between group effect on magnitudes of hip flexion, trunk rotations (forward pitch) and knee flexion during BT. This is the normative inplace movement synergy for BT. This was present in enough subjects, in all groups to attempt this analysis.

v) To correlate performance level with disability, each trial of a subject for V1, V2 and V3, was assigned a performance grade: 4 (good-successful), 3 (fair-successful) and 2 (stumble-unsuccessful) and 1 (fall). This was done for FT as well as BT separately. The composite performance grade score from all the trials was calculated and entered into a Spearman Rho correlation with each subjects' Chedoke walking index scores to obtain correlation coefficients (r values). Similar correlation was done with the 3 m gait speed.

4. RESULTS

The result section addresses three main objectives of this study:

1. To identify and compare performance levels of corrective balance reactions between healthy individuals and stroke individuals
2. To identify the means by which the corrective reactions occurred
3. To correlate performance levels on this balance test to clinical tests.

Comparisons are made between young healthy subjects, and NST and PST sides of the stroke subjects. Only one side of the control group was analyzed i.e. right side stance. Data from the stroke subjects was analyzed both when the non-paretic side was in stance as well as when the paretic side was in stance. The results are presented in 7 sections. These are:

1. Characteristics of Subjects
2. Equivalent balance disturbance between groups
    - Similarity in task performance
    - Onset of platform motion
3. Strategy selection
4. Performance levels
5. Means by which corrective response was achieved
    - Angular kinematics for corrective reactions in sagittal plane
    - Association of performance levels with movement patterns
    - Scaling of the corrective response between group with increasing disturbance
    - Angular kinematics for balance reactions in the frontal plane
    - EMG onset latencies
6. Correlation between overall performance and Chedoke walking index scores
7. Walking Index and gait speed

4.1. CHARACTERISTICS OF SUBJECTS

Table 1 displays the age of the stroke subjects, side of stroke, onset of stroke and the % mass on the paretic leg in standing, Chedoke walking index scores and gait speeds

Table 1 -Characteristics and clinical scores of stroke subjects.

SUBJECTS	SEX	AGE	SIDE (CVA)	POST CVA (Months)	% PARETIC WEIGHT	WI SCORE	GAIT SPEED (m/s) 3 m	25 m
1	F	55	LT	36	43.6	27	1.05	1.16
2	M	67	LT	6	37.5	21	0.75	0.96
3	M	60	LT	12	36.6	17	0.28	0.27
4	M	53	LT	6	38.9	24	0.91	0.93
5	M	61	LT	24	33.33	14	0.26	0.24
6	F	60	LT	10	32.3	17	0.38	0.21
7	F	60	LT	7	40	21	0.49	0.48
8	M	54	RT	10	37.5	23	0.57	0.59
9	F	63	RT	132	42.6	27	0.51	0.52
10	F	65	RT	72	41.9	19	0.34	0.29
Mean	**NA**	**59.8**	**NA**	**31.50**	**38.42**	**21.00**	**0.55**	**0.57**
SD		4.64		40.86	3.76	4.35	0.27	0.34

% Paretic weight = % weight bearing on paretic side of total body weight

WI = Total score of the walking index section from the disability inventory of the Chedoke McMaster Stroke Assessment Scale of the subjects over 3 m and 25 m. The Chedoke impairment scores for the leg and foot ranged between 3 and 5 for all the subjects. The maximum score a subject can get on the impairment score is 7 and the minimum score one can get is 1. The walking index scores ranged between 14-27. The maximum score one can get on the walking index is 30 and minimum is 4. The gait speed range was 0.34 -1.05 m/s. It can be seen from the table that the stroke subjects had less than 50% weight bearing on the paretic side, ranging between 32-47% with a mean of 38 (3.3).

4.2. EQUIVALENCE OF TASK PERFORMANCE AND BALANCE DISTURBANCE

4.2.1 Task Performance

In terms of centre of foot pressure displacements, the leg-lifting task was performed similarly between groups and between sides of the body. Figure 3 shows typical during the leg-lifting task. The platform moves as shown at time zero. It can be seen that there is substantial overlap in the three traces, suggesting there was no difference in displacement of the center of foot pressure during the leg lifting tasks. Table 2a shows the group means and standard deviations for the peak medio-lateral (AX) and antero-posterior (AY) centre of foot pressure displacement for all the three groups from pre-movement base-line to onset of platform motion, i.e. during the task performance. Student t- tests done between CON and NST, CON and PST and NST and PST showed no statistical difference in the magnitudes of both AX and AY. The group means for AY was taken using absolute values of AY displacements. The magnitude of the displacements either forwards or backwards in the antero-posterior plane was small as seen from the group means. For majority of the trials the centre of mass was displaced backwards towards the heel. However for a few trials, the centre of mass was displaced forward towards the toes. Table 2b shows the frequency of trials for backwards and forwards shift of the centre of foot pressure. It was observed that the control subjects lifted the foot off the ground to a higher height (7-20 cm) than most of the stroke subjects in NST and PST (3-15 cm).

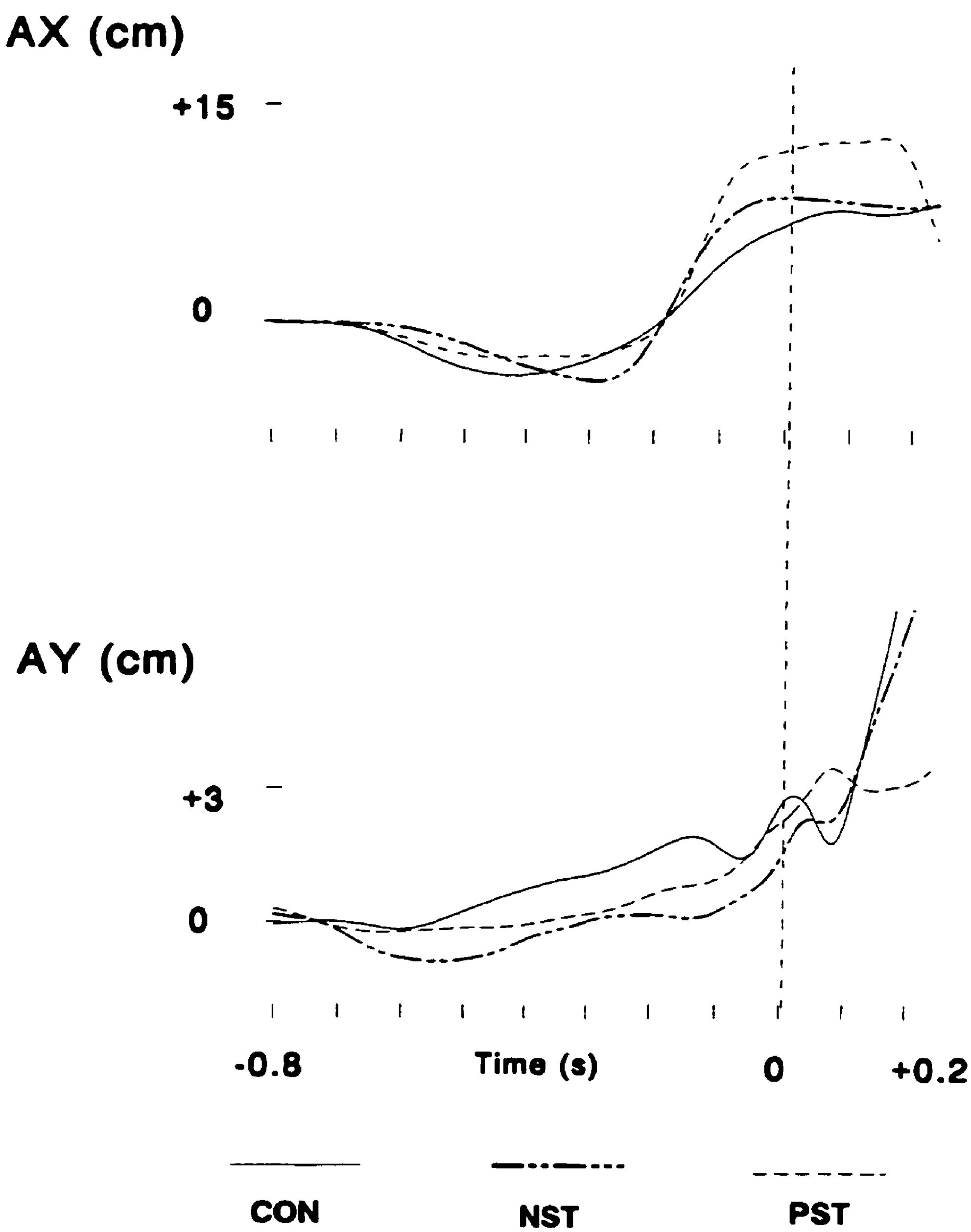


Figure 3: Plots of linear displacements of medio-lateral (AX) and antero-posterior (AY) centre of foot pressure for controls, and for stroke subjects during non- paretic side in stance and Paretic side in stance, during the task of leg lifting. Vertical dashed line at time zero is the onset of platform translation. Positive values for AX indicate displacement towards stance leg. Positive values for AY indicate anterior or forwards displacement. All curves have been offset to zero for display purposes; for each plot the y-value at time 0 was subtracted from each point.

Table 2a - Means and Standard deviations of the centre of foot pressure in the anteroposterior (AY) and mediolateral (AX) planes from all the trials during performance of the task before onset of platform motion.

	CON		NST		PST	
	AY	**AX**	**AY**	**AX**	**AY**	**AX**
Mean	1.858	13.415	1.45	11.07	2.75	13.15
S.D	0.866	2.97	0.93	2.86	1.78	4.16

CON - Controls (n= 54)

NST- Non paretic leg in stance (n=60)

PST-Paretic leg in stance (n=60)

Table 2b - Frequency of direction of movement of AY in CON, NST and PST

	CON	PST	NST
Backward	55	45	56
Forward	8	25	15
Total	63	70	70

4.2.2 Onset of platform Translation

As stated in the methodology the moment about the anterior-posterior axis (MY) was used to trigger the platform into motion. Figure 4 shows typical plots of MY from a control and stroke subject and platform displacement from the linear potentiometer recordings. A steep rise in MY is seen as the subjects unload the swing leg and transfers all the body mass over to the stance side. MY reaches its peak and plateaus once the foot is lifted off the ground and the centre of mass is over the stance side. The platform was programmed to trigger into motion when a set threshold value on MY was reached. A threshold range for MY was determined for each subject during practice trials. The range for all the controls and most of the stroke subjects was -2 to –4V while lifting the right leg, and +2 to – 4V when lifting the left leg. For some stroke subjects the range had to be widened to prevent false triggers and ensure that the platform was triggered into motion when the foot was off the ground. It can be seen from Figure 4 that the platform was set into motion when the MY reached its peak. This was the case for all subjects and trials.

4.3. STRATEGY SELECTION

Corrective balance responses were grouped into three strategies 1) inplace 2) neutral step (touchdown) and 3) primary step. The strategies were determined from the linear displacements of the x and y coordinates of the lateral malleolus (ankle) marker and confirmed by looking at the direction and magnitude of displacement of the centre of mass relative to space. In a previous study Brunham (1996) reported that for an inplace strategy with both feet on the ground the CM-S begins to move in the same direction of the platform within 20-250 ms. A similar finding was observed in this study. For an inplace strategy, the CM-S moved in the same direction as the platform. For a touchdown (neutral step), the CM-S also moved in the same direction as platform motion. However for a primary step to restore balance the direction of the step was opposite to that of platform motion, and thus, the CM-S moved in the direction of the step. Figures 5 and 6 show the different trajectories of linear displacements of the lateral malleolus (x and y) and the CM-S that represents the different strategies observed in this study.

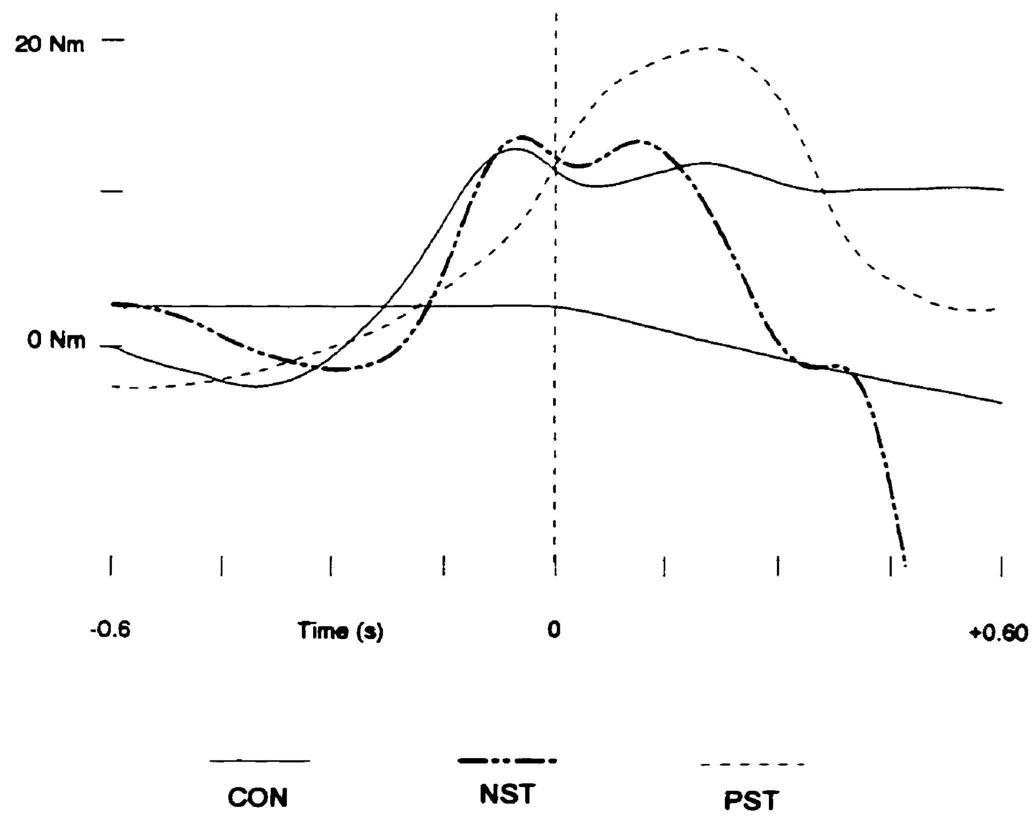


Figure 4: Figure representing traces of moments (MY) about the antero-posterior axis from a control subject, non-paretic side in stance and paretic side in stance from a LCVA subject and record of the linear potentiometer signal, representing displacement of the platform. Negative values for linear potentiometer indicate displacement of platform in the backward direction. Positive values represent shift of MY towards the right side and negative values represent shift of MY towards the left side. For illustrative purposes the MY from PST was inverted. Vertical dashed line at time zero is the onset of platform translation.

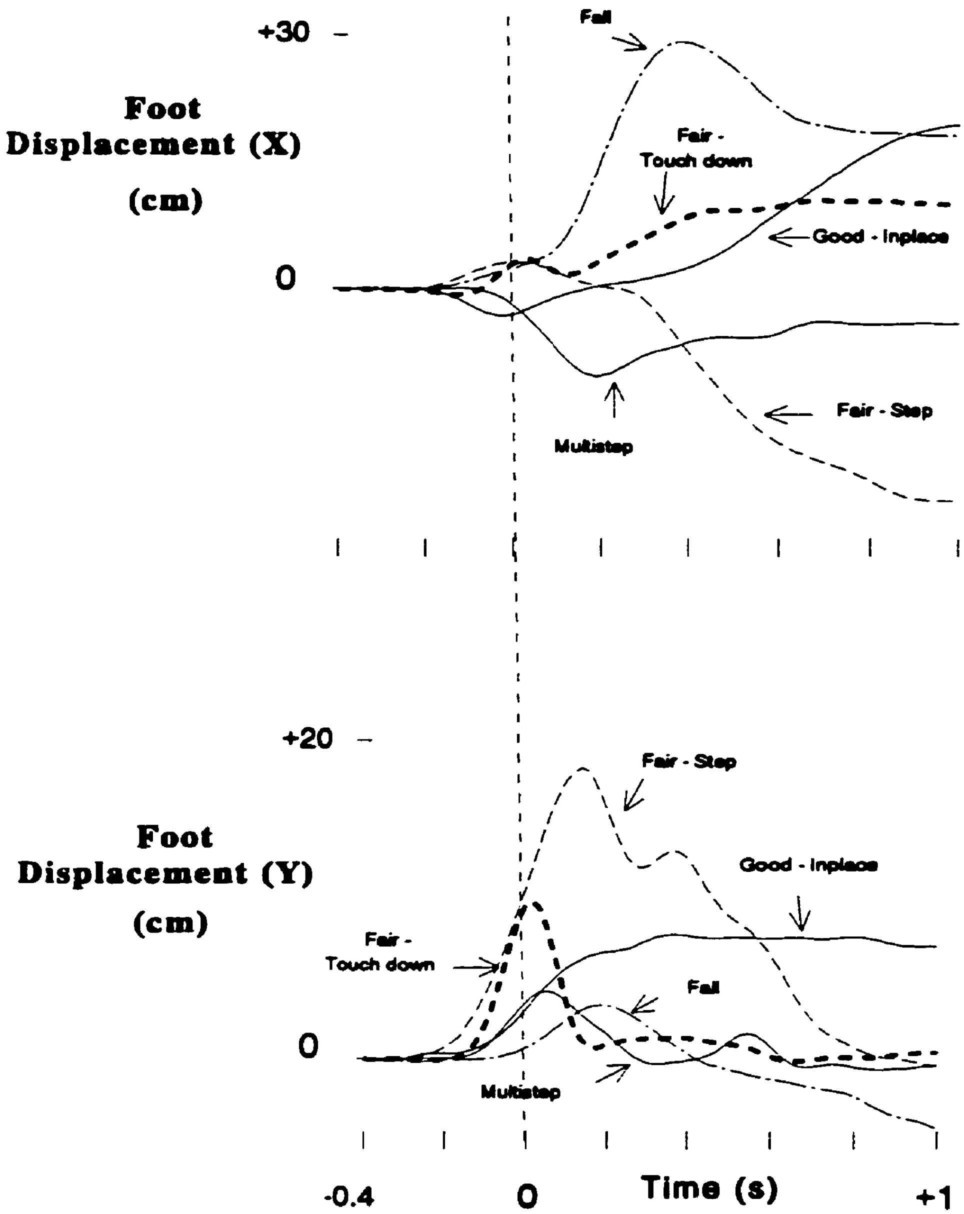


Figure 5: Plots of horizontal (X) and vertical (Y) displacements of the lateral malleolus marker during BT from the swing side for selected trials from a variety of subjects (controls and strokes) having different performance levels and strategies; Good- inplace, Fair- touchdown, Fair-step, Stumble-multiple steps and Fall. Vertical dashed line at time 0 is the onset of platform motion. All curves have been offset to zero for display purposes; for each plot the y-value at time 0 was subtracted from each point. For (X), positive direction is backward displacement and negative direction is forward displacement. For Y coordinate, upward displacement represents upward vertical displacement of the foot.

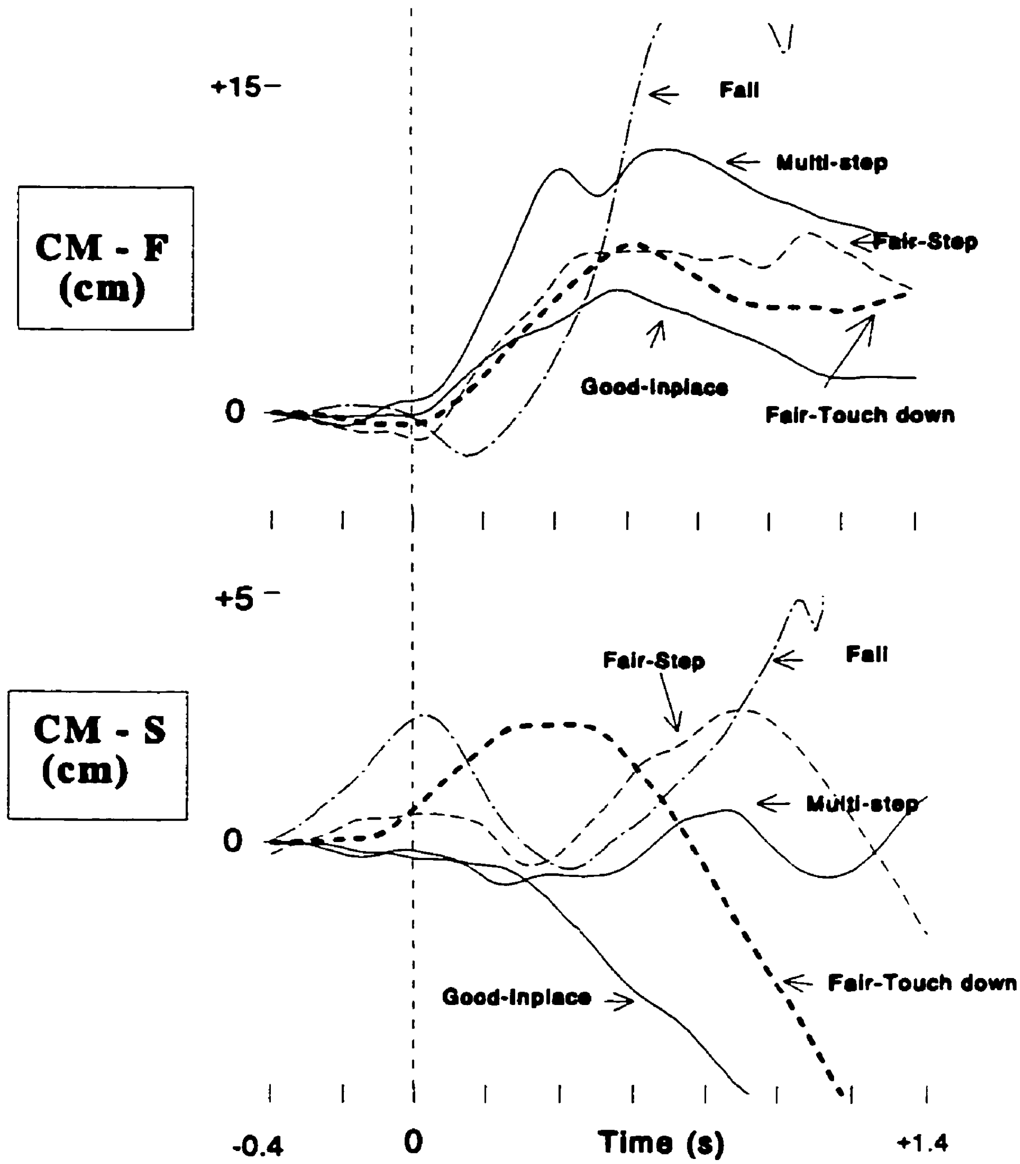


Figure 6: Plots of centre of mass relative to base of support (CM-F) and centre of mass relative to space (CM-S) from the stance side during BT selected from a variety of subjects (controls and strokes), representing the different performance levels and strategies; Good- inplace, Fair- touchdown, Fair-step, Stumble-multiple steps and Fall. Vertical dashed line at time 0 is the onset of platform motion. All curves have been offset to zero for display purposes; for each plot the y-value at time 0 was subtracted from each point. For CM-S positive direction is forward displacement and negative direction is backward displacement. For CM-F positive direction represent COM ahead of the foot and negative direction represent COM behind the foot.

Figure 7 (right panel) presents summary histograms for percentage of total trial (all rates of platform translations combined) in which an inplace strategy was used for CON, NST and PST. Table 3a and b (left sides) presents the frequency of the different strategies exhibited by the three groups over all three velocities for FT and BT. Note the total number of trials for each velocity are nine for the controls and ten for the stroke group, and the total number of trials for BT and FT for all velocities combined was 27 for the control group and 30 for both the stroke groups. All three strategies were present in all the groups (CON, NST, and PST) at least at one velocity. For B3 in both NST and PST and for FT in PST, none of the subjects had an inplace strategy.

BACKWARD TRANSLATION: The control group exhibited mainly the inplace corrective strategy (23/27) to recover balance. There were three touch down's and one step, which occurred at the highest velocity (B3). For both the NPT and PST leg lifts there were substantially more touchdowns and primary steps compared to the controls. There were 15/30 touch down's for NST, and 10/30 touch down's in PST. There were 13/30 steps in the NST, and 19/30 steps in PST. For Chi-square statistical analysis the inplace and touch down strategies were collapsed into one category, and called "inplace". Chi-square test between CON, NST and PST revealed a group difference on selection of inplace versus primary stepping strategy ($p< 0.001$). The individual chi-test results between CON and NST and PST are given in table 4a (left side). Post-hoc analysis was done using 2 X 2 chi-square tests between CON and NST, CON and PST, and PST and NST. There was a significant difference at B1 ($p<0.05$) between NST and PST. 80% of the trials in NST were inplace compared to 20% of inplace trials in PST. There was no significant difference in strategy selection at B2 and B3. There was a significant difference between CON and NST ($p<0.001$) and also between CON and PST ($p<0.001$) at all the velocities with the controls exhibiting a higher percentage of trials with an inplace strategy.

FORWARD TRANSLATION: As can be seen in Table 3b (right panel), the control group exhibited mainly the inplace corrective strategy to recover balance at F1 and F2. There were more steps (7/27) compared to BT, occurring at the highest velocity (F3). Similar to BT there were more touchdowns and primary steps compared to the controls.

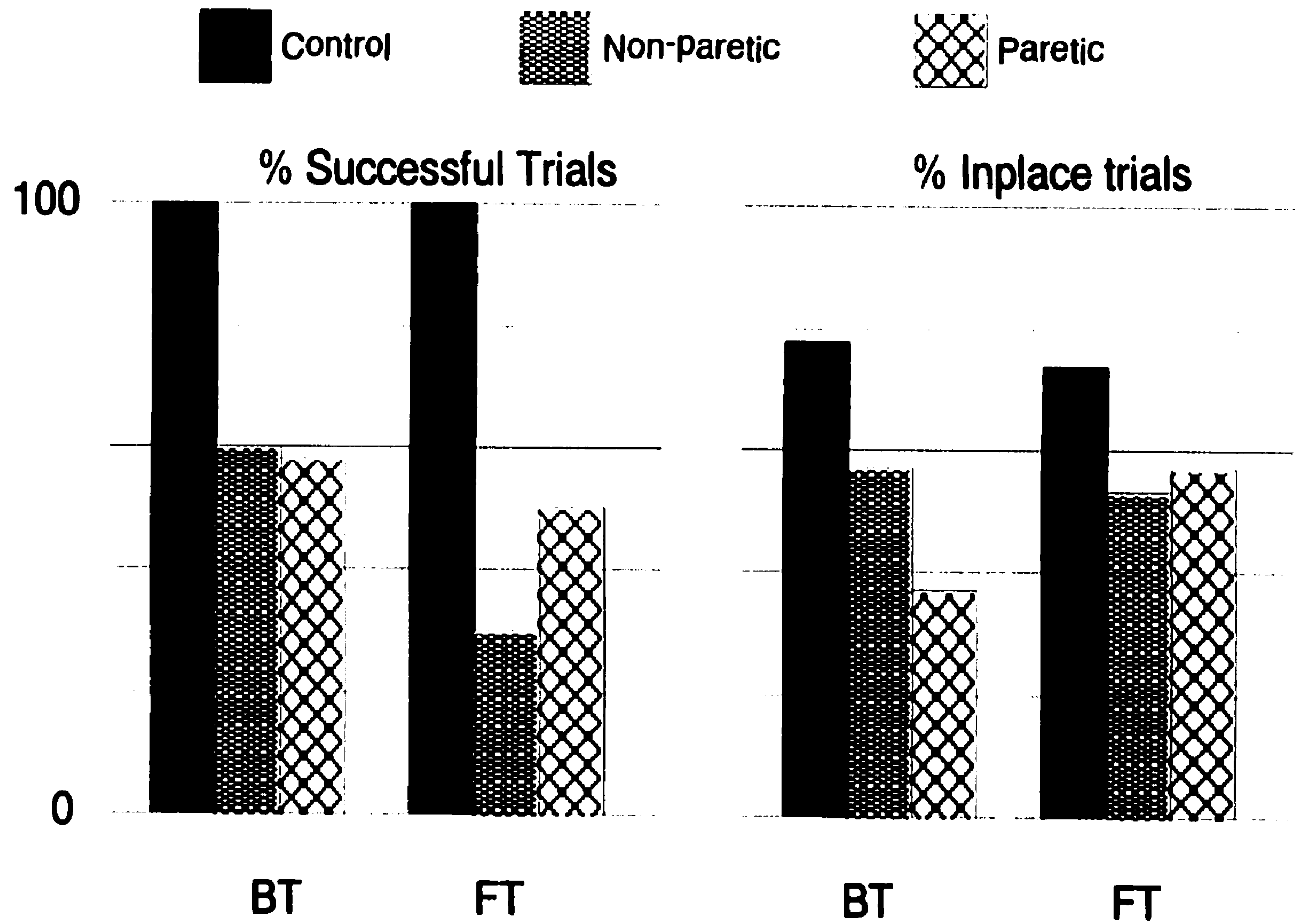


Figure 7: Histogram of percentage of total trials with successful performance (left panel) and inplace corrective strategy (right panel) all velocities combined for controls, non-paretic stance and paretic stance.

Table 3- Strategy and performance Levels by group, platform velocity and direction of disturbance for Controls, Non paretic stance and Paretic stance.

a. BACKWARD TRANSLATIONS

	Strategy	CON	NST	PST	Performance	CON	NST	PST
B1	*Inplace*	8	1	-	*Good*	9	2	-
	Touch dwn	1	7	2	*Fair*	-	5	5
	Step	-	2	8	*Stumble*	-	3	3
					Fall	-	-	2
B2	*Inplace*	8	1	1	*Good*	9	1	3
	Touch dwn	1	4	4	*Fair*	-	3	4
	Step	-	5	5	*Stumble*	-	5	3
					Fall	-	1	-
B3	*Inplace*	7	-	-	*Good*	7	-	1
	Touch dwn	1	4	4	*Fair*	2	7	4
	Step	1	6	6	*Stumble*	-	2	4
					Fall	-	1	1

b. FORWARDTRANSLATIONS

	Stratergy	CON	NST	PST	Performance	CON	NST	PST
F1	*Inplace*	8	-	-	*Good*	9	-	3
	Touch dwn	1	7	8	*Fair*	-	4	3
	Step	-	3	2	*Stumble*	-	4	2
					Fall	-	2	2
F2	*Inplace*	8	1	-	*Good*	9	-	1
	Touch dwn	1	2	6	*Fair*	-	4	6
	Step	-	7	4	*Stumble*	-	4	1
					Fall	-	2	2
F3	*Inplace*	2	1	-	*Good*	2	-	-
	Touch dwn	-	5	4	*Fair*	7	3	5
	Step	7	4	6	*Stumble*	-	3	2
					Fall	-	4	3

n for V1,V2,V3 = 9 (CON)
n for V1,V2,V3 = 10 (NST & PST)

Table 4- Chi- Square results for strategy selection by group, platform velocity and direction of disturbance for Controls, Non paretic stance and Paretic stance (in %).

a. BACKWARD TRANSLATION

	CON	NST	PST	p value
BT	96^{+++}	57^{+++}	37^{+++}	< 0.001

	Strategy	CON	NST	p value
B1	*Inplace*	100	80	0.156
B2	*Inplace*	100^{+}	50^{+}	0.013
B3	*Inplace*	89^{+}	40^{+}	0.027
BT	*Inplace*	96^{+++}	57^{+++}	< 0.001

	Strategy	CON	PST	p value
B1	*Inplace*	100^{+++}	20^{+++}	< 0.001
B2	*Inplace*	100^{+}	50^{+}	0.013
B3	*Inplace*	89^{+}	40^{+}	0.027
BT	*Inplace*	96^{+++}	37^{+++}	< 0.001

b. FORWARD TRANSLATION

	CON	NST	PST	p value
FT	74	53	60	0.261

	Strategy	CON	NST	p value
F1	*Inplace*	100	70	0.073
F2	*Inplace*	100^{++}	30^{++}	0.002
F3	*Inplace*	22	60	0.096
FT	*Inplace*	74	57	0.105

	Strategy	CON	PST	p value
F1	*Inplace*	100	80	0.156
F2	*Inplace*	100^{+}	60^{+}	0.033
F3	*Inplace*	22	40	0.405
FT	*Inplace*	74	60	0.260

Legend:

$^{+}$ P<0.05

$^{++}$ P<0.01

$^{+++}$ P<0.001

For NST there were 14/30 touch down's and 18/30 touch down's in PST. There were 14/30 steps the NST group and 12/30 steps in PST. A 3 X 2 Chi-square test between CON, NST and PST revealed no significant group difference on selection of inplace versus primary stepping strategy ($p=0.2$). The individual chi-test results between CON and NST and PST are given in table 4b (right side). Post-hoc analysis revealed a significant higher number of inplace trials in CON compared to NST ($p<0.05$) and PST ($p<0.01$) at B2. Similar to the controls both NST and PST showed a greater percent of inplace trials at F1.There was no significant difference between NST and PST in strategy selection at all F1, F2 and F3.

4.4. PERFORMANCE LEVELS

Performance levels were categorized by evaluation of four parameters: 1) direction and magnitude of displacement center of mass relative to space; 2) time to initial peak magnitude of CM-F displacement, and magnitude of recovery in CM-F displacement; 3) horizontal displacement of the lateral malleolus marker and 4) vertical displacement of the lateral malleolus marker. Four different and distinct performance levels were identified.

Grade 1 or Good performance was an inplace strategy, with the CM-S moving in the direction of the platform, within 200 ms after onset of platform motion, and having a magnitude of at least 5-7 cm. This can be seen in Figure 6a, CM-S figure. To maintain balance and prevent falling during an inplace movement strategy the centre of mass has to be actively moved back over the foot. With onset of platform motion, the foot will be displaced in the direction of the platform, while the centre of mass would remain stationary due to inertia. Thus the trajectory of CM-F displacement from base-line is initially positive. The CM-F waveform will peak when the platform stops moving. At this time for inplace movement strategies the CM-S is moving towards the displaced foot. Thus the distance between centre of mass and the foot will be getting smaller. In the CM-F waveform this would result in a reversal in the trajectory and a return to pre-movement baseline position. For a good performance level the CM-F waveform will be displaced in the direction of the platform motion, peak at a point in time and return back to base-line. In Figure 6, we can see a typical wave for a Good-inplace performance. Note here the

CM-F reverses direction and starts returning to base-line, but cannot be fully seen as the waveform has been cut off at a particular point in time.

Grade 2 or Fair performances were trials where a neutral step (touchdown or a primary forward/backward step was taken, but where no threat of falling was evident). Trajectories of CM-S displacement were similar for trials with a Good performance (inplace movement strategy) and Fair performance (neutral step or touch down strategy), i.e. after a delay of approximately 200 ms the centre of mass moved in the same direction as the platform. Trajectories of CM-F displacement during neutral steps were also similar to trials graded as Good (inplace) performance. The only difference between a Good-inplace and Fair-neutral touchdown, was that in the touchdown, the swing leg returned to the ground within 300 ms after onset of platform motion as shown by the vertical displacement of the lateral malleolus marker in Figure. 5. During a primary step the foot was rapidly moved in a direction opposite to platform displacement, and also the centre of mass moves in a direction opposite to the platform, within 200 ms of platform onset. The trajectory of CM-F (calculated from markers on the stance side of the body) initially moves in direction of platform as the foot is being displaced in that direction. It plateaus and does not return to pre-movement baseline position. This is because the swing limb takes a step and moves the centre of mass including stance side in a direction opposite to the direction of the stance foot or platform. This change in base of support strategy is effective in restoring upright balance and preventing a fall. A few trials with an inplace strategy but having a delayed onset (>300 ms) of CM-S movement and small peak magnitude between (2-3 cm) of displacement were given a fair performance grade.

Grade 3 or Poor performance, were trials where a stumble (multiple steps) was observed, i.e. more than one step taken by the swing or stance leg in addition to an initial inplace, or touchdown strategy. The secondary steps occurred much later in time and were not in any consistent direction relative to the platform motion. Majority of the multiple steps occurred in the trials where the subjects showed a primary stepping strategy. The secondary steps taken were determined from the horizontal and vertical displacement of the lateral malleolus marker from the swing side. Figure 5a, b shows the traces for a trial where a second step was taken by the swing leg, in addition to the primary step. In Figure 6 we can see that the pattern of the CM-S and CM-F waveforms initially resembles a primary step. However the magnitude and direction later depend on the direction and magnitude of the second step. This study was restricted to just quantifying the number of trials where multiple steps occurred. Any characteristics of the multiple steps were not analyzed, as they occurred late and were the second line of defense.

Grade 4 or Falls were trials where subjects could not maintain balance and would have fallen in not caught by the investigator standing besides them. In most cases center of mass rapidly moves in direction opposite to displacement of the platform. In the CM-F relation ship the center of mass lags behind the foot or even starts moving further away from the foot. Figure 6 shows typical traces from a fall, where the CM-S has a large rapid displacement in the direction opposite to that of platform motion, and the CM-F relationship continues to move in the direction of the platform, without reversing direction or plateauing.

Table 3b (right side) presents frequencies of the different performance levels exhibited by the three groups over all three velocities for FT and BT. Note the total number of trial for each velocity are nine for the controls and ten for the stroke group, and the total number of trials for BT and FT for all velocities combined was 27 for the control group and 30 for both the stroke groups. The performance for control subjects was either good or fair for all the trials for both BT and FT. The stroke subjects exhibited all the 4 performance grades. For statistical purposes good and fair performance levels were collapsed into Successful and stumble and fall into unsuccessful. Figure 7 (left panel) presents a summary histogram of percentage of total number of successful trials (all rates of platform motion combined) for to BT and FT.

BACKWARD TRANSLATIONS: The control subjects had successful performance in 100% of trials as compared to 60% for NST and 57% for the PST. A 3 X 2 Chi-Square analysis between CON, NST and PST showed a significant group difference on performance level ($p < 0.001$). Table 5a (left side) shows the individual Chi-test results for CON, NST and PST. There is no clear velocity effect seen on performance levels. Post –hoc analysis between pairs of groups was done using 2 X 2 chi-square test. There was no significant difference on percentage of successful trials between the NST and PST

Table 5- Chi-square results for performance Levels by group, platform velocity and direction of disturbance for Controls, Non paretic stance and Paretic stance (in %).

a. BACKWARD TRANSLATION

	CON	NST	PST	p value
BT	100^{+++}	60^{+++}	57^{+++}	< 0.001

	Performance	CON	NST	p value
B1	*Success*	100	70	0.073
B2	*Success*	100^{++}	40^{++}	0.005
B3	*Success*	100	70	0.073
BT	*Success*	100^{+++}	60^{+++}	< 0.001

	Performance	CON	PST	p value
B1	*Success*	100^{+}	50^{+}	0.013
B2	*Success*	100	70	0.073
B3	*Success*	100^{+}	50^{+}	0.013
BT	*Success*	100^{+++}	57^{+++}	< 0.001

b. FORWARD TRANSALTION

	CON	NST	PST	p value
FT	100^{+++}	37^{+++}	60^{+++}	< 0.001

	Performance	CON	NST	p value
F1	*Success*	100^{++}	40^{++}	0.005
F2	*Success*	100^{++}	40^{++}	0.005
F3	*Success*	100^{++}	30^{++}	0.002
FT	*Success*	100^{+++}	37^{+++}	< 0.001

	Performance	CON	PST	p value
F1	*Success*	100^{+}	60^{+}	0.033
F2	*Success*	100	70	0.073
F3	*Success*	100^{+}	50^{+}	0.013
FT	*Success*	100^{+++}	60^{+++}	< 0.001

Legend:

$^{+}$ P<0.05

$^{++}$ P<0.01

$^{+++}$ P<0.001

group. There was a significant difference between CON and NST ($p< 0.001$) and CON and PST ($p< 0.001$), with the controls having much greater rate of success.

FORWARDS TRANSLATIONS: The control subjects had successful performance in 100% of trials. The NST group had only 37% success rate and the PST group 60%. Similar to BT, The Chi-square test, showed a significant group difference on frequency of successful trials ($p<0.001$). Individual chi-test results between CON and NST and PST are given in table 5b (right side). A 2 X 2 post-hoc analysis revealed significant differences between CON and PST ($p<0.001$) and CON and NST ($p< 0.001$) with the percentage of successful trials being significantly greater in the controls. There was no significant difference on performance between NST and PST ($p= 0.07$). The difference between % successful performances was quite substantial (NST= 37%, PST=60%), as evident by the p value that is very close to the significance level of 0.05. There was no velocity effect on performance levels in the stroke groups. The percentage of fair performance was substantially higher in the controls group at the highest velocity (F3), as most of the subjects took a step at this velocity. The percentage of total falls was higher in FT compared to BT.

4.5. MEANS BY WHICH CORRECTIVE RESPONSE WAS ACHIEVED

4.5.1 Angular kinematics for balance reactions in the sagittal plane

Figure 2, described above, illustrates the typical patterns of angular displacements to forward (FT) and (BT) backward translations from a healthy control during standing with both feet on the ground. (Brunham, 1996). For BT the active recovery is produced by early hip flexion, mean onset 200 ± 10 ms, followed by ankle plantar flexion, and knee extension. For FT a typical pattern is early knee flexion (170 ± 10 ms), followed by hip extension (350 ± 25 ms). Dorsiflexion coincides with knee flexion. In the present study, all control subjects exhibited these typical patterns on the stance leg during BT and FT. As describe in methods section we could not analyze angular displacements about the ankle, thus the pattern was determined from the knee and hip angular displacements and the trunk segment rotation. Onset times for all the corrective angular displacements could not be obtained in this study, as in some cases the corrective displacement was in the same direction as that of the ongoing task, before platform onset. For example while lifting the leg up during the task, there is some knee flexion on the stance side, thus it was difficult to obtain the corrective knee flexion onset after platform motion, as the displacement was in the same direction as the platform.

The typical patterns exhibited in the controls were identified as "normative" movement patterns. Normative is a relative term and in this study refers to the movement patterns observed with consistency for BT and FT in the controls. Any deviations from these normative patterns, in terms of onset, magnitude and direction of angular displacements were called "abnormal".

BACKWARD TRANSLATION: Figures 8-13 presents typical plots of knee and hip angular displacements and trunk segment rotations during different performance levels, to show the various normative and abnormal patterns present in both groups. These figures also show the CM-F and the linear displacement of the foot to give a clear picture of the performance level associated with the respective movement synergies.

All control subjects exhibited the normal pattern of early hip flexion, trunk flexion (forward pitch of the trunk segment). The range of onset of hip flexion and trunk flexion for B1, B2 and B3 combined, respectively were 150-230 ms and 100-150 ms. A variable pattern of knee angular displacement was observed. In some cases a small knee extension of a few degrees was observed, in other trials a small knee flexion of a few degrees was observed, and in many cases no movement (less than 1 degree) was noted at the knee. Fig. 8 shows the knee and hip angular displacements and the trunk segment rotation for a good normative pattern. The ranges of magnitude of peak angular displacement for hip flexion were 3-8 ° for B1, 3-11° for B2, and 4-17 ° for B3. The ranges of peak trunk segment rotation were 2-9° for B1, 3-12° for B2 and 4-13° for B3. The ranges of magnitude of peak angular knee displacement were 0-2° for B1, 0-5° for B2 and 0-7° for B3. One subject in the control group had excessive knee, hip and trunk magnitudes of displacement and was not included in this control range. The movement patterns in the stroke groups were identified based on presence or absence of individual components of the normative patterns. In case all components of the pattern were present, if the peak magnitudes were outside the range for the controls, or if hip flexion onset was delayed (greater than 250 ms) the pattern was classified as abnormal. The following abnormal patterns were identified:

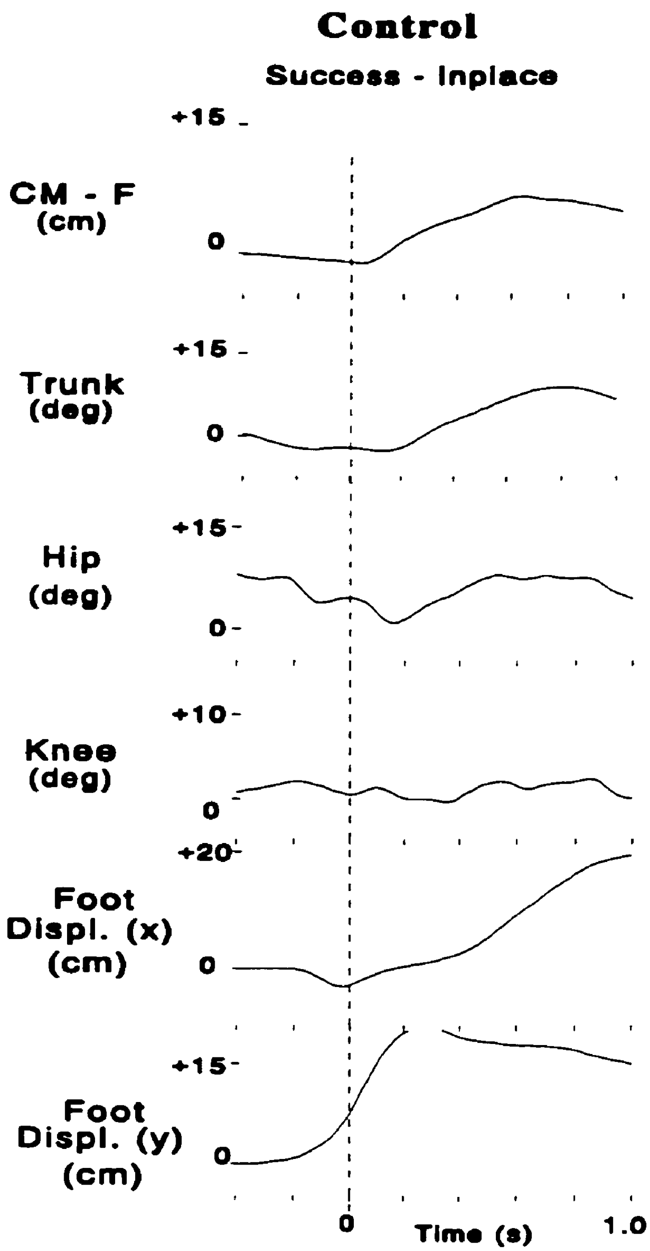


Figure 8: Plots of linear displacements (displ.) of lateral malleolus (foot) marker (x and y coordinates), knee and hip angular displacements, trunk segment rotations and centre of mass relative to foot in sagittal plane during BT for a successful-inplace (good) performance trial of a control subject. The side of the body corresponding to the stance limb is represented. The foot displacement plots are from the swing side of the body. For y-axis zero represents standing still baseline position before stance limb unloading and onset of platform translation. Vertical dashed line at time zero is the onset of platform translation. Positive values for angular displacement represent knee and hip flexion. For the trunk segment positive values represent forward Pitch and negative values backward pitch. For x coordinate of lateral malleolus positive values represent backwards displacement and negative values represent forwards displacement. For y coordinate, upwards displacement represent upward vertical displacement of the foot.

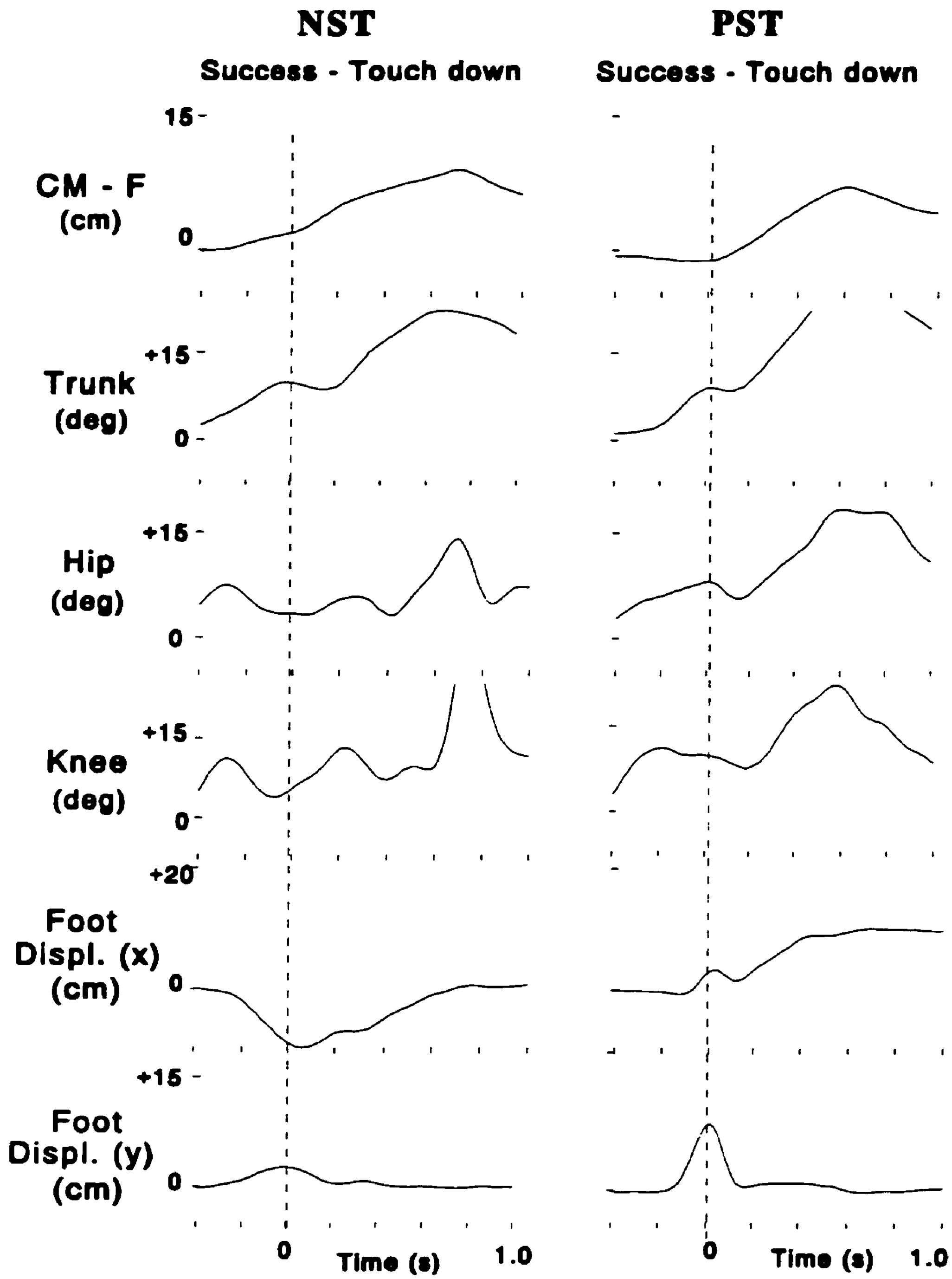


Figure 9: Same as Figure 8, but for a successful- touch down (fair) trial of a RCVA, non-paretic stance and a LCVA, paretic stance during backward translations.

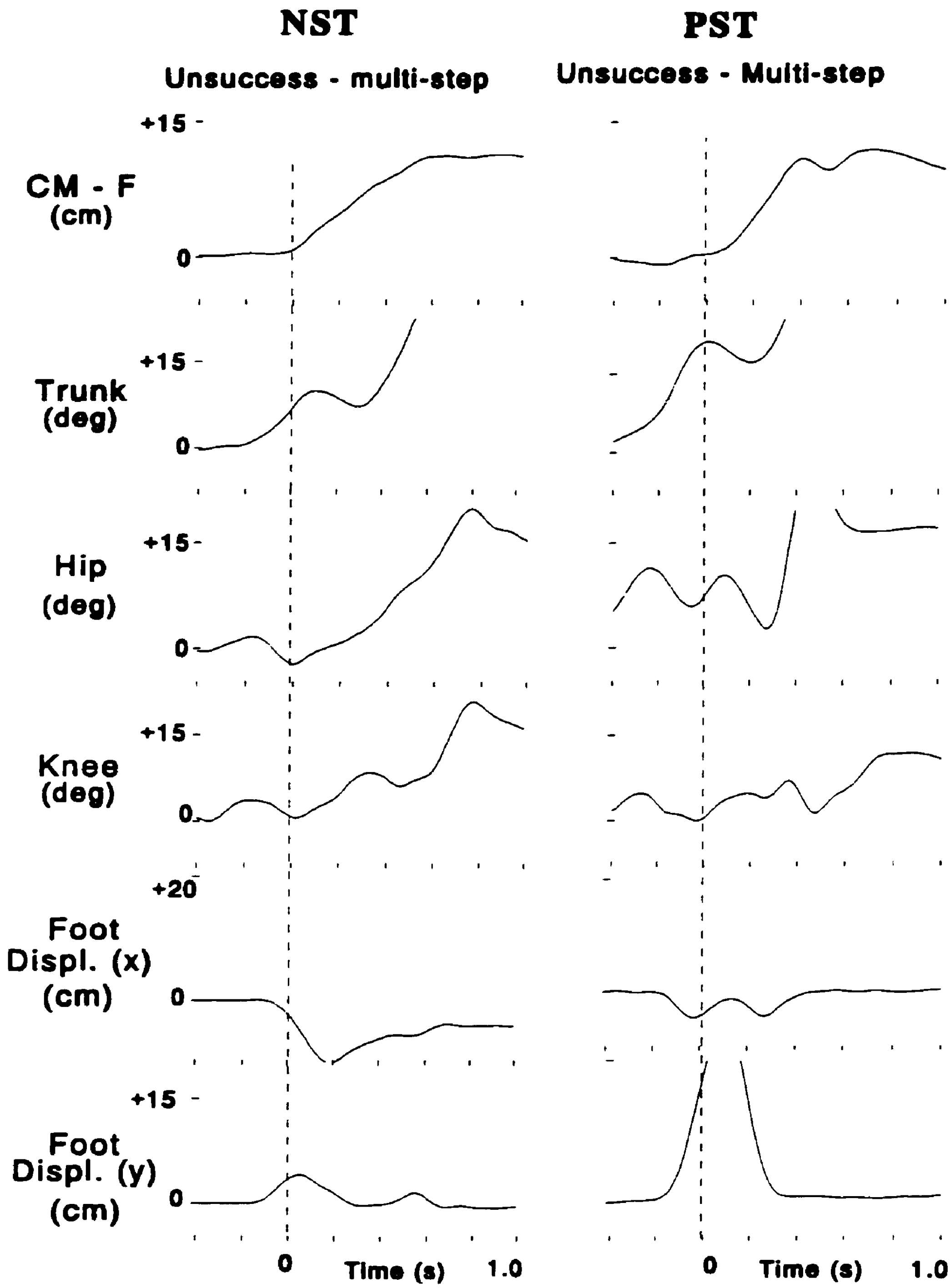


Figure 10: Same as Figure 8, but for an unsuccessful - multi-step (stumble) trial of a LCVA non-paretic stance and a RCVA, paretic stance during backward translations.

1. P1 - This pattern represents excessive hip flexion, and excessive trunk flexion and excessive knee flexion shown in Figure 9 (right panel) and 10 (left panel).
2. P2 - This pattern represents normal hip flexion and trunk segment pitch, but excessive knee flexion.
3. P3 - This pattern represents absent or low magnitude of hip flexion, and/or late hip flexion (>250 ms), with normal or slightly excessive knee and trunk as shown in Figure 9 (left panel).

Fifty three percent (16/30) of the trials in the NST group and 30% (9/30) trials in the PST group showed a normative pattern. The frequency of the three abnormal patterns for NST was 8/30 for P1, 4/30 for P2, and 2/30 for P3. Frequency of abnormal patterns for PST was 9/30 for P1, 8/30 for P2 and 4/30 for P3.

FORWARD TRANSLATION: The normal pattern in the controls for FT was early knee flexion (onset range of 110-150 ms) followed by hip extension (onset range of 225-300 ms). The trunk segment rotated backwards which preceded the hip extension (onset range of 110- 250 ms). Figure. 11 (left panel) show the knee and hip angular displacements and the trunk segment rotation for a normative pattern with a Good performance grade for a control subject. The ranges of magnitude of peak angular displacement for knee flexion were 2-5° for F1, 2-5° for F2 and 3-7° for F3. The ranges of magnitude of hip extension were 2-7° for F1, 2-7° for F2 and 3-18° for F3. The ranges of peak trunk segment rotation were 3-7° for F1, 2-6° for F2 and 6-16° for F3.

As described above for BT, the abnormal patterns in the stroke groups were identified based on presence or absence of individual components of the normative patterns. In case all components of the pattern were present, if the peak magnitudes were outside the range for the controls for knee flexion and hip extension, or the onsets for knee flexion and hip extension were delayed (> than 150 ms and 250 ms respectively) the pattern was classified as abnormal. The following abnormal patterns were identified:

1. P1 - This pattern represents normal or excessive knee flexion and trunk segment rotation, but absent or delayed (>300 ms) and inadequate hip extension. If hip flexion was present instead of hip extension it was included in this pattern. Figure 12 (left panel) represents this pattern.

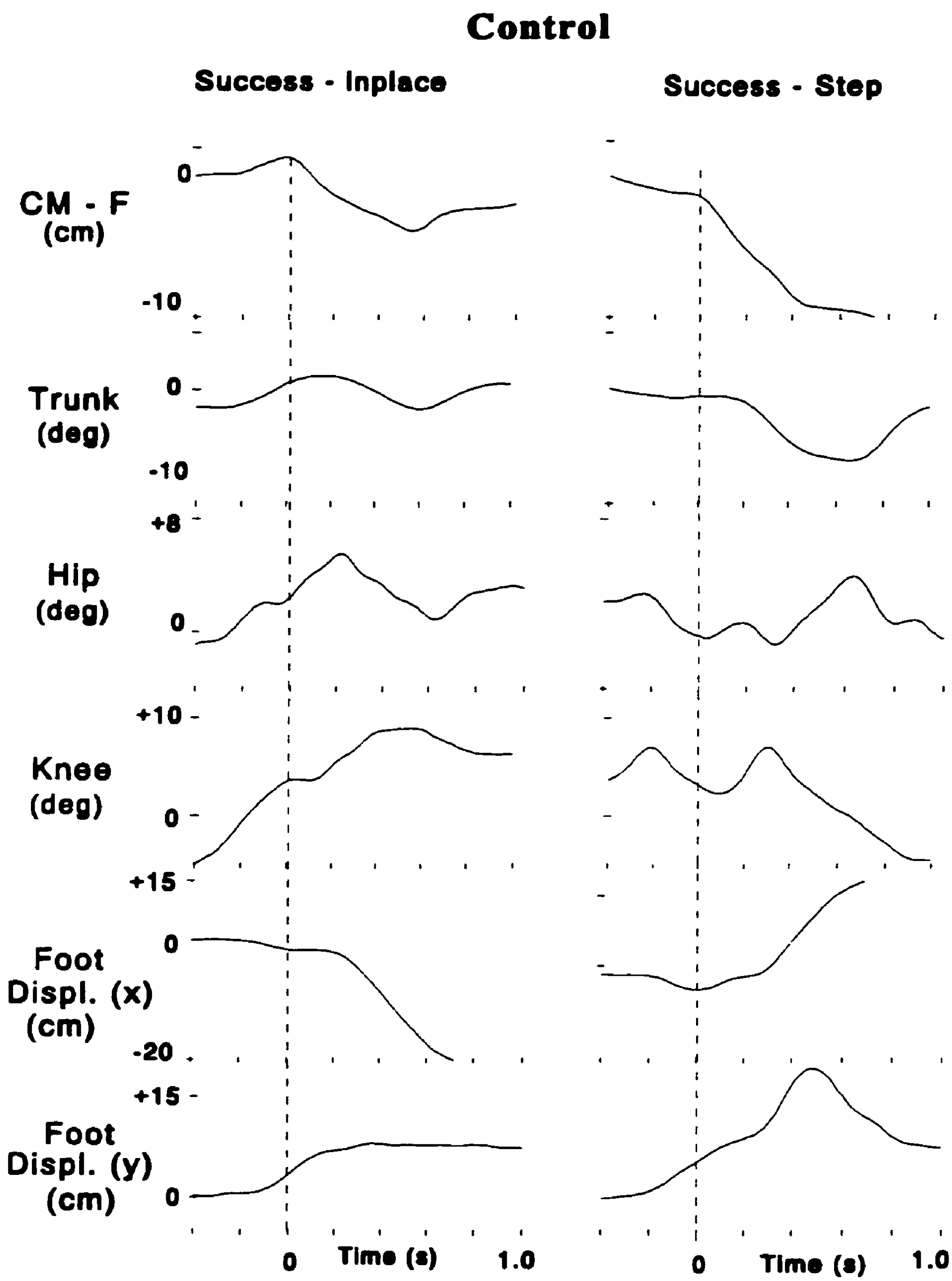


Figure 11: Same as Figure 8, but for a successful -inplace (good) and successful - step (fair) trial of a control subject during forward translations.

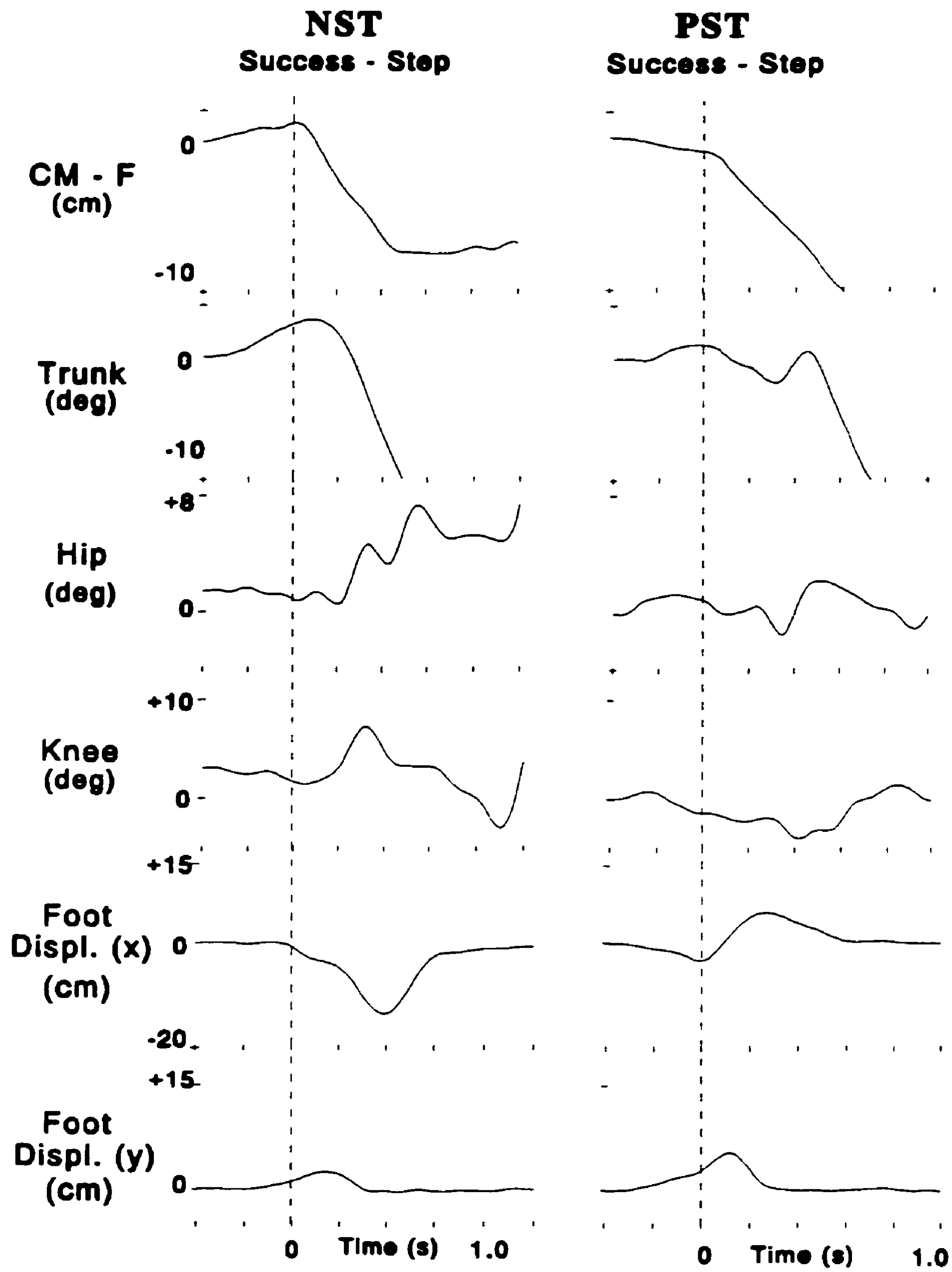


Figure 12: Same as Figure 8, but for a successful - step (fair) trial of a RCVA, non-paretic stance and a RCVA, paretic stance during forward translations.

2. P2 - This pattern represents both absent knee flexion and absent hip extension. Figure 12 (right panel) represents this pattern.
3. P3 - This pattern represents absent knee flexion or inadequate magnitude with delayed (> 150 ms) knee flexion, shown in Figure 13 (right panel)

Note in majority of cases for both NPT and PST in the Stroke subjects the trunk segment was pitched backwards even in absence of hip extension and had magnitudes within normal control ranges. 20% (6/30) of the trials in the NST group and 26% (8/30) trials in the PST group showed a normative pattern. The frequency of the three patterns in the NST group was 15/30 for P1, 7/30 for P2 and 2/30 for P3, and frequency of the patterns in the PST group was 14/30 for P1, 2/30 for P2 and 6/30 for P3.

4.5.2 Association of movement patterns with Performance levels

An association of successful performance to normative movement patterns, and unsuccessful performance to abnormal movement patterns was done for each trial and is presented in Table 6. All the trials with successful performance were divided into two groups one with an inplace or touchdown strategy and second one with a primary stepping strategy. Separate associations were done for these two different strategies. Table 6a (left panel) represents the total number of successful trials (SP) with an inplace strategy and the column labeled "Assoc." represents the number of these SP trials that had a normative movement pattern. Conversely, the right panel represents the total number of unsuccessful trials (UP) with an inplace strategy and the number of these trials that had an abnormal movement pattern (assoc). Table 6b (left panel) is similar to 6a but for successful performance having a primary stepping strategy and Table 6b (right panel) for unsuccessful performance with a primary stepping strategy.

BACKWARD TRANSLATION: As presented in Table 6a, there was a high association of SP to normative patterns only in NST (90% (9/10)). There was a high association of UP to abnormal patterns in PST (100% (4/4)), and a moderate association in NST (71% (5/7)). There was a low association of SP to normative patterns in PST (43% (3/7)). Table 3b shows the associations for SP and UP for a primary stepping strategy. The association of SP to normal movement patterns was 50% both for the NST (4/8) and the PST (5/10)

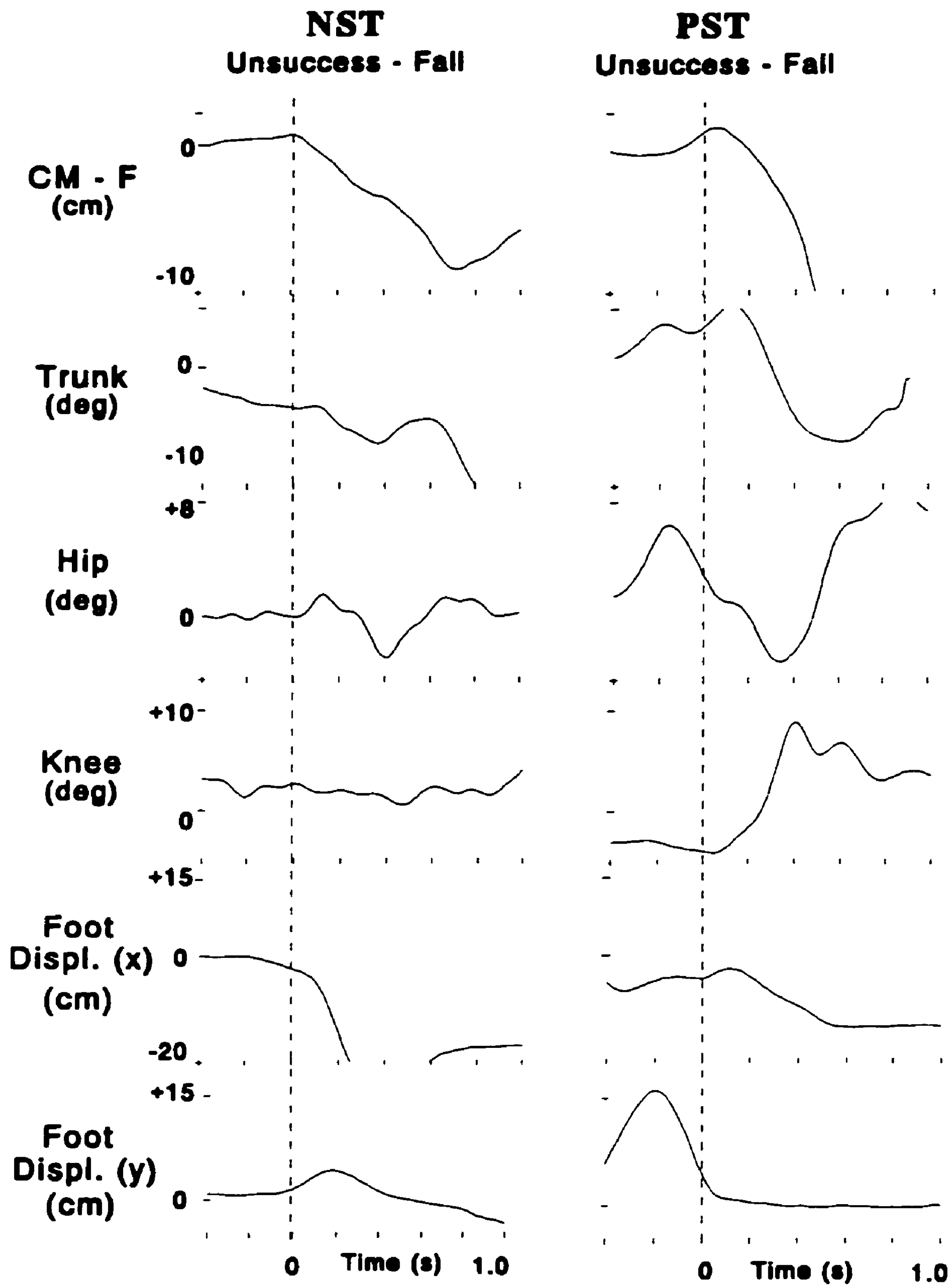


Figure 13: Same as Figure 8, but for an unsuccessful - fall trial of a LCVA the non-paretic and a LCVA, paretic stance during forward translations.

Table 6 - Frequency of successful performance trials (SP) associated with normative movement patterns and unsuccessful performance trials (UP) associated with abnormal patterns.

a. INPLACE STRATEGY

Association of normative movement patterns with successful performance					*Association of abnormal movement patterns with unsuccessful performance*				
	NST		PST			NST		PST	
Velocity	SP	Assoc.	SP	Assoc.	Velocity	UP	Assoc.	UP	Assoc.
B1	6	5	0	0	B1	2	1	2	2
B2	2	2	5	2	B2	3	3	0	0
B3	2	2	2	1	B3	2	1	2	2
	10	9	7	3		7	5	4	4
F1	2	0	4	2	F1	5	4	4	4
F2	0	0	4	1	F2	3	3	2	2
F3	0	0	1	1	F3	6	6	3	3
	2	0	9	4		14	13	9	9

b. STEPPING STRATEGY

Association of normative movement patterns with successful performance					*Association of abnormal movement patterns with unsuccessful performance*				
	NS		PST			NST		PST	
Velocity	SP	Assoc.	SP	Assoc.	Velocity	UP	Assoc.	UP	Assoc.
B1	1	1	5	2	B1	1	1	3	3
B2	2	1	2	1	B2	3	2	3	2
B3	5	2	3	2	B3	1	1	3	3
	8	4	10	5		5	4	9	8
F1	2	1	2	1	F1	1	1	0	0
F2	4	3	3	1	F2	3	3	1	1
F3	3	1	3	2	F3	1	1	2	2
	9	5	8	4		5	5	3	3

n = number of trials for each velocity = 10
N = number of total trials for each group (NST/PST) =30

group. There was a high association of UP to abnormal patterns in NST (80% (4/5)) and PST (89% (8/9)).

FORWARD TRANSLATION: As presented in Table 6aThere was a low association of SP to normative patterns in NST and PST. There were only 2 successful inplace trials in NST and none of them had a normal pattern. There were 4/9 trials with normative patterns (44% association). There was a high association of UP to abnormal pattern in both the NST and PST. The association in NST was 93% (13/14) and in PST was 100%. Table 6b shows the associations for SP and UP for a primary stepping strategy. The associations of SP to normal movement patterns were very similar in both NST and PST, 55% in NST (5/9) and 50% (4/8) in PST group. Association of UP to abnormal patterns for NST was 100% for both, the NST (5/5) and PST (3/3) groups. The associations for stepping strategy were very similar during both BT and FT.

4.5.3 Scaling of corrective response with increasing velocity

A repeated measures ANOVA was used to compare the main effects of two factors, Group (CON, NPT, and PST) and rate of platform motion (V1, V2, and V3) on the magnitude of angular displacements and trunk segment rotation during BT. Although knee flexion is not an active component of the normal pattern, we were interested to look for presence of excessive uncontrolled knee flexion between groups. There were sufficient numbers of trials with a corrective response in all the groups during BT to permit an analysis, i.e. majority of trials had knee, hip and trunk flexion. For FT, there were large numbers of trials with absent hip knee flexion and hip extension for subjects in NST and PST over all velocities. Thus, there were not enough samples to permit an analysis of group difference or the effect of rate of platform motion to look for scaling of corrective responses to FT.

Results of ANOVA of the magnitude of the angular displacements at the knee and hip, and the trunk segment rotation during BT revealed group and velocity differences as shown in figure 14. The magnitude of knee angular displacement was significantly different between the three groups (G) ($p< 0.001$) and between trials or rates of platform motion (T) ($p< 0.05$). There was a significant G* T interaction present ($p<0.05$). As seen from Figure 15, there was substantially more knee flexion for PST compared to the CON

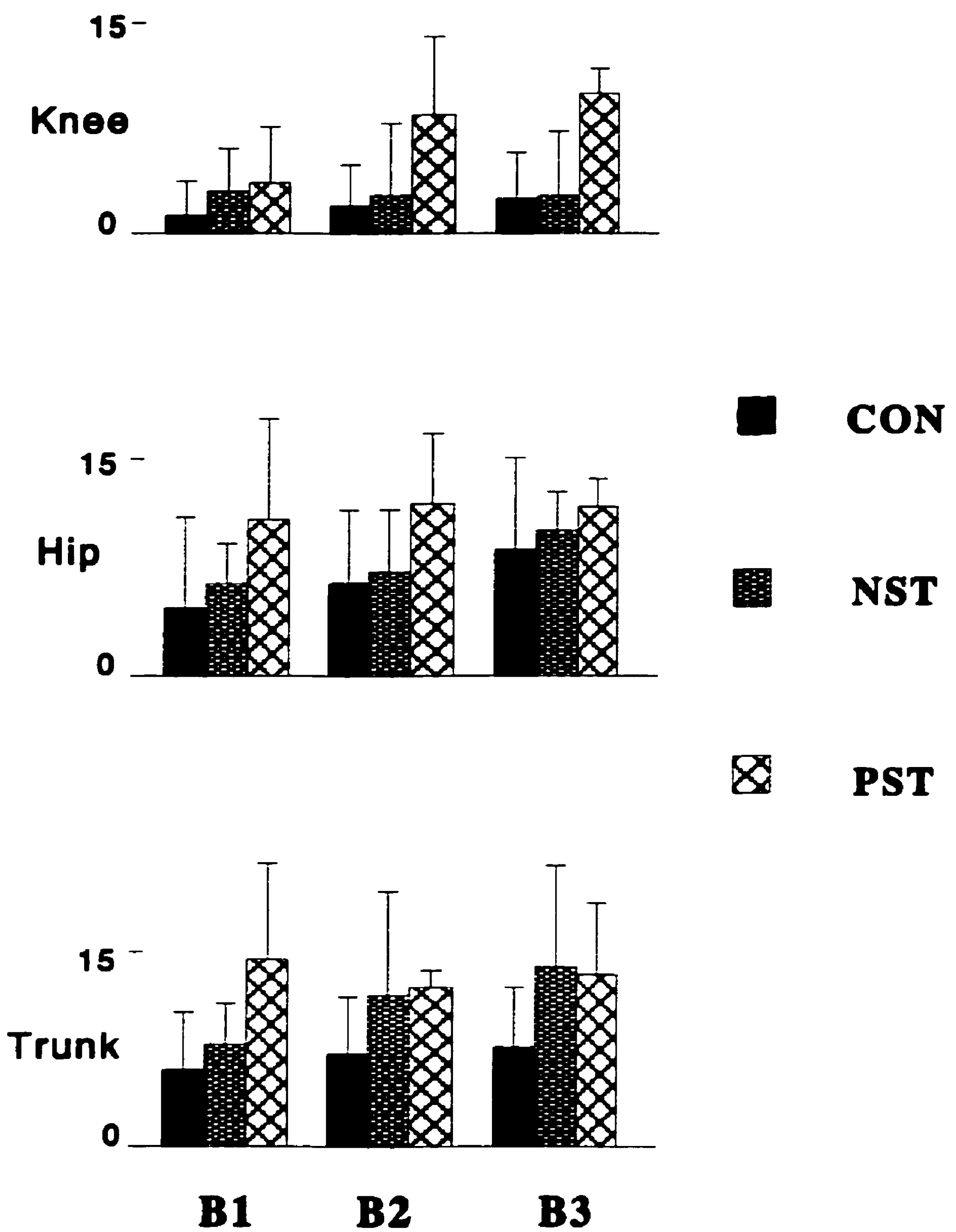


Figure 14: Group means and standard deviations of magnitude of peak angular displacements (stance side) and trunk segment rotation for backward translations for controls, non-paretic stance and paretic stance.

and the NST. There was no statistical difference in knee flexion between CON and NPT. Also the effect of rate of platform motion on difference in magnitude is much greater for PST compared to CON and NST.

There was no between group effect on hip angular displacement ($p = 0.15$), however there was a significant trials effect or rate of platform motion on mean hip magnitude ($p < 0.001$). There was no G*T interaction, thus the magnitude of hip angular displacement increased with increasing velocity/acceleration of platform motion for all groups. This indicates that a scaling of the response for hip flexion was present in all the three groups with magnitude of disturbance as can be seen in Figure 15.

There was a significant G effect on the magnitude of trunk rotation ($p<0.01$), however there was no within group T effect ($p =0.26$). There was no G*T interaction present ($p = 0.4$). Figure 15 shows that the magnitudes of trunk rotation are substantially more for the NST and PST compared to CON and that the magnitudes are similar for individual groups over B1, B2 and B3.

4.5.4 Angular kinematics for balance reaction in the frontal plane

Analyses of angular kinematics in the frontal plane was performed to rule out uncontrolled excessive medio-lateral movements in this plane, which could have been responsible for a potential effect on performance. The hip angular displacement and the trunk segment rotation were analyzed for BT and FT. A corrective response was classified to be present if the pattern of the response was different than that present in the trials without perturbation (T0) within the first 500 ms after platform onset, and/or the magnitude of the angular displacement was at least double of that present in T0. Figure 15 shows typical traces taken from T0, BT and FT trials for CON, NST and PST. It can be noted that while performing the leg lift task in the absence of a platform translation, the trunk segment displaces towards the stance side, and the hip goes into adduction. This was consistent for all subjects and for the vast majority of the trials. Figures 15 a, b and c represent traces of a typical corrective response seen during BT and FT in CON, NST and PST for majority of the trials. It can be seen from Figures 15 a, b and c that responses during BT and FT very closely resembled those from T0 in all the three groups. Figure 15d shows traces from a stroke subject (PST) exhibiting excessive trunk rotation

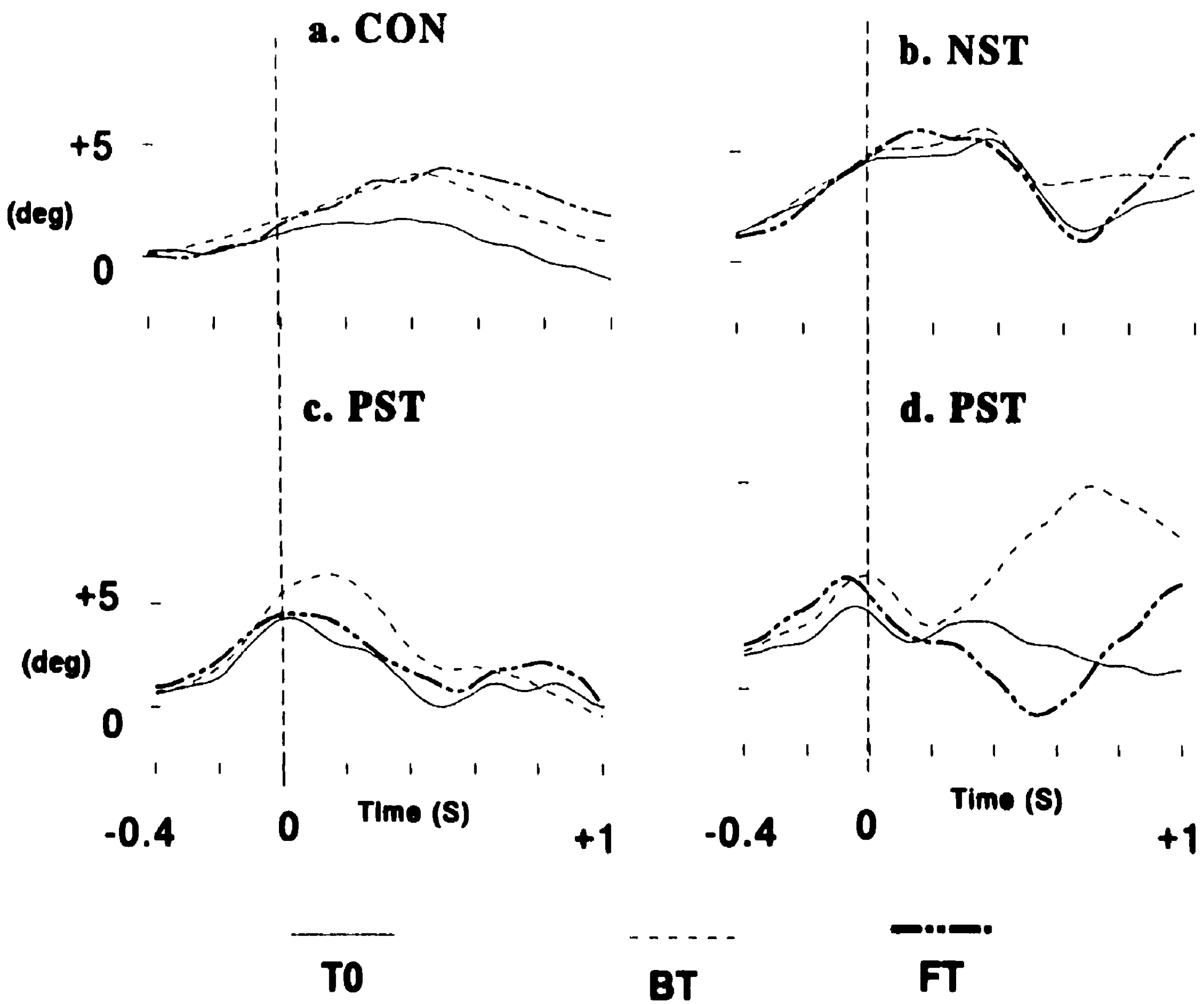


Figure 15: Plots of trunk segment rotation in the frontal plane to BT, FT and no translation (T0) from a control subject, and stroke subjects during non-paretic stance and paretic stance. The side of the body corresponding to the stance limb is represented. For y-axis zero represents standing still baseline position before stance limb unloading and onset of platform translation. Vertical dashed line at time zero is the onset of platform translation. Upward direction is displacement towards stance leg and downward direction is displacement away from stance leg. Fig. 15a, b, and c, show that there was no major correction taking place in trunk segment rotation in the frontal plane during BT and FT compared to T0. Fig. 15d shows a case where a major corrective change was observed in trunk segment rotation during BT and FT compared to T0.

towards the stance side during BT and away from the stance side during FT. Table 7a presents the number of trials with a corrective response and the direction of response for trunk segment rotation during BT and FT for CON, NST and PST. Eleven percent of the trials during BT and 51% trials during FT in CON showed a corrective response in the trunk segment. The corrective response in the control subjects was direction dependent; the trunk segment rotation was away from the stance side during BT and towards the stance side during FT. The responses in both the stroke groups were very variable and inconsistent. During BT, 33%(10/30) of the trials in NST and 47% (14/30) in the PST group showed presence of excess trunk segment rotation. These frequencies were slightly greater than that seen in controls (11%). During FT, 33% (10/30) of trials in NST and 23% (7/30) in PST showed presence of excess trunk segment rotation, which was slightly less than that seen in the controls (51%). The direction of the response was not consistent in the strokes groups and was not direction dependent as seen in the controls. However in majority of trials there was a trend for rotation away from the stance side in both NST and PST for BT and FT as can be seen from table 7a.

Table 7b presents the number of trials with a corrective response and the direction of response at the hip joint during BT and FT for CON, NST and PST. As seen for Trunk segment rotation, the corrective hip angular displacement was also direction dependent in the controls at least at V1 and V2. The number of trials with a corrective response to BT were similar in all groups: 62% (17/27) of trials in CON, 70% (21/30) of trials for NST and 50% (15/30) of trials for PST. In the controls the hip went into abduction after onset of platform motion for all the trials having a response at B1 and B2. At B3 for 44% (4/9) of the trials the hip was abducted and 56% (5/9) of the trials the hip was adducted. Similar to the controls in NST for majority of the trials (70%(16/21)) at all the velocities the hip went into abduction. In PST for 60% (9/15) of the trials the response was abduction and for 40% (6/15) of the trials it was adduction.

Like BT the number of trials with a corrective response to FT were similar in all groups: 48% (13/27) of trials in CON, 53% (16/30) of trials in NST and 43% (13/30) of trials in PST. For the controls, the response at F1 and F2 was of hip adduction. At F3 majority of the trials with a corrective response showed hip adduction (5/7). Similar to BT, in the controls the percentage of trials with a response was greater at the highest

Table 7a : Trunk segment displacement direction and frequency in the frontal plane for Control, Non-paretic stance and paretic stance.

	CON		NST		PST	
Velocity	Direction	Freq	Direction	Freq	Direction	Freq
B1	AS	1	AS	1	AS	2
B2	AS	1	AS	2	AS	2
			TS	1	TS	1
B3	AS	1	AS	5	AS	1
			TS	1	TS	1
F1	TS	4	AS	2	-	-
			TS	1		
F2	TS	3	AS	3	AS	2
			TS	1	TS	1
F3	TS	7	AS	2	AS	2
			TS	1	TS	2

AS = Rotation of trunk away from stance side
TS = Rotation of trunk towards stance side

Table 7b :Hip angular displacement, direction and frequency (Freq) in the frontal plane for Control(CON), Non-paretic (NST) and paretic (PST) stance

	CON		NST		PST	
Velocity	Direction	Freq	Direction	Freq	Direction	Freq
B1	AB	4	AB	5	AB	3
			AD	1	AD	1
B2	AB	4	AB	6	AB	3
			AD	2	AD	2
B3	AB	4	AB	5	AB	3
	AD	5	AD	2	AD	3
F1	AD	3	AB	3	AB	1
			AD	1	AD	4
F2	AD	3	AB	4	AB	1
			AD	3	AD	3
F3	AD	2	AB	2	AB	1
	AD	5	AD	3	AD	3

AD = Adduction
AB = abduction
n= number of trials for each velocity = 10

velocity. Similar to the controls, in PST majority of the trials showed a hip adduction response (77%(10/13)). In the NST 56 % (9/16) of the trials had a hip abduction response and 44% (7/16) had and adduction response. There was no substantial increase in frequency of response seen with velocity in both NST and PST for BT and FT.

4.6. EMG – FREQUENCY AND ONSET LATENCIES

The EMG recordings were analyzed for presence or absence of responses, within a 300 ms time period. When a response was present the onset latency was identified. EMG responses were present in all the three groups both on the stance as well as swing side. Fig. 16 presents typical difference EMG records from a control subject and a RCVA subject during BT. Figure 17 shows typical EMG records from a control subject and a LCVA subject during FT. Table 8 presents the frequencies of occurrence and mean onset latencies of hip abductors, hip adductors, rectus femoris and hamstrings from the stance side for CON, NPT and PST during BT. Table 9 represents the same during FT. Tables 10 and 11 show the frequencies and onset latencies of muscles from the swing side during BT and FT respectively.

<u>**STANCE LIMB**</u>: The frequency of occurrence by muscle was variable between groups during BT as well as FT. There was no clear trend seen in frequency of activation of muscle groups within group or between groups over all three velocities. For example, as shown in table 9, during FT, in RF the frequency of activation was greater in NST and PST compared to controls, however at F2 the frequency in controls was greater than in NST and PST. Also at F1 frequency of NST was greater than PST, however at F2, PST had a higher frequency. At F3 all the three groups had similar activation frequencies in RF.

The onset latencies in PST both for BT and FT, on an average were longer than the controls approximately at least by 100 ms for all the muscles except abductor at B2 and HA and RF at F3. For both BT and FT, the onset latencies of some muscles in NST were longer than the controls but shorter than in PST. For example in HA at B1 and B2, and in RF at F1 and F2. Some muscles in NST had longer onset latencies than the PST, for example as seen in RF at B1 and B2, and in AB, HA and RF at F3.

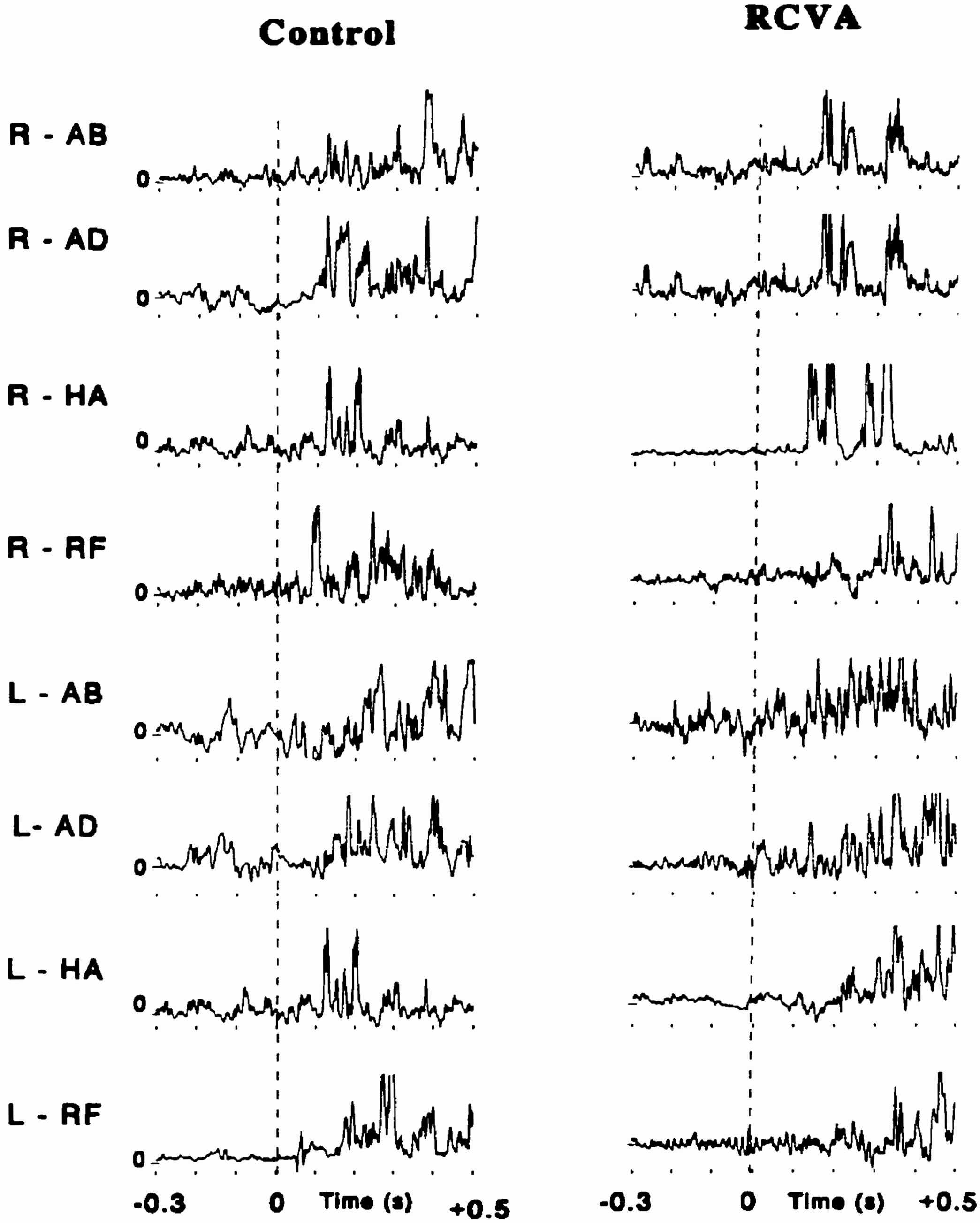


Figure 16: Typical plots representing the difference EMG waveforms of hip abductors (AB), adductors (AD), rectus femoris (RF) and hamstrings (HA) from both sides of the body from a control subject (left panel) and a RCVA (right panel) subject during BT while lifting the left (L) leg. Vertical dashed line at time zero is the onset of platform translation. The y- axes represents arbitrary units in mV.

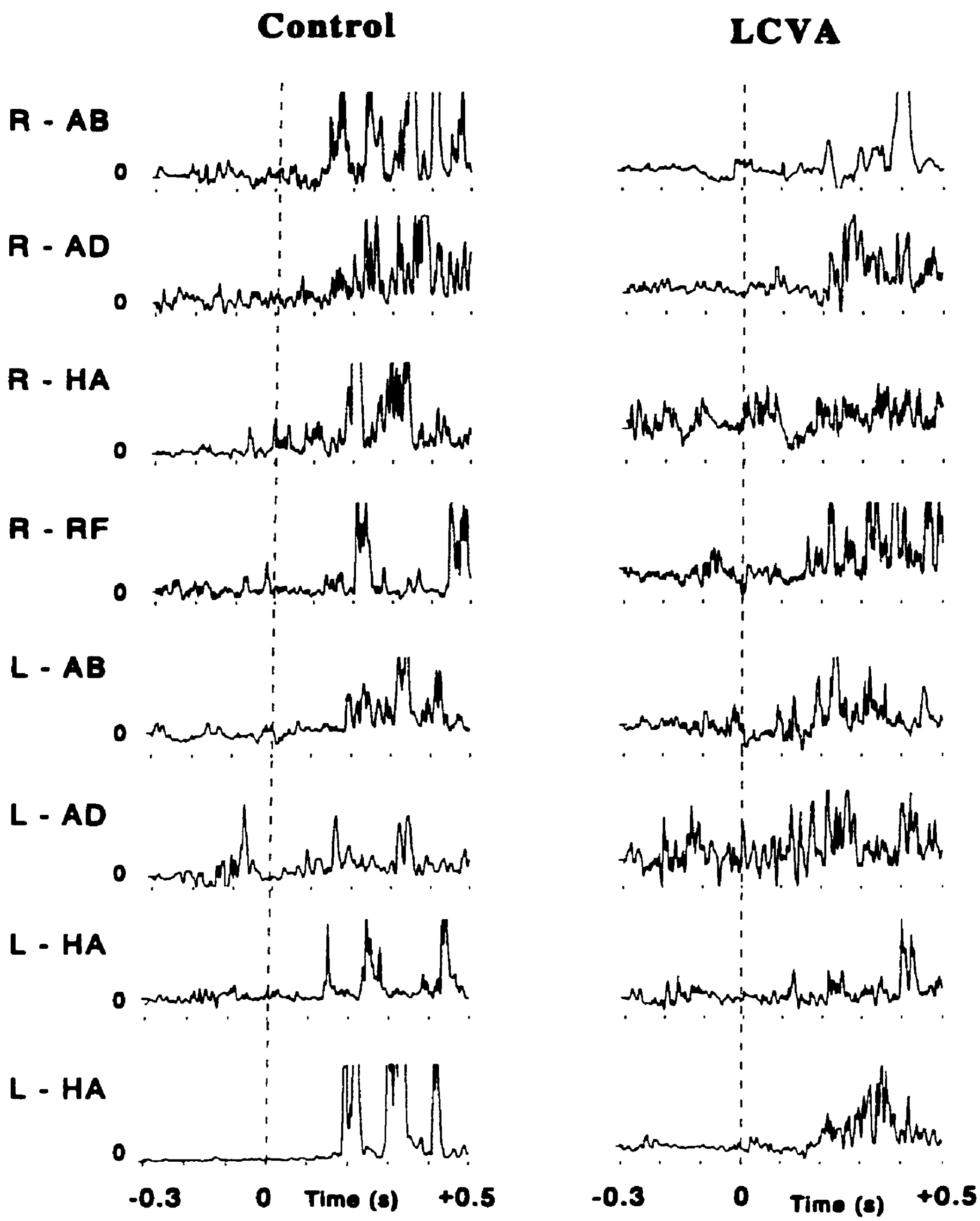


Figure 17: Typical plots representing the difference EMG waveforms of hip abductors (AB), adductors (AD), rectus femoris (RF) and hamstrings (HA) from both sides of the body from a control subject (left panel) and a LCVA (right panel) subject during FT while lifting the left (L) leg. Vertical dashed line at time zero is the onset of platform translation.

Table 8 Mean onset latencies, standard deviations (SD) and frequencies of Control, Paretic and Non paretic stance side muscles during BT.

	CON			NST			PST		
	Mean	S.D	Frequency	Mean	S.D	Frequency	Mean	S.D	Frequency
B1									
AB	0.151	0.063	63	0.309	0.193	62	0.300	0.157	50
AD	0.103	0.044	50	0.262	0.154	62	0.278	0.193	75
HA	0.125	0.028	92	0.160	0.058	62	0.228	0.108	75
RF	0.112	0.049	54	0.363	0.163	38	0.242	0.130	58
B2									
AB	0.182	0.086	63	0.260	0.180	52	0.222	0.151	71
AD	0.085	0.019	58	0.215	0.091	81	0.213	0.109	67
HA	0.102	0.019	100	0.161	0.088	86	0.200	0.096	57
RF	0.107	0.043	50	0.303	0.169	48	0.277	0.172	76
B3									
AB	0.187	0.076	65	0.172	0.125	85	0.302	0.010	54
AD	0.097	0.033	61	0.160	0.060	85	0.270	0.167	85
HA	0.111	0.039	96	0.122	0.021	100	0.208	0.085	77
RF	0.108	0.037	87	0.203	0.122	85	0.292	0.123	62

Table 9- Mean onset latencies, standard deviations (SD) and frequencies of Control, Paretic and Non paretic stance side muscles during FT.

	CON			NST			PST		
	Mean	S.D	Frequency	Mean	S.D	Frequency	Mean	S.D	Frequency
F1									
AB	0.123	0.051	63	0.200	0.090	62	0.250	0.117	50
AD	0.122	0.044	50	0.220	0.130	92	0.263	0.070	75
HA	0.109	0.036	83	0.260	0.170	77	0.284	0.175	50
RF	0.140	0.073	54	0.176	0.090	92	0.291	0.227	67
F2									
AB	0.109	0.040	83	0.284	0.159	68	0.296	0.093	52
AD	0.130	0.070	75	0.248	0.113	95	0.244	0.125	86
HA	0.100	0.020	92	0.195	0.108	74	0.225	0.112	52
RF	0.130	0.060	71	0.170	0.060	58	0.184	0.088	67
F3									
AB	0.153	0.056	92	0.312	0.112	62	0.283	0.099	50
AD	0.147	0.041	54	0.228	0.125	92	0.260	0.074	92
HA	0.129	0.049	92	0.226	0.125	92	0.184	0.140	92
RF	0.145	0.076	92	0.230	0.140	92	0.174	0.063	92

Table 10- Mean onset latencies, standard deviations (SD) and frequencies of Control, Paretic and Non paretic swing side muscles during BT.

	CON			NSW			PSW		
	Mean	S.D	Frequency	Mean	S.D	Frequency	Mean	S.D	Frequency
B1									
AB	0.191	0.175	25	0.362	0.225	50	0.221	0.110	54
AD	0.100	0.028	46	0.390	0.160	67	0.179	0.112	77
HA	0.127	0.049	54	0.233	0.120	67	0.228	0.095	77
RF	0.148	0.110	38	0.200	0.120	67	0.244	0.076	54
B2									
AB	0.105	0.030	33	0.247	0.097	43	0.310	0.104	76
AD	0.080	0.038	42	0.227	0.062	62	0.260	0.096	90
HA	0.125	0.030	50	0.221	0.080	62	0.240	0.114	90
RF	0.180	0.170	29	0.283	0.140	52	0.286	0.130	71
B3									
AB	0.134	0.054	30	0.248	0.133	69	0.279	0.111	77
AD	0.118	0.073	35	0.267	0.142	69	0.210	0.112	92
HA	0.148	0.056	61	0.205	0.050	92	0.208	0.088	100
RF	0.164	0.086	57	0.240	0.130	69	0.214	0.073	92

Table 11- Mean onset latencies, standard deviations (SD) and frequencies of Control, Paretic and Non paretic swing side muscles during FT.

	CON			NST			PST		
	Mean	S.D	Frequency	Mean	S.D	Frequency	Mean	S.D	Frequency
F1									
RAB	0.150	0.090	29	0.202	0.098	62	0.268	0.124	50
RAD	0.130	0.060	25	0.230	0.095	85	0.188	0.079	67
RHA	0.120	0.020	46	0.221	0.126	77	0.219	0.108	50
RRF	0.120	0.045	54	0.242	0.128	92	0.207	0.075	58
F2									
RAB	0.153	0.060	42	0.270	0.117	68	0.206	0.102	57
RAD	0.190	0.044	50	0.217	0.101	95	0.250	0.160	67
RHA	0.190	0.086	63	0.240	0.138	89	0.187	0.110	62
RRF	0.124	0.036	71	0.233	0.131	74	0.214	0.083	90
F3									
RAB	0.164	0.054	79	0.292	0.078	77	0.209	0.084	67
RAD	0.144	0.050	79	0.194	0.067	85	0.195	0.114	75
RHA	0.156	0.053	88	0.233	0.090	85	0.172	0.082	75
RRF	0.147	0.057	83	0.264	0.110	77	0.225	0.089	83

There were few cases where sample sizes were large enough in all the three groups to permit statistical analysis on onset latencies. A one-way analysis of variance (ANOVA) was only done on those muscles with a frequency of occurrence greater than 75% (response present in > seven out of ten trials). Since the sample size of the present study is small and the 75% value was arbitrarily chosen, the results must be interpreted with caution. A one-way ANOVA was performed to look for significant difference in onset latencies between the groups for the HA at B3. The results indicated group difference in onset latencies ($p<0.001$). Bonferroni's Post-hoc analysis revealed a significant difference between CON and PST and NST and PST, but no difference between CON and NST. The onset latencies in both NST and PST were much longer than the controls as can be seen from the group means presented in table 8.

In response to FT, the onset latencies for majority of the muscles in both NST and PST groups were much longer than the CON group. In general, at F3 the frequency of most of the muscles was greater than 90%. A one-way ANOVA by groups was done to look for difference in onset latencies in AD (F2), and HA and RF (F3). The results are as follows:

1. AD (F2): There was a significant group difference in AD (F2) muscle ($p<0.01$). Post hoc-analysis revealed significantly longer latencies in NST ($p< 0.01$) and PST ($p< 0.01$) compared to the controls. There was no significant difference in onset latencies between NST and PST.
2. HA (F3): A significant difference was found in onset latencies between groups ($p<0.05$). Post-hoc analysis revealed a significant difference between CON and NST ($p<0.05$) with CON having longer onset latencies as can be seen from the means presented in table 9. No significant difference was found between CON and PST and NST and PST.
3. RF (F3): A significant difference was found between the three groups ($p<0.05$), with PST having longer latencies than CON ($p<0.05$) and no difference between CON and PST, and NST and PST.

SWING LIMB: For the swing side muscles, the frequency of response tended to be greater in both NST and PST compared to the controls for BT as well as FT, except at F3 where frequency of response was similar for CON, NST and PST for all muscle groups.

For example the frequency range during BT for AB muscle group was 25-30 % for CON, 43-69 % for NST and 50-67 % for PST. The onset latencies of the swing side muscles were much longer in both NST and PST compared to controls for BT as well as FT. For example, the onset latencies of all the muscle groups during BT were in the range of105 – 191 ms for CON, 247- 362 ms for NST and 221-310 ms for PST. An ANOVA to examine difference in onset latencies similar to that done for stance side muscles was not done on the swing side muscles. The reason for this being, in this study the onset latencies of the muscles were grouped together irrespective of whether the swing leg took a step or maintained and inplace response. Thus different muscle groups could be activated potentially for these two very different types of responses.

4.7. CORRELATION BETWEEN OVERALL PERFORMANCE AND WALKING INDEX AND GAIT SPEED

Table 12a presents the gait velocity in ascending order for all the subjects and the corresponding walking index and performance scores for BT and FT. Table 12b displays the correlation coefficients (r value) for each test. For BT the Spearman r value for correlation between performance and gait speed was 0.13 and for performance with walking index was 0.33. For FT the r value for correlation between performance and gait speed was 0.68 and 0.55 for walking Index. The r values for correlation with gait speed and walking index were similar to each other. The correlations were much lower for BT compared to FT for both walking index and gait speed.

Table 12a - Gait speed (3 m), Walking Index, and performance scores for backwards (BT) and forwards translations (FT) for each subject.

SUBJECTS	Gait Speed GS (m/s)	Walking Index Score WI	Performance Score BT	Performance Score FT
5	0.261	14	16	7
3	0.286	17	18	16
10	0.343	19	14	14
6	0.375	17	14	8
7	0.493	21	12	12
9	0.514	27	19	13
8	0.569	23	16	17
2	0.750	21	15	17
4	0.915	24	14	15
1	1.053	27	19	18

Table 12b - r values for correlation of gait speed and walking index with performance during backward (BT) and forward translations(FT)

Spearman Rho rank order correlation

	BT	FT
GS	0.13	0.681
WI	0.33	0.549

5. DISCUSSION

The purpose of this study was to identify the limits of stability in chronic stroke individuals by evaluating corrective balance responses to forward and backward disturbances during a leg-lifting task, and comparing these responses to those present in healthy individuals.

5.1 MAIN FINDINGS

1.The initial leg-lifting task before onset of platform motion was performed similarly between the three groups i.e. controls (CON) and the stroke subjects both during non-paretic stance (NST) and paretic stance (PST). There was no significant group difference in peak magnitudes of centre of foot pressure displacements (AX and AY) during the task performance before onset of platform motion.

2. The control group mainly exhibited an inplace corrective strategy (96% to BT and 74% to FT) to recover balance. Steps were typically observed at higher rates of platform translation in particular F3 (77%). In contrast, a stepping corrective strategy was observed in one half of the trials for stroke subjects (NST + PST). There was no significant difference in strategy selection whether the non-paretic leg was in stance (NST) or the paretic leg was in stance (PST). A strategy switch was evident in the control subjects from inplace at F1 to stepping at F3. There was no clear strategy switch evident in the stroke subjects from V1 throughV3 for BT as well as FT, in general stepping was seen right from V1.

3. The percentage of successful trials in controls (100%) was significantly greater than NST and PST. Performance levels of success in the stroke individuals for BT during non-paretic side stance and paretic side stance were very similar and were 60% and 57% respectively, compared to 100% in the controls. For FT, percentage of success for paretic side stance (60%) was greater than non-paretic stance (37%), however this difference was

not statistically significant (p=0.07). Falls occurred only in the stroke subjects, in particular at higher velocities of FT.

4. Stereotypical multi-segmental normative movement patterns to BT and FT, were present on the stance side in the control subjects. Three abnormal patterns were identified in stroke individuals during BT. The most common abnormal pattern in both NST and PST was excessive hip, knee and trunk segment flexion. Three abnormal patterns were also identified in stroke individuals during FT. The most common pattern in both NST and PST was abnormal control of knee flexion and absent or inadequate hip extension.

5. For BT, there was a high association of successful performance trials to normative patterns in NST, and conversely there was a high association of unsuccessful trials to abnormal patterns in PST for inplace trials. For FT, there was a low association of successful performance trials to normative patterns in both NST and PST and there was a high association of trials with unsuccessful performance to abnormal movement patterns in both NST and PST.

6. Scaling of the active component of the corrective response to BT was seen in the controls, NST and PST i.e. magnitudes of angular displacements about hip increased with increasing platform velocity (B1-B3). Increase in knee flexion (passive component) with increase rate of platform velocity was observed only in PST.

7. The timing of the corrective muscle responses in the stroke subjects was substantially delayed compared to the controls both on the stance and swing side for FT as well as BT.

8. There was a low correlation between walking index ($r = 0.33$) and gait speed ($r = 0.13$) scores and overall performance scores during BT. There was a moderate correlation of these clinical measure scores ($r = 0.68$ for WI, $r = 0.55$ for GS) with performance scores during FT.

5.2 SIMILARITY IN TASK PERFORMANCE AND ONSET OF PLATFORM MOTION

It is a known that stroke subjects have difficulty in weight bearing on the paretic side and have difficulty in weight shifting body weight from the swing to the stance side (Dettman et al,1987; Sackley et al, 1990, Bohannen and Larkin, 1985). Pai et al. (1994), report that, stroke subjects were unable to sufficiently transfer the COM successfully over the stance limb for 26% of the trials to the non-paretic side and 48% of the trials to the paretic side during the single leg flexion task. In the present study, the stroke subjects were able to perform the leg lifting task and displace the COM towards the stance side, as seen by the moment about the anterior-posterior axis and centre of foot pressure records (AX and AY). All the stroke subjects transferred their weight over the stance limb, in a similar fashion to the controls. There was no statistical difference between the controls and the non-paretic side in stance and the controls and the paretic side in stance in the medio-lateral displacement of centre of foot pressure. There was also no difference between NST and PST. There was no obvious difference in the time to peak, and the magnitude of rise in MY between the controls and the stroke subjects. Similar to the present study, Brunt et al, 1995, studied ground reaction forces in stroke subjects from the swing and stance sides of stroke subjects, during gait initiation, a task very similar to this study. Consistent to the findings of the present study, they found that stroke subjects were able to shift their body weight successfully onto the paretic as well as non-paretic stance side.

A major difference in the task performance between the stroke subjects and the controls in the present study was that the stroke subjects did not lift the swing leg as high as the controls. This was true, for both, when the non-paretic side as well as paretic side was in swing. The difference in height of the leg-lift could be due to their motor impairment, restricting maximum voluntarily knee and hip flexion.

5.3 STRATEGY SELECTION AND PERFORMANCE LEVELS

Stroke subjects in this study were not able to maintain an inplace strategy, seen in the control for most of the trials, although some trials during BT and very few trials during FT exhibited an inplace strategy. They on the other hand chose to touchdown (neutral step) or take a rapid step in opposite direction of platform motion, in order to successfully restore their balance. A touchdown was given a successful-fair performance grade instead of a successful-good as the subjects were able to restore balance but were unable to do it with one leg in stance. Trials with an inplace strategy and touchdowns were grouped and called an inplace response, as the basic strategy in both was similar, to move the COM over the base of support. Single steps in the correct direction were also given a successful-fair performance level as subjects were able to rapidly and accurately restore the COM to BOS relationship with a single step. However, the stepping strategy involves a change in the BOS to restore balance.

In the controls, there were steps seen only at the highest velocity, signifying that increasing disturbance levels challenged the nervous system greater, and forced it to cope with the changed environment by adopting a different strategy. Thus as predicted a switch in strategy from an inplace to stepping was evident in the controls at the highest velocity. It appears that rate of platform motion had little effect on strategy selection and performance levels in the stroke individuals. They chose to use the step strategy right from lowest level of disturbance (V1), instead of struggling on the stance leg to generate a normative inplace response in-order to move the COM over the changed BOS. This is one of the major reason for the relatively high success rate seen in the stroke subjects. However as predicted, performance levels of stroke subjects in this study were still significantly reduced compared to the controls, during both non-paretic stance and paretic stance. Since in the task of single leg-lifting, the swing leg is off the ground, it is easier to take a step than trying to maintain balance on single stance. Since stepping resulted in successful performance, the stroke subjects had a good percentage of successful trials. This may be a reason for the higher success rates in the present study compared to a similar study by Brunham (1996) which used the same rates of platform translations during stationary standing. The stroke subjects in their study were successful only for 35-40% of all the perturbed trials. Thus even though the task in the present study was more complex than, theirs (standing with both feet on the ground), the success rates in the stroke subjects were higher in the present study. The stroke subjects in the present study chose to take advantage of task, and adopted the stepping strategy to restore their balance, instead of utilizing the inplace strategy. Brunham (1996) did not observe any stepping responses in both the controls and the stroke subjects. Thus the subjects either exhibited a poor inplace strategy or fell leading to unsuccessful performance. Similar to this study, McIlroy and Maki (1996) observed compensatory stepping responses in elderly and healthy individuals, to relatively fast, unexpected forward and backward perturbations with both feet on the ground. However, in contrast to this study, in their study, both the controls and elderly showed stepping responses for majority of the trials. The accelerations of the platform translations both during BT and FT were much higher than that used in the present study. Thus at that level of disturbance the controls were likely at the limits of their stability and needed to take a step to restore balance. Similar to the present study, in the study by McIlroy and Maki (1996), the elderly group took multiple steps. Multiple stepping in the elderly occurred for 63% of the total trials. The authors could not identify or predict the reasons for multiple steps from kinematic or kinetic analysis of the initial corrective step. In the present study also, we were unable to determine the exact reason for the occurrence of multiple steps based on the kinematic analysis of the trunk and stance leg. It could be either due to absence or breakdown of a normative stance leg pattern, or due to poor voluntary control and coordination on the swing side. It must be noted that in trials where multiple steps were taken, though the stroke subjects did not fall they were at a greater potential risk of falling.

In the present study, most of the steps taken in the controls showed presence or at least remnance of the normative movement patterns on the stance side, used to restore total body balance. In the stroke subjects for 50% of trials where a step was taken, presence of a normative movement pattern was evident. It may be that the nervous system initially tried to cope with an inplace strategy, however, failed and a step was taken in consequence of an aborted inplace strategy. It is also possible that a step was taken right from the beginning and the stance leg movement pattern when a step is taken is similar to the stance leg pattern seen in an inplace strategy. In this study due to the small sample size we were not able to determine the stance leg movement pattern when a step was taken. Future studies need to be done to determine the normative stance leg patterns in rapid corrective stepping responses.

5.5 MOVEMENT PATTERNS FOR CORRECTIVE BALANCE RESPONSES

Stereotypical inplace movement patterns used to restore balance were identified in the controls and stroke subjects on the stance side. These multi-segmental patterns were similar to those seen in healthy individuals, children and stroke individuals during BT and FT with both feet the ground. (Nardone 1993, Szturm and Fallang, 1998, Brunham, 1996). The purpose of these patterns is to re-locate the COM over the translated BOS, restoring balance and preventing a fall. The movement synergies observed during FT and BT were fundamentally different. Hip flexion and ankle plantar flexion is seen during BT and early knee flexion followed by hip extension during FT. In the present study the controls subjects exhibited a few degrees of knee flexion. This knee flexion is a passive component of the synergy, as a result of active trunk/hip flexion (Alexandrov et al, 1998). It is likely that the external forces due to head-trunk-acceleration contributing to a knee flexion moment would be more difficult to control when standing on one leg than two. Normative movement patterns were present in stroke subjects, both on the paretic side as well as non-paretic side for less than fifty percent of the trials.

Abnormal patterns were identified in the stroke subjects during BT and FT. During BT, the normative movement components of hip flexion and forward trunk rotation were evident in most trials. However in many trials excessive trunk flexion was observed along with excessive hip flexion. The majority of the stroke subjects showed excessive knee flexion instead of extension, during paretic side stance. Excessive hip flexion and trunk flexion seen in the stroke subjects, with excessive knee flexion would result in the COM moving more anterior than posterior. This would limit the ability to shift centre of mass backwards over the base of support and thus lead to a higher risk of losing balance (poor performance.)

During FT, the two most common abnormal patterns seen were 1) absent hip extension and excessive knee flexion and 2) absent hip extension and absent knee flexion. In this case the component(s) of the inplace movement pattern was absent. Knee flexion and hip extension together, help to thrust the pelvis forward in or order to bring the COM forward over the translated base of support. Most of the trials with absent normative patterns were associated with a fall, in cases where a strategy switch was not seen, i.e. when a step was not taken. The third abnormal pattern was absent knee flexion, where again one active component of the pattern was missing. Similar to BT, during FT also poor control pelvic-trunk and knee control is evident.

On an average (BT + FT) there was a moderate to high association of normative movement patterns with successful performance, during an inplace response in the stroke subjects. A few trials in the stroke subjects, with a normative movement pattern did not lead to a successful performance. This indicates that there are likely additional factors responsible for performance levels apart from strategy selection and normative movement patterns determined from the stance side. Though normative movement patterns were able to control stance side of the body, deficient control of the swing side of the body or trunk could lead to unsuccessful performance. It would be difficult to control, whole-body balance on one-legged stance after a pathology such as stroke, affecting one side of the body. Cases where the subjects didn't chose to switch to the stepping strategy, and tried to maintain an inplace response, may have resulted in multiple steps or falls. As expected, there was a high association of trials with abnormal patterns and unsuccessful performance during an inplace response.

In stroke subjects, the associations of successful performance with normative stance patterns, as well as unsuccessful performance to abnormal patterns during a stepping strategy were approximately 50%. A single appropriate step taken by the subjects could restore balance and thus prevent an unsuccessful performance, inspite of an abnormal stance pattern. On the other hand in presence of a normative stance pattern, if the single step was not accurately taken it could lead to a stumble. Conversely, some times the stepping may be accurate, but stance side control may break down, which could also result in multiple steps/falls and unsuccessful performance scores. However, as discussed in the previous section, it is not confirmed if the normative stance patterns seen in the controls during an inplace response should also present on the stance side during a stepping response. This could be a major factor affecting associations of performance levels with movement patterns. It was found in this study that by looking at the stance pattern within 400 ms of platform onset it could not be predicted if a primary step would be taken or a multiple step would be taken.

The magnitude of corrective hip and knee angular displacements, in the controls, increased with increasing rate of platform translations during BT. Thus, within the platform velocities/accelerations used in the present study, the controls were able to scale the active (hip flexion) component of the normative movement synergy. Similar to the controls, the stroke subjects (NST + PST) were able to scale the magnitude of hip flexion at least at B1 and B2.

There was no effect of rate of platform motion on knee flexion, (passive component) for the controls and stroke subjects during non-paretic stance. However during paretic side stance, knee flexion increased in magnitude from B1 through B3. Thus some limits in knee joint control during corrective reactions were evident on the paretic side.

5.4 ROLE OF CORTEX IN CORRECTIVE BALANCE REACTIONS

The findings of the present study have identified limitations in corrective balance responses in stroke subjects with cortical lesions. It is known that supraspinal centres, in particular, brain stem sites such as PMRF and its descending pathways (VST and RST) are involved in corrective balance reactions. (Mori et al., 1987; Russel and Zajac, 1979; Lai and Segal, 1990). Animal studies done by Gorrsini and Hiebert, 1994 provide evidence that supraspinal control is required for corrective balance responses during function related tasks. A few human studies (Nashner, 1980; Eng and Winter, 1994; Szturm and Fallang, 1998) have suggested supraspinal control in corrective balance reactions seen in healthy human subjects during perturbed standing. This was based on the presence timing and magnitude of multi-segmental corrective movement patterns and muscle responses. However no physiological studies have directly proved the role of cortex in corrective balance reactions to sudden or unexpected disturbances. Anatomical and electrophysiological studies report presence of bilateral pathways between the cortex and PMRF (Canedo and Lamas, 1993; Lamas et al, 1994; Magni and Willis,1964; Matsuyama and Drew, 1988). The study by Brunham (1996) provided evidence that the cortex was involved to some extent in generation of corrective balance reactions. For example presence of abnormal lower limb movement patterns seen on the non-paretic and paretic sides in subjects with a cortical lesion.

Normative movement patterns present in the stroke subjects both on the paretic side and non-paretic side stance, and association of these patterns with successful performance indicates that the balance regulating mechanism is functional to some extend in these individuals for the present task. It furthers suggests regulation of balance at a sub-cortical level, for example balance regulation centres in the brainstem (Mori et al, 1987, Nashner, 1980). However if balance control was regulated only at the brain stem level (PMRF) then one should expect normative motor strategy and normative movement patterns for majority of the trials at all different levels of platform acceleration/velocity or degrees of difficult similar to that found in controls. Presence of abnormal movement patterns in individuals post-stroke and significantly greater risk of poor performance and/or falls is consistent with a cortical influence in the control of corrective balance reactions. This provides evidence in support of the role of the cortex and cortico-bulbo pathways in corrective balance reactions to unexpected disturbances. Lawrence and Kuypers, 1968, showed that monkeys with combined lesions of the cortico-spinal tract and medial descending brain stem pathways had more difficulty in maintaining and controlling balance during functional activities compared to monkeys with lesions of brain stem pathways alone. This work by Lawrence and Kuypers also suggests that certain tasks require some cortical control for balance reactions.

Secondly in this study, the stroke subjects chose the stepping strategy at the lowest levels of platform motion, which resulted in successful performance. It must be noted that although some of these subjects may not be able to take a rapid voluntary step when asked, when challenged they were able to rapidly step to maintain balance, both with the non-paretic leg as well as the paretic leg. Following a cortical motor lesion a strategy switch was used to reduce the risk of falling. Volitional motor control was most likely required in cases where a second step was taken, to restore balance and prevent a fall. Multiple stepping trials were observed both during non-paretic side stance as well as paretic stance. Following sensori-motor lesions of the cortex these individuals may have reduced ability to effectively or adequately produce and control rapid voluntary multi-segmental stepping movements especially when total body balance is required.

5.6 DIFFERENCE BETWEEN FORWARDS AND BACKWARDS TRANSLATIONS

In the present study, forward translation seemed a more challenging disturbance in controls as well as the stroke subjects than backward translation. This is in agreement with the findings reported by Brunham, (1996). McIlroy and Maki (1996) observed stepping responses in controls and healthy elderly subjects at a lower peak platform acceleration during FT (1.5 m/sec^2) than during BT (2.0 m/sec^2). This implies that an inplace response could be easily more maintained by the subjects at the lower acceleration during BT compared to FT, as during FT, stepping occurred at lower rates of platform motion.

In the present study there were a number of findings that would also indicate that FT is a more difficult direction of disturbance than BT. In the present study, a strategy switch was evident in majority of trials at F3 in the controls, which was not observed during BT at V3. The strokes subjects on average (NST + PST) exhibited a lower percentage of success during FT compared to BT. There were more numbers of falls in the stroke subjects during FT compared to BT. There was a greater percentage of abnormal patterns seen during FT compared to BT. Also there was a higher association of abnormal patterns to unsuccessful trials during FT compared to BT.

The reasons for the differences seen between forwards and backwards translations may be as follows: 1) Forward translations require a more complex synergy to control the COM, which the nervous system may not able to generate efficiently. For example, the multi-link knee flexion, hip extension synergy during FT is difficult to achieve compared to hip flexion synergy of BT. 2) From a bio-mechanical perspective, there is a greater range available for hip flexion than hip extension. Normally there is 90 degrees of hip flexion compared only to 30 degrees of hip extension present when tested passively. Since the movements on the stance side occur in a closed-link pattern (i.e. foot stationary), hip-pelvic and trunk disassociation is required to bring thrust the pelvis forward in order to achieve hip extension. This task is quite complex and difficult to achieve even in some healthy controls. As seen in most stroke subjects, dissociation of trunk segment and hip joint is difficult to achieve. During FT, we observed that in most of the trials, trunk segment extension was present even in absence of hip extension. Trunk segment extension, in absence of hip extension, doesn't allow the pelvis to be thrust forwards to quickly move the COM over the translated BOS. 3) We may be encountering many more daily living activities that displace COM forward in relation to the foot. For example a push on the back is more commonly encountered, than a push on the chest. Thus the body might be accustomed to such kind of disturbances and may have learnt to deal with them effectively. Backward translation is similar to a push on the back, where the COM remains anterior to the feet and thus it may be much easier to restore balance during BT. Therefore we see a higher percent of normative patterns and higher success rates during BT compared to FT. While restoring balance during FT, instead of adopting the more complex inplace synergy it might be easier to take a step and change the base of support. This might be the reason why majority of control subjects also exhibited a stepping strategy at F3, and why the stroke subjects had very few inplace trials during FT compared to BT. During FT majority of the trials in stroke subjects with an inplace strategy were unsuccessful.

5.7 FRONTAL PLANE CONTROL

A few researchers have reported frontal plane instability as potential cause of falls towards the paretic side even during anterior-posterior directed disturbances or movements (Diller and Winberg, 1970; Weinstein et al, 1989; Pai et al, 1994; McKinnon and Winter, 1993; Topper et al, 1993). In the present study, excessive abnormal responses in the frontal plane were observed only for 50% or less of the trials in the controls and as well as stroke subjects. Excessive hip and trunk motion was present in a greater percent of trials in the stroke subjects than controls however the response was not direction dependent. For some of the trials the displacements were towards the stance side and for some trials away from the stance side. This excessive response seen in the controls and the stroke subjects may be due to the corrective steps being taken by them. Stepping responses may have altered the displacement directions in the subjects. As seen for the controls, at F3, the response changed from abduction to adduction. There were maximum stepping responses seen at F3 for the controls. Also, greater stepping responses were observed in the stroke subjects, which could explain the greater percentage trials with excessive response in the strokes compared to the controls. In the present study we did not separate the stepping trials from the inplace ones. Future studies should be done to quantify the normative inplace movement patterns in the frontal plane, to perturbations during the leg-lifting task, as well as quantify the normative stance leg patterns during stepping responses.

5.8 EMG RESPONSES

An attempt was made to determine the timing of the corrective response by looking at the EMG onset latencies of four muscle groups (hip abductors, hip adductors, rectus femoris and hamstrings). In the present study it was evident that the stroke subjects had longer onset latencies than the controls, at least on the paretic side. The range of onset latencies in the controls during FT and BT (80-153 ms) was within the normal ranges sited by previous work in healthy controls (Nardone et al, 1990, Allum et al., 1989, Sztrum and Fallang, 1998). Similar to that seen in the present study, Badke et al, 1983, 1987 report longer onset latencies from the paretic side in stroke subjects compared to healthy controls and no difference in onset latencies between non-paretic side and the left and right sides of the controls. Di fabio and Duncan, 1986, report variable onset latencies both from the paretic side and the non paretic side, with some muscles having longer onset latencies than the controls where as some muscles having shorter onset latencies than the controls. This is consistent with the finding of the present study. Note most of the above studies involving stroke subjects (Badke et al, 1983,1987, and Di-Fabio and Duncan, 1986) did not attempt statistical analysis of onset latencies on individual muscles. It can be assumed that similar to the present study, the problem of large variability in frequency of muscle responses between groups, and small sample sizes made it difficult to perform statistical analysis. In addition, in the present study, balance was restored by means of different strategies. The difference in frequency of trials with and inplace versus a stepping strategy could explain why the frequency of EMG responses was so variable and different between controls, non-paretic side and paretic side. The slightly greater frequency of activation of swing side muscle, in both NST and PST compared to the controls could be explained due to the higher percentage of steps and touchdowns seen in these groups.

5.9 CONTRIBUTION FROM PARETIC AND NON-PARETIC SIDE IN CORRECTIVE BALANCE RESPONSES

No one to our knowledge has examined the contribution of paretic side versus non-paretic side in stance as well as swing for corrective balance responses as yet. This study was able to determine the contribution from the paretic and non-paretic sides to some extent, both in stance and swing conditions. There was no major difference in strategy selection, successful performance and number of falls between trials when the non-paretic leg was in stance (NST) versus when paretic leg was in stance (PST). In addition to the similarity in the movement patterns observed, the frequency of muscle activation and EMG onset latencies between the non-paretic side and paretic side both on the stance and swing sides were within a similar range. It was found that both the stance and the swing sides of the body contributed substantially towards restoring body stability to unexpected platform translations, with corrective steps being more successful sometimes than inplace responses.

Although we saw no statistical difference in performance levels between the non-paretic side stance and the paretic side stance there was definitely a trend to see a higher frequency of successful trials when the paretic leg was in stance compared to when the non-paretic leg was in stance during FT. In addition during FT, there were greater number of multiple steps when the non-paretic leg was in stance (37%) compared to when the paretic leg was in stance (17%). The reason for these may be that when the paretic side was in stance, the non-paretic side was capable of taking an effective corrective step in the opposite direction to restore balance. Conversely when non-paretic leg was in stance the, paretic side may not have the motor control necessary to rapidly take a corrective step appropriate to re-position the BOS. This appears to be consistent with the Chedoke-Mcmaster leg and foot motor impairment scores (range 2-5) which show that there was substantial motor deficit on the paretic side. In contrast to FT during BT there were equal number of multiple steps seen in PST and NST. This may due the fact already mentioned that restoring balance during FT is more difficult than during BT. Thus poor performance of the paretic side became evident in the more challenging task. It is also possible that the difference seen between NST and PST during FT may just be a sample size problem and there may be no real difference present in performance levels between non-paretic side stance and paretic side stance. Future studies with a larger sample size would be able to resolve this problem.

The stroke subjects had abnormal movement patterns both on the NST as well as PST. However there was a trend to see a higher percentage of abnormal patterns during PST trials compared to NST trials. Also there was a higher association of abnormal patterns with unsuccessful trials, a low association of successful trials to normative patterns when paretic limb was in stance. Thus even though the number of successful trials was very similar during non-paretic stance and paretic stance, there were a higher number of trials with abnormal stance patterns when the paretic side was in stance.

During FT, in-spite of a higher trend of success rate in PST compared to NST, there was greater percentage of abnormal patterns seen in PST compared to NST. Presence of uncontrolled knee flexion was seen at all velocities during BT in paretic side stance. Thus though the paretic side, achieved successful performance using different means for example stepping, some limits in automatic balance reactions, are evident from a mechanism point of view after a cortical lesion.

It must be noted that though the performance levels were similar when the non-paretic leg was in stance and when the paretic leg was in stance, the reasons for this could be different. Some limitations on the paretic side were definitely identified, as discussed above. Thus successful performance when the paretic side was in stance could be achieved due to a greater contribution from the non-affected swing side. Conversely, when the non-paretic side was in stance, successful performance could be achieved by a greater stance control.

One would expect normative movement patterns to be present on the non-paretic side (unaffected), similar to that in controls. The abnormal movement pattern observed during non-paretic stance in this study could be explained as follows: Firstly most of the subjects in this study, were chronic stroke survivors, and were not involved in any regular physical training after completion of their out-patient rehabilitation program. The non-paretic side was thus in a de-conditioned state. Secondly since the task in the present study was bilateral, and more complex compared to standing still, there would be a greater need for both sides of the body to contribute fully to recover total body stability. So the non-paretic side may have failed to exhibit normative stance responses by itself. Thirdly, from a neuro-physiology perspective it is well documented in literature that the fronto-parietal cortex has descending and ascending connections with both ipsilateral and contralateral PMRF (Berrevoets and Kuypers, 1975; Drew and Rossignol, 1986; He and Wu, 1985; Kuypers, 1958). Thus some limits in balance control could be evident even on the ipsilateral side of the lesioned cortex (non-paretic/unaffected side of the body).

Current evidence suggest that the intact cortex can take over the functions of the damaged cortex (Chollet et al, 1991; Weiller et al, 1992; Weder et al, 1994; Cao et al, 1998; Cramet et al, 1999). This could explain the successful trials and normative stance patterns present during paretic side stance. Secondly, as mentioned above, the descending pathways from the intact ipsilateral cortex may contribute towards the corrective response.

5.10 CORRELATION WITH CLINICAL MEASURES

Self-paced gait speed was tested in this study, which is typically used in the literature study, and subjects were allowed to use their walking aids, thus walking was cautious and slow and most of the subjects relied with heavily upon their assisstive devices. Secondly gait speed was tested indoors, which could be very different than daily outdoor walking. A few researchers (Sackley et al, Dettman et al., 1987; Olney et al, 1994, Malouin et al, 1994) have found moderate correlation of balance tests (Berg balance Scale and Fugly-Meyer) with gait speed. The ranges of gait speeds in those studies were very similar to that in the present study. A moderate correlation of performance scores with both gait speed as well as walking index was observed during FT, however no correlation was observed during BT. Pai et al, (1994), found a poor correlation of gait speed with % successful trials in their anticipatory balance control test. As reported by Pai et al, (1994), gait speed by itself cannot predict balance control or stability requirements of subjects. In the present study, some subjects obtaining a high performance score had very low gait speeds. When tested in the unexpected environment, these subjects had the ability to make rapid balance corrections, however during self paced walking, tended to spend more time in double support phase, and rely heavy upon the assistive device to transmit the body weight. On the other hand, some subjects having moderate to low performance scores had fast gait speeds. It must be noted that these subjects could walk fast due to their assistive device. When subjected to the balance test, most of the subjects could not either generate appropriate inplace movement synergies to control the COM to BOS relation ship, or cold not take an accurate single step to change the BOS to accommodate the stationary COM at least during FT. Thus this gave them a low performance score.

Walking index has five items walking: 1) walking indoors 25m (note no time limit) 2) walking outdoors over rough ground, ramps and curbs for 140m, 3) similar outdoors walking but for 900m, 4) walking up and down stairs and 4) a 2-minute walking distance, scored according to age. No one to our knowledge has done a correlation of walking index with other balance measures like Berg Balance Scale or Fugyl-Meyer

Balance Scale. Similar to correlation with gait speed, a moderate correlation of walking index scores with performance levels was observed during FT in the present study and a low correlation was observed during BT. Though the walking index allows the individual to walk outdoors, as in daily living, it is not a time-based scale. Individuals can take as long as they wish to complete the 140 m or 900 m walk, and individuals are allowed to take rest as well as use their assisstive device. Besides the examiner always walks besides them, so that they are psychologically more secure.

The difference in correlations between BT and FT can be explained as follows: BT is not as challenging a balance disturbance compared to FT. More complex movement control is required for the corrective response to FT as already discussed. Also the range of performance scores for BT was small (12-19), compared to FT (7-18). Thus there was a lot of clustering of performance scores during BT, which could also give a low correlation value. Lastly, this difference may just be a sample size problem. As sample size in the present study was small.

5.11 CLINICAL SIGNIFICANCE

A traditional view exists that weight-bearing asymmetry leading to poor postural responses on the paretic side is the main cause of balance impairment post-stroke. (Sackley, 1990; Collen, 1995; Dickstein and Dvir, 1993; Difabio and Badke, 1990; Badke and Duncan, 1983) Training is concentrated on the weight bearing and weight shifting tasks towards the paretic side in stance, so as to obtain normative weight distribution and perhaps more normative postural responses from the affected side. (Summway-Cook, 1988; Dettman et al, 1987; Sackley, 1991; Duncan and Badke, 1987). It would be incorrect to judge balance or postural deficits by assessing and training only one side of the body, especially when functional activities like outdoors walking, stair climbing other activities of daily living usually involve bilateral body control. This study showed that both inplace and stepping responses were required for corrective responses and unsuccessful performance seen in the stroke subjects was due to both improper swinging abilities as well as breakdown of normative movement patterns on the stance side.

Stroke subjects could be taught to use change in base of support strategy if they have difficulty generating adequate muscle forces for a normative inplace strategy. Change of base of support strategy might be a better solution to chose to control balance, in the stroke subjects instead of trying to actively move the COM and failing. One must be cautious to train stroke subjects to achieve accurate stepping movements, foot placements and good voluntary control about the knee, hip, and trunk and ankle, in order for them to succeed with the stepping strategy and prevent a stumble (multiple-steps) or a fall. One should also focus more on the swinging capabilities of paretic side. An increase in step length could be achieved by training the swing leg stepping. This can also help increase gait speed.

It should be noted that there are many situations where stepping is better than inplace, but there are also situations where inplace responses are needed and should also be trained. Thus rehabilitation training should also focus on training specific components of the normative movement patterns, for example trying to achieve active hip extension, with knee flexion during single stance support conditions. Such training can produce normative stance movement patterns in the stroke subjects, allowing them to control active COM displacement effectively. Therapists should concentrate on training concentric plus eccentric muscle work around hip joint. This would aid in preventing excessive hip flexion seen in stroke subjects during BT. It would also aid in achieving active hip extension required to recover balance during FT. Training for concentric and eccentric muscle work about the knee joint could help control for the excessive uncontrolled knee flexion observed in stroke subjects during paretic side stance. Once, individual movement patterns can be achieved voluntarily by subjects, these need to be practiced during unexpected disturbances. This would allow in re-enforcing these learned movement patterns during sudden disturbances encountered in daily living activities.

This study, in addition also found that the both the paretic side and the non-paretic side contributed during corrective balance responses. It is suggested that training should focus on tasks involving bilateral lower-extremity movements, with equal concentration on paretic side as well as non-paretic side. Researchers have found that it is not automatically possible to transfer standing balance skills to balance skills required for functional activities suck as walking, gait initiation and stepping (Winstein et al, 1989; Malouin et al, 1992, Hocherman et al, 1984). Therapy must try and concentrate on training dynamic functional tasks, instead of training standing balance in relatively static situations. Balance training should concentrate on training both voluntary movements as well as stability components. For example, body weight support harness (Visintin et al, 1998; Hesse et al, 1994) training can be one way of training both balance and voluntary movements during a functional task. The body weight support harness provides the required amount of stability to an individual, allowing the therapist and individual to focus on factors such as multi-segmental stance and swing movements, accurate foot placement and speed. As the voluntary aspects of stepping movements improve, the amount of support provided by the harness can be reduced to train then train for the balance aspects during these movements.

Owings et al. (2000) showed that assessment of static postural tasks does not allow one to predict how an individual would perform to sudden dynamic balance disturbances. The present balance test on the movable platform apparatus allows testing for whole body balance, as well as contribution from individual sides during unexpected disturbances. Thus the ability of the paretic side in stance as well as the ability of the paretic side to step and vice versa can be tested. The present balance test could serve as a laboratory test of dynamic balance assessment. Subjects could be assessed on the platform test, and performance grades can be recorded. Subjects could then be re-tested after appropriate balance training to look for an improvement in the performance grades. To cut down on the cost of using the computer controlled platform, unexpected anterior posterior perturbations could also be delivered by the treating physiotherapist by suddenly accelerating and decelerating a stationary tread mill belt. Balance tests and balance training should mainly include forward translations, as they appear to be more challenging to subjects than backward translations.

6. CONCLUSION

Balance impairment is a major functional problem after stroke and the risk of falls and stumbles are also considerably greater after stroke. This is consistent with the findings of the present study. In the present study, corrective balance reactions were present in all the healthy controls and stroke. Stumbles and falls however were present only in the stroke subjects.

A classification of movement strategies and performance levels exhibited during corrective balance reactions to anterior-posterior perturbations has been established in the present study. Majority of the stroke subjects chose the change in base of support strategy at all levels and direction of balance disturbances in contrast to the control subjects who exhibited an inplace response majority of the times. This stepping corrective strategy with non-paretic or paretic limb often resulted in successful performance.

The limitations in performance levels, and high association of unsuccessful performance with abnormal movement patterns observed in stroke subjects points towards a cortical modulation of balance responses. The switch in strategy observed in the controls, and the secondary steps observed in the stroke subjects also suggests that cortical influence is necessary for generating the most appropriate responses to deal with unexpected conditions and restore body stability.

The present study showed that, the non-paretic and paretic side in the stroke subjects both when in stance as well as in swing, contributed to restore whole body balance. Balance assessment and training in the stroke population should thus focus on both sides of the body and both types of functional tasks, restoring COM to BOS relationship using stance movement patterns as well as using stepping responses.

Similar to a few other studies, the present study also indicates that, walking ability and self-paced gait speed can only partially predict dynamic balance recovery during functional activities.

7. LIMITATIONS AND FUTURE IMPLICATIONS

This study had a few limitations, which could be overcome and dealt effectively in future studies. The following limitations are identified and future implications suggested:

1) Heterogeneity of subjects. We tried our best to recruit subjects having similar levels of motor impairment and disability by restricting ourselves to subjects having a stroke in the distribution area of the middle cerebral artery. However the time period for completion of a Masters' thesis being limited, it was difficult to get such a homogeneous subject group. Thus the inclusion criteria for subjects was cortical stroke in the region fronto-parietal region.
2) Sample size of this study was small This relates to the recruitment of post-stroke individuals who matched the inclusion criteria and the time constraints of this study.

Recruitment of more subjects through a multi centre study is required. Also there is a need to separate LCVA subjects from RCVA in order to look for difference in effect of left and right cortical lesions on corrective balance responses.

3) Age matched control subjects should be recruited to exclude any age-related difference in performance levels between controls and stroke subjects.
4) There was a slight variability in how high the foot was lifted off the ground in the stroke subjects compared to the controls. Two of the stroke subjects had difficulty in performing the task, and were not be able to lift their foot completely off the platform. However the platform was programmed to trigger into motion only when complete weight transference over to the stance leg had been achieved, thus the response would still be obtained in a single stance situation.
5) In order to compare characteristics of the primary stepping strategy between controls and healthy individuals, step characteristics need to be quantified i.e. step onset, step duration, step displacement. Also control studies to get normative data for stance patterns during stepping patterns in sagittal and frontal plane need to be done.
6) In this study the CM-S displacement in the sagittal plane was analyzed only from the stance side, the centre of mass displacement on the swing side of the body should also be evaluated since, both the sides of the body are participating in the response. The centre of mass displacements during the movement response in the frontal plane should also be looked at to determine control of body stability in this plane.
7) EMG analysis was only restricted to frequency and onset latency analysis in the present study. Analysis of EMG patterns for particular strategies during BT and FT and analysis of EMG magnitudes could be done in future studies.
8) Kinetic analysis on the swing and stance side should be done to give a better understanding of the motion dependent, gravitational and internal forces acting on the both sides of the body. This would enable a better interpretation of the kinematic patterns observed and EMG muscle activation patterns.

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APPENDIX 1

Chedoke-McMaster Stroke Assessment
SCORE FORM Page 1 of 4
IMPAIRMENT INVENTORY: SHOULDER PAIN AND POSTURAL CONTROL

POSTURAL CONTROL: Start at Stage 4. Starting position is indicated beside the item or underlined. No support is permitted. Place an X in the box of each task that is accomplished. Score the highest Stage in which the client achieves at least two Xs.

SHOULDER PAIN			POSTURAL CONTROL		
1	☐	constant, severe arm and shoulder pain with pain pathology in more than just the shoulder	1	☐	not yet Stage 2
2	☐	intermittent, severe arm and shoulder pain with pain pathology in more than just the shoulder	2 Supine	☐	facilitated log roll to side lying
			Side lying	☐	resistance to trunk rotation
			Sit	☐	static righting with facilitation
3	☐	constant shoulder pain with pain pathology in just the shoulder	3 Supine	☐	log roll to side lying
			Sit	☐	move forward and backward
			Stand	☐	remain upright 5 sec
4	☐	intermittent shoulder pain with pain pathology in just the shoulder	4 Supine	☐	segmental rolling to side lying
			Sit	☐	static righting
			Sit	☐	stand
5	☐	shoulder pain is noted during testing, but the functional activities that the client normally performs are not affected by the pain	5 Sit	☐	dynamic righting side to side, feet on floor
			Sit	☐	stand with equal weight bearing
			Stand	☐	step forward onto weak foot, transfer weight
6	☐	no shoulder pain, but at least one prognostic indicator is present • Arm Stage 1 or 2 • Scapula malaligned • Loss of range of shoulder movt - flexion/abduction $< 90°$ or external rotation $< 60°$	6 Sit	☐	dynamic righting backward and sideways with displacement, feet off floor
			Stand	☐	on weak leg, 5 seconds ☐ sec
			Stand	☐	sideways braiding 2 m
7	☐	shoulder pain and prognostic indicators are absent	7 Stand	☐	on weak leg: abduction of strong leg
			Stand	☐	tandem walking 2 m in 5 sec
			Stand	☐	walk on toes 2 m

☐ STAGE OF SHOULDER PAIN

☐ STAGE OF POSTURAL CONTROL

Chedoke-McMaster Stroke Assessment
SCORE FORM
IMPAIRMENT INVENTORY: STAGE OF RECOVERY OF ARM AND HAND

ARM and HAND: Start at Stage 3. Starting position: sitting with forearm in lap in a neutral position, wrist at 0° and fingers slightly flexed. Changes from this position are indicated by underlining. Place an X in the box of each task accomplished. Score the highest Stage in which the client achieves at least two Xs.

Stage	ARM	Stage	HAND
1	☐ not yet Stage 2	1	☐ not yet Stage 2
2	☐ resistance to passive shoulder abduction or elbow extension ☐ facilitated elbow extension ☐ facilitated elbow flexion	2	☐ positive Hoffman ☐ resistance to passive wrist or finger extension ☐ facilitated finger flexion
3	☐ touch opposite knee ☐ touch chin ☐ shoulder shrugging > ½ range	3	☐ wrist extension > ½ range ☐ finger/wrist flexion > ½ range ☐ <u>supination, thumb in extension</u>: thumb to index finger
4	☐ extension synergy, then flexion synergy ☐ shoulder flexion to 90° ☐ <u>elbow at side, 90° flexion</u>: supination, then pronation	4	☐ finger extension, then flexion ☐ thumb extension > ½ range, then lateral prehension ☐ finger flexion with lateral prehension
5	☐ flexion synergy, then extension synergy ☐ shoulder abduction to 90° with pronation ☐ <u>shoulder flexion to 90°</u>: pronation then supination	5	☐ finger flexion, then extension ☐ <u>pronation</u>: finger abduction ☐ <u>hand unsupported:</u> opposition of thumb to little finger
6	☐ hand from knee to forehead 5 x in 5 sec. ☐ <u>shoulder flexion to 90°</u>: trace a figure 8 ☐ <u>arm resting at side of body:</u> raise arm overhead with full supination	6	☐ <u>pronation</u>: tap index finger 10 x in 5 sec ☐ <u>pistol grip</u>: pull trigger, then return ☐ <u>pronation:</u> wrist and finger extension with finger abduction
7	☐ clap hands overhead, then behind back 3 x in 5 sec ☐ <u>shoulder flexion to 90°</u>: scissor in front 3 x in 5 sec ☐ <u>elbow at side, 90° flexion</u>: resisted shoulder external rotation	7	☐ thumb to finger tips, then reverse 3 x in 12 sec ☐ bounce a ball 4 times in succession, then catch ☐ pour 250 ml. from 1 litre pitcher, then reverse

☐ STAGE OF ARM

☐ STAGE OF HAND

Chedoke-McMaster Stroke Assessment
SCORE FORM **Page 3 of 4**

IMPAIRMENT INVENTORY: STAGE OF RECOVERY OF LEG AND FOOT

LEG: Start at Stage 4 with the client in crook lying. FOOT: Start at Stage 3 with the client in supine. Test position is beside the item or underlined. If not indicated, the position has not changed. Place an X in the box of each task accomplished. Score the highest stage in which the client achieves at least two Xs. For "standing" test items, light support may be provided but weight bearing through the hand is not allowed. Shoes and socks off.

Stage	Position	LEG	Stage	Position	FOOT
1		☐ not yet Stage 2	1		☐ not yet Stage 2
2	Crook lying	☐ resistance to passive hip or knee flexion ☐ facilitated hip flexion ☐ facilitated extension	2	Crook lying	☐ resistance to passive dorsiflexion ☐ facilitated dorsiflexion or toe extension ☐ facilitated plantarflexion
3		☐ <u>abduction</u>: adduction to neutral ☐ hip flexion to 90° ☐ full extension	3	Supine Sit	☐ plantarflexion > ½ range ☐ some dorsiflexion ☐ extension of toes
4	Sit	☐ hip flexion to 90° then extension synergy ☐ bridging hip with equal weightbearing ☐ knee flexion beyond 100°	4		☐ some eversion ☐ inversion ☐ <u>legs crossed</u>: dorsiflexion, then plantarflexion
5	Crook lying Sit Stand	☐ extension synergy, then flexion synergy ☐ raise thigh off bed ☐ hip extension with knee flexion	5	Stand	☐ <u>legs crossed</u>: toe extension with ankle plantarflexion ☐ <u>sitting with knee extended</u>: ankle plantarflexion, then dorsiflexion ☐ <u>heel on floor</u>: eversion
6	Sit Stand	☐ lift foot off floor 5 x in 5 sec. ☐ full range internal rotation ☐ trace a pattern: forward, side, back, return	6		☐ <u>heel on floor</u>: tap foot 5 x in 5 sec ☐ <u>foot off floor</u>: foot circumduction ☐ <u>knee straight, heel off floor</u>: eversion
7	Stand	☐ <u>unsupported</u>: rapid high stepping 10 x in 5 sec ☐ <u>unsupported</u>: trace a pattern quickly; forward, side, back, reverse ☐ <u>on weak leg with support</u>: hop on weak leg	7		☐ heel touching forward, then toe touching behind, repeat 5 x in 10 sec ☐ <u>foot off floor</u>: circumduction quickly, reverse ☐ up on toes, then back on heels 5 x

☐ **STAGE OF LEG**

☐ **STAGE OF FOOT**

Chedoke-McMaster Stroke Assessment
SCORE FORM Page 4 of 4
DISABILITY INVENTORY

SCORING LEVELS

NO HELPER	Independence	
	7 Complete Independence	(Timely, Safely)
	6 Modified Independence	(Device)
HELPER	Modified Dependence	
	5 Supervision	
	4 Minimal Assist	(Client = 75%)
	3 Moderate Assist	(Client = 50%)
	Complete Dependence	
	2 Maximal Assist	(Client = 25%)
	1 Total Assist	(Client = 0%)

		Score
1.	Supine to side lying on strong side	☐
2.	Supine to side lying on weak side	☐
3.	Side lying to long sitting through strong side	☐
4.	Side lying to sitting on side of the bed through strong side	☐
5.	Side lying to sitting on side of the bed through the weak side	☐
6.	Remain standing	☐
7.	Transfer to and from bed towards strong side	☐
8.	Transfer to and from bed towards weak side	☐
9.	Transfer up and down from floor and chair	☐
10.	Transfer up and down from floor and standing	☐
11.	Walk indoors - 25 meters	☐
12.	Walk outdoors, over rough ground, ramps, and curbs - 150 meters	☐
13.	Walk outdoors several blocks - 900 meters	☐
14.	Walk up and down stairs	☐
15.	Age appropriate walking distance for 2 minutes (2 Point Bonus)	☐
	Distance ☐ meters	
	Total Score	☐

Walking aids:

- walker ☐
- 4 point cane ☐
- 1 point cane ☐
- brace ☐

To score Bonus:
for age less than 70 years distance must be > 95 meters or greater
for age 70 years or greater distance must be > 85 meters or greater

COPY FREELY - DO NOT CHANGE