AGE-RELATED EFFECTS ON BALANCE RECOVERY DURING PERTURBED LOCOMOTION

by
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A THESIS
Presented to the Faculty of Graduate Studies
In Partial Fulfilment of the Requirements
for the Degree of
Master of Science (Rehabilitation)

MASTER OF SCIENCE

Faculty of Medicine
School of Medical Rehabilitation
Division of Physical Therapy
University of Manitoba

February 1999

THE UNIVERSITY OF MANITOBA

FACULTY OF GRADUATE STUDIES
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ABSTRACT

It has been proposed that the neural process organizing balance and locomotor activities is integrated. During disturbances while walking, output of the locomotor movement generator would produce changes in the stepping movement to restore stability and maintain forward progression. Studies show that muscle responses to sudden disturbances while walking are phase dependent (Nashner, 1980; Eng, Winter and Patla, 1994; Tang, Woollacott and Chong, 1998). The corrective muscle responses, along with interlimb dynamics, are responsible for those parameters of stepping movements which are modulated. The parameters within the stepping movement which can be modulated are pattern, duration and magnitude. The purpose of this study was to examine which parameters are modified, in young and elderly groups when the relationship between the centre of mass (CM) and the base of support is suddenly changed during walking.

Nine active healthy young subjects and nine active healthy elderly subjects participated in this study. Each subject was perturbed through backward translations (BT) and forward translations (FT). The perturbations were presented just after right heel-off (PT1) or just after right mid-swing (PT2). Each subject performed 8 trials with no perturbations (NPT), 8 trials with FT at right PT1, 8 trials with FT at right PT2, 8 trials with BT at right PT1 and 8 trials with BT at right PT2. The timing of early corrective muscle response was determined bilaterally for anterior tibialis (TA), gastrocnemius (GA), hamstrings (HA) and quadriceps (QU) muscles. The task of walking performed during FT or BT was compared to the NPT condition and evaluated using the analysis of displacement and velocity of the CM relative to space (CM(S)). Timing and divergence in the trajectories of the angular displacement at the ankle, knee and hip bilaterally, trunk segment rotation and CM displacement relative to the lateral malleolus (CM(R)) during FT and BT were compared to the NPT condition. This was determined for both the young and the elderly. Magnitude and duration of the different phases in the trajectories of angular displacements at the ankle, knee and hip bilaterally, trunk segment rotation and CM(R) displacement were compared between NPT, FT and BT for both the young and the elderly.

Corrective muscle responses were observed in all muscles. The timing and patterns of muscle responses were similar in both groups. Some muscles showed an increase in activity relative to the NPT condition and others showed a decrease. The onset latency and sign of the muscles were specific to the direction of the disturbance and where in the gait cycle the disturbance occurred. In both groups, there was a change in swing/stance duration and swing distances which, as compared to the NPT condition, was opposite in direction for BT and FT. The BT condition exhibited shorter duration and distance while the FT condition exhibited longer duration and distance. In both groups the change in trajectories of angular displacements and CM(R) displacement that occurred in response to the FT or BT was a phase shift with no significant change in pattern or magnitude. The phase shift was evident during right PT1 and PT2. During BT, where the foot was displaced backward relative to the CM, an initial increase in the rate of angular displacement at the ankle, knee and hip resulted in a phase lead. During FT, where the foot was displaced forward relative to the CM, a decrease in the rate of angular displacement which resulted in a phase lag was observed. These findings were consistent with the assertion that corrective muscle responses are organized into locomotor like patterns. The process organizing balance adjustments and locomotor activities is integrated and independent of where in the gait cycle the disturbance occurrs.

TABLE OF CONTENTS

ACKNOWLEDGEMENTS

This research program was carried out under the supervision of Dr. Tony Szturm. I would like to sincerely acknowledge Dr. Szturm for his guidance, encouragement and support thoughout the program. The generous sharing of his expertise, and his enthusiasm were appreciated.

I am grateful to the members of my committee, Dr. Dean Kriellaars and Dr. Pat Nance, for their invaluable instruction. Their assistance and willingness to provide support in completing this thesis were appreciated.

My friends, family and collegues provided assistance to me at various times throughout my study. I am especially grateful to my employers and colleagues Ross Noble and Doug Corder for their encouragement and for allowing me time away from work to finish my thesis. I would like to extend my gratitude and appreciation to Heather Parrott-Howdle and Pam Homenick for their help, friendship and continual support. Their knowledge of computers and editing help was invaluable. I am thankful that my parents Eric and Ivy Riediger taught me a strong work ethic and for their love and support throughout the years.

A special thanks to the people who volunteered their time to participate in this study. Without their cooperation and assistance the study would not have been possible.

LIST OF FIGURES

LIST OF TABLES

CHAPTER 1 - REVIEW OF LITERATURE

1.1 INTRODUCTION

Balance impairment is commonly implicated as a major contributing factor to falls in the elderly (Studenski and Rigler, 1996; Maki, Holliday and Topper, 1994; Gu et al, 1996; Lord et al, 1994; Maki, 1996; Maki and McIlroy, 1996). Changes in the neuromuscular system which lead to falls can additionally cause injuries of varying severity from soft tissue injuries, such as hematoma and lacerations, to fractures and death. Fractures occurring in the elderly population as the result of a fall are common in the neck, humerus, wrists, vertebrae and ribs (Salgado et al, 1994). The actual incidence of falls is unknown. However, the incidence of falls leading to injury is known, since those injuries leading to immediate emergency attention are generally of the variety which are reported formally and systematically (Gryfe, Amies and Ashley, 1977).

The issue of "fear of falling" has a relationship to quality of life as it may cause a decline in the level of independence of an individual, thus limiting mobility and reducing social interaction (Salgado et al, 1994) (Tinetti, Speechley and Ginter, 1988) (Murphy and Isaacs, 1982) (Myers et al, 1996) (Vellas et al, 1987). Fear of falling may additionally contribute as a risk factor for falls due to the fact that a decrease in movement will result in a reduction in physical conditioning (Walker and Howland, 1991). Walker and Howland completed a pilot study which was intended to assess the incidence of falls and the prevalence of fear of falling among community-living elderly people. They interviewed 115 people, 24% men and 76% women, whose mean age was 78 years; 53% of the population reported experiencing a fall within recent years and 69% of these individuals reported that they suffered an injury in the fall. They also reported that fear of falling, independent of reported risk factors, contributed to a limitation in activity ($p=0.06$).

Falls can also be associated with mortality. Winters et al (1996) reveal that a 1991 study implemented by Statistics Canada reported 185 deaths per 100,000 in the over-80 population. Gryfe, Amies and Ashley (1977) provide a long-term study of falls among the elderly during a five-year period for an active, independent, ambulatory, institutionalized population. The subjects in this study were a population of individuals over the age of sixty-five. (This study did not have the same problems as studies of open communities of the aged.) Studies involving open communities are complicated by limitations on reporting and observation. These investigators found 668 incidents of falls per 1000 and thus determined that 45% of the subjects had at least one fall. Half of the falls resulted in no injury, 28% resulted in trivial soft tissue injury, and 17.5% resulted in severe morbidity. The limitation of this study is that the residence was designed to minimize falls. For example, the residence had better illumination, grab rails in the corridors and safety floor coverings to minimize the risk of falling (Gryfe, Amies and Ashley, 1977). It can be inferred then that in an independent, non-institutionalized population the percentage of falls could rise dramatically. Falls cover a multiplicity of very different event, such as falling down stairs, falling when rising from a chair, and tripping (Hadley, Radebaugh and Suzman, 1985). However, epidemiological studies show that falls take place most often during walking (Tinetti, Speechley and Ginter, 1988). Pursuant to these facts it is very important to study the reasons for falls in the elderly population in an attempt to develop programs to reduce such accidents. In turn, these efforts would have the additional benefit of reducing healthcare costs and generally increasing the quality of life of our aging population.

Normal age-related changes in different components of the nervous system and musculoskeletal system can increase the risk of falls in the elderly by reducing the ability of the postural control system to function optimally (for review – Craik and Oatis, 1995; Rubenstein et al, 1988; Woollacott and Tang, 1997). Balance, or body stability, is a prerequisite to performing daily living skills and maintaining the mobility necessary for independence. Stability is comprised of a complex interaction of sensory information, central organization, and efferent neurological responses to appropriately directed mechanical responses (Studenski and Rigler, 1996). Sensory system changes include visual, vestibular and proprioceptive changes. With aging the visual system exhibits reduced acuity, contrast sensitivity (low and intermediate spatial frequencies), depth perception and dark adaptation (Sekuler and Hutman, 1980). The vestibular system evinces a loss of labyrinthine hair cells, ganglion cells and nerve fibres. Additionally, there is a reduction in the vestibular ocular reflex (Paige, 1991). Proprioceptive or joint-position sense is also found to deteriorate with advanced age (Skinner, Barrack and Cook, 1982). Physiological changes in the elderly also include alterations in cutaneous touch (Kenshalo, 1986).

The aging effect on the central and peripheral nervous systems results in the progressive loss of neurons, depletion of neurotransmitters within the basal ganglia, and involuntary changes in the dendritic tree of motor neurons in the spinal cord (Scheibel, 1985). This slows information processing and decreases nerve conduction (Stelmach and Worringham, 1985). The musculoskeletal system is prone to a reduction in muscle strength (Myers and Gonda, 1986) which coincides with a decrease in the size and number of fibres and motor neurons with age. The effect of this is to slow contraction, and such a factor may be responsible for the reduction in the rapid force production needed for postural responses. Hakkinen et al (1996) found this to be true. In their investigation they tested forty-eight healthy men and women: twenty-four had a mean age of fifty years and twenty-four had a mean age of seventy years. This investigation considered the elements of muscle cross-sectional area, percentage of body fat, maximum voluntary force and EMG activity of the knee extensors during bilateral and unilateral isometric contractions. The results showed that the maximum bilateral knee extensor force values were related to the cross-sectional area of the muscle and that both values decrease with age. The women over seventy demonstrated the lowest values. It was also found in this investigation that the peak normal maximum muscle strength is evident between the ages of twenty and thirty years. There is then a slight decrease in strength over the next twenty years. With advanced age, starting about the sixth decade, there is the most dramatic decrease in muscle mass and strength. Muscle stiffness is responsible for decrease in the flexibility and range of motion noted with age. This physiological change may impact on the prospects for compensatory postural responses (Judge and Ounpuu, 1996). Judge and Ounpuu concurrently noted postural changes associated with loss of intervertebral disc height with displacement of the centre of mass relative to the pelvic girdle, but this finding was not substantiated with data.

Since the average age of Canadians is increasing due to low fertility rates and increased life expectancy (Foot and Stoffman, 1996), there is a growing interest in age-related changes in gait and their relationship to balance and falls. It is known that falls in the elderly cause injuries, and research has largely attempted to quantify the extent of the problem. In the above mentioned studies there is no direct evidence to show that these age-related changes in the peripheral and central nervous systems cause falls. It is, however, implied that changes in the postural control system are possibly an important cause of not only falls but general instability and consequently loss of independence due to fear of falling.

1.2 POSTURAL CONTROL DEFINED

Posture is a multidimensional task. Postural control, body stability or balance is often defined as the ability to maintain and control the position and motion of the total body centre of mass (TBCM) relative to the base of support (Maki, 1996) (Winter, Patla and Frank, 1990) (Mackinnon and Winter, 1993) (Winter and Eng, 1995) (Jian et al, 1993) (Winter et al, 1993) (Maki and McIlroy, 1996). This is achieved by generating muscle forces to counter weight and reaction forces in each body segment (Putnam, 1991). The postural goal changes during ambulation due to the fact that as we move from point *A* to point *B* we must accelerate the centre of gravity forward. In essence we start this motion by initiating a forward fall which forces us to control the TBCM relative to a narrow and changing base of support (Winter, 1991).

With regards to postural control, the nervous system must deal with a number of different task requirements and stability conditions. Stability must be maintained during voluntary movements when the base of support is stationary or when the base of support is moving. This system also restores body stability during unexpected balance disturbances or sudden loss of balance. In either case balance or body stability is achieved when there is an equilibrium of internal and external forces acting about each body link. The external forces are as follows: a) gravity-dependent or weight forces which accelerate body segments; b) motion-dependent forces or joint reaction forces which reflect the inertial effect of acceleration of one body segment on other distal and proximal body segments, and c) ground reaction forces acting on the foot segment during weight bearing. The internal forces are: a) active muscular forces under the control of the nervous system; b) passive visco-elastic forces of tissues spanning a joint, and c) bony contact forces.

In general, there are two basic mechanisms for balance control. Initially we must consider the feed forward voluntary control system, which is an anticipatory or predictive response thought to play a significant role to ensure stability during voluntary movements. In this case one can plan the stability requirements in advance of the movement. For example, an individual is able to modify walking patterns before contact with an obstacle that suddenly appears in the path of walking. Patla and Rietdyk (1993) proposed in their research that there would be a change in limb trajectory or locomotion patterns at the EMG and kinetic level while subjects step over obstacles of different height and width. These authors found that the trajectory was modulated for obstacle height but not width. It is not positively known what centres are responsible for producing these preprogrammed responses but they are thought to be cortical, as preparatory responses precede the onset of the prime mover of the focal voluntary movement. (This mechanism of postural control will not be examined in this research paper.) The second system which is responsible for the restoration of body stability following sudden unexpected postural disturbances has been referred to as an automatic or involuntary system. These muscle responses leading to restoration of equilibrium occur within 100 ms following the onset of the balance disturbance. Due to the short latency of these balance reactions this system has been considered a feedback control system.

Early work by Nashner (1976) enlisting EMG and force plate analysis defined consistent stereotypical single-link or ankle synergies. Nashner found that subjects who were translated in the anterior/posterior direction exhibited the first postural adjustments at the ankle in 100-120 ms following the onset of the disturbance. A subsequent study by Nashner (1977) using the same methodology found that during stance muscles in the legs are activated in a fixed distal to proximal pattern. It should be noted that these studies were limited to slow movements with a peak velocity of 13 cm/sec. Szturm and Fallang (1993) examined forward and backward translations (FT, BT) at varying degrees of difficulty (increasing acceleration/velocity). They discovered that, for both forward and backward translations, multi-segmental movement synergies were required for fast or large perturbations. In regards to FT, the strategy included knee flexion followed by hip extension, and then dorsiflexion. During BT, the pattern consisted of hip flexion, plantarflexion of the ankle, and then knee extension. It was also demonstrated that the basic pattern and timing of muscle responses of lower limb muscles and angular displacement did vary as a function of the acceleration of platform displacement; additionally, the magnitude of the corrective balance responses did increase with increasing platform acceleration. Within a wide range of disturbance magnitudes, balance reactions were scaled in magnitude.

It is largely unknown what neural centres are responsible for automatic balance control. Due to the short latency of the balance reaction it has been theorized that the brain stem is involved. It was proposed by Horak and Nashner (1986) that postural responses are centrally determined. The central neural network may function to regulate global parameters such as position/velocity of centre of foot pressure/total body centre of mass (CFP/TBCB) (Nashner, 1977; Horak and Nashner, 1986; Horak, Diener and Nashner, 1989; Crenna and Frigo, 1991; Macpherson, Rushmer and Dunbar, 1986).

In regard to different modes of sensory information, there is no channel that can act singularly to trigger a balance reaction (Stelmack and Worringham, 1985). The postural mechanism requires multiple sources of sensory information to determine task requirements if there is a threat to balance. The sensory information is directed to integrative centres within the nervous system wherein the most rapid and efficient postural response is determined. In consideration of these factors postural control involves three neural subsystems or processes. The first is comprised of the sensory receptor systems which include the visual system, the vestibular system and the somatosensory system (e.g. muscle spindles, joint and cutaneous afferents), which are responsible for proprioceptive and exteroceptive input. Each system has its own frame of reference. The vestibular receptors have the only fixed reference frame, detecting head position relative to gravity, vector or vertical. The visual system has a relative reference frame. The proprioceptive component of the somatosensory system (i.e. muscle spindles) provides information regarding relative orientation and motion between the body and the environment. The exteroceptive component of the somatosensory system (i.e. cutaneous afferents of the feet) supplies information about external reference points such as the location of the support surface and the distribution of foot pressure. It has been demonstrated during quiet standing that all sensory information can play important roles in the selection of postural movement strategies (Horak, Nashner and Diener, 1990). Black et al (1988) noted that there are two components required for quiet stance. The first is accurate sensory and perceptual information about body movements relative to external references (such as vertical); the second component is accurate motor information to maintain or correct the centre of mass (COM) position and motion relative to earth vertical.

A number of investigators have applied sensory conflict conditions designed to examine the related roles of different sources of the orientation and motion cues given by the different sensory receptors while maintaining standing balance. For example, healthy adults and subjects with bilateral vestibular deficit were exposed to misleading sensory information as the investigators rotated the support surface and/or the visual surround in proportion to ankle-angular displacements and body sway (Black et al, 1988; Peterka and Black, 1990a). The experimenters provided misleading sensory information from at least one system. The results of these studies indicated that when visual and somatosensory inputs were available, the vestibular deficit patient could maintain sway within usual limits. However, during sensory conflict situations without visual and somatosensory reference, there was significantly greater body sway of patients with unilateral or bilateral vestibular deficit as compared to healthy control subjects, and often these patients fell.

Woollacott, Shumway-Cook and Nashner (1986) studied the role of sensory organization in postural control in healthy young adults and compared the results to healthy elderly subjects. The elderly population included eight females and four males between the ages of 61 and 78; the young population included eight females and six males between the ages of 19 and 38. The subjects had no history of neurological disturbances. Postural stability was measured under six sensory conflict conditions and a performance ratio was thus determined. The performance ratio was a numerical integral of the rectified anterior/posterior (A/P) sway possible without loss of balance. The six conditions were as follows: a) normal fixed support; b) normal support, eyes closed; c) normal support, visual servoed; d) support servoed, eyes open; e) support servoed, eyes closed, and f) support servoed, vision servoed. It was discovered that the elderly generally had a reduced ability to balance under the influence of conflicting sensory information (i.e., inappropriate visual or somatosensory information). Under conditions in which the support inputs were normal, the sway index of the elderly was slightly but not significantly higher than the young adults. In some cases individuals in the elderly group lost their balance. Six out of twelve elderly individuals lost their balance when the support surface was servoed and their eyes were closed as opposed to two who lost their balance when both the support and the vision were servoed. These results indicate that sensory processing of conflicting orientation and motion cues differ with age.

A number of studies have examined balance reactions to sudden disturbances in healthy elderly subjects. Investigators have enlisted the moving platform paradigm during perturbed standing when comparing balance reactions to unexpected disturbances between a population of young and elderly healthy subjects. Peterka and Black (1990b) studied EMG latencies, body sway, amplitude and timing of changes in centre-of-pressure (COP) displacement to sudden forward and backward translation. The subjects included 214 individuals between the ages of 7 and 81 years. This study did not give information about the numbers of elderly subjects included in the tests, except by indicating that the ages were uniformly distributed over the entire age range. The investigators also suggested that these subjects were selected from the normal, healthy population. These investigators found a small increase in EMG latencies and time to reach the peak amplitude of COP response with increasing age, but the amplitude of COP exhibited no change. Any differences were not statistically significant. It must be noted that the examiners used slow, small platform forward and backward translations where it has been shown that only ankle synergies were needed. Basically, the findings indicated no difference between the young and the elderly. Woollacott, Shumway-Cook and Nashner (1986), in the second portion of their study mentioned previously, investigated a healthy control group and a healthy elderly group with no history of neurological disturbance during forward/backward translations and rotations. Peak velocity platform translations were 30 cm/sec attained in .25 sec and peak velocities of rotations were 20 deg/sec attained in .25 sec. These investigators examined the timing and the amplitude of muscle responses to postural perturbations. It was found that in the elderly there was an increase in absolute latency of distal muscle responses. The onset latency was 109 (±9) for the elderly group and 102 (±6) for the young adult population. This difference in onset latency is slight but significant. Also, five out of twelve in the elderly group exhibited temporal reversals, i.e., activation of proximal muscles before distal muscles. In this study the disturbances appear to be slow and small but the parameters are not well defined. These investigators did not provide the mean age of the subjects in the elderly population or the distribution of ages in this population. It is, therefore, not known how many of the subjects were sixty-one years or older. This fact is of utmost importance since other studies suggest that significant changes only start to occur in the sixth decade (Hakkinen et al, 1996). Woollacott et al (1986) discovered a meaningful effect between groups in this study, whereas Peterka et al (1990b) found a less dramatic effect; this is due to the fact that in the Peterka study the sample size was much larger.

In a similar study, Keshner, Allum and Honegger, (1993) studied 10 healthy young subjects and compared them to 12 healthy elderly subjects during support surface rotations. There were some subtle differences that were noted and they were as follows: a) delayed muscle response for the soleus and the tibialis anterior in the order of 10-20 ms, for both eyes open and eyes closed, which were said to be significant. The other muscles that were studied were the neck extensor muscles. These latencies were also said to be significant, but only for eyes open. The magnitude of the muscles in the lower extremities showed reduced responses in the elderly population, while the response magnitude in the extensor muscles of the neck, although variable, were greater than in the young population; b) magnitude of the ankle torque change was not significantly different between the young and the elderly; and c) no difference in the corrective trunk angular acceleration. It should be noted that in this study no subjects lost their balance.

McIlroy and Maki (1996) examined age-related differences in stepping responses to fast platform translations. Fourteen healthy adult subjects, five between the ages of twenty-two and twenty-eight years and nine between the ages of sixty-five and eighty-one years, were examined. They studied spatial and temporal characteristics of compensatory stepping response to relatively fast unpredictable platform perturbations in both forward and backward directions. The timing measures were defined with respect to the onset of platform acceleration. The spatial and temporal characteristics included: onset latency, time to foot-off, time to foot contact, unloading phase (onset of swing leg unloading to foot-off), swing duration, step length, medial/lateral and anterior/posterior swing velocity (step length and swing duration), centre of motion (COM) displacement (M/L and A/P), and velocity of COM at time of foot contact. The onset of stepping was defined by the onset of unloading of the swing leg and the onset of M/L asymmetry. Force plate data and kinematics were used to evaluate responses to 600 ms platform translation (300 ms acceleration and 300 ms deceleration). In the forward direction the peak acceleration was 1.5 ms squared, the maximum velocity was 45 cm/sec and the displacement was 13.5 cm. In the backward direction the peak acceleration was 2.0 m/sec squared, the maximum velocity was 6 cm/sec and the displacement was 18 cm. These trials included five forward and five backward translations implemented randomly. The results showed that ninety-eight percent of the trials in both the young and the elderly population resulted in a stepping response. Temporal characteristics did not differ dramatically between populations. In the first step the temporal parameter that showed a statistically-significant, age-related difference was the medial/lateral asymmetry onset (P=0.04). The young population showed a shorter latency. Group differences in spatial characteristics were modest. For example, the elderly exhibited an increase in step length 3 cm in the anterior/posterior direction and 1 cm in the medial/lateral direction. (The authors did not state whether the difference is statistically significant.) Forty-nine percent of the trials which revealed stepping responses featured multi-stepping responses. Multiple stepping was twice as common in the elderly population (63% as opposed to 35% in the younger group)(P=0.01), and this largely consisted of lateral stepping. In 13% of the trials individuals in the elderly group had to grab the handrails, which is a response which did not occur in the younger population. The authors concluded that multiple stepping might be a predictor of risk of falling in the elderly population. The fact that the elderly used the handrail may be more of a fall predictor as only 35% of the young used multi-stepping for balance. This study is limited to simple force plate and time measurements in investigating the responses to fast platform translations during standing. It is, however, examining the 'second line of defence' response (which is to step and move the base of support under the COM), rather than moving the COM back over the displaced base of support, as occurs when the first line of defense is utilized.

Gu et al (1996) inquired as to how 24 young (mean age 26) healthy subjects and 15 elderly (mean age 72) healthy subjects restored balance after a perturbation. The investigators looked at age-related changes in whole body dynamics, for resting and maintaining balance. The subjects were studied during the following tasks: a) while standing on a platform which accelerated in the forward direction; b) while standing for ten seconds on an 11 cm wide support beam; c) while standing on a support beam accelerating in the forward direction, and d) while standing on a springboard which was designed to decrease proprioception at the ankle. For each flat translation and beam translation trial the platform accelerated 100 ms at 1.67 m/sec sq., followed by 100 ms of constant velocity. Body segment motion was measured directly. Foot support surface reaction, or centre of foot pressure (CFP), was measured directly from a force plate. The following variables were calculated using measurements from a seven-link biomechanical model: a) the centre of mass displacement; b) the centre of foot pressure (CFP) displacement, and c) vertical ground-reaction force, location, peak angular momentum and joint torque. The results indicated that in all four tasks, with both age groups factored, the CFP was larger in amplitude and had a higher frequency variation than the COM excursions. This was determined to be a significant finding. The elderly, compared to the younger population, exhibited larger amplitude excursions of both the COM and CFP. This was not always statistically significant. The COM excursions were significantly larger in the elderly beam-standing subjects. The CFP excursion was significantly larger in both beam standing and translation. The elderly, compared to the young, had smaller average time intervals between peaks. This indicates that the elderly exhibited a significantly higher frequency of both COM and CFP excursions. The age-related CFP frequency differed in all tasks but the COM frequency was significant in the flat translation and springboard standing; generally speaking, this suggests that the elderly developed larger support surface reactions (i.e., foot A/P shear forces, foot vertical force and foot torque). The elderly subjects on average enlisted larger mean maximum joint torque changes from reference in order to restore balance. This was seen more so in beam standing, wherein it was significant in 9 out of 12 cases. The younger subjects recovered postural stability more rapidly as they recovered with fewer oscillations in the excursions. The results imply that elderly subjects with no neurological or musculoskeletal impairment do in fact differ from healthy young subjects during perturbed stance. The authors also suggest that the disturbance in this study was of a high-end variety; however the target velocity in this study was 16 cm/sec, which is a low-end or relatively small disturbance.

Challenges to balance during the gait cycle are much different than the challenges during standing. In short, while standing the goal is to maintain the COM within the narrow base of support. The goal of walking is to maintain the forward progression of the TBCM in the sagittal plane while regulating the upper body balance. This is difficult due to the fact that the large mass of the head, arms and trunk are located two-thirds of the body height above the ground (MacKinnon and Winter, 1993) (Winter et al, 1993). With gait initiation the COM moves forward outside this base. The safest period during steady state gait is double support, and the more difficult period to stabilize is a single support position. During single support position the COM moves forward along the medial border of the foot and this yields an inherent state of instability.

All the studies mentioned previously provide us with valuable information regarding the effects of age on balance control. However, it is questionable whether there is a direct relationship between standing balance and balance control during gait. For instance, many researchers found that it is not automatically possible to transfer standing balance skills to balance skills necessary for walking, as evidenced by studies involving stroke patients (Winstein et al, 1989; Malouin et al, 1992).

The focus on gait studies has primarily been concerned with a young and healthy population, and from such investigations information has been gathered in the areas of kinematics, electromyographics and kinetic profiles. Patla et al (1993) developed the 'stepping response test', which is based on a common response to a disturbance. The subject population included 12 young and 23 elderly individuals. The subjects stood on a force plate with pressure-sensitive mats positioned at the front, sides and back of the plate. A cue box with three lights was positioned at eye level. When the top light was on the subject stepped forward; the side light indicated to the subject to step sideways and the bottom light was a stimulus for the subject to step back. The time between directions varied and the different stepping directions varied as well. The experimenters measured reaction time, weight transfer time, and the time histories and peak magnitudes of the ground reaction force. The results indicated that the elderly subjects had an increased reaction time. They also took longer to transfer weight and initiate limb lift-off in all directions, especially in the forward direction. The elderly had statistically significant lower peak vertical forces during steps made in the forward direction, and they took longer to reach peak force in all three directions in comparison to the younger population.

There is little known about corrective balance reactions during unexpected mechanical perturbation while walking. Mechanically-perturbed human gait is not often examined as it is difficult to control the timing and magnitude of the perturbation. Nashner (1980) used the moving platform paradigm to examine the earliest adjustment in activity during unexpected platform movements. The platform movements included forward and backward translations, elevation and depression of the force plate, and rotations. This investigator studied different phases of gait by incorporating different delays (no delay, 175 ms, 350 ms, 575 ms) in the platform movement. These delays represented heel strike, the beginning of single support, mid-stance, and the beginning of double support phase. Nashner (1980) studied EMG from bilateral gastrocnemius and tibialis anterior and analyzed the forces and movements associated with each perturbation. It was found that there was an increase in muscle activity of the perturbed leg similarly observed in perturbed standing. The response was pronounced in the early phase of gait and significantly reduced in the later stance phase. Nashner (1980) concluded that locomotor centres use the same postural synergies during perturbed standing as during perturbed gait. This can be challenged mechanically since the task of regulating standing balance and balance during gait is very different. The finding that balance demands differ between different phases of gait is supported by many investigators. Eng, Winter and Patla (1994) studied how humans recover from a trip during the early and late-swing phases of walking. In this case the investigators looked at distinct movement strategies that occurred after the swing foot encountered an object. The subjects included ten young healthy men who were instructed to walk at a self-directed pace, stepping on two consecutively placed force plates. There were thirty trials in total, which included ten early-swing perturbations, ten late-swing perturbations and ten trials with no perturbations. A thin, flexible metal strip which caused the trip was triggered to rotate 90 degrees upon weight acceptance on the first force plate. Sixteen muscles were recorded, which included eight from each of the stance and the swing leg. The force plate data provided temporal measures of the gait patterns. Temporal and kinematic parameters included step length and duration, maximum-swing hip flexion, maximum-swing knee flexion, maximum-swing ankle dorsiflexion, and maximum trunk angle. The average profile of the joint angles from the ten unperturbed trials was subtracted from the data provided by the perturbed trials. It was found that early-swing perturbation yielded an elevating strategy of the swing leg, whereas the late swing perturbation effected a lowering strategy. The elevating strategy, or extensor strategy, consisted of an extensor pattern of the stance leg and a flexor pattern of the swing leg. The lowering strategy was accomplished by the inhibitory response of the vastus lateralis of the swing leg and/or an excitatory response of the swing leg biceps femoris, which caused rapid lowering of the foot to the ground and thus a short step. The onset latencies were also determined in this regard. In the elevating trials an early latency group was found wherein these muscles responded in 60 to 80 ms, in addition to a late latency group, which included those muscles which responded within 110 to 130 ms. The gluteus medius, medial gastrocnemius and peroneus longus of the stance leg and the biceps femoris of the swing leg were activated in the early response latency. The rectus femoris of the swing leg was the main muscle activated during the late response latency. The collaborative effect was to increase the height of the COM and the amount of time before foot contact. In response to the late perturbation there was a significant difference between the muscle latencies. In general, the early response at 70 to 80 ms included the gluteus medius, medial gastrocnemius and the peroneus longus of the stance leg. The late response at 90 to 100 ms shows an inhibition of the vastus lateralis and an excitation of the biceps femoris of the stance leg. The result of these muscle responses was to rapidly lower the leg to the ground along with shortening the step length. This study shows: a) early activation of a number of muscles between 60 to 120 ms is critical in generating functionally-appropriate corrective adjustments, and b) corrective responses (lowering vs. elevating) are phase-dependent, i.e., different patterns of muscle activation are associated with a different movement synergy, depending on the moment in the gait cycle the disturbance or perturbation was applied. It can thus be stated that noting the 'trip' or perturbation imposed at different times of swing phase will result in different stability requirements and require different corrective adjustments. This is consistent with a number of animal experiments which have examined corrective balance reactions during different types of perturbations during walking (Wand, Prochazka and Sontag, 1980; Forssberg, 1979). The results indicate that all subjects (human or animal) were able to compensate rapidly to accommodate the disturbance and that the corrective movements are adapted to the locomotor activity; that is to say the corrective movement responses observed were integrated with or into the ongoing locomotor activity in order to overcome the obstacle, restore stability and to maintain the forward progression of the body.

Schillings, Van Wezel and Duysens (1996) studied stumbling reactions in human subjects, but in this case the individuals were walking on a treadmill. With the use of a treadmill in this study the timing of the perturbation and the walking velocity was controlled by the investigators. In this case, the obstacle was dropped at a preprogrammed delay after right or left heel-strike. The obstacle interrupted early swing and the drop was controlled by an electromagnet. The measurement time interval started 100 ms prior to the electromagnetic trigger. The five subjects walked on the treadmill while wearing glasses to block downward sight, which eliminated the possibility of the subjects seeing the obstacle. An EMG was recorded from the biceps femoris and rectus femoris of the left leg while an electrical goniometer measured the left knee joint angle. Markers were placed on the left shoulder, hip, and lateral epicondyle of the femur, the fibular head, lateral malleolus, calcaneus and first metatarsal bone. In all five subjects the stumbling reaction was such that the ipsilateral foot was lifted over the obstacle. An increase in plantarflexion at the ankle occurred 200 ms after the collision, along with an increase in knee flexion. The stumbling reaction was the same between trials. The first change in the knee-joint angle occurred with the passive effect of the foot striking the obstacle, and this occurred 48 ms after the perturbation. The second change in the corrective response to the perturbation occurred at 174 ms from time zero, which coincides with the increase in biceps femoris activity at 125 ms. (Time zero is the point at which the foot strikes the obstacle.) The plantarflexion changed to dorsiflexion to prepare for heel strike. It must be noted that when using a treadmill there is no forward propulsion of the subject. It is unclear what effect this has on the experimental outcome.

In the past, studies of both young and elderly gait have focused on simple measures which include cadence, stride length and stance time. It was found that with increasing age the elderly have a reduction in gait velocity at a rate of 12 to 16% per year, while their cadence was unchanged. The elderly walked slower and spent more time in stance. For men there was an increase in stance time from 59% in twenty-year-olds to 63% in seventy-year-olds. There was also a reduction in stride length from 12% to 18%; this indicates an increase in double support. In general, it was concluded that older persons take shorter steps, spend less time in single support walk with their pelvis rotated in the anterior direction, with hips slightly flexed and with feet toed out (Judge and Ounpuu, 1996; Winter et al, 1990; and Winter, 1991).

At the kinematic level, many variables have been measured in several studies using a wide variety of techniques. Kinematics includes such joint measures as linear horizontal/vertical positions, velocities and accelerations along with angular positions, velocities and accelerations. Winter (1991) discusses some of the kinematic changes that have been documented in the elderly. His subjects were eighteen fit and healthy elderly volunteers who walked with natural cadence in eight repeated trials. In turn, these volunteers were statistically compared to eleven young adults. It was found that the head A/P acceleration for the elderly was higher, which was attributed to a decreased stable platform for the visual system in the elderly. The heel contact velocity in the volunteer elderly population was higher, which may increase the possibility for slipping during weight acceptance. At the ankle, the angular change during the critical push-off period was reduced for the elderly. The older subjects initiated push-off at 12.35 degrees dorsiflexion and ended with 12.53 degrees plantarflexion. This means that there was a net angular change of 24.9% in the elderly subjects, compared to 29.3% for young adults. This suggests the elderly have diminished lower push-off power which may be accentuated by natural weakening of the muscle and/or inherent instability. At the knee, the angular changes are the same for young adults and the elderly except for terminal swing – the young adult extends to .5 degrees flexion and the elderly to 5.3 degrees flexion. This is seen as a shorter step length in the elderly. At the hip, the dynamic range of hip angle was larger for the elderly at 40 degrees versus 32 degrees for the young adult. It was found that the joint angular velocity was the same between groups, with the exception that the elderly were less variable at all of the joints.

1.3 SUMMARY

Balance impairment is commonly implicated as a contributing factor to falls in the elderly (Alexander et al, 1992; Studenski and Rigler, 1996; Gu et al, 1996; Lord et al, 1994; Maki and McIlroy, 1996). Normal age-related changes in the nervous system and musculoskeletal system reduce the ability of the postural control system to function optimally.

Most studies investigate incidents of falls and the severity of the injury. These studies provide general information about an individual's functional ability but they fail to provide important information regarding the reasons why or how balance is impaired in the elderly. With regard to postural control, the nervous system must deal with a number of stability conditions. Stability must be maintained during voluntary movements and restored during unexpected disturbances.

To date, most studies examining postural control have focused on maintaining or restoring standing balance with a stationary base of support. However, the postural requirements of many activities of daily living extend well beyond the demands of tasks featuring a fixed, stationary base of support. The challenge of controlling the motion of the centre of body mass with a moving and reduced base of support is quite different in comparison to tasks originating from a firm support base. Statistics show that the majority of falls in the elderly population occur during their main daily activity. That activity is walking. In effect, few studies have examined postural control during locomotion and/or tested for age-related effects on recovery of body stability during locomotion.

What is known from studies of standing, stationary subjects is twofold. There is evidence of age-related delays in the early responses of some muscles to sudden platform displacements, which may be the result of a decrease in sensitivity of portions of the sensory system (Allum and Pfaltz, 1985). In addition, there is an age-related slowing of reaction times or an age-related delay in automatic postural responses after the onset of a perturbation (McIlroy and Maki, 1996). Age-related changes in muscular onset latencies that impact balance reactions and movement speeds could influence functional ability, such as the progression of the COM over the base of support, i.e., swing phase and new placement of the swing leg may be strongly associated with a risk of falling (Maki, 1996).

The general focus of gait studies has been concerned with active young and healthy subjects. These studies have described the normal patterns of muscle activity, movement sequences and joint torque involved during gait initiation or steady state gait; subsequently this information has been gathered and consolidated in the area of kinematics, electromyographics and kinetic profiles. However, the fact remains that further experimental investigation is warranted as far as muscle response and an elderly population are concerned.

1.4 STUDY PURPOSE AND OBJECTIVES

The purpose of this study is to examine factors influencing changes in balance reactions, for young and elderly subjects, during the task of walking.

The objectives of this study were as follows:

1) Determine and compare performance levels of nine young and nine elderly subjects responding to unexpected balance disturbances during different phases of gait.

2) Identify and characterize the main response features of automatic balance reactions to unexpected disturbances produced by sudden forward and backward platform translations during different phases of gait (at the time of heel-off and mid-swing). Specifically the results shall identify: a) the timing of early muscle activity of the lower extremities; and b) the duration and magnitude of angular displacement about the ankle, knee and hip of the swing and stance legs.

CHAPTER 2 - METHODOLOGY

2.1 SUBJECTS

The subjects included nine healthy active young adults between twenty-five and thirty-eight years of age, and nine healthy elderly subjects between seventy-two and eighty years of age. The young adult subjects were volunteers recruited from the University of Manitoba, colleagues or employees, friends and relatives. Five of the elderly subjects were volunteers recruited from a local square dancing group; the remainder of the active elderly subjects were relatives of friends. All subjects in this study gave informed consent. They were fully informed of the requirements of the study, recording procedures and what was expected of them. The University of Manitoba Faculty of Medicine ethics committee approval was sought and granted prior to enrolling the subjects. After a short discussion with each individual, it was evident that the subjects had no history of neurological or musculoskeletal problems that would affect their balance and orientation. They had no history of episodes of dizziness nor did they undergo any ear surgery and were not taking any medication at the time of testing. None of the subjects reported a history of falling. All elderly subjects were functionally independent and living at home. All elderly subjects were also participating in at least one recreational activity, such as, square dancing, mall walking club, or lawn bowling.

2.2 EXPERIMENTAL SET-UP AND PROCEDURES

The subjects were weighed (kg) when they presented. Each subject was asked to wear running shoes, black shorts and T-shirts with cutouts to allow placement of reflective joint markers and EMG electrode placements. See Figure 1 for an illustration of the general experimental setup along with EMG electrode and marker placement. Marker placement was of utmost importance to ensure accurate data, as the markers identify the position of the axis of rotation of the individual joints and define the end points of each segment. To define the end points of a segment two markers were necessary, and to define the axis of

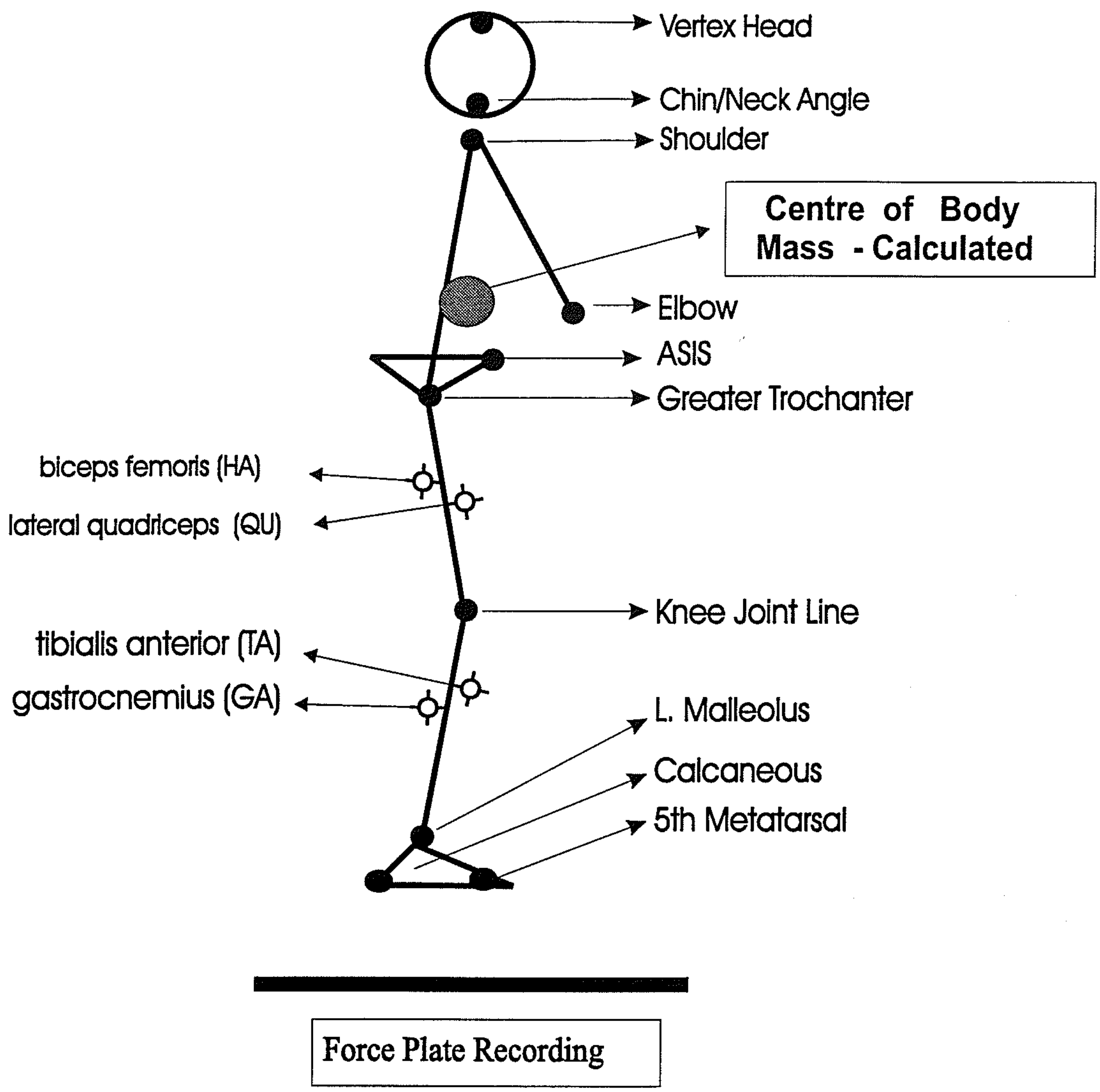

Figure 1 Illustration of the general experimental set up along with EMG electrode and marker placements.

the angle of rotation three markers were required. Markers were placed on both the right and left sides of the body. The centre of rotation of the hip joint was located transversely at the anterior-superior margin of the greater trochanter. The joint centre of the knee was located transversely over the lateral epicondyle of the femur. The transverse ankle joint was located 5 mm inferior to the medial malleolus. The lateral landmark was located 3 mm inferior and 8 mm anterior to the tip of the lateral malleolus. The two markers placed on each foot were positioned on the posterior aspect of the calcaneus and the lateral aspect of the fifth metatarsal head. An important pelvic landmark was the anterior superior iliac spine (ASIS). The centre of rotation of the shoulder joint was generally located in the middle of the humeral head, although it was difficult to ascertain as, throughout shoulder movement, translation occurs. The elbow joint marker was located on the lateral condyle and the wrist marker dorsally and midway between the radial and ulnar styloid. Two other important markers located only on the right side of the body were the vertex marker placed over the temple and the chin-neck angle marker.

Surface electrodes were placed over the muscle bellies of eight muscle groups. Care was taken to ensure that the skin surface was clean and that there was the same bilateral placement of the electrodes. Pro-wrap was used to secure the electrodes and to provide consistent electrode placement throughout the experiment. The muscle groups included the bilateral gastrocnemius (GA), bilateral anterior tibialis (TA), bilateral quadriceps (QU), and bilateral hamstrings (HA). The electrodes were connected by cables to miniaturized lightweight pre-amplifiers (30 g).

A safety harness similar to a parachute harness was used to ensure the subjects' safety in the course of the trials. Straps on the harness were attached to low-friction rollers which were free to move on two parallel metal bars, and in turn the bars were secured to joists on the ceiling of the lab. The straps were positioned such that they allowed freedom of movement, but were nonetheless tight enough to prevent ground contact in case of a fall. The subjects stood with their feet parallel and arms bent to 90 degrees at the elbow in the sagittal plane. The bending of the arms was necessary to ensure that the ASIS and greater trochanter markers were not lost to the camera during testing. This position is similar to arm position during a brisk walk. Care was taken to duplicate foot position each trial. The subjects were asked to stand with weight evenly distributed and to focus on a spot on the curtain ahead of them. A researcher stood beside the subjects at all times but did not aid them unless necessary; it was classified as a fall if the researcher aided the subject in maintaining their balance.

The subjects were randomly translated forward (FT) and backwards (BT) at two velocities over four trials while walking independently on a computer driven moveable platform. The peak linear velocities of the platform translations were V1=20 cm/s and V2=30 cm/s. For the purpose of this thesis only the translations at 30 cm/s were analyzed. The data from the trials at 20 cm/s will be used for future studies. The peak velocity of the platform translation was 30 cm/s reached in 100 ms. The amplitude of translation was set at 9 cm. The subjects were instructed to "stand still" before each trial, and they were then instructed to start walking after hearing a verbal cue to "go". At that time the subjects proceeded when it was comfortable for them to do so. They were also instructed to walk briskly at their own preferred pace, but always leading with their right leg. All the subjects were right hand dominant but it was not assessed whether the right leg was the preferred lead leg. In two of the blocks (one block equals ten trials) the subjects initiated walking while standing on the force plate and took 2 to 3 steps before stopping, i.e., before the end of runway. In the other two blocks the starting position of the subjects was behind the force plate, wherein the first step was taken directly onto the plate. In this case the subjects took 3 to 4 steps before reaching the end of the runway. Force sensing resistors (Interlink, CA) were positioned on the right plantar surface of the heel of the running shoe and on the plantar surface of the running shoe which coincided with the first metatarsal head. The heel sensor signalled heel-off/on and the toe sensor signalled toe-off/on.

The platform motion was controlled by a H3401 motor control unit (Industrial Devices Corporation) interfaced to an Amiga 2000 computer with digital to analog converter. The platform displacement was determined from recordings of a linear potentiometer housed in the cylinder of the motor. The signal from the heel sensor was used to trigger the computer, which initiated platform movements, and activate two LED lights in view of the two video cameras which were used to record the subjects' movements. The computer which was used to control the drive motor for platform translation was programmed to introduce a delay between the trigger signal (indicating heel-off) and the onset of platform motion. For this study two different delays between heel-off and onset of platform translations were used. The two delays included a 70 ms (PT1) and a 270 ms (PT2) delay after heel-off. The PT1 delay, which approximated initial swing, was the shortest delay. The PT2 delay following heel-off approximated mid-swing. In total the subjects were asked to participate in 40 walks arranged in 4 blocks of 10 trials. Two blocks originated on the force plate (ON blocks) and two blocks started off (behind) the force plate (OFF blocks). The ON and OFF blocks were alternated. In each block there were two trials without perturbations to provide baseline information. For each block the trials included two without translation (NPT), four forward (PT1 at V1, PT1 at V2, PT2 at V1 and PT2 at V2), and four backward (PT1 at V1, PT1 at V2, PT2 at V1 and PT2 at V2). The order of trials within each block was varied to avoid any ordering effect and/or anticipation on the part of the subjects. The order of trials was the same for each subject. Both sides of the body were videotaped simultaneously. The subjects were allowed to rest between blocks (3 to 5 minutes), as endurance was not an outcome measure. The control box for the LED lights also generated a square wave pulse at the time the lights were activated, and this video-sync pulse was collected on the 16 channel, 12-bit A/D converter (RC Electronics) used to collect EMG, motor-linear potentiometer signals and force plate signals. The force plate used was an AMTI OR-6 Biomechanics platform. This platform simultaneously measured 3 ground-reaction forces and 3 moments (*Fx(N), Fy(N), Fz(N) / Mx(Nm), My(Nm), Mz(Nm)*). From these calibrated values the centre of force (C of F) application about the x and y axes was calculated.

2.3 DATA RECORDING

EMG: EMG signals from the four muscle groups in both legs were amplified and processed using two four-channel differential EMG amplifiers (Biosys). The EMG signals were collected for a time period consisting of four seconds, which included 700 ms before the onset of platform motion and 300 ms after. The pre-amplifiers amplified the raw EMG signal 100 times. Subsequently, the signals were band-pass filtered at 10 Hz to 3000 Hz, full wave rectified, and then low-pass filtered at 50 Hz.

<u>Videotaping</u>: The subjects were filmed on the left side with a Sony (model CCD-V801) video camera and also on the right side using a Sony (model CCD-TR81) video camera; the shutter speeds were 1/150 s and 1/250 s respectively. The field of view (FOV) was well within the full frame. Each camera was placed on a rigid stationary box at a fixed distance and each lens was positioned perpendicularly to the midpoint of the force plate in the sagittal plane. The right camera was positioned 6 m from the midpoint of the force plate and the left camera was situated 5 m from the force plate. To highlight the reflective markers a spotlight was focused on both sides of the subjects. A blinder was added behind the greater trochanter markers to reduce the glare that otherwise occurred when the subjects walked through the midpoint of the camera focal plane. This ensured that the marker was in view throughout each trial.

The x and y coordinates of the body markers on the sides of the subjects were digitized from the video recordings with the Peak 5 Performance system. The sampling rate of the video system was 60 images per second. One hundred and forty video images were digitized for each platform translation, 20 before and 120 after onset of platform translation. The x/y coordinates of each marker from the two cameras were referenced to a common 2-sided earth fixed-reference marker. For the right camera the reference marker was located in the left upper portion of the FOV and for the left camera the reference marker was located in the right upper portion of the FOV.

A calibration rod was placed at the mid-point of the force plate where it was viewed by each camera. The calibration rod provided a scaling reference to the coordinate data and so was used as a scaling factor when digitizing each marker location. The scaling factor related pixel units to real units.

Raw x/y coordinate data was filtered at 5 Hz using a recursive Butterworth digital filter.

A custom software package was used to calculate point and segment kinematics from the set of x and y coordinates. These parameters included linear x and y displacement, segment angular displacement with respect to earth horizontal and joint angular displacement for ankle, knee, hip and hip/trunk. The ankle angle was defined by the intersection of the line from the fifth metatarsal head marker to the ankle marker and the line from the ankle to the knee marker. The knee angle was defined by the intersection of the line from the ankle marker to the knee marker and the line from the knee marker to the greater trochanter. The hip angle was formed by the intersection of the line segment from the knee marker to the greater trochanter and the line from the greater trochanter to the ASIS. The hip-trunk angle was formed by the line intersecting the segment from the knee to the greater trochanter and the line from the greater trochanter to the shoulder. These anatomically-based angles produced positive angles when moving counterclockwise relative to the positive x and y axes, and negative angles when moving clockwise relative to the positive x and y axes.

The end-point coordinate data and the anthropometric data obtained from Chandler (1975) was used to compute the location of the x and y coordinates of the total body centre of mass displacement (TBCM) in the sagittal plane.

2.4 DATA ANALYSIS

2.4.1 General Characteristics

The duration of the first two half-cycles of gait (swing phase and stance phase) for the right leg (which was the initial swing leg) and the left leg was determined. This study looked at recovery from the perturbation in both the first half and the second half-cycles. From this information it was possible to determine whether recovery occurred completely in the first half-cycle or whether corrections were also present in the second half-cycle. This provided a measure of effects of a perturbation on the duration of the gait cycle, thus clarifying how quickly the young and elderly groups recover respectively from such perturbations.

Step length was also determined during all trials. We measured the step length of the initial swing leg and the swing of initial stance leg (second half-cycle), and this also provided a measure of the effect of perturbations on the gait cycle.

2.4.2 Electromyography

The EMG analysis was limited to the onset times of early muscle responses to the balance disturbance; for this purpose the methods outlined by Eng, Winter and Patla (1994) were employed, as described below. The onset of the EMG signals was quantified for each of the eight muscle groups on the right side relative to the start of the platform movement. Seven muscle groups were used on the left side instead of the original eight, as the lead from the swing leg GA muscle was not functioning properly. This portion of the study focussed on the timing of early muscle responses that represent automatic balance reactions; for this reason, onsets within a time period of 200 to 250 ms from the initiation of platform motion were recorded. The EMG records of individual perturbed trials were subtracted from the average of the unperturbed trials to obtain the difference waveforms. As a result, early muscle responses from the ongoing EMG that were present in the first 200 to 250 ms after the onset of platform motion were identified. The onset latency was defined when muscle activity was greater or less than 2 SD (standard deviations) from the baseline of the difference waveforms. Time zero was set to the onset of the platform displacement.

2.4.3 Total Body Centre of Mass Displacement (CM)

The CM variables which were quantified are as follows:

a) The peak magnitude of the centre of mass relative to space (CM(S)) and the centre of mass relative to the base of support (CM(R)) displacement at first right (R) heel-contact, which was at the end of the first half of the cycle (initial swing phase).
b) The magnitude of the CM(S) and CM(R) displacement from R heel-contact to R toe-off was the end of the second half-cycle (stance phase).
c) The velocity of CM(S) reached at R heel-contact and R toe-off. The velocity of the CM during gait initiation has been examined by Jian et al, (1993). They found that the velocity of the CM in the first half-cycle of gait to be 90% of steady state velocity, and the velocity of the CM at right toe-off or the second half-cycle was 100% of steady state. These variables

were expressed as a percentage of the average of the eight unperturbed walking trials (four walks from the ON blocks and four walks from the OFF blocks).

2.4.4 Kinematics

In order to identify the general movement synergy(ies) used to restore balance, deviations in the trajectory of the angular displacement relative to the unperturbed trials of both the swing and the stance legs were identified.

For both perturbed and unperturbed trials in the first and second half-cycles, the following points from trajectories of angular displacements (or segment rotations in case of trunk) were quantified. These represented peaks or transition from flexion to extension and vice versa:

a) time and magnitude to maximum angular displacement in flexion and extension at the hip, knee and ankle; similarly, the maximum trunk segment angular displacement was quantified.

b) duration to maximum displacement for hip flexion/extension, knee flexion/extension and ankle dorsi/plantarflexion.

2.5 STATISTICAL ANALYSIS

Statistical analysis of the data determined if any comparisons made regarding the automatic balance responses were statistically different, and related the outcomes to the objectives of this study. The independent variables in this study were Group (G) and Trial (T) conditions (NPT, FT, BT). The two groups were the Young and the Elderly subjects. The trial condition NPT means no perturbation. An alpha of 0.05 was used.

The first thing that had to be determined was if the task of walking was performed in the same manner between the Elderly group and the Young group. The unpaired t-test was used to determine statistically significant differences of the dependent variables between the two independent groups in the study. The dependent variables used were: a) R/L swing/stance duration and L/R swing distances; b) CM variables as stated in the results section (i.e., peak magnitude of the CM(S) displacement and velocity); and c) timing and magnitude of angular displacements at the ankle, knee, and hip during the swing and stance phases of the gait cycle, for both the right and left side.

Prior to any interpretation of results it had to be determined if the postural disturbance due to the sudden platform displacements produced similar initial biomechanical effects in the body in both groups. Specifically, it had to be determined if timing and magnitude of the initial body disturbance was the same for both groups and if the body disturbance changed between FT and BT conditions. From an angular displacement perspective it was shown how the normal trajectory of the joint angle changed from walks during forward and backward translation. It was determined when these changes occurred and if the changes in trajectory were due to the disturbance. If the changes were due to the disturbance, they were within 50 ms of the disturbance. An unpaired t-test was used to determine statistically significant differences of the dependent variables between the two independent groups in the study. The dependent variables used were: a) linear displacement of the heel marker in the X direction; b) CM variables as stated in the results section (i.e., peak magnitude of the CM(R) displacement); and c) joint angular displacement at the ankle, knee and hip.

A split plot analysis of variance (ANOVA) was used to examine G, T and G*T on a set of dependent variables that represent timing, duration and amplitude modulation, as one or more than one of these variables could be responsible. Both swing and stance were considered in an attempt to see how long it took the body to return to normal after a disturbance. Dependent variables used were: a) R/L swing/stance duration and R/L swing distances; b) CM variables as stated in the results section (i.e., peak magnitude of the CM(S) displacement and velocity); and c) timing and magnitude of angular displacements at the ankle, knee and hip during the swing and stance phases of the gait cycle.

An unpaired t-test was used to examine group differences in the onset of corrective muscle responses to FT and BT.

The Systat statistical package (version 5) was used for all statistical procedures.

CHAPTER 3 - RESULTS

3.1 GROUP COMPARISON DURING NPT CONDITION

A group comparison of the results for the NPT condition was important to determine whether the walking task without perturbations was performed the same between the Young and Elderly groups. The main result was that there was very little difference between the Young and the Elderly group in how they performed the experimental task of walking. This was evident from evaluation of both the left and the right side of the body.

3.1.1 Swing/Stance Duration and Swing Distance

Table 1 presents group means (SD) of swing and stance duration and distances. This table includes the right and left side of the body for both the Young and the Elderly groups.

With the exception of right swing distance, there were no statistical differences in swing distance and swing/stance duration between groups. Right swing distance showed a G effect ($p<0.05$), with the Elderly group having a shorter step length than the Young group. The group means however reflected a tendency for the Elderly to walk with a reduced duration and shorter distance relative to the Young in the NPT condition.

3.1.2 Angular Displacement

Tables 2A and 2B present group means (SD) of timing and Tables 3A and 3B present group means (SD) of magnitudes for all conditions including the NPT condition. Figures 2A, 2B, 2C and 2D present the ensemble group averages of angular displacements for all conditions including the NPT condition. As evident in Figures 2A, 2B, 2C and 2D, the ensemble average waveforms of the Young group and the Elderly group are similar in appearance.

Table 1 Group means (SD) of swing/stance duration (seconds) and swing distance (meters), and results (p-values) of Split Plot ANOVA; G, T, and G*T effects. The table includes BT, NPT and FT for PT1 and PT2. "Y" indicates Young group and "E" Elderly group.

PT1	Y-BT	Y-NPT	Y-FT	E-BT	E-NPT	E-FT	G	T	G*T
R Swing Distance	0.58 (0.06)	0.61 (0.05)	0.70 (0.05)	0.53 (0.08)	0.57 (0.12)	0.63 (0.12)	0.05	0.001	NS
R Swing Duration	0.44 (0.02)	0.50 (0.03)	0.54 (0.03)	0.42 (0.05)	0.47 (0.03)	0.50 (0.05)	NS	0.01	NS
R Stance Duration	0.58 (0.50)	0.59 (0.06)	0.59 (0.05)	0.56 (0.05)	0.56 (0.07)	0.58 (0.06)	NS	NS	NS
L Swing Distance	1.22 (0.18)	1.11 (O.33)	1.11 (0.26)	1.22 (0.17)	1.14 (0.12)	1.08 (0.20)	NS	NS	NS
L Swing Duration	0.63 (0.08)	0.60 (0.14)	0.53 (0.14)	0.62 (0.12)	0.58 (0.44)	0.53 (0.14)	NS	0.01	NS
L Stance Duration	0.33 (0.07)	0.45 (0.06)	0.52 (0.04)	0.39 (0.06)	0.44 (0.07)	0.50 (0.08)	NS	0.001	NS

PT2	Y-BT	Y-NPT	Y-FT	E-BT	E-NPT	E-FT	G	T	G*T
R Swing Distance	0.59 (0.05)	0.63 (0.05)	0.67 (0.05)	0.56 (0.09)	0.61 (0.12)	0.65 (0.12)	NS	0.001	0.033
R Swing Duration	0.43 (0.02)	0.50 (0.03)	0.54 (0.02)	0.44 (0.07)	0.49 (0.03)	0.52 (0.04)	NS	0.017	NS
L Swing Distance	1.22 (0.15)	1.11 (0.33)	1.17 (0.29)	1.18 (0.22)	1.14 (0.12)	1.10 (0.23)	NS	NS	NS
L Swing Duration	0.60 (0.06)	0.60 (0.14)	0.56 (0.15)	0.50 (0.11)	0.58 (O.04)	0.52 (0.12)	NS	NS	NS
R Stance Duration	0.57 (0.04)	0.60 (0.06)	0.64 (0.07)	0.55 (0.04)	0.56 (0.07)	0.62 (0.09)	NS	0.004	NS
L Stance Duration	0.39 (0.09)	0.45 (0.06)	0.49 (0.08)	0.41 (0.10)	0.44 (0.07)	0.46 (0.10)	NS	0.001	NS

Table 2A Group means (SD) of absolute onset times and time to peak angular displacements (seconds) for the right and left sides. Results (p-values) of Split Plot ANOVA are presented on right side of table. The table includes BT, NPT and FT for PT1 condition. "Y" indicates Young group and "E" indicates Elderly group.

PT1	Y-BT	Y-NPT	Y-FT	E-BT	E-NPT	E-FT	G	T	G*T
R-SWING									
onset hip flex	-0.04 (0.04)	-0.02 (0.04)	-0.04 (0.04)	-O.O3 (0.03)	-0.04 (0.02)	-0.05 (0.04)	NS	NS	NS
time peak dorsiflex	0.06 (0.004)	0.06 (0.005)	0.06 (0.005)	0.06 (0.06)	0.07 (0.007)	0.06 (0.007)	NS	NS	NS
time peak knee flex	0.14 (0.04)	0.16 (0.05)	0.14 (0.04)	0.14 (0.03)	0.16 (0.03)	0.14 (0.04)	NS	NS	NS
time peak hip flex	0.24 (0.03)	0.24 (0.05)	0.24 (0.04)	0.29 (0.12)	0.27 (0.05)	0.30 (0.10)	NS	NS	NS
RIGHT STANCE									
time peak plantarflex	0.43 (0.04)	0.52 (0.03)	0.59 (0.04)	0.48 (0.05)	0.49 (0.06)	0.51 (0.06)	0.01	0.01	NS
time peak knee ext	0.41 (0.05)	0.42 (0.03)	0.42 (0.05)	0.47 (0.04)	0.51 (0.03)	0.53 (0.06)	0.01	NS	NS
onset hip ext	0.37 (0.05)	0.39 (0.10)	0.39 (0.05)	0.35 (0.07)	0.38 (0.10)	0.41 (0.13)	NS	NS	NS
time peak knee flex	0.57 (0.07)	0.63 (0.06)	0.65 (0.07)	0.53 (0.11)	0.88 (0.13)	0.73 (0.09)	NS	0.001	NS
time peak dorsiflex	0.88 (0.015)	0.92 (0.17)	0.97 (0.15)	0.83 (0.19)	0.92 (0.14)	1.03 (0.17)	NS	0.001	NS
time peak knee ext	0.83 (0.07)	.88 (.015)	0.95 (0.07)	0.87 (0.10)	0.95 (0.013)	0.98 (0.17)	NS	NS	NS
time peak hip ext	0.99 (0.08)	1.04 (0.1)	1.1 (0.07)	0.95 (0.09)	1.01 (0.13)	1.06 (0.16)	NS	0.001	NS
LEFT STANCE									
time peak dorsiflex	0.33 (0.05)	0.44 (0.08)	0.54 (0.11)	0.38 (0.07)	0.46 (0.06)	0.51 (0.16)	NS	0.001	NS
LEFT SWING									
onset knee flex	0.38 (0.09)	0.40 (0.08)	0.42 (0.09)	0.46 (0.10)	0.51 (0.08)	0.44 (0.10)	NS	NS	NS
onset hip flex	0.52 (0.05)	0.60 (0.06)	0.64 (0.05)	0.49 (0.08)	0.56 (0.09)	0.06 (0.10)	NS	0.001	NS
time peak plantarflex	0.64 (0.04)	0.72 (0.05)	0.81 (0.04)	0.73 (0.07)	0.77 (0.06)	0.86 (0.10)	NS	0.001	NS
time peak knee flex	0.74 (0.05)	0.81 (0.06)	0.89 (0.05)	0.83 (0.07)	0.88 (0.07)	0.94 (0.11)	NS	0.001	NS
time peak hip flex	0.86 (0.05)	0.94 (0.06)	1.02 (0.05)	0.90 (0.10)	0.94 (0.07)	1.15 (0.12)	NS	0.001	NS

Table 2B Same as Table 2A, except for PT2 condition.

PT2	Y-BT	Y-NPT	Y-FT	E-BT	E-NPT	E-FT	G	T	G*T
R-SWING									
onset hip flex	-0.05 (0.09)	-0.02 (0.05)	-0.05 (0.08)	-0.05 (0.03)	-0.04 (0.02)	-0.02 (0.08)	NS	NS	NS
time peak dorsiflex	0.06 (0.12)	0.05 (0.04)	0.05 (0.11)	0.08 (0.04)	0.08 (0.07)	0.09 (0.03)	NS	0.03	NS
time peak knee flex	0.17 (0.12)	0.16 (0.05)	0.16 (0.12)	0.15 (0.04)	0.16 (0.03)	0.16 (0.03)	NS	NS	NS
time peak hip flex	0.31 (0.09)	0.31 (0.05)	0.31(0.10)	0.36 (0.07)	0.35 90.05)	0.36 (0.09)	NS	NS	NS
RIGHT STANCE									
time peak plantarflex	0.38 (0.11)	0.41 (0.03)	0.48 (0.10)	0.47 (0.05)	0.48 (0.06)	0.49 (0.05)	NS	0.004	0.02
time peak knee ext	0.38 (0.07)	0.41 (0.03)	0.44 (0.07)	0.52 (0.08)	0.56 (0.07)	0.58 (0.07)	NS	NS	NS
onset hip ext	0.41 (0.13)	0.39 (0.10)	0.36 (0.08)	0.42 (0.12)	0.38 (0.10)	0.38 (0.08)	NS	NS	NS
time peak knee flex	0.59 (0.09)	0.63 (0.06)	0.65 (0.11)	0.71 (0.09)	0.71 (0.13)	0.72 (0.12)	NS	NS	NS
time peak dorsiflex	0.67 (0.11)	0.81 (0.17)	0.97 (0.14)	0.78 (0.17)	0.93 (0.14)	0.99 (0.13)	NS	0.001	NS
time peak knee ext	0.93 (0.10)	0.95 (0.11)	1.05 (0.15)	1.16 (0.11)	1.08 (0.13)	1.38 (0.09)	NS	NS	NS
time peak hip ext	0.99 (0.12)	1.04 (0.11)	1.05 (0.13)	0.98 (0.11)	1.01 (0.13)	1.03 (0.11)	NS	0.04	NS
LEFT STANCE									
time peak dorsiflex	0.40 (0.10)	0.42 (0.08)	0.45 (0.09)	0.34 (0.10)	0.40 (0.06)	0.36 (0.10)	NS	NS	NS
LEFT SWING									
onset knee flex	0.39 (0.11)	0.40 (0.05)	0.40 (0.10)	0.50 (0.06)	0.50 (0.06)	0.51 (0.07)	NS	NS	NS
onset hip flex	0.51 (0.10)	0.52 (0.06)	0.54 (0.09)	0.56 (0.06)	0.56 (0.09)	0.58 (0.07)	NS	NS	NS
time peak plantarflex	0.66 (0.07)	0.71 (0.05)	0.75 (0.10)	0.72 (0.06)	0.75 (0.06)	0.75 (0.07)	NS	NS	NS
time peak knee flex	0.75 (0.07)	0.81 (0.06)	0.85 (0.07)	0.83 (0.07)	0.87 (0.07)	0.92 (0.08)	NS	0.05	NS
time peak hip flex	0.90 (0.10)	0.93 (0.06)	0.95 (0.12)	0.98 (0.09)	0.97 (0.07)	0.96 (0.09)	NS	NS	NS

Table 2B Same as Table 2A, except for PT2 condition.

PT2	Y-BT	Y-NPT	Y-FT	E-BT	E-NPT	E-FT	G	T	G*T
R-SWING									
onset hip flex	-0.05 (0.09)	-0.02 (0.05)	-0.05 (0.08)	-0.05 (0.03)	-0.04 (0.02)	-0.02 (0.08)	NS	NS	NS
time peak dorsiflex	0.06 (0.12)	0.05 (0.04)	0.05 (0.11)	0.08 (0.04)	0.08 (0.07)	0.09 (0.03)	NS	0.03	NS
time peak knee flex	0.17 (0.12)	0.16 (0.05)	0.16 (0.12)	0.15 (0.04)	0.16 (0.03)	0.16 (0.03)	NS	NS	NS
time peak hip flex	0.31 (0.09)	0.31 (0.05)	0.31(0.10)	0.36 (0.07)	0.35 90.05)	0.36 (0.09)	NS	NS	NS
RIGHT STANCE									
time peak plantarflex	0.38 (0.11)	0.41 (0.03)	0.48 (0.10)	0.47 (0.05)	0.48 (0.06)	0.49 (0.05)	NS	0.004	0.02
time peak knee ext	0.38 (0.07)	0.41 (0.03)	0.44 (0.07)	0.52 (0.08)	0.56 (0.07)	0.58 (0.07)	NS	NS	NS
onset hip ext	0.41 (0.13)	0.39 (0.10)	0.36 (0.08)	0.42 (0.12)	0.38 (0.10)	0.38 (0.08)	NS	NS	NS
time peak knee flex	0.59 (0.09)	0.63 (0.06)	0.65 (0.11)	0.71 (0.09)	0.71 (0.13)	0.72 (0.12)	NS	NS	NS
time peak dorsiflex	0.67 (0.11)	0.81 (0.17)	0.97 (0.14)	0.78 (0.17)	0.93 (0.14)	0.99 (0.13)	NS	0.001	NS
time peak knee ext	0.93 (0.10)	0.95 (0.11)	1.05 (0.15)	1.16 (0.11)	1.08 (0.13)	1.38 (0.09)	NS	NS	NS
time peak hip ext	0.99 (0.12)	1.04 (0.11)	1.05 (0.13)	0.98 (0.11)	1.01 (0.13)	1.03 (0.11)	NS	0.04	NS
LEFT STANCE									
time peak dorsiflex	0.40 (0.10)	0.42 (0.08)	0.45 (0.09)	0.34 (0.10)	0.40 (0.06)	0.36 (0.10)	NS	NS	NS
LEFT SWING									
onset knee flex	0.39 (0.11)	0.40 (0.05)	0.40 (0.10)	0.50 (0.06)	0.50 (0.06)	0.51 (0.07)	NS	NS	NS
onset hip flex	0.51 (0.10)	0.52 (0.06)	0.54 (0.09)	0.56 (0.06)	0.56 (0.09)	0.58 (0.07)	NS	NS	NS
time peak plantarflex	0.66 (0.07)	0.71 (0.05)	0.75 (0.10)	0.72 (0.06)	0.75 (0.06)	0.75 (0.07)	NS	NS	NS
time peak knee flex	0.75 (0.07)	0.81 (0.06)	0.85 (0.07)	0.83 (0.07)	0.87 (0.07)	0.92 (0.08)	NS	0.05	NS
time peak hip flex	0.90 (0.10)	0.93 (0.06)	0.95 (0.12)	0.98 (0.09)	0.97 (0.07)	0.96 (0.09)	NS	NS	NS

Table 3A Group means (SD) of magnitude of angular displacements (degrees) for the right and left sides. Results (p-values) of Split Plot ANOVA are presented on right side of table. The table includes BT, NPT and FT for PT1 condition. "Y" indicates Young group and "E" indicates Elderly group.

PT1	Y-BT	Y-NPT	Y-FT	E-BT	E-NPT	E-FT	G	T	G*T
RIGHT SWING									
peak dorsiflexion	8 (3.5)	8 (3.1)	9 (3.4)	7 (3.6)	7 (3.9)	6 (3.7)	NS	NS	NS
peak knee flexion	43 (5.1)	40 (5.1)	40 (5.3)	43 (5.5)	44 (5.9)	42 (5.9)	NS	NS	NS
peak knee extension	21 (4.4)	27 (4.1)	28 (4.7)	23 (4.2)	27 (4.9)	28 (4.8)	NS	NS	NS
peak hip flexion	31(5.7)	33 (5.2)	33 (5.8)	32 (6.4)	32 (6.1)	32 (6.6)	NS	NS	NS
peak plantarflexion	9 (3.7)	8 (3.3)	9 (3.9)	8 (3.8)	10 (3.6)	12 (4.1)	NS	NS	NS
RIGHT STANCE									
peak dorsiflexion	12 (3.8)	11 (3.8)	11 (4.2)	12 (4.3)	12 (4.1)	13 (4.7)	NS	NS	NS
peak knee flexion	4 (3.1)	11 (3.8)	4 (3.1)	3 (2.7)	3 (2.9)	10 (7.5)	NS	NS	0.02
peak hip extension	44 (6.7)	44 (6.9)	43 (6.9)	31 (6.5)	32 (6.2)	30 (6.7)	0.02	NS	NS
LEFT STANCE									
peak dorsiflexion	11 (3.7)	13 (3.5)	15 (3.9)	14 (4.7)	12 (4.1)	11 (4.8)	NS	NS	NS
peak knee flexion	10 (3.4)	9 (3.4)	9 (3.5)	11 (4.5)	11 (4.9)	11 (5.3)	NS	NS	NS
peak knee extension	7 (1.6)	6 (1.6)	7 (2.5)	9 (3.1)	8 (2.7)	8 (2.8)	NS	0.01	NS
peak hip flexion	9 (1.5)	10 (3.0)	10 (2.8)	10 (2.9)	14 (3.6)	11 (3.1)	NS	NS	NS
LEFT SWING									
peak plantarflexion	21 (4.5)	19 (4.2)	17 (4.8)	21 (5.1)	17 (4.7)	12 (5.2)	NS	0	0.02
peak knee flexion	55 (4.1)	55 (5.7)	56 (6.1)	50 (5.5)	47 (5.8)	43 (5.9)	NS	NS	0.02
peak hip flexion	29 (5.5)	28 (4.5)	29 (4.6)	30 (5.5)	28 (5.1)	29 (5.4)	NS	NS	NS

Table 3B Same as Table 3A, except for PT2 condition.

PT2	Y-BT	Y-NPT	Y-FT	E-BT	E-NPT	E-FT	G	T	G*T
RIGHT SWING									
peak dorsiflexion	6 (3.1)	6 (2.8)	6 (3.2)	6 (3.3)	5 (3.1)	5 (3.2)	NS	NS	NS
peak knee flexion	41 (3.5)	41 (5.1)	42 (10.6)	44 (6.6)	43 (5.9)	44 (6.0)	NS	NS	NS
peak knee extension	25 (4.9)	25 (3.7)	27 (4.1)	29 (4.5)	27 (4.9)	28 (4.8)	NS	NS	NS
peak hip flexion	33 (5.6)	33 (5.2)	33 (5.5)	32 (5.9)	32 (6.1)	32 (5.7)	NS	NS	NS
peak plantarflexion	8 (3.4)	8 (3.3)	9 (3.6)	11 (3.6)	10 (3.6)	10 (3.4)	NS	NS	NS
RIGHT STANCE									
peak dorsiflexion	9 (3.7)	10 (3.8)	8 (3.7)	13 (4.2)	12 (4.1)	12 (4.1)	NS	NS	NS
peak knee flexion	10 (2.9)	10 (2.6)	10 (3.1)	14 (6.2)	3 (2.9)	6 (5.5)	NS	NS	0.02
peak hip extension	33 (5.4)	33 (5.1)	32 (5.7)	32 (6.1)	32 (6.2)	32 (5.8)	NS	NS	NS
LEFT STANCE									
peak dorsiflexion	14 (3.5)	13 (3.5)	13 (3.3)	12 (3.1)	12 (4.1)	12 (3.4)	NS	NS	NS
peak knee flexion	9 (2.2)	9 (2.1)	9 (2.3)	9 (2.9)	9 (2.7)	9 (2.9)	NS	NS	NS
peak knee extension	6 (1.7)	6 (1.5)	6 (1.7)	9 (2.5)	8 (2.7)	8 (2.8)	0.03	0.01	NS
peak hip flexion	10 (2.7)	10 (2.4)	9 (2.6)	12 (3.2)	11 (3.1)	8 (3.2)	NS	NS	NS
LEFT SWING									
peak plantarflexion	19 (4.3)	19 (4.2)	25 (3.0)	21 (4.8)	17 (4.7)	17 (4.6)	NS	0	NS
peak knee flexion	53 (5.5)	55 (5.7)	56 (5.8)	50 (5.2)	47 (5.8)	43 (5.4)	NS	NS	NS
peak hip flexion	30 (4.2)	28 (4.5)	25 (3.0)	30 (5.6)	28 (5.1)	28 (5.5)	NS	NS	NS

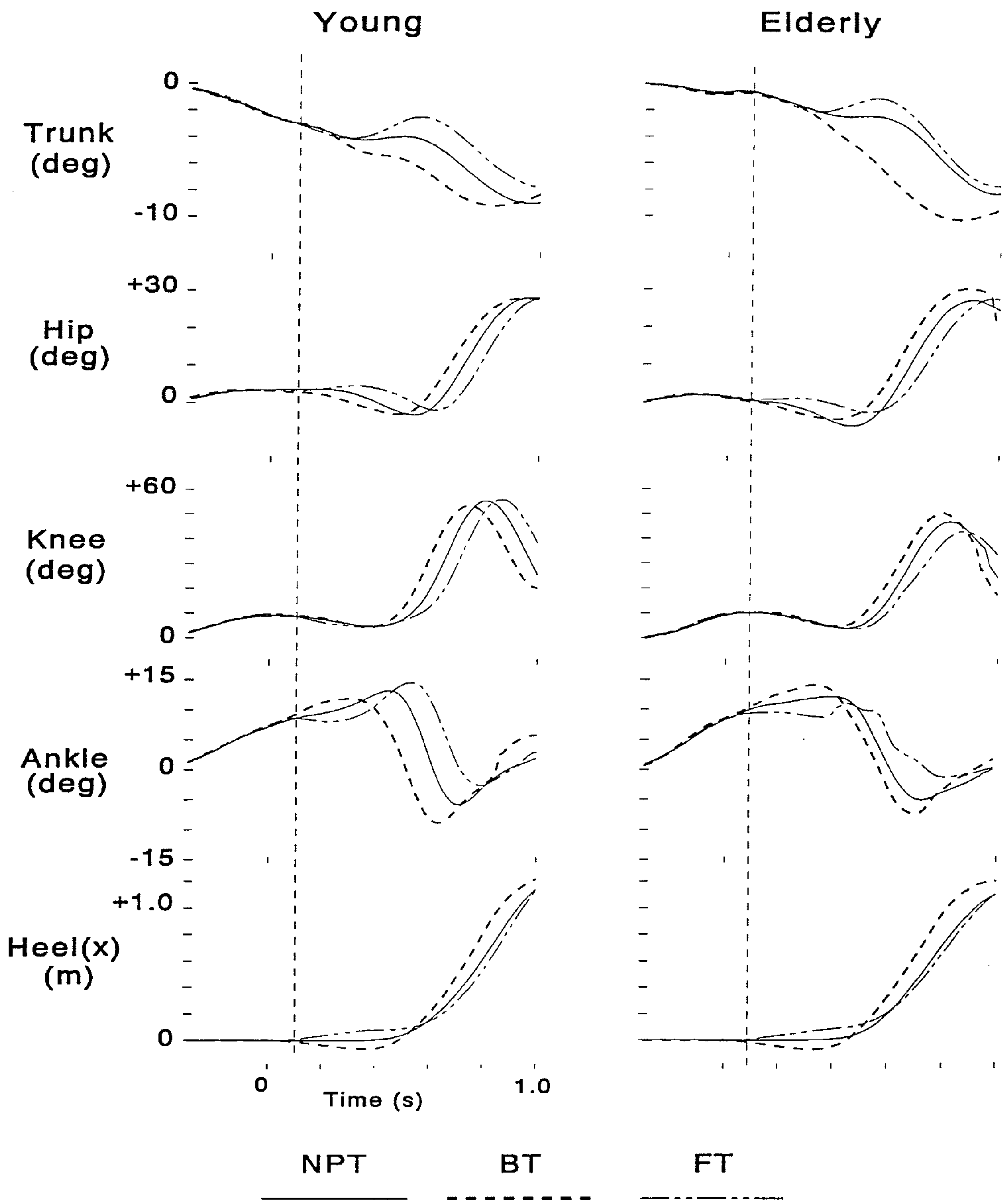

Figure 2A Left side PT1: Presented are group average plots of displacement of the heel marker (x-coordinate), angular displacement at the ankle, knee and hip, and trunk segment rotation for each condition (NPT, FT, and BT). Young group (left panels) and Elderly group (right panels). For y-axis, zero represents standing still baseline position; before right heel off and onset of platform translation. Time zero is heel off and the dashed vertical line represents start of platform translation. For angular displacements, positive values are flexion/dorsiflexion and negative values are extension/plantarflexion. For trunk segment, positive values represent backward rotations and negative values are forward rotations.

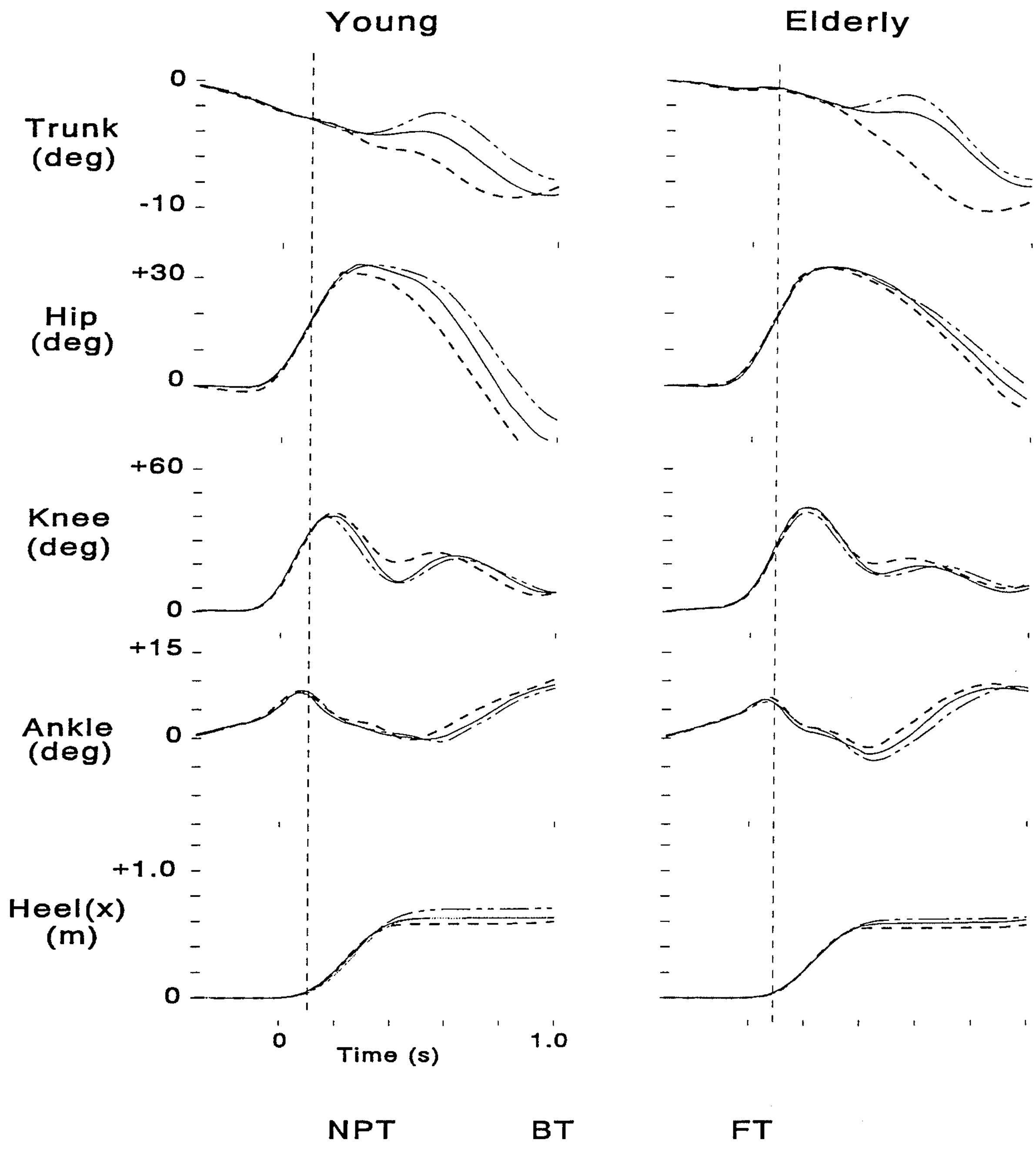

Figure 2B Same as Figure 2A except, Right side PT1.

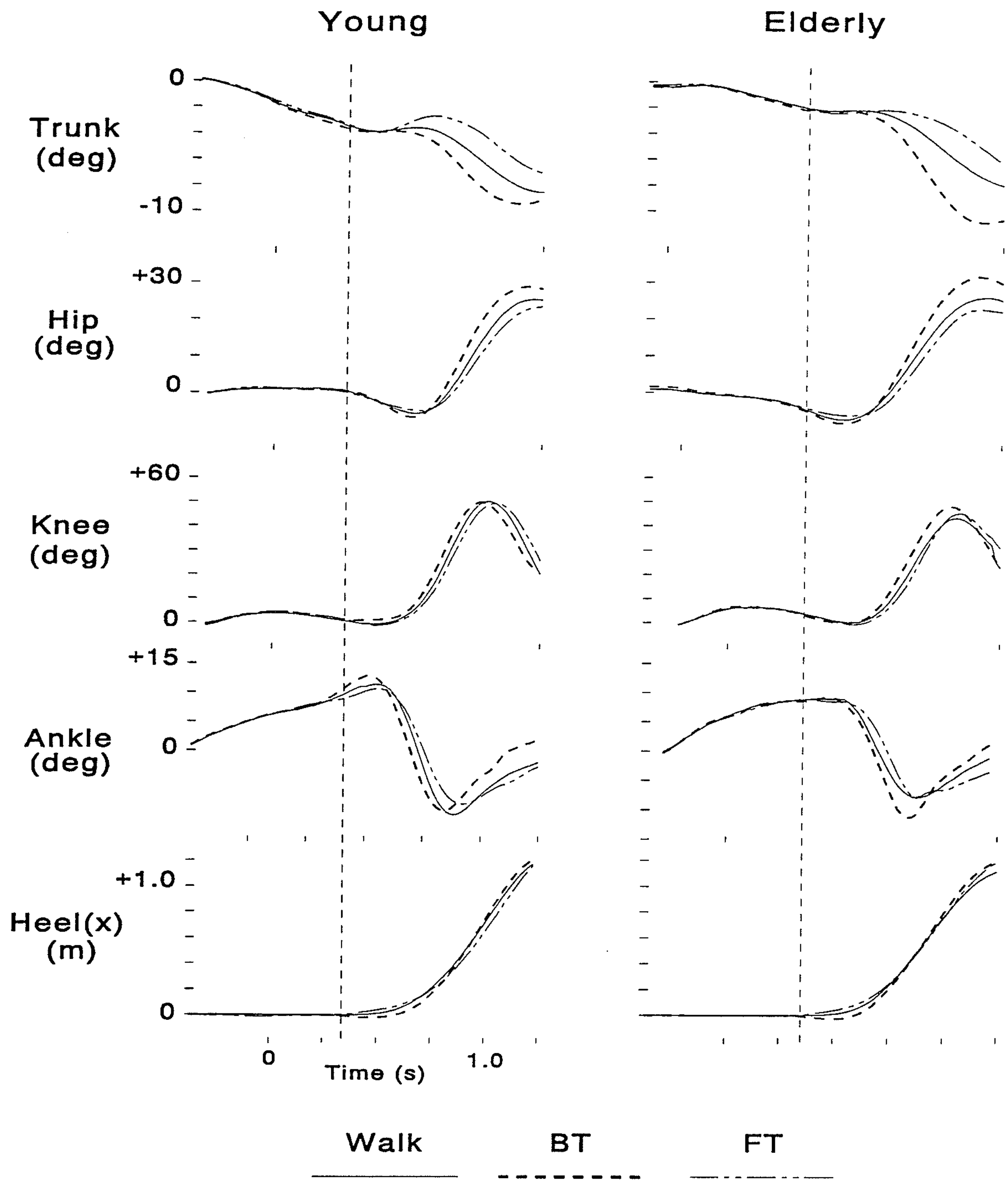

Figure 2C Same as Figure 2A except, Left side PT2.

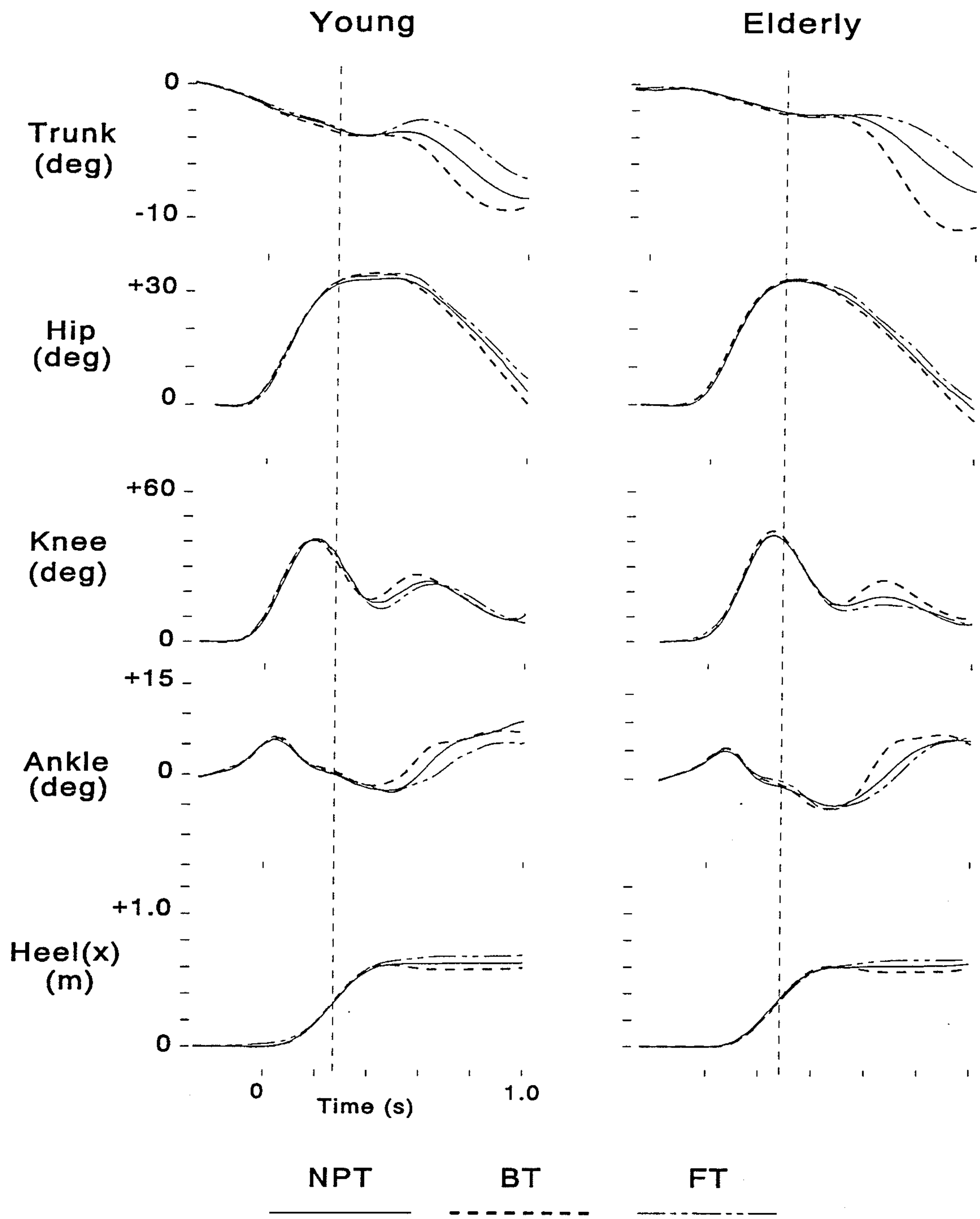

Figure 2D Same as Figure 2A except, Right side PT2.

The exception to this was time to peak plantarflexion in R swing and time to peak knee extension in R stance. The mean time to peak plantarflexion in the Elderly group was 485 ms as compared to 521 ms for the Young group. The mean time to peak knee extension in the Elderly group was 95 ms as compared to 163 ms in the Young group. R time to peak knee extension was the only parameter that showed a statistically significant G effect ($p<0.01$).

The results indicate that the peak magnitudes of the different phases of angular displacements during swing and stance were similar in both the Young and Elderly groups. There was no statistical difference in these dependent variables between groups, with the exception of a significant G*T effect on magnitude of peak knee flexion in R stance ($p<0.02$), and L swing ($p < 0.02$).

3.1.3 EMG

<u>Left Side (Stance):</u> Figure 3 presents the ensemble group averages (8 trials per subject for 9 subjects) of muscle activity during walking with no perturbation (NPT condition). The pattern of activity for TA and HA was similar between the Young and the Elderly groups. The activity of TA decreased to the end of stance and then a burst of activity was noted at the beginning of the swing phase which lasted approximately 250 ms for both groups. The HA activity was virtually the same as TA activity. There was no activity of the GA muscle before the onset of heel-off. Activity began for both groups approximately 150 ms after heel-on. This increased in magnitude until the end of stance and decreased during swing. For the Young group the low point of GA activity was at 600 ms and at 700 ms for the Elderly group. There was a burst of QU muscle activity beginning at 230 ms before heel-on in both groups. This activity was reduced during stance for both groups. The Young group showed a minor increase in activity starting at mid-stance and a burst of activity 80 ms after the start of swing or 500 ms from heel-off. The Elderly burst was earlier (420 ms after heel-off) at the start of swing.

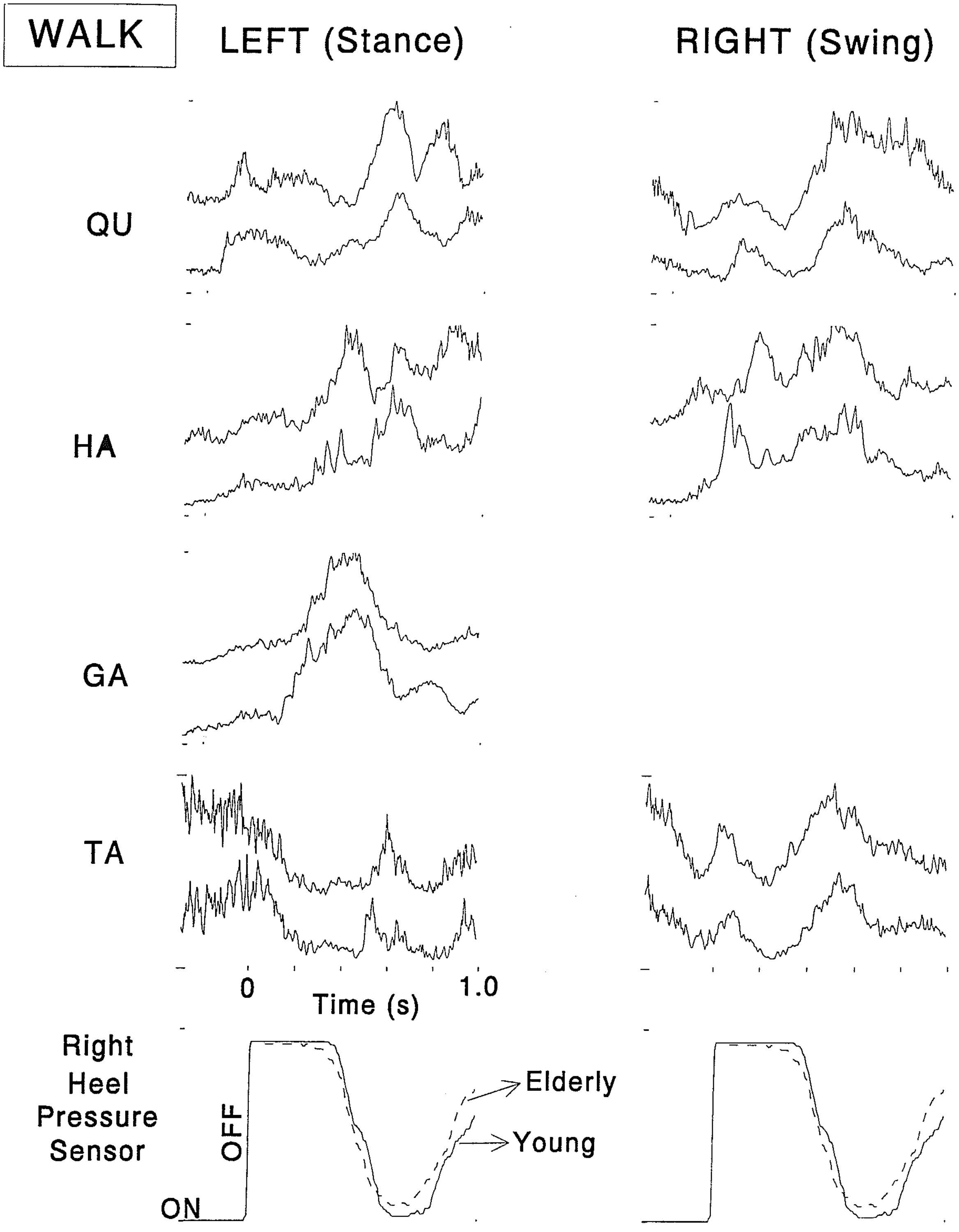

Figure 3 Group ensemble average records of muscle activity in TA, GA, HA and QU, during walks with no perturbations (NPT condition). For each muscle bottom trace is ensemble average of the Young group and top trace is of Elderly group. Also presented is group average plots of right heel pressure sensor signals for Young and Elderly groups. "ON" represents heel in contact with platform or stance and "OFF" foot in swing. Time zero is time of heel off.

Right Side (swing): Figure 3 presents the ensemble group averages (8 trials per subject for 9 subjects) of muscle responses during NPT condition for the right side. The Young and the Elderly ensemble average of TA, HA and QU was virtually the same. At heel-off there was an increase in TA activity which peaked for the Elderly group at 50 ms and for the Young group at 100 ms. This activity then decreased until it reached 250 ms in both groups, after which TA activity again increased into stance. The peak of this activity was at 550 ms after heel-off for both groups. This activity was then reduced to baseline activity. The HA muscle in both groups showed no real activity before heel-off. At heel-off the Young group showed a sudden increase in activity which occurred in the Elderly group 50 ms later. In both cases this activity lasted for 150 ms. The HA again became active for both groups during swing at 300 ms after heel-off. This HA activity peaked during stance 500 ms after heel-off and was gradually reduced to before heel-off levels. At heel-off the QU muscle of the Young group became active, while the QU of the Elderly group was active just before heel-off. In both groups the QU peaked at 80 ms and showed a reduction in activity by 250 ms. The QU activity was immediately increased in both groups until it peaked at 500 ms after heel-off. The activity was then reduced.

3.2 BALANCE DISTURBANCE

The passive effects of sudden platform translation on the body were evident from the linear displacement of the heel marker, CM(R) displacement, and angular displacements. Changes in direction of these waveforms occurred immediately or within 50 ms of the onset of platform motion, i.e., before the onset of the earliest muscle response. The passive component due to platform translation can be seen from the ensemble average waveform as the point where there is a divergence of the waveform from the NPT condition. These passive effects help define the type and direction of the balance disturbances. When comparing corrective balance responses between groups one needs to be sure that the timing and magnitude of the balance disturbances are the same in both groups.

3.2.1 Right Side (Swing)

Figures 4B and 4D present the ensemble group averages on the R side (8 trials per subject for 9 subjects) of the CM(R) displacement and the linear displacement of the heel marker x-coordinate. Figure 4B represents the PT1 condition and Figure 4D represents the PT2 condition. Figures 2B and 2D present the ensemble averages on the R side (8 trials per subject for 9 subjects) of the angular displacements at the ankle, knee, hip and trunk segment for NPT, FT and BT. Figure 2B represents the PT1 condition and Figure 2D represents the PT2 condition.

There was no divergence from the NPT condition in the first 100 ms after the onset of platform motion during FT and BT for: a) linear displacement of heel marker in the x direction; b) CM(R); and c) joint angular displacement at the ankle, knee and hip and trunk segment rotation.

3.2.2 Left Side (Stance)

<u>*Linear Displacement of Heel Marker x- coordinate*</u>

Figures 2A, 2C, 4A and 4C present the ensemble group averages (8 trials per subject for 9 subjects) of linear displacement of the x coordinate for NPT, FT and BT. The divergence in FT and BT waveforms relative to NPT occurred immediately after onset of platform motion. During PT1, the maximum average difference between BT and FT trials, which occurred in the first 150 ms after the onset of platform motion, was similar for both groups (Young group 12 cm and Elderly group 13 cm). During PT2, the maximum average difference between BT and FT trials, which occurred in the first 150 ms after the onset of platform motion, was the same for both groups (Young group 7 cm and Elderly group 7 cm).

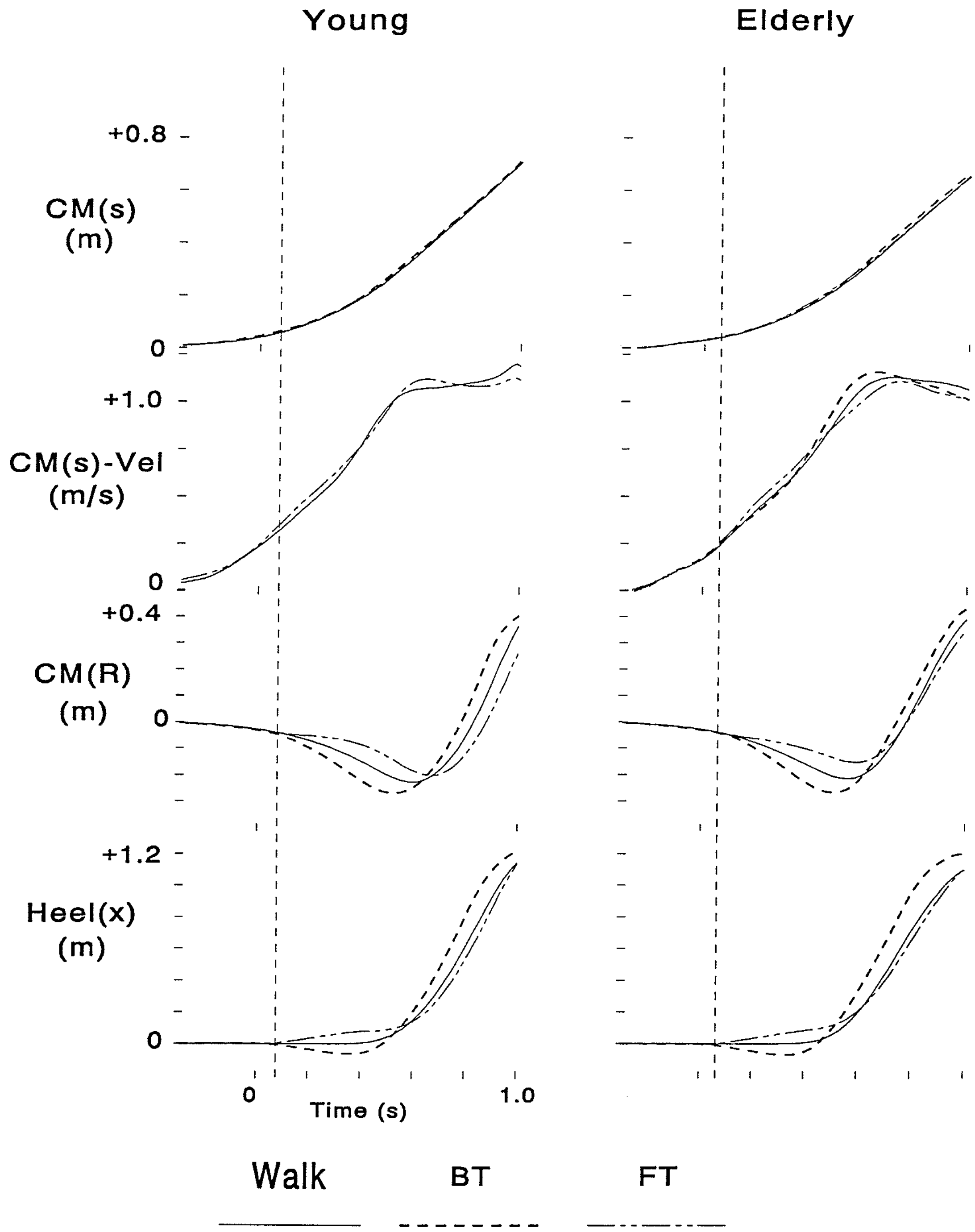

Figure 4A Left side PT1: Presented are group average plots of CM(s) displacement, velocity of CM(s) displacement (CM(s)-VEL), CM(R) displacement and linear displacement of the heel marker for each condition (NPT, FT, and BT). All plots are of x-coordinates. Young group (left panels) and Elderly group (right panels). For y-axis, zero represents standing still baseline position; before right heel off and onset of platform translation. Time zero is heel off and the dashed vertical line represents start of platform translation. For CM(R), positive values represent CM behind the foot and negative values are CM ahead of foot.

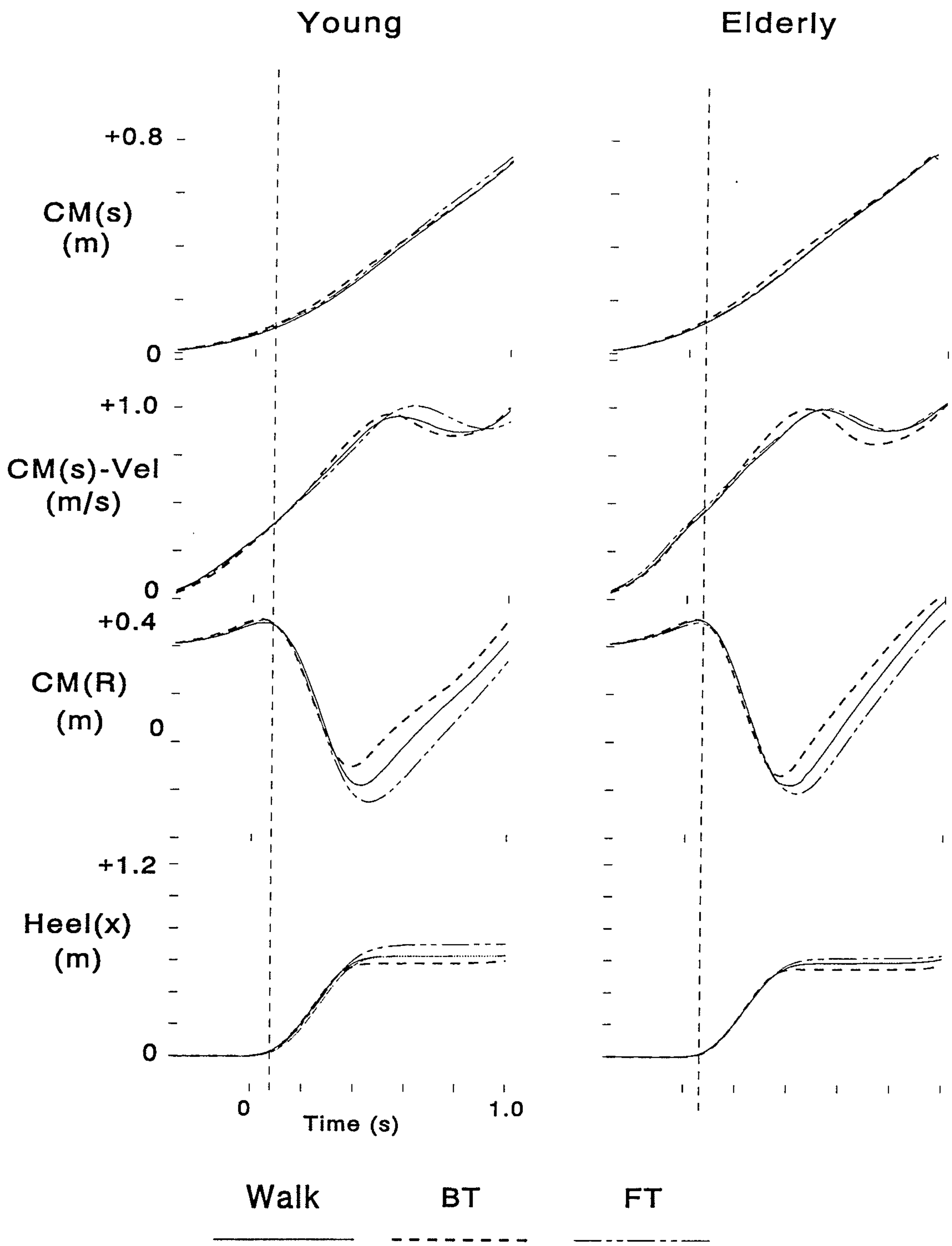

Figure 4B Same as Figure 4A except, Right side, PT1.

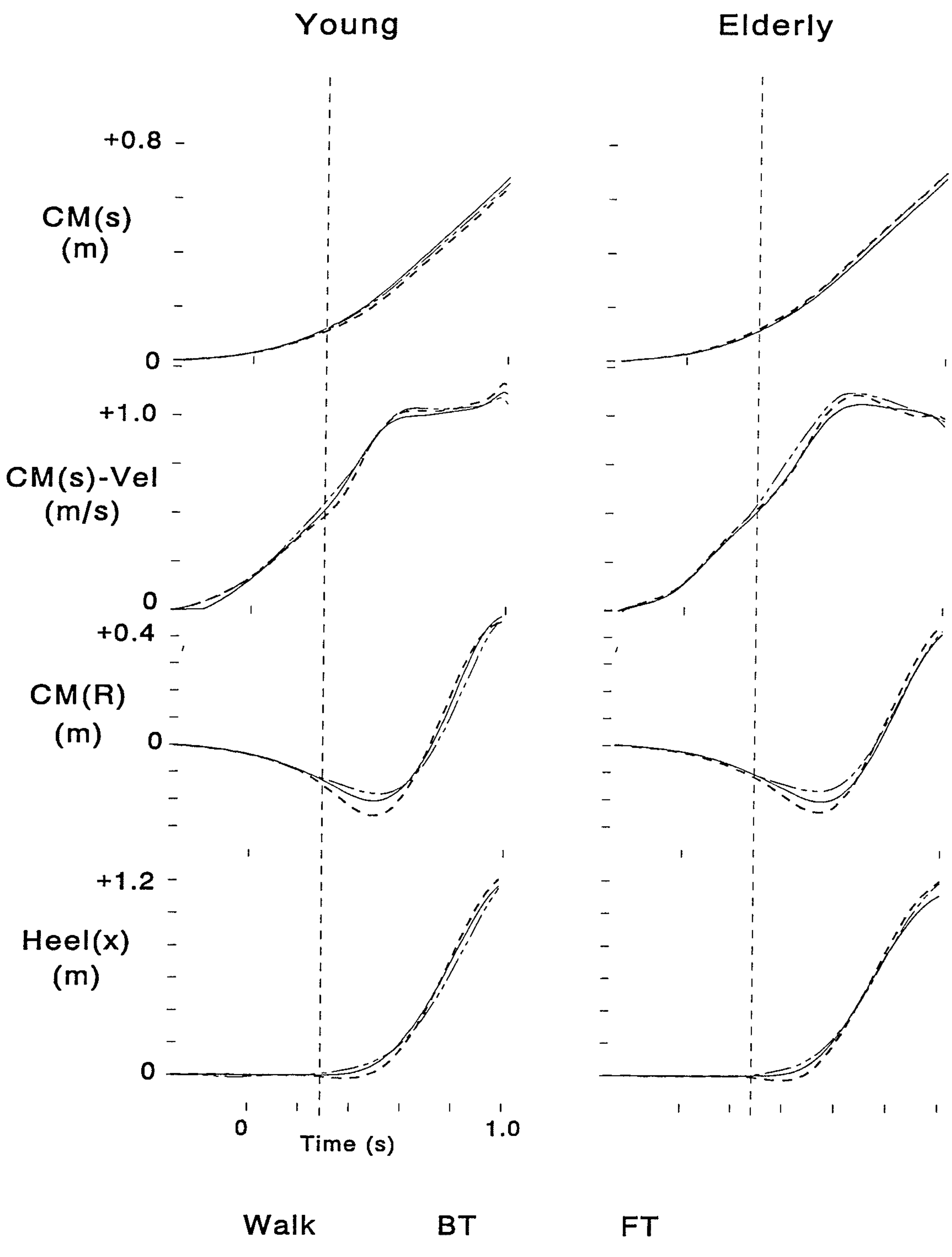

Figure 4C Same as Figure 4A except, Left side, PT2.

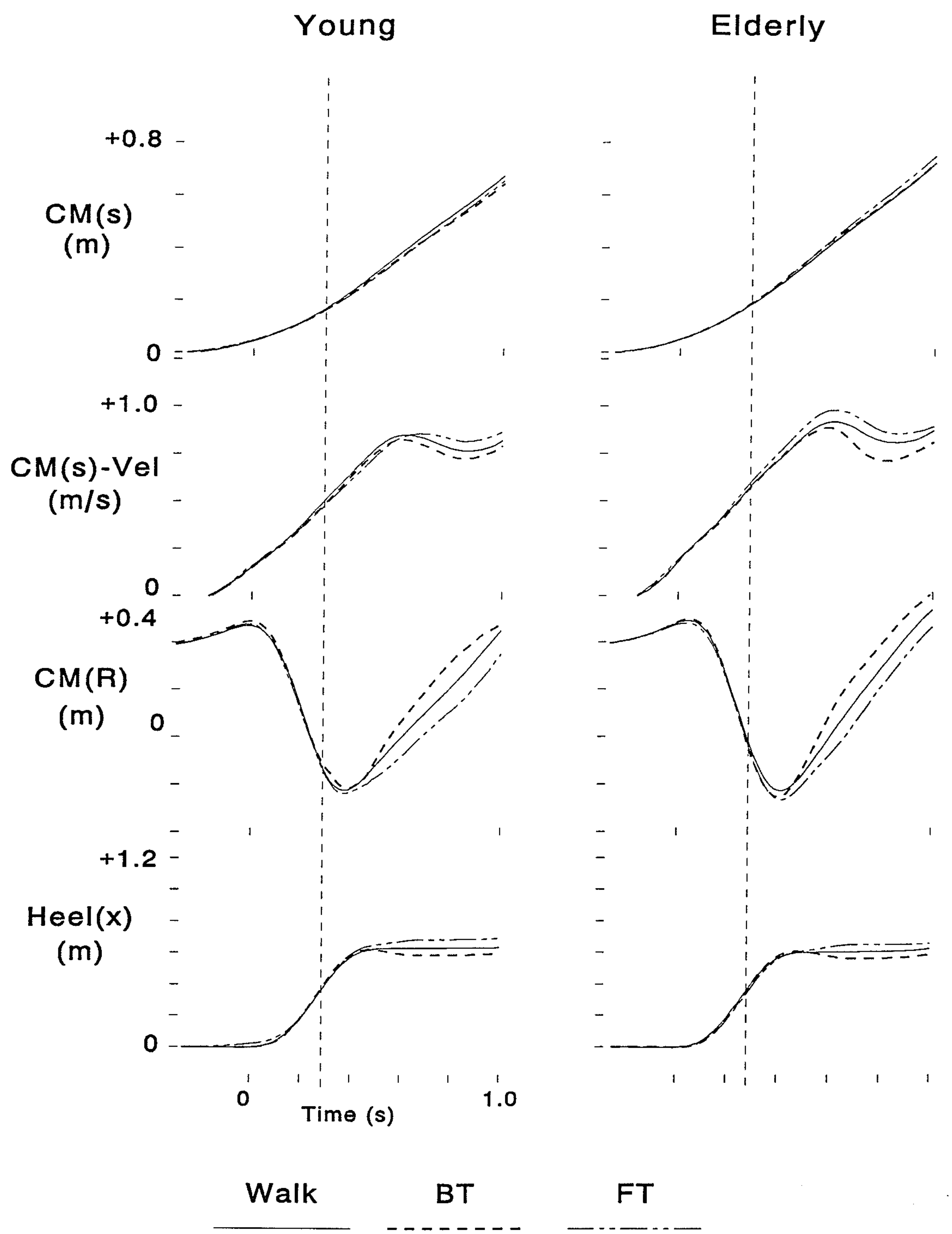

Figure 4D Same as Figure 4A except, Right side, PT2.

CM(R) Displacement

Figures 4A and 4C present the ensemble group averages (8 trials per subject for 9 subjects) of CM(R), displacement relative to the foot for NPT, FT and BT. There was an immediate divergence in CM(R) between NPT, BT and FT. Divergence of the waveform during BT and FT from the NPT condition were opposite in direction. For BT, CM(R) was behind the foot and for FT the CM(R) was ahead of the foot. For both PT1 and PT2 there was no significant group difference in time to initial peak CM(R) or magnitude of initial peak CM(R) displacement. During PT1, the maximum average difference between BT and FT trials, which occurred in the first 200 ms after the onset of platform motion, was similar in both groups (Young group 14 cm and Elderly group 17 cm). During PT2, the maximum average difference between BT and FT trials, which occurred in the first 200 ms after onset of platform motion, was also similar in both groups (Young group 9 cm and Elderly group 8 cm).

During PT1, there was a significant T effect in the time to peak CM(R) displacement ($p<0.01$) and the magnitude to initial peak CM(R) displacement ($p<0.01$). No group effect was observed. During PT2 there was a significant T effect in magnitude to initial peak CM(R) displacement ($p<0.01$), but no significant difference in time to initial peak CM(R) displacement. No group effect was observed in PT2.

Angular Displacement

Figures 2A and 2C present the ensemble group averages (8 trials per subject for 9 subjects) of angular displacements at the ankle, knee, hip and trunk segment for NPT, FT and BT.

Early divergence in angular displacement within the first 50 ms after the onset of platform motion relative to the NPT condition was observed during BT and FT at the ankle and hip. No such passive components were observed at the knee joint and trunk segments. The ankle passive component started immediately following the onset of platform motion (i.e., 70 ms after heel-off). The hip passive deviation started 30 ms after the onset of platform motion. This slight delay showed an inertial effect.

As mentioned before, the passive effects help to define the type and direction of the balance disturbance. For example, at the ankle, FT resulted in a decrease in dorsiflexion relative to the NPT condition, while during BT the opposite occurred, which was an increase in dorsiflexion. At the hip, FT resulted in a decrease in hip extension relative to the NPT condition, while BT resulted in an increase in hip extension relative to the NPT condition. This was evident in varying degrees in both the PT1 and the PT2 conditions.

During PT1, the maximum average difference between BT and FT trials in the ankle angular displacement was similar in both groups (four degrees for the Young group and five degrees for the Elderly group). This occurred in the first 150 ms after the onset of platform motion. During PT2, the maximum average difference between BT and FT trials in the ankle angular displacement was also similar for both groups (three degrees for the Young group and three and one-half degrees for the Elderly group). This also occurred in the first 150 ms after the onset of platform motion.

During PT1, the maximum difference between BT and FT trials in hip angular displacement was similar in both groups (four degrees for the Young group and three degrees for the Elderly group). This occurred in the first 150 ms after the onset of platform motion. During PT2, the maximum difference between BT and FT trials in the hip angular displacement was the same for both groups (one and one-half degrees for the Young group and one and one-half degrees for the Elderly group). This also occurred in the first 150 ms after the onset of platform motion.

3.3 COMPENSATORY RESPONSE

3.3.1 Swing/Stance Duration And Swing Distance

Table 1 presents the group means (SD) of swing duration and swing/stance duration along with the results of split plot ANOVA, i.e., G, T and G*T effects, for swing and stance duration and distances. This table includes the right and left side of the body and both the PT1 and PT2 conditions.

Left Side (stance) - PT1: The left stance duration showed a T effect ($p<0.001$). The duration of left stance was decreased during the BT condition and increased for the FT condition relative to the NPT condition. There was no G or G*T effect in left stance duration. The left swing duration also showed a T effect ($p<0.01$). In this case the duration was increased in the BT condition and reduced in the FT condition. There was no G or G*T effect in left swing duration. There was no significant T or G effect in the left swing distance (the second half cycle for the left leg).

Left Side (stance) - PT2: The left stance duration showed a T effect ($p<0.001$). The duration of left stance was decreased during the BT condition and increased for the FT condition. There was no G or G*T effect in left stance duration. The left swing duration and left swing distance showed no T or G effect.

Right Side (swing) - PT1: The right swing duration showed a T effect ($p<0.012$). The duration of right swing was decreased during the BT condition and increased for the FT condition relative to the NPT condition. There was no G or G*T effect in right swing duration. The right stance duration (second half cycle) showed no significant T or G effect. The right swing distance showed a G effect ($p<0.05$) and a T effect ($p<0.001$). The distance was decreased in the BT condition and increased in the FT condition. In the Young group the decrease in distance in the BT condition was minimal relative to the Elderly group.

Right Side (swing) - PT2: The right swing duration showed a T effect ($p<0.017$). The duration of right swing was decreased during the BT condition and increased for the FT condition. There was no G or G*T effect for right swing duration. The right stance duration showed a T effect ($p<0.004$). The duration of right stance was decreased during the BT condition and increased for the FT condition. There was no G or G*T effect for right stance duration. The right swing distance showed a T effect ($p<0.001$) and a G*T effect ($p<0.033$). The distance was decreased in the BT condition and increased in the FT condition for both groups.

3.3.2 CM Displacement

Figures 4A, 4B, 4C and 4D present the ensemble group averages (8 trials per subject for 9 subjects) of CM(S) and CM(R) displacements for NPT, FT and BT.

CM(S) Displacement

As evident in the figures, CM(S) displacement or velocity did not change between NPT, FT and BT conditions. There was a minor but not significant difference in initial peak CM(S) velocity on the right side during stance for both PT1 and PT2 perturbations in both groups. The time to peak CM(S) and the time to peak CM(S) velocity showed no T or G effect.

CM(R) Displacement

Left Side (stance) - PT1: As evident in Figure 4A, there was a reverse in direction of CM(R) displacement from the initial passive phase during FT and BT at about 250 ms after the onset of platform motion. This compensatory change in CM(R) displacement to FT and BT appears as a phase shift relative to the NPT condition. For FT, CM(R) was further behind the foot than the NPT condition and for BT, CM(R) was ahead of the foot relative to the NPT condition. There was a T effect in time to initial peak CM(R) displacement ($p<0.01$), and magnitude to initial peak CM(R) displacement ($p<0.01$). There was no G effect in time to, or magnitude of, initial peak CM(R) displacement.

Left Side (stance) - PT2: As evident in Figure 4C, there was a reverse in direction of CM(R) displacement for the initial passive phase during FT and BT at the end of stance in both the Young and the Elderly groups. For both groups the time when this change in direction occurred was about 180 ms after the perturbation. For FT, CM(R) was further behind the foot than in the NPT condition; for BT, CM(R) was ahead of the foot relative to the NPT condition. There was little phase shift in CM(R) displacement between FT, NPT and BT during the stance phase. There was no T or G effect in time to initial peak CM(R). There was a T effect in magnitude to initial peak CM(R) displacement ($p<0.01$) but no G effect.

<u>Right Side (swing) - PT1:</u> As seen in Figure 4B, the earliest divergence in CM(R) during FT and BT relative to the NPT condition was observed at about 240 ms after the onset of platform motion. By the end of swing there appears to be a phase shift in CM(R) displacement between FT, NPT and BT. Relative to NPT the CM(R) displacement for BT was behind the ankle and for FT it was ahead of the ankle. At the largest point of divergence the difference was 22 cm for the Young and 20 cm for the Elderly. Statistically there was a T effect in time to initial peak CM(R) displacement ($p<0.01$) and magnitude to initial peak CM(R) displacement ($p<0.001$). There was no G effect in either time to peak CM(R) displacement ($p<0.01$) or magnitude to initial peak CM(R) displacement.

<u>Right Side (swing) - PT2:</u> As described above in right side PT1, a similar phase shift in CM(R) displacement between FT, NPT and BT was observed (see Figure 4D). The divergence began at the beginning of right stance in both the Young and the Elderly groups. At the largest point of divergence the difference was 22 cm for the Young and 17 cm for the Elderly group. There was no T or G effect in either time to initial CM(R) displacement or magnitude to peak CM(R) displacement.

Based on the results of CM(S) and CM(R), there was virtually no difference in task performance between the Young and the Elderly groups. All subjects were able to perform the task of walking when perturbed at heel-off and mid-swing.

3.3.3 Angular Displacement

<u>Left Side (stance) - PT1</u>: Figure 2A presents group ensemble averages of angular displacement waveforms of ankle, knee, hip and trunk segment for FT, NPT and BT in the PT1 condition. Table 2A presents group means (SD) of onset times and time to peak magnitudes of the different phases of walking.

There was a significant T effect (FT, BT and NPT) in the time to peak dorsiflexion during stance ($p<0.001$). Relative to the NPT condition, peak dorsiflexion and, consequently, onset of ankle plantarflexion occurred earlier during BT and later during FT. During the NPT condition, peak dorsiflexion occurred at 435 ms in the Young and 445 ms in the Elderly. During BT, peak dorsiflexion occurred at 325 ms in the Young and 382 ms in the Elderly. During FT, peak dorsiflexion occurred at 538 ms in the Young group and at 505 ms in the Elderly group. There was no G or G*T effect in time to peak dorsiflexion. There was no T or G effect in magnitude of initial peak dorsiflexion in stance. A trend toward a magnitude change in peak dorsiflexion was evident from the ensemble average waveforms presented in Figure 2A. The NPT condition showed a similar magnitude between the Young and the Elderly groups. The BT condition showed a reduced magnitude of dorsiflexion in the Young group as compared to the Elderly group. For the FT condition, there was an increase in the magnitude of dorsiflexion in the Young group and a decrease magnitude of dorsiflexion in the Elderly group relative to the NPT condition.

During the swing phase (second half cycle) the time to peak plantarflexion showed a T effect ($p<0.001$) but no G effect. In both the Young and the Elderly groups, peak plantarflexion occurred earlier during BT and later during FT, relative to the NPT condition (see Table 2A). The magnitude of the plantarflexion phase in swing showed a G*T effect. Further investigation indicated a T effect in magnitude for the Elderly group ($p<0.01$), but not in the Young group. Relative to the NPT condition, magnitude of plantarflexion was greater during BT and less than during FT.

As evident from Figure 2A, the divergence at the knee began at about 200 ms after onset of the perturbation at the end of stance. There was no G or T effect in magnitude of knee flexion or extension during stance. Knee flexion in left swing started about 350 ms after the disturbance, which was statistically the same for the Young and the Elderly. There was no T effect in time of onset of knee flexion in swing, but there was a T effect ($p<0.001$) in the time to peak knee flexion. No G or G*T effect was found for time of onset of knee flexion in swing, or time to peak knee flexion in swing. Therefore, there was a change in duration of left knee flexion during the swing phase for both the Elderly and Young groups during FT, NPT and BT conditions. There was no T effect for magnitude of knee flexion but there was a G*T effect ($p<0.02$). Further statistical analysis by group revealed a T effect ($p<0.05$) in magnitude of knee flexion in the Elderly, but not the Young group. In the

Elderly group the magnitude of peak knee flexion was larger during BT and smaller during FT relative to the NPT condition.

As evident in Figure 2A, both the Young and Elderly groups showed a phase shift in hip angular displacement between FT, NPT and BT conditions. Divergence and, consequently, phase shifting occurred in the hip immediately after PT1. The onset of hip flexion showed a T effect ($p<0.001$) but no G or G*T effect. In both the Elderly and the Young groups the BT condition revealed a decrease in onset time and the FT condition showed an increase in onset time relative to the NPT condition. The time to peak hip flexion also showed a T effect ($p<0.001$) but no G or G*T effect. In both groups, the BT condition showed a decrease in time to peak hip flexion and an increase in time to hip flexion in the FT condition. There was no T, G or G*T effect on the magnitude of hip flexion, although the Elderly group showed a tendency towards an increase during BT, when comparing the ensemble average waveforms.

The compensatory response of the trunk segment was described using the group ensemble average waveforms, as a statistical analysis was not performed. The reason for this was that there was no obvious minimum and maximum points to quantify the trajectories in the first 500 ms from onset of platform motion. As evident in Figure 2A, the plots of the trunk segment rotation of both groups show an appropriate phase shift relative to the NPT condition. During BT, both groups showed a phase lag; during FT, both groups showed a phase advance. The BT condition showed a decrease in magnitude of forward trunk segment rotation and the FT condition showed an increase in magnitude of forward trunk segment rotation relative to the NPT condition. The divergence in waveforms began at about 220 ms after the onset of platform movement in both groups, which represents the compensatory response. The magnitude of peak backward trunk segment angular displacement was similar between FT, NPT and BT. The Elderly group showed waveforms similar to the Young group, except there was no flexion peak for BT. The Elderly group also showed a tendency toward an increase in backward trunk segment rotation magnitude during BT and a decrease during FT condition relative to NPT.

<u>Left Side (Stance) - PT2:</u> Figure 2C presents the ensemble average waveforms of angular displacements at the ankle, knee, hip and trunk segment rotation for both groups during

NPT, FT and BT conditions. As described above for PT1, divergence of the waveforms representing NPT, FT and BT conditions was evident during PT2. With the exception of the passive phase at the ankle, the divergence began at approximately 200 ms after the onset of the platform motion. This was similar for both groups. As with PT1 (see Figure 2A), the divergences in NPT, FT and BT waveforms took on the appearance of a phase shift. This was a similar pattern and magnitude of angular displacement, but with trajectories of the BT condition occurring earlier than the NPT condition. The FT condition occurred later. At the ankle there were no T, G or G*T effects in: a) time to peak dorsiflexion (early stance phase); b) magnitude of dorsiflexion (early stance phase); or c) magnitude of plantarflexion (early swing phase). There was a significant T effect ($p<0.01$) in time to peak plantarflexion (early swing phase). Peak plantarflexion occurred earlier in BT as compared to the NPT condition, and in FT it occurred later. There was no G effect in time to peak plantarflexion.

At the knee there was no T, G or G*T effect in the following: a) time to peak knee flexion (stance phase); b) magnitude of knee flexion or extension (stance phase); or c) time of onset of knee flexion (start of knee flexion). There was a T effect ($p<0.05$) in the time to peak knee flexion in swing but no G or G*T effect. Therefore, there was a change in duration for both the Young and Elderly groups for left knee flexion during swing. During BT there was a decrease in knee flexion duration relative to the NPT condition and an increase in duration during FT. There was no T, G or G*T effect on magnitude of knee flexion (swing phase). However, the ensemble average waveform showed a tendency towards a magnitude effect in the Elderly group; i.e., greater during BT and less during FT.

There was no significant T, G or G*T effect on: a) onset of hip flexion (early swing phase); b) time to peak hip flexion; c) magnitude of peak hip flexion; or d) time to peak hip flexion. However, as evident in Figure 2C, both groups showed a tendency towards: a) an increase hip flexion during BT and a decrease in hip flexion during FT as compared to NPT condition; and b) a decreased time to peak hip flexion during BT and an increased time to peak hip flexion during FT as compared to the NPT condition.

Similar to the findings for PT1 described above, there was a phase shift in trunk segment rotation between the FT, NPT and BT conditions. This began about 200 ms after onset of platform motion. See Figure 4B for information regarding trunk segment angular displacement. Backward trunk segment rotation began earlier during BT and later during FT as compared to the NPT condition. This was evident in both groups.

<u>Right Side (Swing) - PT1:</u> There was no T, G or G*T effect on the following parameters obtained from the ankle angular displacement: a) time to peak dorsiflexion (early swing); or b) magnitude of peak dorsiflexion (early swing). The Young and the Elderly groups showed, on the ensemble average waveform (Figure 2B), a phase shift starting at early peak plantarflexion during stance, 30 ms after the onset of platform motion. The time to peak plantarflexion showed a T ($p<0.01$) and G ($p<0.01$) effect. Relative to the NPT condition in both groups, peak plantarflexion occurred earlier during BT and later during FT. The Elderly group, in all conditions, showed a decrease in time to peak plantarflexion relative to the Young group, as seen in Table 2. The time to peak dorsiflexion in stance showed a T effect ($p<0.001$) but no G or G*T effect. Relative to the NPT condition, peak dorsiflexion in stance occurred earlier during BT and later during FT (Table 2). There was no T, G or G*T effect on magnitude of the ankle plantarflexion (early stance) or on magnitude of dorsiflexion (late stance).

There was no T, G or G*T effect on the following parameters obtained from knee angular displacements: a) time of onset of right knee flexion in early swing; b) time of peak knee flexion or peak knee flexion magnitude in mid-swing; or c) magnitude of peak knee extension in late swing. The time to peak knee extension in late swing did not show a T effect, but a G effect ($p<0.01$) was observed. The time to peak knee flexion in stance showed a T effect ($p<0.01$) but no G effect. Relative to the NPT condition, peak knee flexion in stance occurred earlier during BT and later during FT. There was no T, G or G*T effects on: a) time to peak right knee extension (late stance); or b) magnitude of peak knee extension (late stance).

There was no T, G or G*T effects on: a) time to peak hip flexion; b) magnitude of peak hip flexion; or c) time to onset of hip extension. However, as evident in Figure 2B, the divergence of hip angular displacement between the FT, NPT and BT conditions began shortly after onset of hip extension; i.e., about 200 ms after the onset of the platform displacement. The magnitude of peak hip extension showed a G effect ($p<0.02$). Relative to the NPT condition, the Elderly group showed a much smaller magnitude than the Young group.

Right Side (Swing) - PT2: There was no T, G or G*T effect on the following parameters: a) time to peak dorsiflexion (in swing); or b) magnitude of peak dorsiflexion (in swing). The Young and the Elderly groups showed, on the ensemble average waveform (Figure 2D), a phase shift starting at early peak plantarflexion during stance, about 200 ms after the perturbation in both groups. The right time to peak plantarflexion showed both a T($p<0.004$) effect and a G*T ($p<0.02$) effect. Relative to the NPT condition, in both groups peak plantarflexion occurred earlier during BT and later during FT. The Elderly group in all conditions showed an increase in time to peak plantarflexion relative to the Young group, as seen in Table 3. The time to peak dorsiflexion in stance showed a T effect ($p<0.001$), but no G or G*T effect. Relative to the NPT condition, peak dorsiflexion in stance occurred earlier during BT and later during FT (Table 2B). The magnitude of ankle plantarflexion showed a G effect ($p<0.04$), but no T or G*T effect. Relative to the NPT condition and the Young group, the Elderly group showed a smaller magnitude of ankle plantarflexion. The ensemble average waveform (Figure 2D) showed a tendency towards a difference in plantarflexion between trials in the Young group.

There was no T, G or G*T effect on the following parameters obtained from the knee angular displacements: a) time of onset of knee flexion (in swing); b) time of peak knee flexion (in swing) or peak knee flexion magnitude (in swing); or c) time to peak knee extension (late swing). When looking at the ensemble average waveform (Figure 2D) of knee extension in late swing, both the Young and Elderly groups showed a trend towards a magnitude difference for BT and FT conditions. This was not a statistically significant difference, as seen in Table 3B. At this point both groups showed a small phase shift. There was no T, G or G*T effect in the time to peak knee flexion in stance. The ensemble average waveforms (Figure 2D) showed a tendency for a magnitude difference in peak knee flexion in stance for both groups. Relative to the NPT waveform, the BT condition showed an increase in magnitude and the FT condition showed a decrease in magnitude. There was no T, G or G*T effect in the following: a) time to peak knee extension (in stance); or b) magnitude of peak knee extension (in stance).

There was no T, G or G*T effect on the following parameters: a) time to peak hip flexion; b) magnitude of peak hip flexion; c) time to peak hip extension (see Table 2B for means (SD)); or d) magnitude of peak hip extension (see Table 3B for means (SD)). The divergence in the hip ensemble waveform (Figure 2D) was 230 ms after the perturbation in the Young group and 70 ms after the perturbation in the Elderly group. The divergence in both cases showed a small a phase shift.

3.3.4 EMG

Figures 5A, 5B, 5C and 5D present the ensemble group averages of different waveforms (four perturbed trials for nine subjects) for BT-PT1, FT-PT1, BT-PT2 and FT-PT2 respectively. Table 4 presents the group means (SD) of onset latencies of corrective muscle responses for BT and FT during PT1 and PT2.

<u>Left Side (stance):</u> All muscles exhibited early responses from the onset of platform motion for both FT and BT. The range of mean onset latencies was from 78 ms to 275 ms for the left stance PT1 condition. Some of the responses were excitatory, i.e., there was an increase in activity relative to the NPT trials, and some were inhibitory, i.e., there was a decrease in activity relative to the NPT trials. For both FT and BT the pattern of corrective muscle response observed for PT1 was different than PT2. For example, during BT-PT1, left TA was inhibitory while during BT-PT2 the initial corrective response was excitatory. The initial corrective response in HA during BT was excitatory for PT1 while for PT2 it was inhibitory. During FT left GA in the PT1 condition exhibited initial excitation and during PT2 condition the left GA exhibited inhibition. The initial corrective response in HA during FT was inhibitory for PT1 and excitatory in PT2.

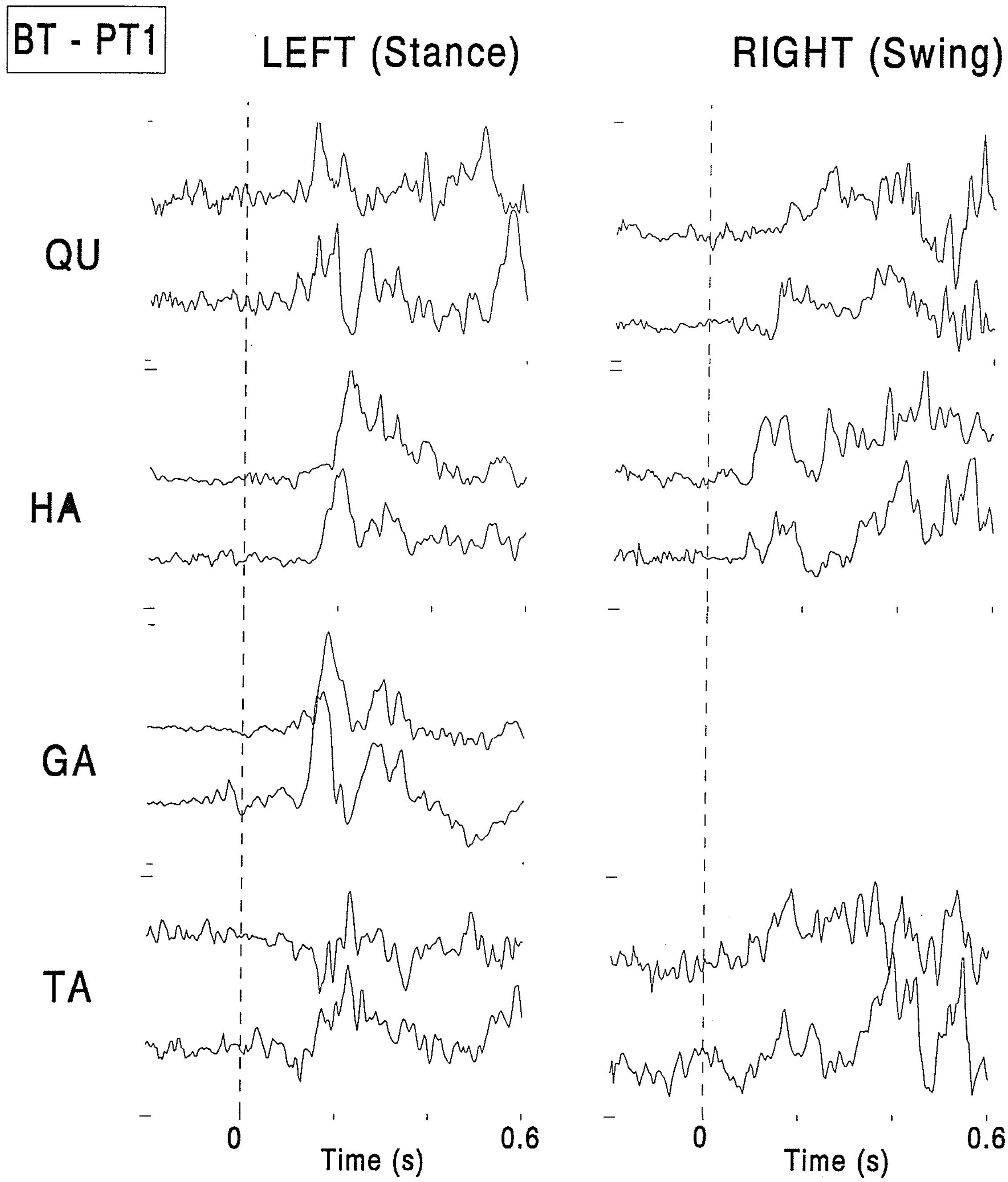

Figure 5A Backward translation (BT), PT1: Group averages of the difference EMG waveforms of TA, GA, HA and QU. For each muscle bottom trace is average of the Young group and top trace is of Elderly group. Vertical dashed line (time zero) is onset of platform translation.

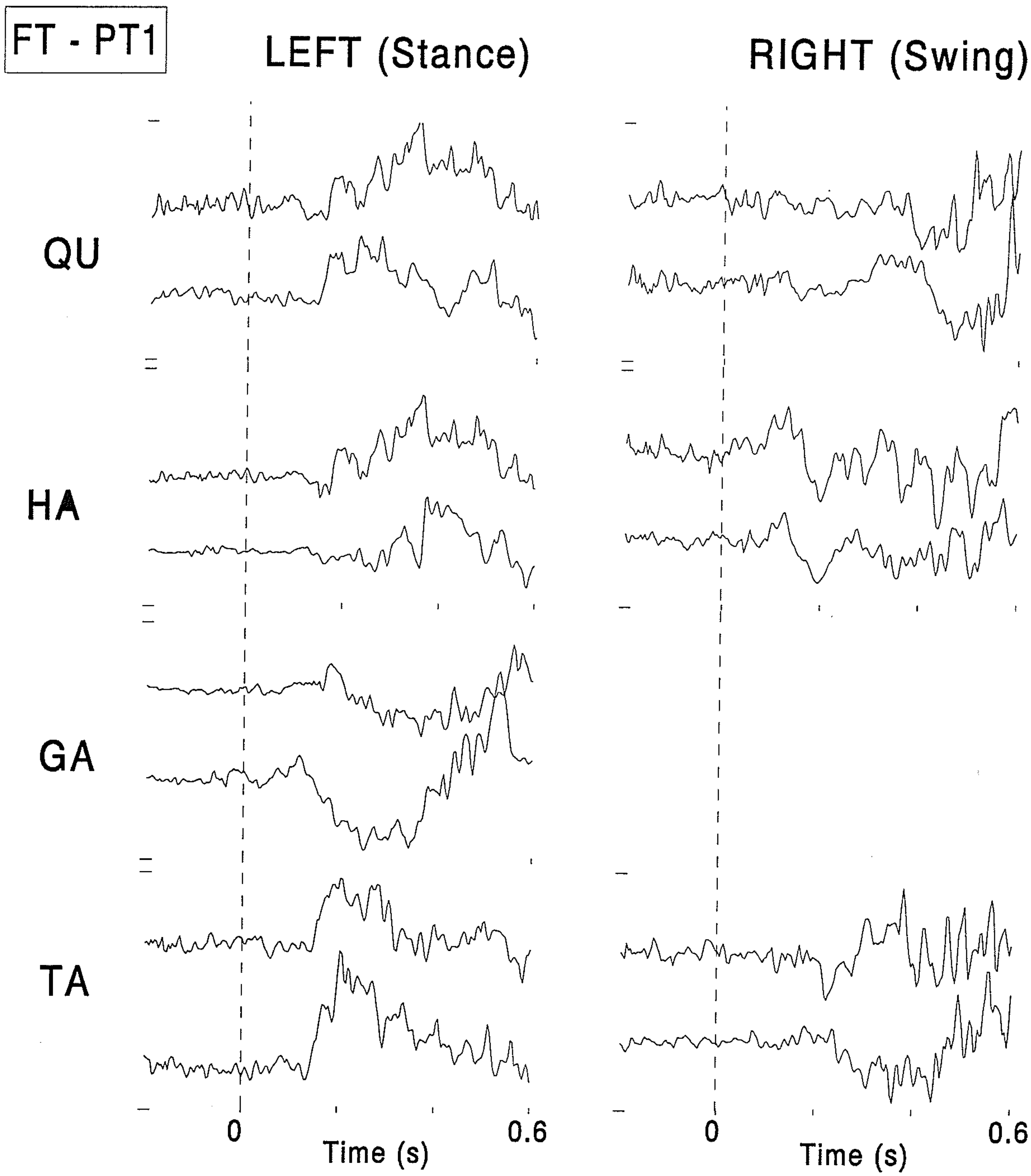

Figure 5B Same as Figure 5A, except for FT, PT1.

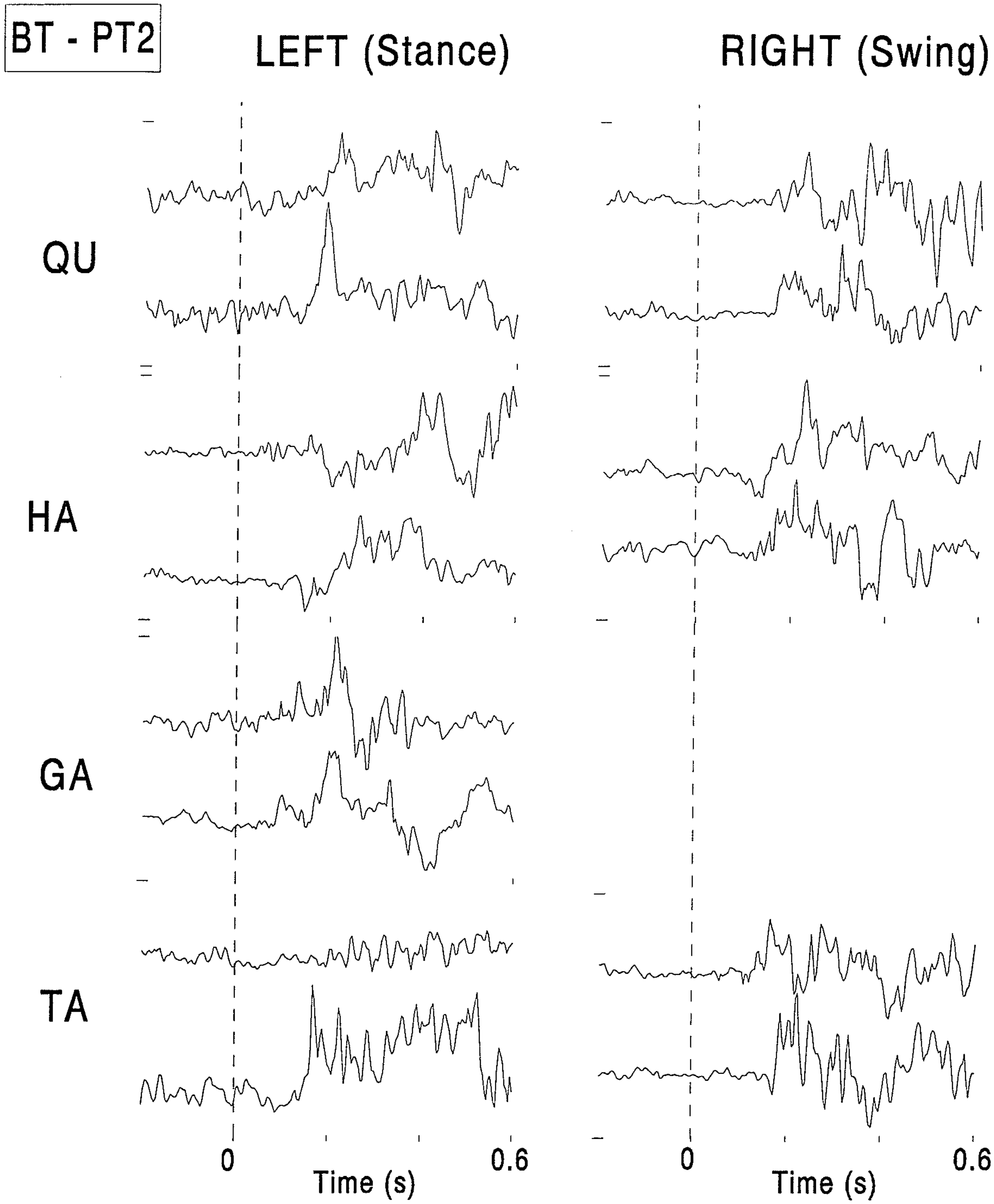

Figure 5C Same as Figure 5A, except for BT, PT2.

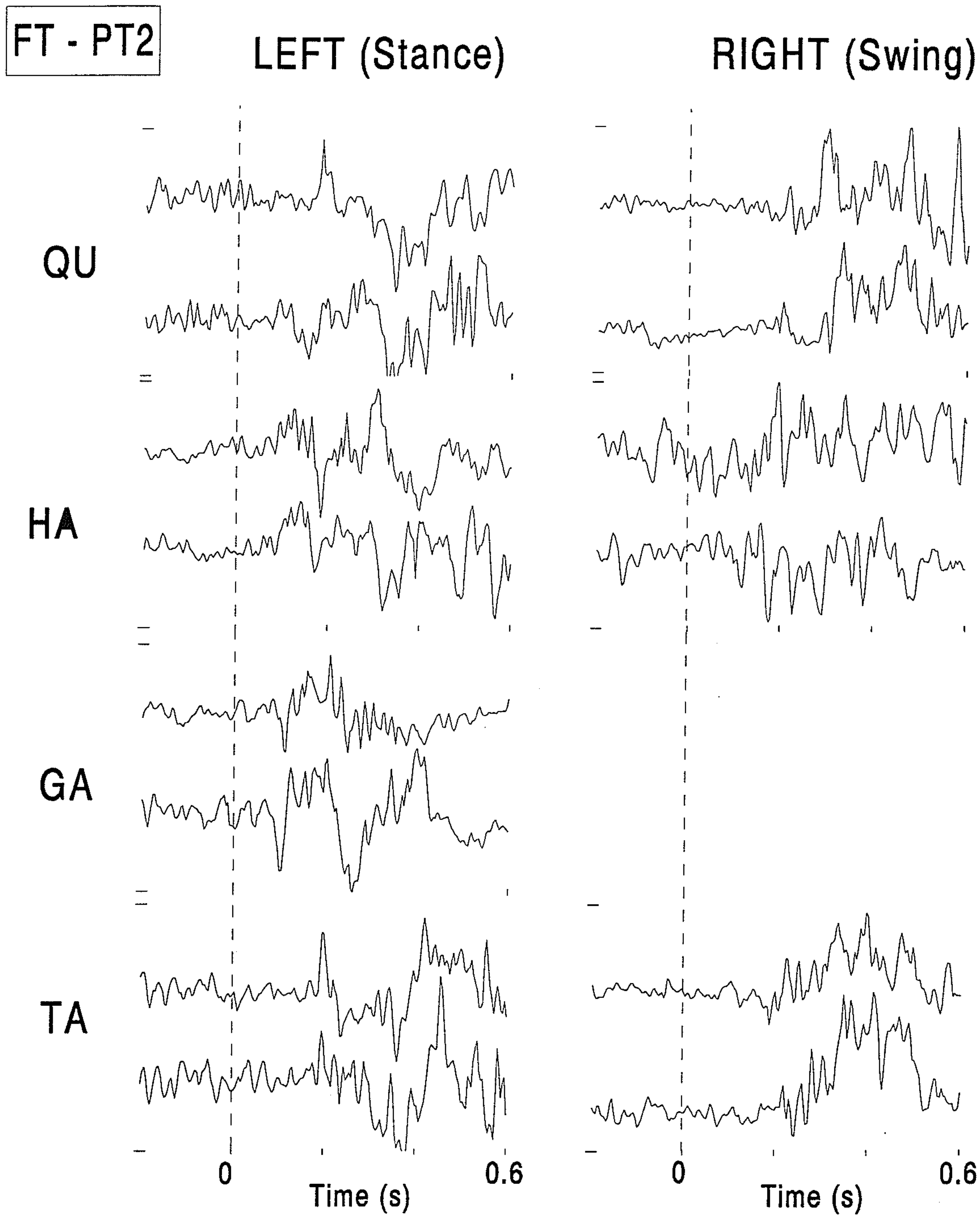

Figure 5D Same as Figure 5A, except for FT, PT2.

TABLE 4 Group means (SD) of onset latencies for GA, TA, QU, and HA responses. Right side is "R", Left side is "L". The letter "E" indicates excitatory responses and "I" inhibitory responses.

BACKWARD TRANSLATIONS

MUSCLE	PT1 YOUNG	PT1 ELDERLY	PT2 YOUNG	PT2 ELDERLY
R-TA (Swing)	E - 109 (21)	E - 113 (29)	E - 164 (29	E - 119 (33) *
R-HA (Swing)	E - 71 (11)	E - 78 (12)	E - 149 (21)	I - 95 (19) ++ E - 152 (26)
R-QU (Swing)	E - 131 (18)	E - 145 (22)	E - 152 (23)	E - 163 (29)
L-TA (Stance)	I - 89 (24)	I - 105 (28)	E - 139 (27)	E - 212 (31) *
L-GA (Stance)	E - 124 (17)	E - 118 (21)	E - 83 (11)	E - 99 (13)
L-HA (Stance)	E - 152 (21)	E - 161 (23)	I - 117 (16)	I - 141 (24)
L-QU (Stance)	E - 81 (18)	E - 103 (23)	E - 116 (24)	E - 129 (31)

FORWARD TRANSLATION

MUSCLE	PT1 YOUNG	PT1 ELDERLY	PT2 YOUNG	PT2 ELDERLY
R-TA (Swing)	I - 228 (31)	I - 181 (37)	E - 225 (39)	I - 132 (234) ++ E - 277 (41)
R-HA (Swing)	E - 89 (19) I - 132 (28)	E - 79 (17) I - 158 (33)	I - 158 (26)	E - 151 (31) ++
R-QU (Swing)	E - 268 (38)	I - 340 (41) ++	E - 294 (43)	E - 268 (39)
L-TA (Stance)	E - 130 (19)	E - 133 (23)	E - 177 (21) I - 275 (31)	E - 172 (27) I - 221 (37)
L-GA (Stance)	E - 88 (12) I - 113 (17)	E - 136 (19) * I - 168 (22) *	I - 76 (11) E - 99 (17)	I - 81 (14) E - 209 (33) *
L-HA (Stance)	I - 122 (23) E - 275 (31)	I - 92 (21) * E - 152 (25) *	E - 93 (19)	E - 89 (16)
L-QU (Stance)	E - 131 (21)	E- 144 (26)	I - 102 (23)	E - 151 (29) ++

* Group Effect $p < 0.05$
++ Different sign (excitatory vs inhibitory response)

PT1 Condition - BT: The pattern of muscle responses in BT was similar in both groups. There was no G effect in onset latency for the four muscles. The TA muscle in both groups showed an early inhibitory phase. The mean onset latencies were 89 ms for the Young group and 102 ms for the Elderly group. In both groups the initial inhibitory phase was followed by an excitatory phase. In both groups GA showed excitatory phase. The mean onset latencies were 124 ms for the Young and 118 ms for the Elderly group. An excitatory phase was observed in the HA muscle in both groups. The mean onset latencies were 152 ms in the Young group and 161ms in the Elderly group. Although approximately 60 ms earlier than the HA muscle, the QU also exhibited an excitatory phase in both groups. The mean onset latencies were 81 ms for the Young group and 102 ms for the Elderly group.

PT1 Condition - FT: During FT, both groups exhibited an early excitatory burst in TA. The Young group showed a burst at 130 ms and the Elderly exhibited the burst at 133 ms. There was no G effect in onset latency. The excitatory TA response decreased from mid-stance to heel-off, at which time it reached NPT trial levels for both groups. The pattern of muscle responses observed in GA was a brief excitatory phase followed by an inhibitory phase. This pattern was observed in both groups. However, there was a significant G effect ($p<0.01$) in onset latency for both the excitatory and inhibitory phases. The Young group exhibited earlier onset latencies than the Elderly group for both the excitatory phases. The mean excitatory onset latency for the Young group was 88 ms and was 136 ms for the Elderly group. The mean inhibitory onset latency for the Young group was 113 ms and was 168 ms for the Elderly group. This inhibitory GA muscle response was followed by an excitatory response for both groups which lasted well into swing phase of gait. The pattern of muscle responses observed in HA was an initial inhibition followed by an excitatory phase. This pattern was observed in both groups. However, there was a significant G effect ($p<0.01$) in onset latency for both the inhibitory and excitatory phases. The Elderly group exhibited earlier onset latencies than the Young group. The mean inhibitory onset latency for the Elderly group was 92 ms and was 122 ms for the Young group. The mean excitatory onset latency for the Elderly was 152 ms and 275 ms for the Young. The QU muscle exhibited an excitatory response in both groups. There was no significant G effect in onset of the early QU response. The Young mean onset latency was 131 ms and the Elderly mean onset latency was 144 ms.

PT2 Condition - BT: During BT, the TA muscle showed an excitatory response. The onset latency of this excitatory phase was significantly greater in the Elderly group as compared to the Young group ($p<0.01$). The mean onset latencies were 139 ms in the Young group and 212 ms in the Elderly group. The initial GA corrective response was excitatory activity. The mean onset latencies were 83 ms in the Young group and 99 ms in the Elderly group. The HA showed a small inhibitory response in both groups. The mean onset latencies were 117 ms for the control group and 141 ms for the Elderly group. This was followed by a larger excitatory response. When looking at the QU muscle, both groups showed an excitatory phase. The mean onset latencies were 116 ms for the Young group and 125 ms for the Elderly group.

PT2 Condition - FT: During FT the pattern of corrective muscle response during PT2 was different than during PT1, as all muscles for both groups except TA exhibit a different sign. For example, GA exhibited an initial excitatory phase in the PT1 condition and an inhibitory phase in the PT2 condition. The pattern of muscle response observed in TA was an initial excitatory response followed by an inhibitory phase. The pattern was observed in both groups and the statistical analysis revealed no G effect in onset latency. In both groups the GA muscle showed an early inhibitory response followed by an excitatory response. There was no G effect in onset of the initial inhibitory response, but the Young group had a significantly earlier onset of excitatory phase as compared to the Elderly group ($p<0.01$). The mean onset latency of the excitatory phase was 99 ms in the Young group and 209 ms in the Elderly group. The HA exhibited an excitatory phase immediately following the perturbation in both groups. The mean onset latency for the Young group was 93 ms and was 89 ms for the Elderly group. This excitatory response lasted throughout stance and into swing for both groups. There was no G effect in onset latency for HA. The QU muscle in the Elderly showed a definite excitation phase with a mean onset latency of 151 ms. This was followed by an inhibitory response starting at 193 ms. The QU muscle in the Young group showed only the inhibitory phase. The mean onset latency was 102 ms.

Right Side (swing): All muscles exhibited early responses from the onset of platform motion for both FT and BT. The range of mean onset latencies was from 71 ms to 340 ms for the right swing condition. Some of the responses were excitatory and some were inhibitory. For example, during BT-PT1, right HA was excitatory in both groups while during BT-PT2 the initial corrective response was inhibitory in the Elderly group. During FT, right TA in the PT1 condition exhibited inhibition in both groups; during the PT2 condition the right TA exhibited excitation in the Young group, and initial inhibition followed by excitation in the Elderly group.

PT1 Condition - BT: During BT, the pattern of corrective muscle responses was similar in both groups. There was no G effect in onset latency. The TA muscle in both groups showed an initial excitatory phase. Mean onset latencies were 109 ms for the Young group and 113 ms for the Elderly group. The TA excitation in the Young group was reduced to NPT condition 530 ms after the perturbation, while the Elderly group did not reach this level until much later. An excitatory phase was observed in the HA muscle in both groups immediately following the perturbation. The mean onset latencies were 71 ms in the Young group and 78 ms in the Elderly group. In both groups QU exhibited an excitatory phase. The mean onset latencies were 131ms for the Young group and 145 ms for the Elderly group.

PT1 Condition - FT: During FT, the pattern of muscle responses was similar in both groups. There was no G effect in onset latency. Both groups exhibited an inhibitory response in TA. The Young group showed an onset latency at 228 ms and the Elderly at 181 ms. The pattern of muscle responses observed in HA was an initial excitation followed by an inhibitory phase. This pattern was observed in both groups. The mean excitatory onset latency for the Young group was 89 ms and was 79 ms for the Elderly group. The mean inhibitory onset latency for the Young group was 132 ms and 158 ms for the Elderly group. The QU muscle in the Elderly group exhibited an inhibitory response. The mean onset latency was 340 ms. The QU muscle in the Young group showed an excitatory phase with mean onset latency of 268 ms.

PT2 Condition - BT: During BT, for all muscles except TA, there was no G effect in the onset latency. The TA muscle showed an excitatory response in both groups. The onset latency of this excitatory phase was significantly greater in the Young group as compared to the Elderly group ($p<0.01$). The mean onset latencies were 164 ms in the Young group and 119 ms in the Elderly group. The initial HA corrective response was excitatory in the Young group and inhibitory in the Elderly group. The mean onset latencies were 149 ms in the Young group and 95 ms in the Elderly group. The HA inhibitory phase in the Elderly was followed by an excitatory response. The mean onset latencies of this excitatory response was 152 ms. When looking at the QU muscle both groups showed an excitatory phase. The mean onset latencies were 152 ms for the Young group and 163 ms for the Elderly group.

PT2 Condition - FT: During FT, the initial TA response was excitatory for the Young group and inhibitory for the Elderly group. The mean onset latency was 225 ms in the Young group and 132 ms in the Elderly group. The Elderly group exhibited an excitatory response following the initial inhibitory response. The onset of this excitatory response was 277 ms. The HA exhibited an inhibitory phase in the Young group and an excitatory phase in the Elderly group immediately following the perturbation. The mean onset latency for the Young group was 158 ms and for the Elderly group 151 ms. There was no G effect in onset latency for HA. The QU muscle in both groups showed a definite excitation phase with a mean onset latency of 294 ms for the Young group and 268ms for the Elderly group.

CHAPTER 4 - DISCUSSION

The purpose of this study was to determine the postural control capabilities for restoring balance in response to sudden disturbances while walking. Walking balance response in healthy elderly subjects was compared to healthy young controls. This was accomplished by fulfilling a series of objectives.

4.1 MAIN FINDINGS

The main findings of the comparison of the task of walking between groups are as follows:

a) the group ensemble averages of angular displacement waveforms of the young and elderly groups were virtually the same in appearance;
b) there was very little difference in the onset times and peak magnitudes of the different phases of angular displacements for the ankle, knee and hip between the right and left sides; and
c) the timing and pattern of muscle activity in TA, HA, QU and GA on the right and left sides during the walks were virtually the same in both groups.

The platform movement during FT and BT induced in all subjects a similar initial disturbance. Trajectories of CM(R)in the first 200 ms were similar in timing and magnitude between the two groups. There was no G effect in onset and magnitude of passive phase of angular displacement at the ankle and hip.

Successful corrective balance responses to FT and BT were present in all trials in both groups. There were no significant G or T effects on:

a) magnitude of initial peak CM(S) displacement;
b) the timing to initial peak velocity; and
c) magnitude of initial peak CM velocity.

The results do show a statistical difference in right swing distance for the first step between the young and elderly groups. PT1 showed G ($p<0.05$) and T ($p<0.001$) effects; PT2 showed T ($p<0.001$)and G*T ($p<0.033$) effects. However, these differences are due to the small sample size and the fact that subject stature (e.g., height) was not taken into account.

The presence of normal balance reactions in both groups was established by:

a) the timing and pattern of the CM(R) displacement;
b) the timing, pattern and magnitude of the active phase of angular displacement; and
c) the timing and pattern of muscle responses bilaterally.

Walking balance was recovered by distinct multi-segment stereotypical movement patterns involving all limb segments on both sides. Deviations from the NPT condition to restore balance were similar for FT and BT, except relative to the NPT condition, where they were opposite in direction. The fundamental change in kinematic trajectories or main divergence that occurred in response to FT and BT was a phase shift in trajectories with no significant change in magnitude. This phase shift was evident during perturbations at PT1 and PT2. This suggests that motor strategy is the same during PT1 and PT2. The corrective response was observed in both groups.

This study demonstrated that active healthy elderly subjects were able to elicit successful rapid balance reactions during FT and BT. There were some differences between the young and elderly at least in EMG patterns and onset latencies.

4.2 NEURAL CONTROL OF REACTIVE BALANCE ADJUSTMENTS DURING LOCOMOTION

In the study by Tang, Woollacott and Chong (1998), two hypotheses concerning balance control during walking were presented. The first hypothesis, which was based on results of Nashner's (1980) study of perturbed over-ground walking, states that muscle inputs from changes in ankle joint displacement (i.e., BT versus NPT condition) is the primary sensory input for triggering the corrective balance reaction. This hypothesis is not accurate, as no one class of sensors (i.e., proprioceptors from ankle muscles) can define the type and direction of the balance disturbance (i.e., CM relative to the base of support).

The second hypothesis presented, which was based on the results of the Dietz, Quintern and Berger (1985) treadmill perturbation study, states that muscle activity from distal muscles alone is sufficient to restore balance in this type of perturbation. Along very similar lines Nashner (1980) also states that perturbations required reactions to change the rate of forward progression, and the only way to do this is to change the torsional moment about the ankle of the supporting leg. In both of these studies, early activation of the stretched muscle (i.e., soleus and GA) was shown on the stance leg, along with inhibition in TA. Activation of the soleus and GA helps to restore the relationship of the CM(R) on the stance side. These muscles are certainly not the only muscles that are required to stabilize the whole body and restore the relationship of CM(R). For example, in this study corrective responses at the knee and hip on both the stance and swing limbs were observed. These cannot be accounted for by a short latency corrective response change in the ankle muscles. The information provided above indicates that the hypotheses proposed by Nashner (1980) and Dietz, Quintern and Berger (1985) are limited.

A number of studies on healthy young humans (Nashner, 1980; Dietz, Quintern and Berger, 1985; Tang, Woollacott and Chong, 1998; Eng, Winter and Patla, 1994) have shown that the changes in normal walking EMG patterns in response to sudden disturbances are rapid in onset (60 to150 ms) and movement specific (i.e., patterns of corrective muscle responses were different when the same disturbance was presented at different phases of the gait cycle). Nashner (1980); Tang, Woollacott and Chong (1998); and Dietz, Quintern and Berger (1985) evaluated patterns of corrective muscle responses to disturbances presented at early stance, mid-stance and late stance, while Eng, Winter and Patla (1994) describe early swing and late swing. The results of the present investigation for heel-off and mid-swing are consistent with these findings. From a biomechanical perspective this phase dependence of muscle activation patterns is obvious. For example, the CM(R) and the joint angular position in the current study were different during heel-off and mid-swing. Since this is true, the muscles around each joint must respond differently at different periods of the gait cycle to produce or control the different movements and to maintain a state of dynamic stability.

A number of animal studies also show that reflexive responses to the same stimulus results in different responses, depending on which point in the gait cycle the stimulus is presented (Wand, Prochazka and Sontag, 1980; and Forssberg, 1979). A number of animal studies have shown the effect of specific classes of peripheral afferent signals in the modulation of the locomotor rhythm generator. For a review of these studies see Van de Crommert , Mulder and Duysens (1998). During induced locomotion in reduced animal preparations there are three sources of sensory signals which can produce a switch from a flexor burst to an extensor burst. These sources are: spindle/golgi tendon organs, cutaneous sensors from the feet, and hip muscle spindles (Duysens and Pearson, 1976; Kriellaars et al, 1994). In addition, 1a and 1b afferents from the lower limb muscles can entrain locomotor rhythm, cause changes in the duration/magnitude of extensor or flexor muscle bursts (Pearson, Ramirez and Jiang, 1992). The analysis in the current investigation has shown immediate changes (in the order of 50 to 100 ms before onset of corrective muscle responses) in joint angular displacements at the ankle and hip during FT and BT compared to NPT. Therefore there would be a differential activation of spindles and golgi tendons following perturbation. Since early divergence in CM(R) has been observed in the present study, it would be expected that early changes in CFP or cutaneous inputs from the feet would also be seen.

Nashner (1980) stated that the phasic activation of the movement generator for locomotion, based on feedback signals from the peripheral afferents, produces the parametric changes in the stepping movement necessary to restore total body stability and maintain the forward progression. Nashner (1980) presented a model to describe how balance is restored during perturbed walking. He proposed that the processes organizing both balance adjustments and locomotor activities are closely related or integrated. He also proposed that, together with the biomechanical properties of the limb segments, the short latency corrective muscle contractions dictate which parameters of the stepping movement will be adaptively modulated during the unexpected disturbance. In the model he postulated a number of adaptive or modifiable elements within the stepping movement generator; in particular rate, duration and magnitude. The adaptive elements refer to those components of the network or modules that receive peripheral feedback in order to make adjustments for unexpected disturbances. The adaptive elements or parameters of the stepping movement generator that are modifiable during a sudden unexpected disturbance within the appropriate time frame were not examined.

The main findings of the current study are as follows:

a) Early corrective muscle responses were observed in all muscles examined. Some showed increased activity relative to the NPT condition and others showed a decrease in activity. The muscle responses were of relatively long duration, specific to the type of disturbance and where in the gait cycle the disturbance was presented;
b) There was a change in swing duration and distance which, as compared to NPT condition, was opposite in direction for BT and FT; and
c) From a kinematic perspective, the basic parameter modulated was phase of the trajectory with no affect on pattern or magnitude. For situations where the foot was displaced backward relative to the CM (BT), an initial increase in the rate of angular displacement, resulting in a phase lead in the trajectory, was observed. For situations where the foot was displaced forward relative to the CM (FT), an initial delay or decrease in the rate of angular displacement, which resulted in a phase lag in the trajectory, was observed. This phase shift was also evident in the CM(R).

These findings would be highly consistent with the view that short latency corrective muscle responses are organized into locomotor like patterns as Nashner (1980) suggested as it was proposed that the process organizing balance adjustments and locomotor activities are closely related or integrated. This basic phase shift in angular trajectories following the disturbances was present at PT1 and PT2; the same disturbance was presented at different phases of gait. Thus motor strategy observed in the current study was not dependent on the point in the gait cycle where the disturbance occurred. The disturbance was the same in the PT1 and PT2. This was different than Eng, Winter and Patla (1994) in that these investigators show different strategies at different points in gait cycle. A possible reason for the conflicting results between the two studies was that the perturbation in each of the studies was different. In the study by Eng, Winter and Patla (1994), the subjects foot contacted the obstacle and so the foot motion was physically stopped. In the study the foot was moved forward or backward.

4.3 AGING EFFECTS ON BALANCE CONTROL

The ability of healthy elderly individuals to maintain postural stability during quiet standing and perturbed standing has been documented in the literature. A number of investigators enlisted the moving platform paradigm during perturbed standing when comparing balance reactions to unexpected perturbations between a population of healthy young and elderly subjects. Each of the studies incorporated different types of balance disturbances and of these studies looked at slightly different parameters. For the most part the investigators looked at patterns and onset latencies of corrective muscle responses. Some investigators studied kinematic variables such as amplitude and timing changes in the center of foot pressure (COP), displacement and CM, and body sway. In general the investigators found age-related delays in the early corrective responses of some muscles to sudden platform displacements. Peterka and Black (1990b) and Woollacott, Shumway-Cook and Nashner (1986) who studied translations and rotations, and Keshner, Allum and Honegger (1993) who studied rotations, found an increase in response latencies of only the distal ankle muscles in the elderly as compared to young healthy adults. There were some conflicting results as Woollacott, Shumway-Cook and Nashner (1986) and Keshner, Allum and Honegger (1993) showed statistically significant increases in response latencies in the distal muscles but in the proximal muscles only some muscles showed delays while others did not. Peterka and Black (1990b) with a larger study group, showed small but not statistically significant increases. Woollacott, Shumway-Cook and Nashner (1986) studied translations while Keshner, Allum and Honegger (1993) studied rotations, which made it difficult to compare the two studies due to the different types of perturbations. The controversy then was between Woollacott, Shumway-Cook and Nashner (1986) and Peterka and Black (1990b), as both studies had similar types of disturbances. Woollacott, Shumway-Cook and Nashner (1986) used a small sample size and Peterka and Black (1990b)a large sample size. Woollacott, Shumway-Cook and Nashner (1986) used a p level of .05 and showed an onset time of TA muscle was 102(6) ms for the young population and 109(9) ms for the elderly population. It was evident that, even though Woollacott, Shumway-Cook and

Nashner (1986) stated the results were significant, the actual difference in onset latencies between the young and the elderly was small in ms.

The current study recorded 7 muscles in response to FT and BT at PT1 and PT2, which add up to 28 muscle responses. In the 28 muscle responses their was a statistically significant G effect in onset latency in five out of the 28 and a sign change in three out of the 28 observed. Of the muscles that showed a significant group effect some were in the form of delayed elderly responses and some responses occurred earlier in the elderly. Thus the majority of the muscle responses (20/28) were not significantly different between groups. The results of the current study are consistent with Peterka and Black's (1990b)and Woollocott, Shumway-Cook and Nashner's studies in that Peterka and Black found no significant differences in onset latencies and Woollocott, Shumway-Cook and Nashner found a minor but significant difference in onset latencies of the distal muscles only.

Gu et al (1996) looked at CM displacement during standing translations. The results indicated that the elderly exhibited a higher frequency of both CM and a larger support surface reaction. The elderly recovered postural stability less rapidly and with more oscillations in sway. In the current study, CM displacement during walking was also investigated. The results indicated that the elderly did not exhibit a different CM(S) or CM(R) displacement than the young. The reason for the difference in results was probably due to the fact that the tasks in the two studies were different.

McIlroy and Maki (1996) looked at age-related differences in spatial and temporal characteristics of compensatory stepping during FT and BT perturbations. Their study investigated platform translations of high acceleration/velocity greater than in the current study. The main finding of the McIlroy and Maki study was that, for most of the many parameters they analyzed, there was no G effect. The parameter that showed a significant G effect was the responses in the frontal plane which the current study did not examine. Future researchers should complete a study similar to the present study, but should also include data analysis in the frontal plane.

The majority of parameters studied so far in the limited set of experimental tasks show very little difference between the healthy young and the healthy elderly in performance and the means by which the task was performed.

4.4 ROLE OF CORRECTIVE MUSCLE RESPONSES IN RESTORING TOTAL BODY STABILITY

Muscle onset latency suggests that a polysynaptic response and postural activity from the leg and thigh muscles are key to balance control during perturbations during heel-off and mid-swing. Bilateral activation of leg muscles is coordinated to overcome balance threats. With these muscle responses, walking velocity is minimally disrupted and joint trajectories return to a usual path within the gait cycle.

<u>Left Side (stance) - PT1:</u> Translating the platform moves the foot backwards during BT and forwards during FT, but because of inertia the proximal segment lags behind. This is what leads to a change in CM(R). This also leads to an increase in plantarflexion during FT and a decrease in plantarflexion during BT.

At the ankle, the passive response during BT was an increase in dorsiflexion and during FT a decrease in dorsiflexion. Statistically, the time to peak dorsiflexion was earlier in the BT and delayed in the FT relative to NPT condition. In other words, there was a decrease in duration of the BT dorsiflexion phase and an increase in duration of the FT dorsiflexion phase. The muscles' responses observed during BT were a decrease in TA activity and an increase in GA relative to the NPT activity. For the FT condition there was an increase in TA activity and a decrease in GA activity. The pattern of muscle activation would result in a decrease in the early dorsiflexion phase and thus earlier onset plantarflexion during BT, and an increase in dorsiflexion phase during FT. The onset latencies of activity of the TA muscle in BT and GA in FT were earlier in the young than the elderly. This was evident as the time to peak dorsiflexion was earlier in the young group.

There was no immediate divergence of knee angular displacement for FT and BT in either group, compared to NPT trials. The corrective response at the knee for FT was that knee flexion occurred later compared to the NPT condition. The corrective response at the knee for BT was that knee flexion occurred earlier compared to the NPT condition. Thus, there was a phase shift in knee flexion beginning at late stance in preparation for push-off and swing phase. For FT, inhibition of HA compared to NPT was observed. This could result in a delay or phase shift knee flexion. Late HA excitation occurred to produce later knee flexion. For BT, excitation of the HA compared to NPT was seen to allow knee flexion to occur earlier. For both groups the QU exhibited excitation in both the FT and BT condition. Early onset of knee flexion for BT and delay onset of knee flexion for FT were observed. Early QU excitation during FT would assist in delayed knee flexion, i.e., eccentric contraction. However it is not clear what the role of early QU excitation would be during BT, as earlier knee flexion was required. Here other factors such as external forces must be taken into consideration to explain the trajectory of knee angular displacement. This may be considered in future studies.

Left Side (stance) - PT2: At the ankle, the passive response during FT was a decrease in dorsiflexion. Statistically the time to peak dorsiflexion was delayed in the FT condition. In other words there was an increase in duration of the FT dorsiflexion phase. The muscles needed to overcome the results of the passive component of the perturbation and delayed onset of the plantarflexion phase for the FT condition would be an increase in TA activity and a decrease in GA activity. This was observed in both groups. After this initial GA inhibitory phase an excitatory phase was then observed. Similarly the TA excitatory was followed by an inhibitory phase relative to NPT in both groups. Both groups exhibited an early excitatory phase in GA activity during BT. This was not accompanied by an inhibitory phase in TA as was seen in the PT1 condition; instead, a TA excitation was observed. For the young group the onset of TA excitatory phase was approximately 50 ms after the onset of the GA excitatory phase. In the elderly group the TA excitatory phase was significantly delayed as compared to the young group. From the kinematic plots (Figure 3B) the corrective response observed at the ankle was an increase in the rate of plantarflexion and a decrease in time to peak plantarflexion. What was required to make this change was an early GA activation. Based on kinematic results, the role of increased TA activity is not clear.

At the time the HA and QU muscles are responding to the FT, the knee was going into flexion. The difference in knee angular displacement between FT and BT, as seen in Figure 3B, was a decrease in rate of knee flexion and for the young an increase in time to peak knee flexion. The elderly did not show this increase time to peak knee flexion for FT as compared to NPT. There was also no difference in magnitude of knee flexion between FT and NPT. There were two ways that the body might decrease the rate of knee flexion and thus phase shift the trajectory without a magnitude effect. The two possibilities were a decrease in HA and an increase in QU during FT as compared to NPT. The opposite was observed in the current study, as there was an increase in HA and a decrease in QU. This activity pattern can not explain the changes that occurred at the knee joint. This requires a kinetic analysis, as motion dependent and gravity dependent forces are involved with such a complex motion.

In BT the main trajectory change at the knee for PT2 was similar to that seen during PT1, but not as marked as shown in Figure 3B. The effect was to increase the rate of knee flexion during BT as compared to the NPT and also to decrease the time to peak knee flexion for BT compared to NPT. This was statistically significant as there was a T effect. An early increase in HA activity was needed to produce the increased rate of knee flexion; however, an early HA inhibition followed by HA excitation and a QU excitation actually took place. Again, the role of HA and QU cannot easily explain the kinematic changes observed during BT.

<u>Right Side (swing) – PT1:</u> The ankle was plantarflexing when the platform began to move. Statistically the time to peak plantarflexion was earlier in the BT condition and delayed in the FT condition relative to the NPT condition. In other words, there was a decrease in duration of the BT plantarflexion and thus there was earlier dorsiflexion compared to the NPT condition. In the FT condition there was an increase in duration of the plantarflexion and thus later dorsiflexion occurred. In the BT condition, divergence started during early stance, which was 250 ms after the onset of platform motion; in the FT condition, divergence started 350 ms after the platform motion. One would expect to see an increase in TA activity during the BT, and an inhibition of TA during the FT active phase. In both the young and the elderly groups during BT there was an increase in TA activity relative to the NPT condition. In both the young and the elderly groups during FT there was an inhibition of TA activity relative to the NPT condition.

For the BT condition there was less knee extension during early stance than the NPT condition, and in the FT condition there was no significant difference in knee extension as compared to the NPT condition. The muscle activity necessary to reduce the knee extension in the BT condition are QU inhibition and HA excitation. In both the young and the elderly groups there was an increase in QU and HA activity relative to the NPT condition. This would contribute to knee joint stiffness which may be necessary for knee stability during BT, as during BT the foot comes in contact with the ground sooner than in the NPT condition. Given the similar trajectories of knee angular displacement for the FT and NPT conditions, it is unclear as to the role of the corrective responses in HA (excitation followed by inhibition) and QU (excitation).

Right Side (swing) - PT2: At the start of platform movement the ankle was moving into plantarflexion. Divergence in trajectories of the ankle angular displacement for BT and FT as compared to the NPT condition did not begin until early stance. At this point the ankle was moving into dorsiflexion. Here the rate of ankle dorsiflexion was increased during BT and decreased during FT compared to the NPT condition. Thus, one would expect to see an excitatory response in TA during BT and an inhibition of TA during FT. In both the young and the elderly groups during BT there was an increase in TA activity relative to the NPT condition. In both the young and the elderly groups during FT there was also an excitatory TA response. In both groups the onset of the TA excitatory response during FT was 70 to 90 ms later that for BT. In the elderly group the excitatory TA response was preceded by an initial period of inhibition.

The knee was moving into extension at the start of platform motion. Divergence in trajectories of the knee angular displacement for BT and FT as compared to the NPT condition did not begin until early stance. During BT there was a greater amount of knee flexion than in the NPT condition, and during FT there was an increase in knee extension as compared to the NPT condition. In both the young and the elderly groups during BT there was an increase in QU and HA activity relative to the NPT condition. This could contribute to knee joint stiffness, which may be necessary for knee stability during the knee flexion phase as, in this case, the foot comes in contact with the ground sooner than the NPT condition. Initially during BT the elderly group exhibited an inhibition in HA which cannot be explained kinematically. During the FT condition both the elderly and the young exhibited excitation of QU but only the young group showed a reduction in HA activity. This pattern of muscle activity would contribute to the greater knee extension observed during FT. The elderly exhibited HA excitation which can not be explained kinematically.

4.5 CONCLUSIONS

The present findings are consistent with the integrated model of balance control during walking proposed by Nashner (1980), regardless of whether the subjects were young or elderly. The kinematic parameter most often altered in the present study was the phasing of the trajectories of angular displacement. The vast majority of angular displacements were not magnitude modulated in either group. Both the young and elderly demonstrated a phase shift between BT and FT. The phase modulation was observed when sudden platform translations were presented at heel-off and mid-swing.

Given the type and level of disturbance used in the present study, there were very few age-related effects in corrective responses to sudden balance disturbances during walking. Some notable differences were present which may reflect the differences in the way the task was performed. For example, in the ankle the elderly exhibited a greater onset time to peak dorsiflexion in left stance, which may account for some of the group difference in EMG results (i.e., TA/GA activation patterns and onset latencies).

Future studies should include: a) kinetic analysis, as in some cases the EMG cannot explain the kinematic changes in angular displacement seen in the present study; b) gait perturbation studies with analysis of body disturbances and corrective responses in the frontal plane; and c) gait perturbation studies on elderly subjects who have a documented history of falling or balance impairment.

CHAPTER 5 - LIMITATIONS AND ASSUMPTIONS

This study was not without limitations. One major limitation resided in the fact that this investigation only collected information in the sagittal plane; thus, important information as to changes that occurred in the frontal plane were overlooked. Secondly, the sample size was small. Thirdly, the walking area was small and this naturally implied the subjects never reached steady state gait. This also meant the trials started at different phases of gait. Another important limitation of this study was that it was unknown as to whether the two velocities chosen challenged the healthy groups enough, yet would not be too difficult for individuals with neurological difficulties, as this investigation was a baseline study for future investigations in the field.

There were a number of known assumptions that were made. The first was that the response evoked by the unexpected movement provided in the study was a true representation of an automatic balance reaction caused by a perturbation that occurs in daily living. Secondly, the subjects in this study were representative of individuals in the healthy active young and healthy active elderly general population. It was also assumed that the links were rigid, the body segment parameters used mirrored these groups, the axis of rotation was a frictionless pin joint, and the centre of mass of the segments were positioned on the line between the joint markers. It was assumed the fluid effects of air (in other words, wind resistance) were negligible. It was also assumed that the contact forces of the harness were non-existent.

CHAPTER 6 - REFERENCES

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