MODULATION OF VARIOUS STEPPING MOVEMENT PARAMETERS DURING PERTURBED WALKING AT SPECIFIC PERIODS OF THE GAIT CYCLE

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ABSTRACT

Epidemiological studies showed that a large percentage of the falls in the elderly occur while they are walking. A large proportion of these falls involves falling laterally. Balance control during perturbed walking is less studied, especially in the frontal plane. Information regarding mechanisms of balance control during perturbed walking in the frontal plane is lacking. In Nashner's study (1980), he proposed that the neural process organizing balance adjustments and locomotor activities were integrated. He also speculated that together with the biomechanical properties of the limb segments, the short latency corrective muscle contractions indicated which parameters of the stepping movements would be modulated during unexpected disturbance. There are a number of modifiable parameters within the stepping movement generator, i.e., rate, duration, and magnitude.

This study investigated the corrective responses of the healthy adults in both sagittal and frontal planes during unexpected disturbances while walking. The purpose of the study was to examine which parameters are modified in both sagittal and frontal planes. Ten active healthy adults participated in the study. Walking was perturbed using a movable platform installed in the middle of a walkway. The types of balance disturbances included backward translation and forward translation in the sagittal plane. These were presented at different points of the gait cycle, specifically during the single support phase, just after right heel off and just after heel strike. The corrective balance responses of bilateral gluteus medius, adductor magnus, rectus femoris, hamstrings, anterior tibialis and gastrocneumius were analyzed. The trajectories of center of mass (CM) relative to space CM(S) were quantified to compare task performance levels between perturbed walking and unperturbed walking condition. The timing, phase changes in the trajectories of angular displacement, CM displacement relative to foot CM(F) were examined in sagittal and frontal planes.

The present results and those of the previous study (Szturm and Riediger, 1999) demonstrate that the change in trajectories of angular displacements, trunk segment rotation, and CM(F) displacement which occurred in response to FT and BT relative to NPT was a phase shift with no significant change in pattern or magnitude. This phase shift was evident for perturbations during heel-off and midswing of the initial step and just after heel off in the second step. These responses occurred on both sides of the body. Corrective muscle responses were observed in all muscles. One primary EMG finding of this study was the definite trend for EMG responses to BT to be excitatory while those for FT to be inhibitory. In the sagittal plane, the change in trajectories of angular displacements, trunk segment rotation, and CM(F) displacement that occurred in response to FT and BT relative to NPT was a phase shift with no significant change in pattern or magnitude. In frontal plane, the change in the trajectories of hip angular displacements, trunk and shank segment rotations which occurred in response to BT and FT relative to NPT also was a phase shift, with no significant change in magnitude. This phase shift was evident for perturbations during the initial step (PT-HO) or during the second step (PT-HS).

These findings were highly consistent with the statement that the early muscle responses are organized into locomotor like patterns, and that the processes organizing balance adjustments and locomotor activities are integrated.

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CHAPTER I - REVIEW OF LITERATURE

1.1. INTRODUCTION

Falls are a major health care concern for the individual, family and health care system, especially falls in the elderly. The incidence of falls and fall-related injuries has been reported in many epidemiological studies. Approximately one-third of the older adults over the age of 65 falls each year (Gryee et al., 1977; Blake et al., 1988; Tinetti et al., 1988; Campbell et al., 1989). Falls in the elderly often lead to serious physical injuries, including bone fractures (Speechley and Tinetti, 1991; Maki and McIlroy, 1996; Judge and Ounpuu, 1996). Falls are also the leading cause of injury-related death and hospitalization in persons aged 65 years and over (Baker, 1985).

The literature reports that most of the falls among elderly adults occur while they are walking (Overstall et al., 1977; Tinetti, Speechley and Ginter, 1988). Thirty to seventy percent of these falls are due to trips, slips, and missteps (Gabell et al., 1985; Lord et al., 1993; Topper et al., 1993). Maki et al. (1994) interviewed 59 residents aged 62 to 96. During a 1-year prospective monitoring period, 120 falls were self-reported. Thirty-two percent of falls were "CM falls", which were perturbation to center of mass (CM), like push, collision, upper extremity movement or bending. Fifty-four percent of falls were "BS falls", which had perturbation to base of support (BS), such as a trip or tangle of feet/legs, a slip, and over-stepping the curb. This group of researchers (1990) using video cameras recorded 25 falls in a period of 15 months in the lobby areas of a geriatric center.

A large proportion of falls occurs in the lateral direction. A prospective study of residents of seniors' apartments showed this to be the case in 42 of 95 falls (Maki et al., 1994; Topper et al., 1993). The elderly people had difficulty in controlling lateral stability (Maki et al., 1994). The older adults often stepped laterally to recover balance even when balance was perturbed in the antero-posterior direction (McIlroy et al., 1996). They used wider stride to increase stability (Murray et al., 1969). The individuals who feared falling tended to show decreases in stride length or increase in the period of double support. These changes may increase stability by minimizing the forward excursion of the CM beyond the BS provided by the stance foot (Maki, 1997).

This evidence supported that an impairment in the medial-lateral direction during walking in the elderly individuals. To date, our knowledge about balance responses to unexpected disturbance during walking is from a few studies about young healthy subjects (Nashner, 1980: Dietz et al., 1986; Eng et al., 1994; Tang et al., 1998). The analysis of corrective balance reaction during walking in the medial-lateral direction is lacking. It is important to explore the mechanisms for balance control in the frontal plane during walking.

1.2. THE DEFINITION OF BALANCE CONTROL

The human body is a multi-link system, where the stability of each link or body segment is achieved by maintaining a balance of external and internal forces acting at each joint (Putnam, 1991). External forces are gravity-dependent forces, motion-dependent forces (or joint reaction forces), and ground reaction forces. Internal forces are muscular forces, bony contact forces, and visco-elastic forces of tissues spanning a joint.

In quite standing, balance control is often defined as the ability to maintain and control position and motion of total body center mass (CM) relative to the base of support (BS) (Winter, Palta, and Frank, 1990; MacKinnon and Winter, 1993; Maki and McIlroy, 1996; Pai and Patton, 1997). Pai and Patton (1997) used a single inverted pendulum model with a foot segment to predict the feasible CM movements during which balance can be maintained. The upper and lower boundaries of the feasible velocity-position region were determined under different constraints which included size of the base of support and simulated slippery conditions where the surface coefficient of friction was reduced to an extremely low value. Forward falls would be initiated if the states exceeded the upper boundary, and backward falls if below the lower boundary. The upper boundary ran from a velocity of 1.2 h/s (body height per second) at 2.4 foot lengths behind the heel to 0.45 h/s over the heel. The lower boundary was from a velocity of 0.9 h/s at 2.7 foot length behind the heel to zero over the heel. The other main finding in this study was the effect of reduced friction on the boundaries of feasible velocity-position region. The friction levels (μ) were less than 0.82 dramatically limited the upper and lower boundaries of the feasible region.

The task of regulating balance during walking is an extremely complex motor control problem. The balance control tasks used during human walking are to maintain control over the position and motion of the CM relative to a changing BS, and to control the placement of the swing leg (Judge and Ounpuu, 1996). During steady state walking, the center of mass is outside the base of support for approximately 75-80% of the stride (Patla, 1992). The single support phase is the most difficult period in which to maintain stability (Maki et al., 1996). During single support phase, the CM moves forward along the medial border of the foot within BS. During the double support phase, the CM is between the stance feet within BS.

The potential instability in the frontal plane is from the high upper body inertia, the small base of support, and long single support periods (MacKinnon & Winter, 1993). The researchers described a whole-body inverted pendulum model to study the control of balance in the frontal plane during human gait. They examined the resultant joint moments, joint acceleration, and gravitational forces acting about the supporting foot and hip during steady gait. Whole body balance was ensured by the center of mass (CM) passing medial borders of the supporting foot oscillating less than 2 cm. At heel contact, the new supporting foot was placed a mean of 7.7 cm lateral to the CM and passed the midline plane of progression in mid-double support. The medial acceleration of the CM was primarily generated by a gravitational moment about the supporting foot.

1.3. CORRECTIVE BALANCE RESPONSES

The normal central nervous system (CNS) provides basic balance objectives: 1) to maintain balance at rest or during voluntary movement; and 2) to restore body stability when there is an unexpected disturbance or sudden loss of balance. There are two mechanisms of balance responses, in general: 1) a feed-forward mechanism which is a preparatory or anticipatory process for pre-planning of stability requirement during voluntary movement; and 2) a feed-back mechanism to deal with the unexpected or sudden loss of balance often referred to as corrective balance response. The main focus in this research was to examine corrective balance responses.

For the normal regulation of balance control, three neural processes are needed: 1) sensory receptor systems, i.e. somatosensory, vestibular, visual sensory systems which provide information regarding body orientation and motion of the body segment relative to each other and to the environment; 2) central nervous system (CNS) processes to organize and integrate the different sources of spatial/motion information from the somatosensory, vestibular, visual sensory systems; and 3) the motor centers which select and execute the goal-orientated compensatory reactions or adjustments (Dietz, 1992; Palta, et al., 1992).

Multiple sources of sensory information are needed to determine the threat to balance, i.e. body segment positions/motion relative to each other and to space. The vestibular, visual and somatosensory sensory systems have unique frames of reference and thus provide different types of orientation and motion information. These three systems are responsible for proprioceptive and exteroceptive input. The somatosensory system provides proprioceptive and exteroceptive information. The muscle spindles and joint receptors provide the proprioceptive information regarding motion relative to adjacent segments, i.e. internal reference system. The cutaneous afferents of the feet provide the exteroceptive information, such as foot pressure and location of support surface. The vestibular input is always available under normal conditions. Vestibular receptors have the only earth-fixed reference frame, detecting head position relative to gravity. The visual system can provide accurate spatial information regarding body relation with environment and environment features.

A number of studies of animals and humans have used the moving platform paradigm to suddenly disturb standing balance (Nashner, 1976; Horak and Nashner, 1986; Woollacott et al., 1986; Rushmer et al., 1987; Diener, Horak and Nashner, 1988; Nardone et al., 1990; Keshner, Allum, and Honegger, 1993; Allum et al., 1995; Szturm and Fallang, 1998). The experimental approach is to translate or rotate the support surface, which results in a sudden change in the position of the base of support relative to CM. This requires a rapid corrective response to restore standing balance.

In the studies of humans, different researchers recorded muscle activities from different muscles. But all the results of different studies showed that muscle activities occur within a range of 60-170 ms following the onset of the balance disturbance (Horak and Nashner, 1986; Woollacott et al., 1986; Allum et al., 1995; Szturm and Fallang, 1998). These onset latencies of muscle responses are longer than stretch reflex latencies (40-50 ms) (Allum et al., 1995; Nardone et al., 1990), but shorter than voluntary reaction times (180-250 ms).

1.4. PHYSICAL CONSTRAINTS MOVEMENT SYNERGIES

The corrective balance responses are dependent on physical constraints. Various physical constraints challenge the different levels of stabilizing capabilities of the balance control system, and thus they require different muscle activation patterns and movement synergies.

Different levels of acceleration, not just simple target velocity of platform, result in single or multi-link movement synergies. Small and slow amplitude disturbances produce an "ankle strategy" or a single link movement synergy (Horak and Nashner, 1986; Diener, Horak and Nashner, 1988). Large and fast magnitude perturbation produces multi-segmental movement synergies involving movement about all joints (Nardone et al., 1990; Szturm & Fallang, 1998). If the magnitude of perturbation is even larger and faster, stepping responses will be observed (McIlroy and Maki, 1996).

A distal to proximal pattern of muscle activation in response to horizontal (forward or backward) platform translations was reported in several early studies of small and slow magnitude disturbances (Nashner 1976, 1977; Horak and Nashner, 1986; Diener, Horack and Nashner 1988). The "ankle strategy" was used when the muscle activity began in the ankle joint muscles, and these early ankle muscles were primarily responsible for restoring balance. In response to backward translations (BT), only dorsal body side muscles were activated. It was found that gastrocnemius/soleus muscles were activated, followed by hamstrings and paraspinals, but the muscle activity on the anterior side of body was absent or minimal. In response to forward translations (FT), only anterior body side muscles were active. Tibialis anterior, followed by quadriceps, then abdominals, were activated, while muscle activity on the posterior side of body was absent or minimal. The body behaved as an inverted pendulum, rotating about the ankles with no appreciable motion about the knees and hips. This activation pattern restored balance by moving the CM forward or backward back to its normal position centered over the displaced base of support.

To restore balance in response to fast or large disturbances, different and more complex movement synergies were used. The multi-segmental movement synergies involved concurrent motion about the ankles, knees, hip and trunk to restore balance (Nardone et al., 1990; Szturm and Fallang, 1998).

Larger and faster support surface movements resulted in stepping responses (McIlroy and Maki, 1996). The peak acceleration of platform in FT was 1.5 m/s^2, with maximum velocity 0.45 m/s and displacement 0.135 m. The peak acceleration of platform in backward direction was 2.0 m/s^2, with maximum velocity 0.6 m/s and displacement 0.18 m.

Another physical constraint, which changes movement synergies, is the extent of support surface. In Horak and Nashner's study (1986), the normal ankle synergy was not observed when the platform was translated forwards or backwards while subjects stood on a narrow beam. The "Hip strategy" was observed. During the hip synergy, a proximal-to-distal order of muscle activity was observed, whereas the ankle muscles were unresponsive. In BT, abdominals followed by quadriceps were activated, and in FT paraspinal muscles were active followed by hamstring muscles. The researchers noticed that the magnitude of the horizontal shear force changed during hip synergy as compared to ankle synergy. Standing on a narrow surface reduced the contact area not contact forces, and thus there would be less total friction between foot and ground. Gu et al. (1996) compared balance responses between healthy young and elderly subjects. Twenty-four healthy young (mean ages 26) and fifteen healthy elderly adults (mean ages 72) performed quite standing on the platform or beam, which accelerated forward. All subjects stood upright with eyes closed during every task. In each flat translation and beam translation, the platform accelerated at 1.67 m/s^2 and 0.89 m/s^2 for 100 ms, followed by 100 ms target velocity period and then an equivalent deceleration. This acceleration of platform in this study could be considered low. Support surface reactions [vertical and anteroposterior(AP)] were measured directly by force plate. A seven-link biomechanical model was used to analyze response dynamics. Center of mass (CM) displacement, center of reaction (or center of foot pressure, CFP) displacement, and joint moment data were calculated by using measurements from this multi-link biomechanical model. The results of this study showed that there were relatively small, but perturbation-specific differences in responses of elderly subject to perturbations of standing. The mean maximum CM excursions were significantly larger in the elderly in beam standing. The mean maximum CFP excursions were found significantly larger in the elderly adults during both beam standing and beam forward translation. During both flat standing and beam standing perturbation, the elderly used larger reactive joint moments and developed larger CFP displacement. In both young and elderly groups, the foot AP shear force was significantly reduced during the narrow beam translation as compared to the normal support surface translation condition. In the elderly group, the AP shear forces in beam standing and beam perturbation were significantly larger than in the young group. This suggested that the elderly tend to use more trunk motion to maintain balance, especially in the narrow beam standing and narrow beam perturbation.

1.5. CENTRAL DETERMINED CORRECTIVE BALANCE RESPONSES

It is commonly accepted that the corrective balance responses are centrally programmed not dependent on short latency segmental reflexes. The central neural network regulates balance parameters, such as the position/velocity of the center of foot pressure or center of body mass relative to the base of support (Horak and Nashner, 1986; Diener, Horak and Nashner, 1988; Nardone et al., 1990; Allum et al., 1995). But how and what neural centers are involved and responsible for this neural network are largely unknown.

Gorassini et al. (1994) investigated the corrective reactions in intact cats when one hind-limb unexpectedly walked into a hole in the support surface. They found that the test animal removed the foot from the hole quickly and the knee and ankle showed strong flexion. The onset latency of flexor muscle activity was in the range of 35-65 ms. The support of the contralateral hind limb was maintained by a prolongation of stance. The researchers suggested that these corrective responses are initiated via supraspinal pathways. To further examine the contribution of supraspinal mechanisms, Hiebert et al. (1994) [continuation of the experiment of Gorassini et al. (1994)] compared the responses of intact and chronic spinal cats. The chronic spinalized cats had recovered the ability to step with hind limbs on a moving treadmill. The onset of a corrective response was the first visible flexion motion at any of hip, knee, or ankle after the entrance of the limb into the hole. Compared to spinal cats, the intact cats showed shorter onset and larger amplitude of flexion responses of the hind limb which entered the hole (70-150 ms, control, compared to 130-350 ms, spinal cats). Also, the flexion motions of spinal cats were too weak to successfully withdraw the leg from the hole. It was concluded that supraspinal structures were required to generate functionally appropriate corrective responses. Both intact and spinal cats maintained support by the contralateral leg when the foot entered the hole. The stance phase of the contralateral leg was prolonged and extension increased. Thus, they suggested that during the corrective response, it was at the spinal level where the contralateral leg maintained support. This provided direct evidence for supraspinal structures in restoration of balance following unexpected disturbances.

1.6. AGE-RELATED CHANGES IN THE CORRECTIVE BALANCE RESPONSES TO PERTURBED STANDING

The age-related changes in corrective balance responses to perturbed standing were studied by sudden platform displacements translations (Woollacott et al., 1986; Peterka et al., 1990) and rotations (Keshner et al., 1993), and body pulls (Luchies et al., 1994).

Ankle synergy, which results from small and slow perturbation, was observed in the study by Woollacott et al. (1986), Peterka et al. (1990) and Keshner et al. (1990). The consistent findings in all these studies were that EMG onset latencies of early corrective response were slight (within 20 ms), but significantly increased in distal muscles (TA or G) in the elderly group.

Luchies et al. (1994) examined the age-related stepping responses to sudden backward pulls at the waist. At small disturbance level, the young adults mostly responded by taking a single step, whereas the elderly responded by taking multiple steps. The elderly took significantly shorter and earlier steps than the young.

McIlroy and Maki (1996) investigated age-related differences in balance control using larger and faster support surface translations that required a stepping response to restore balance. Balance reactions were examined in response to 600 ms platform translations. The peak acceleration of platform in forward direction was 1.5 m/s^2, with maximum velocity 0.45 m/s and displacement 0.135 m. The peak acceleration of platform in backward direction was 2.0 m/s^2, with maximum velocity 0.6 m/s and displacement 0.18 m. Two groups of subjects were 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. A number of forward and backward platform translations were randomly presented in the study. The spatial and temporal characteristics of stepping responses to the unexpected platform perturbations were measured. The timing measures were defined with respect to the onset of platform acceleration. The spatial and temporal dependent variables included: onset latency of stepping, time to foot-off, time to foot contact, unloading phase duration, swing duration, step length, swing velocity, center of mass (CM) displacement in medial/lateral and anterior/posterior direction, and velocity of CM at time of foot contact. The onset of stepping was defined as the onset of mediolateral asymmetry and the onset of unloading of the swing limb. The onset of ML asymmetry was defined to be the onset of divergence of the left and right vertical ground reaction forces (divergence greater than 2% of body weight within 20 ms). It was found that ninety-eight percent of the trials resulted in stepping responses in both young and the elderly groups. Temporal characteristics did not differ statistically between these two groups. In first step the only temporal parameter which showed a statistically significant, age-related difference was the medial/lateral asymmetry onset ($P=0.04$). Sixty-three percent of the trials in the elderly presented multiple stepping to regain balance, while thirty-five percent of trials revealed it in the young. The frequency in the elderly was almost twice that of the young adults. In the elderly group, these extra steps were often directed more in the lateral direction, although the perturbation was given in the antero-posterior direction. The foot stepped laterally as much as 28 cm during the second step. The researchers suggested that the stepping response, which resulted from the errors or delays in the foot placement, allowed the elderly to adjust the new BS to match the changing CM during perturbation. The presence of multiple steps may be due to the inability of creating the rapid and effective adjustment of the BS. The researchers concluded that stability in lateral direction is more affected in the elderly than in young adults, and that the elderly population may have difficulty in controlling CM in medial-lateral direction in response to sudden disturbance of standing balance.

1.7. CORRECTIVE BALANCE RESPONSES TO PERTURBATIONS DURING WALKING

A number of animal experiments have examined short latency segmental reflex responses during different phases of gait (Forssberg, 1979; Wand et al., 1980). The reflex responses to the same external stimulus results in different responses when presented at different points in the gait cycle.

It has been determined that three different types of sensory inputs can cause a switch between the alternating flexor and extensor bursts during induced locomotion (see Van De Crommert et al., 1998 for review). These three specific sensory sources are: a) proprioceptive afferents (group Ia muscle spindles, and group Ib golgi tendon organs) in extensor muscles (Whelan et al., 1995; Pearson et al., 1992; Guertin et al., 1995); b) exteroceptive afferents (gourp II cutaneous afferent) from the feet (Duysens et al., 1976); and c) afferent signals from hip muscle spindles and hip joint (Grillner et al., 1978;

Anderson and Grillner, 1983; Kriellaars et al., 1994). Animal studies which have examined the effects of peripheral afferent signals on the rhythm generator at spinal or brain level have showed that group Ia and Ib from the lower limb muscle can entrain the locomotor rhythm (Pearson et al., 1993), change the duration and/or magnitude of extensor muscle burst (Duysens et al., 1980), and delay onset of flexor burst (Conway et al., 1987). Other studies have shown that during the stance phase, group Ib and group Ia afferents from extensor muscles prolong the extensor burst (Whelan et al., 1995; Guertin et al., 1995) and inhibits the flexor half center, thus delay the flexor burst (Duysens et al., 1980). At the end of stance phase, group Ia afferents from flexor muscles activate the flexor half center and initiate the swing phase.

A phase-dependent response to perturbations applied during human locomotion has been documented. A number of studies on healthy young adults have examined the corrective balance responses to perturbations during walking in the sagittal plane (Nashner, 1980; Dietz et al., 1986; Eng, Winter & Patla, 1994, 1997; Tang et al., 1998).

Nashner (1980) studied the corrective muscle response of tibialis anterior (TA) and gastrocnemius (G) during platform translation. The perturbation during walking were randomly performed at: 1) heel strike; 2) beginning of the single-support phase; 3) crossover from negative to positive shear force; and 4) beginning of the double-support phase. In response to perturbations at heel strike, there were muscle responses in the stretched ankle muscles that began approximately 95 to 110 ms after the onset of platform movement and lasted for about 100 to 400 ms. During the forward translation of the platform at heel strike, the activity of the ipsilateral TA was excitatory. During the backward translation of the platform at heel strike, the activity of both ipsilateral G and

TA was excitatory. The strongest balance responses in this study were found when the platform movement was imposed at heel strike and at the beginning of the single-support phase.

Tang et al. (1998), in a similar study to Nashner (1980), confirmed that earlier onset, longer burst duration and greater magnitude muscles response presented during the phase-dependent perturbed gait. This consisted of an early activation of bilateral anterior leg muscles and both anterior and posterior thigh muscles. These responses were of short latency (90-140 ms), high magnitude (four to nine times muscle activity during normal walking), and relatively long duration (70-200 ms).

Dietz et al. (1986) examined the muscle response of G, TA, rectus femoris (RF), and biceps femoris (BF) following an obstruction of the forward swing leg at the different phases while walking on a treadmill. Eng, Winter, and Patla (1994, 1997) investigated corrective balance responses to foot obstruction during early and late swing phases of walking. All these studies supported that the changes of EMG patterns in response to sudden disturbance were rapid in onset (60-150 ms) and movement specific. The muscle activation pattern, magnitude, onset latency and the signs of excitatory or inhibitory responses were specific to the phase of gait in which the disturbance was presented.

Nashner (1980) stated phasic activation of the movement generator for locomotion, via feedback signals from peripheral afferents, produced parametric changes in stepping movement to restore total body stability and maintain the forward progression of the body. He hypothesized that the processes organizing the balance adjustments and locomotor activities were closely related or integrated. He also speculated that together with the biomechanical properties of the limb segments, the short latency corrective muscle contractions indicated which parameters of the stepping movement would be modulated during unexpected disturbance. There were a number of modifiable parameters within the stepping movement generator, i.e. rate, duration and magnitude. It is unknown which modifiable parameters of the stepping movement generator are modifiable during sudden or unexpected disturbance in the appropriate time frame. This question was addressed in this study by exploring the mechanisms of corrective response to perturbed walking in a specific period of gait cycle.

The studies described above about the phase-dependent responses to perturbations during walking were based on sagittal plane. Little attention has been paid to the frontal plane.

CHAPTER II – STUDY PURPOSE AND OBJECTIVES

Epidemiological studies showed that a large percentage of the falls in the elderly occur while they are walking. A large proportion of these falls involves falling laterally. The direction of falls suggests an age-related affect on the medial-lateral stability during walking. McIlroy and Maki's study (1996) suggested that the more frequent use of lateral steps implies greater lateral instability in the elderly people when their balance is disturbed. Balance control during perturbed walking is less studied. A few studies on healthy young adults and animals have examined the corrective balance responses to perturbations during walking in the sagittal plane (Nashner, 1980; Eng, Winter and Palta, 1994, 1997; Tang et al., 1998). However, information regarding normal mechanisms of balance control during perturbed walking in frontal planes is lacking.

2.1. PURPOSE

The purpose of this study was to examine the corrective balance responses in both the frontal and sagittal plane during perturbed walking. This research was directed to:

1) providing baseline data for future studies on age-related effects on M-L stability during walking;
2) providing a better understanding of the corrective balance responses required during unexpected disturbances at different periods in the gait cycle.

2.2. OBJECTIVES

1. To compare task performance levels between perturbed (PT) walking and the non-perturbed (NPT) walking condition. The types of disturbance used included backward translation (BT) and forward translation (FT), presented at different points in the gait cycle, in particularly during single support phase, just after right heel off (PT-HO), and just after right heel strike (PT-HS). As the goal of this task was to maintain the rate of progression of center of mass (CM), the peak CM displacement relative to space CM(S) as a dependent variable was quantified.

2. To further examine which parameters of the stepping movement in the sagittal plane would be modulated during walking disturbances, and presented at different periods of the gait cycle. The following dependent variables, which represent different aspects of the corrective balance response, were analyzed:

   a) timing or onset latency, occurrence frequencies, and pattern of early corrective muscle response in bilateral gluteus medius (hip abductor, AB), adductor magnus (AD), rectus femoris (RF), hamstrings (HA), tibialis anterior (TA) and gastrocneumius (GS);

   b) timing, magnitude and phase changes in trajectories of angular displacements at the ankle, knee and hip, and rotation of trunk segment; and

   c) timing, magnitude and phase changes in trajectories of CM displacement relative to the foot (CM(F)).

3. To quantify the type of disturbance and stability requirements in the frontal plane that are associated with the sagittal-directed perturbation while walking. The following dependent variables were quantified:

a) CM(S) displacement in frontal plane

b) angular displacements at the hip, and trunk, shank segment for NPT, BT, and FT condition.

4. To identify which parameters of the stepping movement in the frontal plane would be modulated during the walking disturbances. The following dependent variables that represent different aspects of the corrective balance response were analyzed:

a) timing, magnitude and phase changes in trajectories of angular displacements at the hip, and rotation of trunk, shank segment;

b) timing, magnitude and phase changes in trajectories of CM(S) in the frontal plane; and

c) the trajectories of ground reaction force in medio-lateral direction (Fx)

CHAPTER III - METHODOLOGY

The Motor control and Neurological Dysfunction Laboratory in the School of Medical Rehabilitation provided the research environment to undertake this research task. The University of Manitoba, Faculty of Medicine, Ethics Committee approval was granted prior to enrolling the subjects. Ethical approval form is attached in Appendix A.

3.1. SUBJECTS

Participating in this study were ten (five female, five male) healthy and active adults between twenty to thirty-five years of age. The exclusion criteria for all subjects included any history of neurological deficit, musculoskeletal, or orthopedic diseases. The subjects were recruited from the students and staff of the University of Manitoba. The subjects did not report any history of neurological deficit, musculoskeletal, or orthopedic diseases would affect their balance and orientation. All of subjects reported being activity in at least one sport.

3.2. MOVABLE PLATFORM APPARATUS

The perturbed walking was performed using a movable platform in a 5 m long wooden walk way. The movable platform was constructed to provide forward and backward support surface translations. An AMTI biomechanical force plate was firmly mounted into the movable platform.

The platform motion of translation was controlled by electrical DC motors/linear actuator (model H105B, Industrial Devices Corporation, 35 Pamaron Way, Novato, CA 94949). A factory installed linear potentiometer was mounted inside the cylinder of motor. After calibration, the linear potentiometer signals were used to determine linear displacement of the platform translations. The displacement data was low-pass filtered at 20 Hz using a fourth order Butterworth type zero phase lag digital filter and then differentiated with respect to time to obtain velocity. An Amiga 2000 computer equipped with a digital to analog (D/A) converter was interfaced to the motor control units to control platform motion.

One output of the motor control unit generated a square wave pulse at initial of motor cylinder movement. This pulse activated two LED lights for 100 ms in view of three video cameras, and this signal was also collected on the analog to digital (A/D) converter for synchronization of video and analog signlas (EMG, force plate signals, motor linear potentiometer signals, and heel/toe pressure signals). An IBM compatible computer equipped with a 16 channel, 12 bit, analog to digital (A/D) converter (RC Electronics Inc., 6464 Hollister Ave, Goleta CA, USA) was used to collected all these signals.

An AMTI force plate was used. The outputs were the ground reaction forces in the antero-posterior (sagittal) plane (Fy), mediolateral (frontal) plane (Fx) and the horizontal plane (Fz). The force plate provided the three moments Mx, My, and Mz.

The following equipment and procedures were used to trigger onset of platform motion during two specific points in the gait cycle.

Perturbation 1: sudden platform translations just after right leg heel-off (PT-HO). An interlink electronics force-resistive pressure sensor was attached to the heel of the right foot. The amplifier of this pressure sensor generated a zero voltage signal when foot was on the ground, i.e. pressure sensor loaded. It generated a 2 Volt signal at the moment that no force was on the pressure sensor, i.e. at heel off. The output signal of amplifier was used to trigger the platform movement with a mechanical latency of 51 ms after heel off.

Perturbation 2: sudden platform translations just after right leg heel strike (PT-HS). The vertical ground reaction force signal (Fz) of force plate was used to trigger the platform motion. The platform was programmed to begin translation 51 ms after onset of heel strike as indicated by 100 N rise in Fz.

The platform motion for both forward and backward translations in this study was 30 cm/s, reached in 100 ms, followed by a plateau period of constant velocity for approximately 300 ms before deceleration. The amplitude of the platform forward and backward translation was 11 cm.

3.3. PROCEDURES

Every subject signed the consent forms and was told about the requirement of the study. Each subject was asked to wear running shoes, black shorts with two cutouts for placement of reflective markers at hip joints, and T-shirts. Then the subjects were weighed before the experiment.

The subjects were fitted with a parachute safety harness, which was attached overhead to ceiling-mounted support hooks.

The subjects were instructed to stand with erect posture, feet parallel, and arms bent at elbows. The bent arms during walking ensured that arm movement would not interfere with recording of markers at the anterior superior iliac spine of the pelvis (ASIS) and the greater trochanter of the hip.

The subjects were asked to look straightforward, walking normally, and leading with right leg. Care was taken to keep the standing foot position at the same place for every trial. A researcher stood beside the subjects at all times but did not assist them unless necessary.

There were a total of 72 walks per subject, which were arranged in 6 blocks of 12 trails. To minimize prediction of perturbation trails, forward and backward support surface translation occurred only 3 of 12 trials in each block, separately. In block 1, 2 and 5, the perturbation of the platform was timed at heel off. In block 1 and 5, the subject initiated walking while standing on the force plate; in block 2, the subject stood behind the force plate and took the first step onto the platform. In block 3, 4 and 6, the perturbation of the platform was timed at heel strike, in all these blocks, the subject started walking behind the force plate. These gave us the force signals for left stance while subject was on the force plate, and for the right stance while subject was behind the force plate.

3.4. DATA RECORDING AND CALCULATION

3.4.1. Video Based Motion Analysis

The subjects were videotaped using three synchronized video cameras, positioned on both sides of sagittal plane and frontal plane. One camera (model Sony CCD-V801), which was connected to a video cassette recorder (VCR) (model Sony SLV-R5UC), was placed on the left side with a sagittal view. One camera (model Panasonic AG-450), which was connected to a VCR (model Panasonic AG-7350), was placed on the right side of sagittal view. The third camera (Sony CCD-TR66) was placed to provide a frontal view. The shutter speed chosen for these three cameras was 1/250 s, 1/250 s and 1/4000 s, respectively. All cameras were placed on a stationary box at a fixed distance from the platform. To highlight the reflective markers, three spotlights were focused on three sides of the subjects. Prior to each experiment, a calibration rod was filmed by each camera individually. The light reflective markers provided the reference points for the end points of the body segments and axes of rotations. Circular markers of 2.5cm diameter were attached to the skin or clothes over the following landmarks (Figure 1):

*the lateral aspect of the forehead--vertex head

*the zygomatic angle of the mandible--chin-neck angle, and bilateral of

*the middle of the mid-point of the lateral aspect of humeral head --shoulder

*the lateral condyle --elbow

*the anterior superior iliac spine (ASIS)--pelvic crest/L4

*the antero-superior margin of greater trochanter --hip

*the lateral epicondyle of the femur --knee

*the lateral malleolus --ankle

*the posterior aspect of the calcaneous --heel

*the lateral aspect of the fifth metatarsal head --toe

The markers placed on the acromion were viewed in the frontal plane camera for the transverse shoulder joint center. A reflective marker was used as fixed earth reference. This marker was placed on the fixed rod which attached to the ceiling and was behind the subject. This provided a common coordinate system for right and left side of cameras. The peak 2D video motion analysis system (Peak Performance Technologies Inc. 7385 S. Revere Parkway, Suite 601, Englewood, Colorado) was used to digitize the x and y coordinates of the centroid of each marker. The sampling rate of the video system was 60 Hz (60 images per second). For each trial, one hundred and forty video images were digitized, fifteen before the onset of platform movement, as determined by the sync light, and one hundred and twenty-five after the onset. The coordinate data was then low-pass filtered, 5 Hz, using a fourth order Butterworth zero phase lag digital filter. The x and y coordinates of the common earth fixed reference marker were subtracted from coordinate data of each body marker to obtain a common coordinate system for right and left saggital plane video data.

A custom software package was used to calculate time-series data from the raw coordinate data in right and left sagittal plane, and frontal planes. These included the following data:

a) linear displacement (meter) of lateral malleolus marker x- and y- coordinate;

b) joint angular displacement (degree) of the ankle, knee and hip joint;

c) angular displacement (degree) of shank and trunk segment;

d) CM(S) displacement and CM(F) displacement which was CM displacement relative to the foot (using lateral malleous marker).

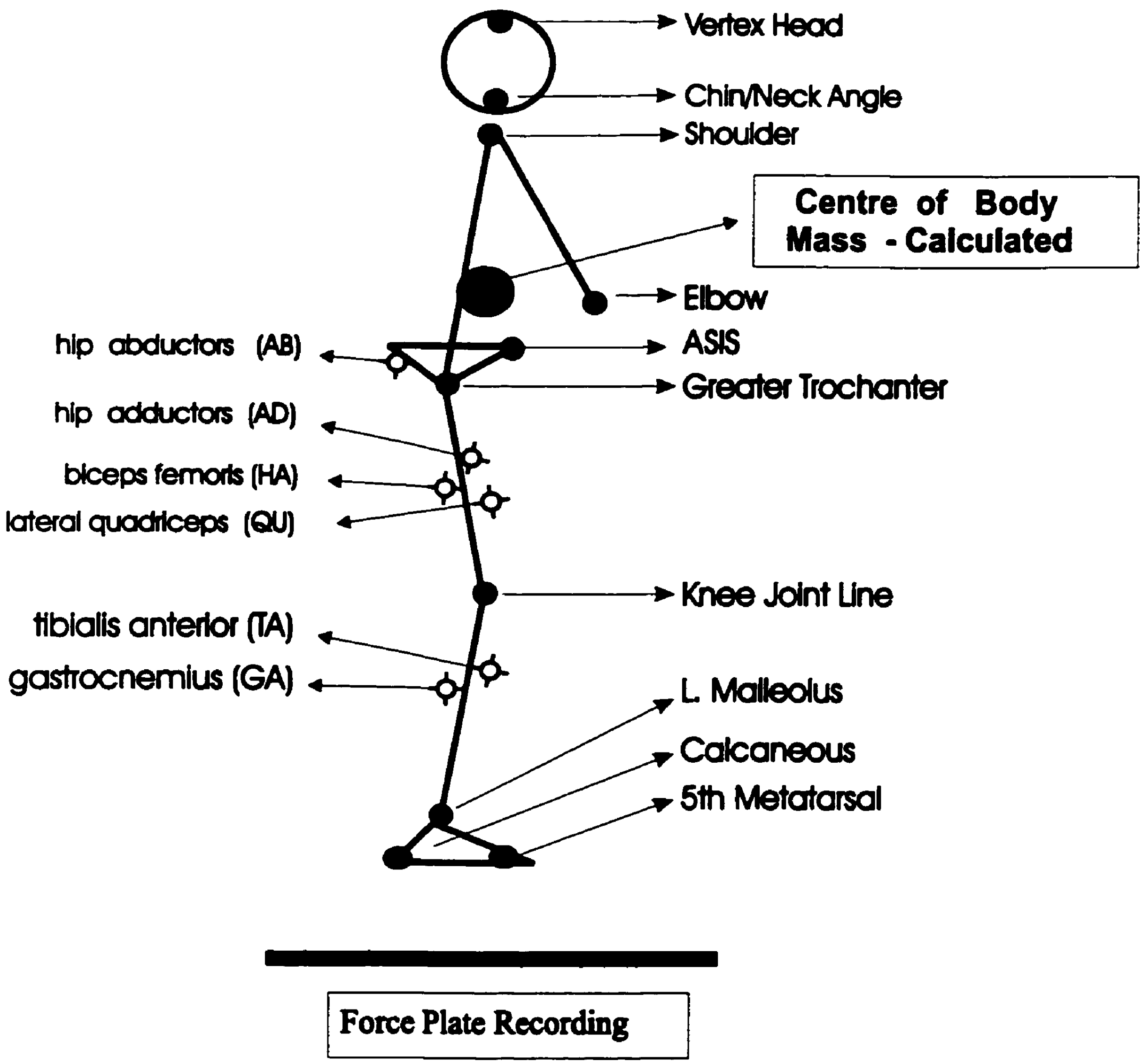

Figure 1. Illustrates the general experimental set up along with EMG electrode and marker placements.

The definition of each angle is as follows:

a) Ankle angle: the intersection of the line from the fifth metatarsal marker to lateral malleolus marker and the line from the lateral malleolus to the knee marker

b) Knee angle: line from lateral malleolus-knee and the line from knee-greater trochanter

c) Hip angle: line from knee-greater trochanter and the line from greater trochanter-ASIS

d) trunk segment: line from ASIS to shoulder

e) shank segment: line from lateral malleolus to knee

The method of CM(S) calculation followed Winter's (1980). The body center of mass displacement was calculated based on:

a) the end-point coordinate data;

b) the anthropometric data (Chandler 1975); and

c) the subject's mass.

The value of CM(F) was obtained from the coordinates of CM(S) and lateral melleolus marker. Same method was used in sagittal right, left plane and frontal plane.

3.4.2 Electromyographic (EMG)Data

EMG surface electrodes were placed over the muscle bellies of eight muscle groups in the first four blocks, then switched to four muscle groups in the last two blocks. The muscle groups included gluteus medius (hip abductor, AB), adductor magnus (AD), rectus femoris (RF) and hamstrings (HA) bilaterally. In block 5 and 6, the electrodes for HA and AD were switched to same body side of tibialis anterior (TA) and gastrocnemius(GS). Care was taken to ensure that the skin surface was clean and the electrodes were placed bilaterally. The placement of the electrodes was followed the locations described by Winter (1991). The electrodes were connected to lightweight differential pre-amplifiers (30 grams) by cables. The electrodes and pre-amplifiers were taped to the skin to provide consistent placement throughout the experiments. The pre-amplifiers were connected to the main amplifiers by cables that were attached at the back of subject, assuring that the cables did not interfere with the subjects' walking.

Differential EMG signals were amplified and processed by using two 4-channel, EMG amplifiers. (Biosys Inc.) The pre-amplifiers amplified the raw EMG signal 100 times. Then, the EMG was band-pass filtered (10 Hz to 1KHz), rectified and low-pass filtered at 50 Hz. All EMG filters were analogue RC type filters (-3 dB).

All signals (EMG, force plate signals, linear potentiometer and motor sync pulse) were recorded on an IBM compatible computer equipped with a 16 channel, 12 bit, analog to digital (A/D) converter (RC Electronics Inc., 6464 Hollister Ave, Goleta CA, USA). For each platform movement, a 4 second sweep of data was collected at a sampling rate of 333 Hz, which included a 700 ms period before onset of platform motion, and 3300 ms after the onset. The data were stored on computer for analysis.

3.4.3. Force Plate

The force plate simultaneously measured the ground reaction forces in antero-posterio (sagittal) plane (Fy), mediolateral (frontal) plane (Fx), and horizontal plane (Fz). It also measured the three moments Mx, My, and Mz. A custom software package was used to calculate the center of foot pressure (CFP).

3.5. DATA ANALYSIS

3.5.1. GENERAL GAIT CYCLE PARAMETERS

Swing/stance duration and swing distance were obtained from x and y coordinates of heel and toe markers. We examined both right and left side of body and both in PT- HO and PT- HS conditions.

a) PT- HO condition: right swing and stance, and then left stance and swing

b) PT- HS condition: right stance and swing, and then left swing and stance

3.5.2. EMG

The methods of Eng et al. (1994) and Tang et al. (1998) were employed in this study to determine the onset latency of corrective muscle responses. Each subject ensemble averaged EMG activity during the walking trials with the no perturbation (NPT) condition was calculated for each muscle, with the HO and the HS conditions separately. EMG records of individual perturbed trial were subtracted from the ensemble averaged of unperturbed trials to produce a difference waveform. From the difference waveform, the following variables were quantified:

a) Onset latency: The time interval between the start of the platform motion and the beginning of the first detectable change in difference waveform exceeding the mean pre-movement difference waveform level (first 700 ms of the signal) by at least 2 standard deviations (SD) was taken as the onset latency of muscle activation. The expected onset was within a time period of 200-250 ms from the initiation of platform motion. The onset of EMG signals was quantified for each of the twelve muscle groups relative to the start of the platform movement.

b) Sign of response: There were two kinds of muscle response sign, excitatory and inhibitory response relative to unperturbed walking. An excitatory response was an increase in muscle activity that exceeds 2 SD above the pre-movement muscle activity level. An inhibitory response was seen when muscle responses fall below the mean pre-movement muscle activity.

a) Occurrence frequencies: The frequencies of presence (excitatory or inhibitory response) or absence of EMG corrective response were analyzed.

3.5.3. KINEMATICS

1) CM(S) displacement and velocity: The task performance level was based on the position and velocity of CM(S) in the sagittal plane. The peak CM(S) displacement at the end of first half cycle for PT- HO and PT- HS condition was determined for each trial (NPT, FT and BT).

The performance levels during the BT and FT will be compared to the NPT condition.

2) CM(F) displacement: The timing, magnitude, and duration of the different phases in trajectories of CM(F) displacement during NPT, BT and FT in PT- HO and PT- HS conditions was determined. This was done in the sagittal plane.

a) The timing of the initial divergence in trajectories of CM(F) displacement during FT and BT as compared to NPT condition was determined.

b) The duration and magnitude of the different phases of CM(F) displacement during the swing and stance periods of gait was quantified.

3) Angular displacement: The timing, magnitude, and duration of the different phases in trajectories of angular displacements at the ankle, knee and hip joint, and trunk segment in the sagittal plane, and at the hip joint and rotation of trunk, shank segment in the frontal plane during NPT, BT and FT in PT- HO and PT- HS conditions was determined.

a) The timing of corrective responses was determined. This was done by looking at the time of divergence in trajectories of angular displacement and trunk, shank segment rotations during FT and BT as compared to NPT condition.

b) The duration and magnitude of the different phases of angular displacement and trunk segment rotation during the swing and stance periods of gait was quantified. For example, during swing phase, the hip angular displacement has two phases, flexion and then extension. Similarly at the ankle, dorsiflexion and then plantarfelxion were observed.

4) Fx (M-L) magnitude: The Fx (M-L) magnitude in frontal plane (medio-lateral) was analyzed. The timing of the initial divergence in medio-lateral (M-L) force during FT and

BT as compared to NPT condition was determined. The magnitude of the divergence was also quantified.

3.6. STATISTICAL ANALYSIS

A paired T-test was performed to examine the magnitude difference of each time point in the waveforms of NPT, BT, and FT. From this analysis, a divergence in the trajectories of the kinematic and kinetic waveforms between BT, FT, and NPT were compared statistically, significance was determined to be at a level of p less than 0.05. Eight trials per subjects for ten subjects were used in this analysis.

There were three different levels of trial conditions; FT, BT and NPT in this study. A one-way repeated ANOVA was performed to examine the difference of "trials effect" (NPT, BT, and FT) on the following variables:

1) The general gait cycle parameters, swing\stance duration and swing distance in the sagittal plane were compared.

2) Performance level was based on the data analysis of CM(S) displacement in the sagittal planes. The peak CM displacement relative to space CM(S) was compare.

3) In order to further examine which parameters of the stepping movement in the sagittal plane would be modulated during walking disturbances, the following dependent variables, which represented different aspects of the corrective balance response, were compared:

a) magnitude and duration of the difference phases in trajectories of angular displacements at the ankle, knee and hip, and rotation of trunk segment; and

b) magnitude and duration of the different phases in trajectories of CM displacement relative to the foot (CM(F)).

4) To identify which parameters of the stepping movement in the frontal plane would be modulated during the walking disturbances, the following dependent variables were quantified by statistical analysis:

a) magnitude and duration of the difference phases in trajectories of angular displacements at the hip, and rotation of trunk, shank segment; and

b) magnitude of divergence in trajectories of force in medio-lateral direction (Fx).

To compare which parameters of movement were modulated for disturbances presented at two different periods of the gait cycle, HO and HS, we separately examined the HO and HS conditions.

CHAPTER IV-RESULTS

4.1. BALANCE DISTURBANCE

Passive mechanical effects on the body due to sudden platform translations would occur before onset of the earliest EMG response. This would be within 100 ms of the onset of platform motion. These passive effects were observed in the following waveforms; a) linear displacement of lateral malleolus marker x-coordinate, b) CM(F) displacement, and c) angular displacement at ankle, knee, hip, trunk and shank segment rotation.

4.1.1. Linear Displacement of Lateral Malleolus Maker x-coordinate

The bottom two plots in Figure 2 and 3 present ensemble group averages (ten subjects, eight trials per subject) of linear displacement of right and left lateral malleolus markers x-coordinate for NPT, BT, and FT in PT-HO and PT-HS conditions. As seen from Figure 3, an early divergence in the trajectories of linear displacement of lateral malleolus marker x-coordinate between NPT, FT and BT was observed in stance legs, right leg in PT-HS and left leg in PT-HO condition. Table 1 presents times at which divergence in waveforms of NPT, BT, and FT were statistically significant different. As presented in Table 1, a significant divergence in the waveforms started at about 20 ms after the onset of platform motion. Divergence in the waveforms for BT and FT relative to NPT were opposite in direction. The maximum difference in magnitude between BT

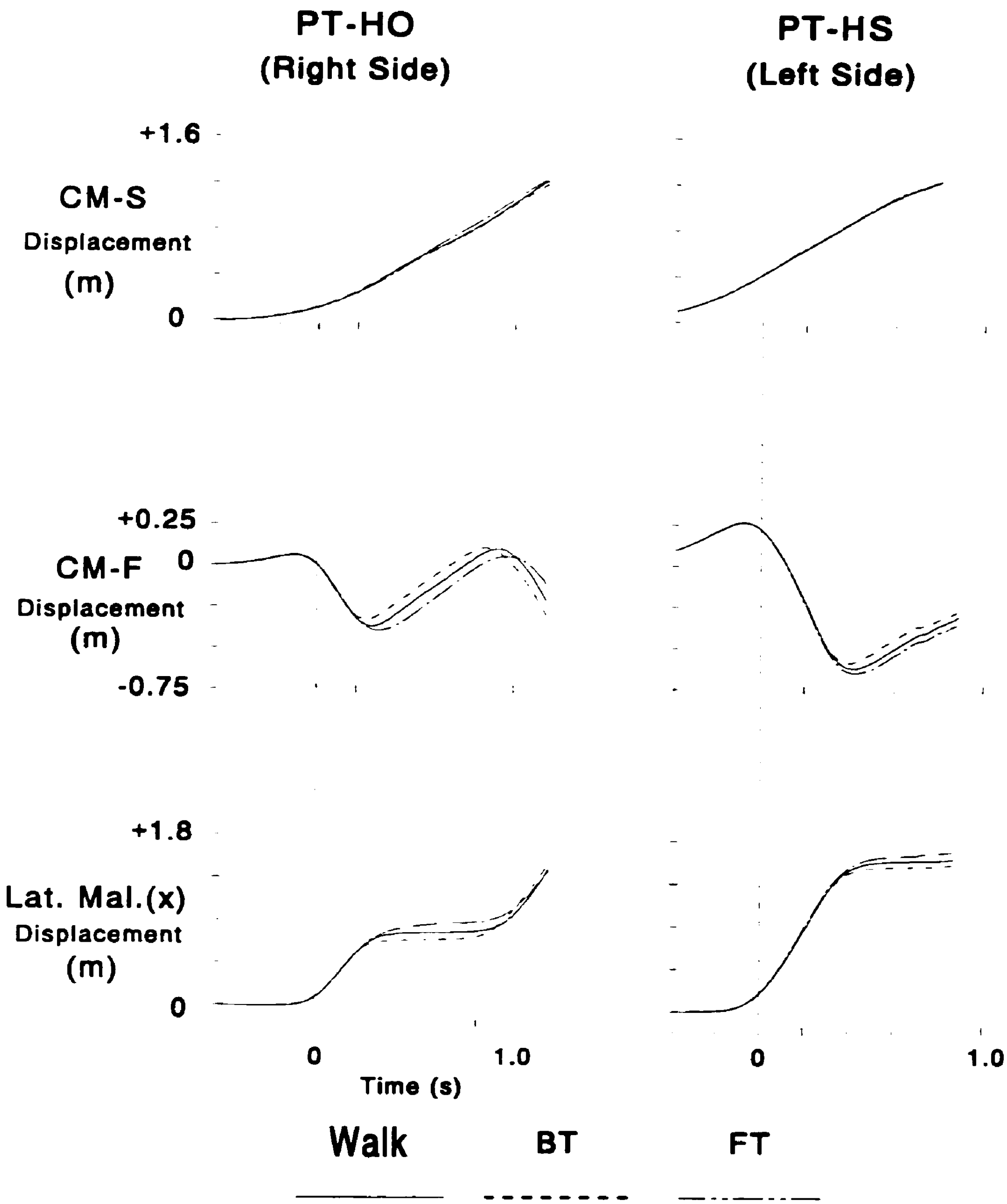

Figure 2. Group average plots of CM-S and CM-F displacement, and linear displacement of the lateral malleolus marker for NPT, BT, and FT conditions. The side of body corresponding to the swing limb is presented, i.e. sagital right plane during right heel-off trials (PT-HO) and sagital left plane during right heel strike trials (PT-HS). All plots are of x-coordinates. For y-axis, zero represents standing still baseline position, before right heel off and onset of platform translation. Vertical dashed line at time zero is onset of platform translation. For CM-F, positive values represent CM behind the foot and negative values are CM ahead of foot.

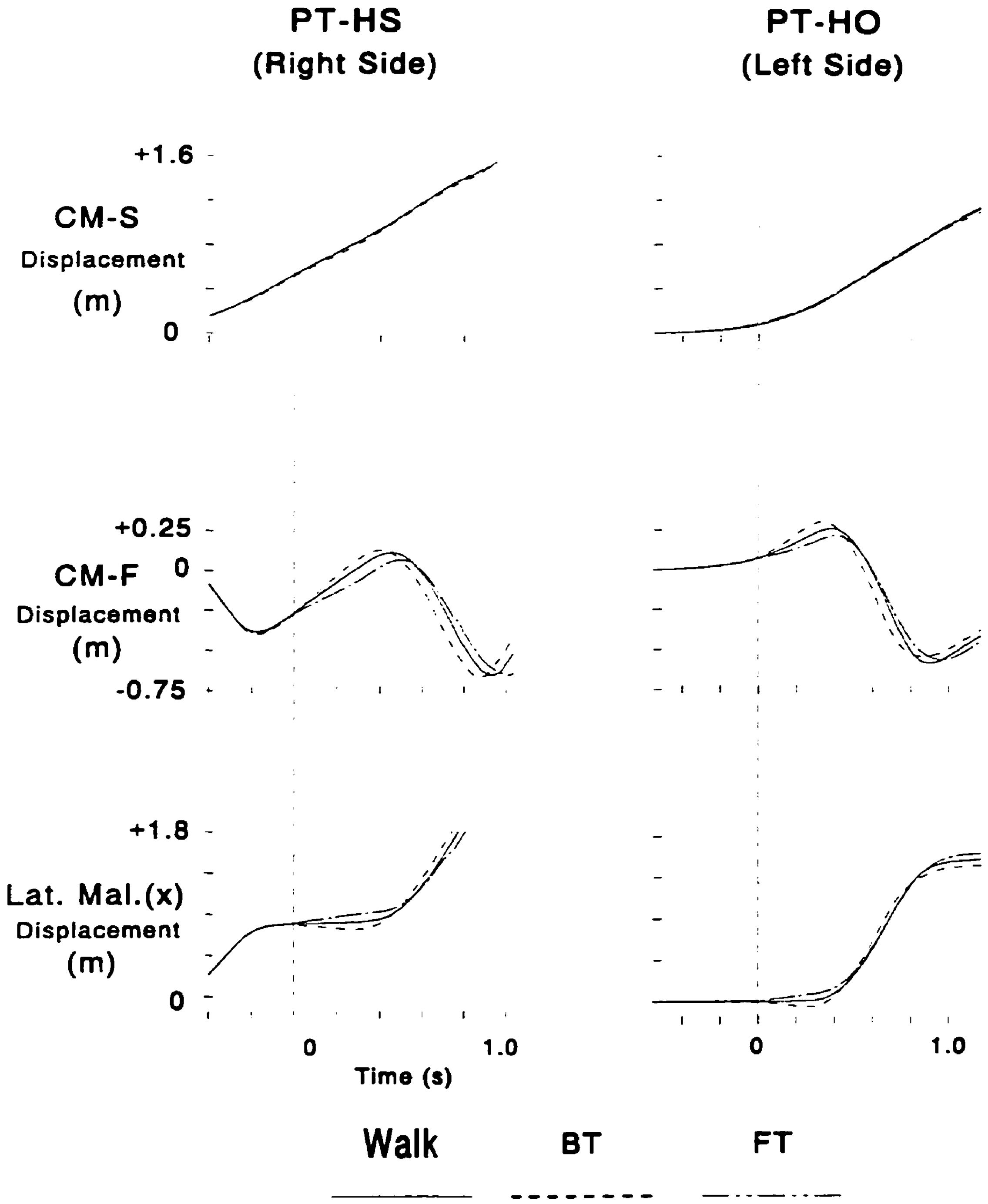

Figure 3. Group average plots of CM-S and CM-F displacement, and linear displacement of the lateral malleolus marker for NPT, BT, and FT conditions. The side of body corresponding to the stance limb is presented, i.e. sagital left plane during right heel-off trials (PT-HO) and sagital right plane during right heel strike trials (PT-HS). All plots are of x-coordinates. For y-axis, zero represents standing still baseline position, before right heel off and onset of platform translation. Vertical dashed line at time zero is onset of platform translation. For CM-F, positive values represent CM behind the foot and negative values are CM ahead of foot

Table 1 Time period (sec) within which the p-value of group results of paired t-test of each time point in kinematic and kinetic waveforms for BT/ FT compared to NPT condition was less than 0.05. In the table, (p) represents passive component, (a) represents active component

		PT-HO				PT-HS			
		BT(p)	BT(a)	FT(p)	FT(a)	BT(p)	BT(a)	FT(p)	FT(a)
Sagital	trunk segment angle		0.19		0.24		0.29		0.20
	R-hip angle		0.34		0.60	0.00	0.44	0.12	0.49
	R-knee angle		0.20		0.32		0.14		0.42
	R-ankle angle		0.84		0.90	0.00	0.25	0.12	0.35
	R-linear displ. Lat. Malleolus		0.25		0.32	0.02	0.54	0.04	0.72
	L-hip angle	0.10	0.35	0.14	0.40		0.24		0.27
	L-knee angle		0.27		0.30		0.36		
	L-ankle angle		0.27	0.07	0.29				0.00
	L-linear displ. Lat. Malleolus	0.02	0.89	0.02	0.97		0.40		0.42
Frontal	R-hip angle		0.27		0.25		0.35		0.39
	R-trunk segment angle		0.40		0.19		0.10		0.35
	R-shank segment angle		0.19		0.20		0.40		0.14
	L-hip angle		0.29		0.20		0.27		0.25
	L-trunk segment angle		0.44		0.10	0.07	0.57		0.00
	L-shank segment angle		0.35		0.67		0.14		0.12
Sagital	CM-F right side		0.22		0.27		0.17	0.04	0.67
	CM-F left side	0.04	0.49	0.05	0.99		0.39		0.34
Frontal	CM-S		0.30		0.42		0.65		0.60
	CM-F		0.24		0.25		0.30		0.19
Force	Fx		0.31		0.43		0.38		0.58
	Fz		0.33				0.32		0.41

and FT, which occurred in the first 200 ms after the platform motion, was 17 cm in right side during PT-HS and 11 cm in left side during PT-HO condition.

No early significant divergence in the trajectories of linear displacement of lateral malleolus marker x-coordinate within first 100-150 ms of onset of platform motion was observed in swing legs i.e. the right leg in PT-HO condition and left leg in PT-HS condition.

4.1.2. CM(F) Displacement

Figure 2 presents the group average (ten subjects, eight trials per subject) plots of CM(F) displacement for NPT, BT, and FT for the side of the body in swing phase during platform translations; sagittal right plane in HO condition and sagittal left plane in HS condition. Figure 3 presents those for stance side; sagittal right side in HS and sagittal left side in HO condition. As evident in Figure 2, there was no early divergence observed in trajectories of CM(F) displacement between NPT, BT and FT when the leg was in swing at time of disturbance. A passive component was observed only in stance side of body. As evident in Figure 3 and presented in Table 1, a significant divergence in the trajectories of CM(F) displacement for NPT, BT, and FT began at about 40 ms of onset of platform motion. Divergence in CM(F) displacement for BT and FT relative to NPT was opposite in direction. In both PT-HO & PT-HS for CM(F) BT actually displace the foot backwards relative to CM and vice versa for FT.

4.1.3. Angular Displacement

Figure 4 and 5 present the ensemble group averages (ten subjects, eight trials per subject) of angular displacements at the ankle, knee, hip, and trunk segment rotation for NPT, BT, and FT in sagittal plane. Figure 6 and 7 present ensemble group averages (ten subjects, eight trials per subject) of hip angular displacement and trunk and shank segment rotations for NPT, BT, and FT in frontal planes. In a similar fashion to linear displacement and CM(F) displacement described above, an early and significant divergence in trajectories of angular displacements between NPT, BT and FT was observed in the stance. For the sagittal plane, as evident in Figure 5, the passive components due to platform motion were observed in ankle and hip angular displacement for right side during PT-HS, and left side during PT-HO. No such early divergence was observed at the knee joint and trunk segments in the sagittal plane. In addition, no passive components were observed in the frontal plane for hip angular displacements or trunk and shank segment rotations.

Table 1 presents time values at which divergence in angular displacement waveforms were statistically significant. The early divergence in ankle angular displacement between NPT, BT and FT occurred within 50 ms from onset of platform motion. This was the case for the left ankle in PT-HO and right ankle in PT-HS. A decreased dorsiflexion in BT, and an increased dorsiflexion in FT, relative to NPT condition was observed. The early divergence in hip angular displacement between NPT, BT and FT for PT-HO and PT-HS occurred within 100 ms after platform motion. There was an increase of hip extension in BT, a decrease of hip extension in FT, relative to NPT condition.

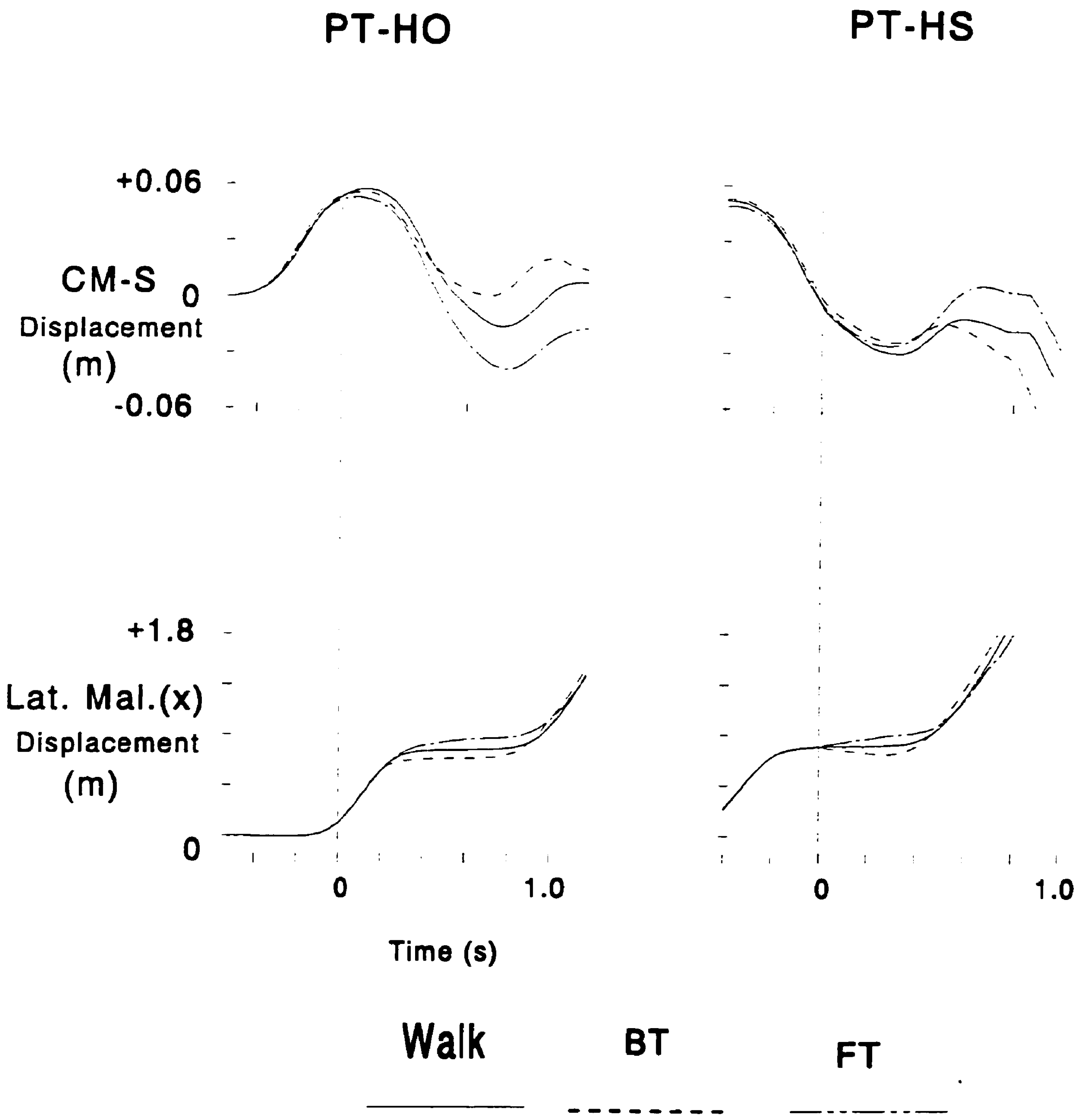

Figure 4. Group average plots of CM-S horizontal displacement in frontal plane along with linear displacement of the lateral malleolus marker in sagital plane for NPT, BT, and FT conditions during right heel-off trials PT-HO and right heel strike trials (PT-HS.. For y-axis, zero represents standing still baseline position, before right heel off and onset of platform translation. Vertical dashed line at time zero is onset of platform translation.

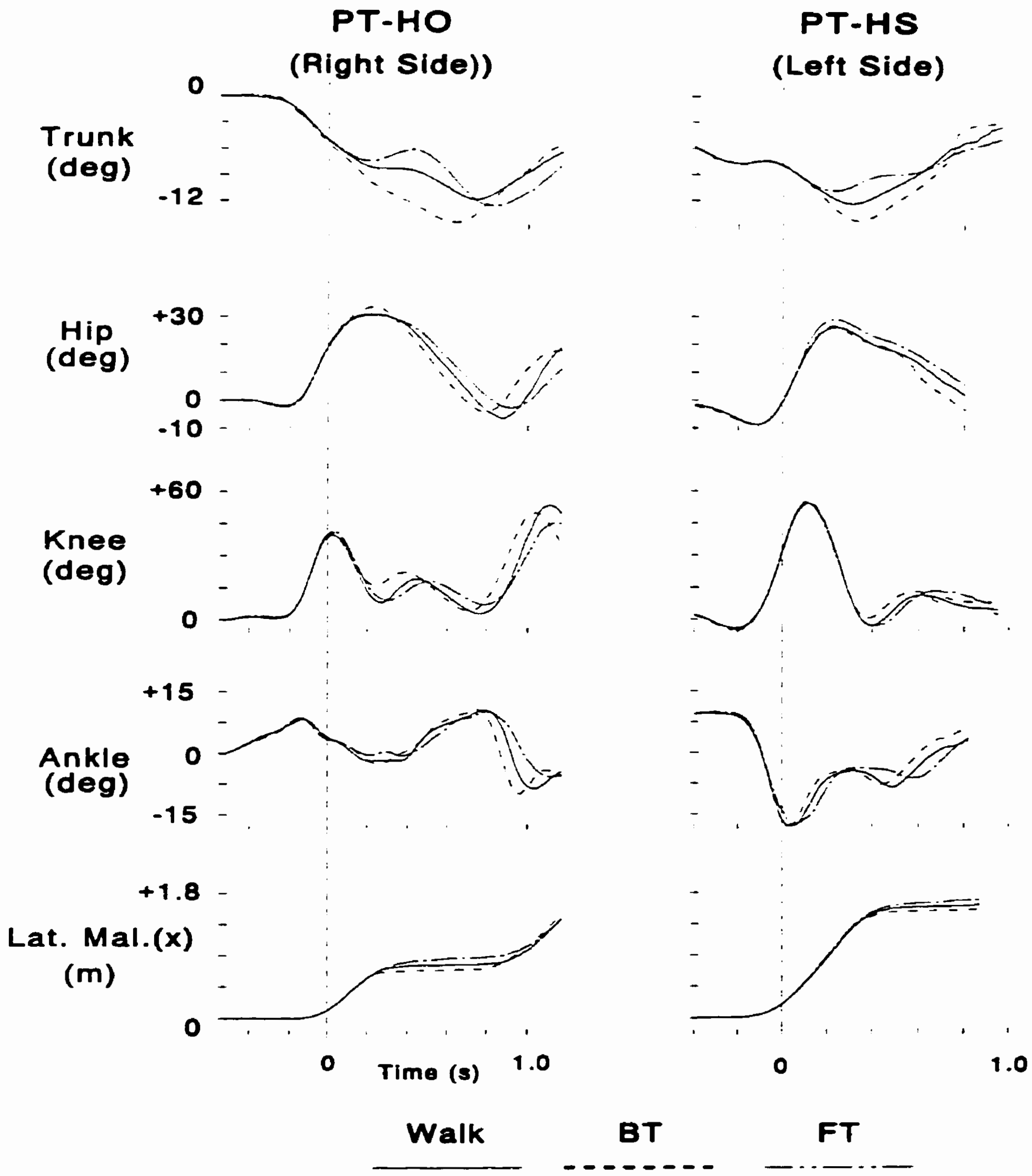

Figure 5. Group average plots of linear displacement of lateral malleolus marker (x-coordinate, meters), angular displacement at the ankle, knee and hip(degrees), and trunk segment rotation (degrees) for NPT, BT, and FT conditions. The side of body corresponding to the swing limb is presented, i.e. sagital right plane during right heel-off trials (PT-HO) and sagital left plane during right heel strike trials (PT-HS). For y-axis, zero represents standing still baseline position, before right heel off and onset of platform translation. Vertical dashed line at time zero is the onset of platform translation. For angular displacement, 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.

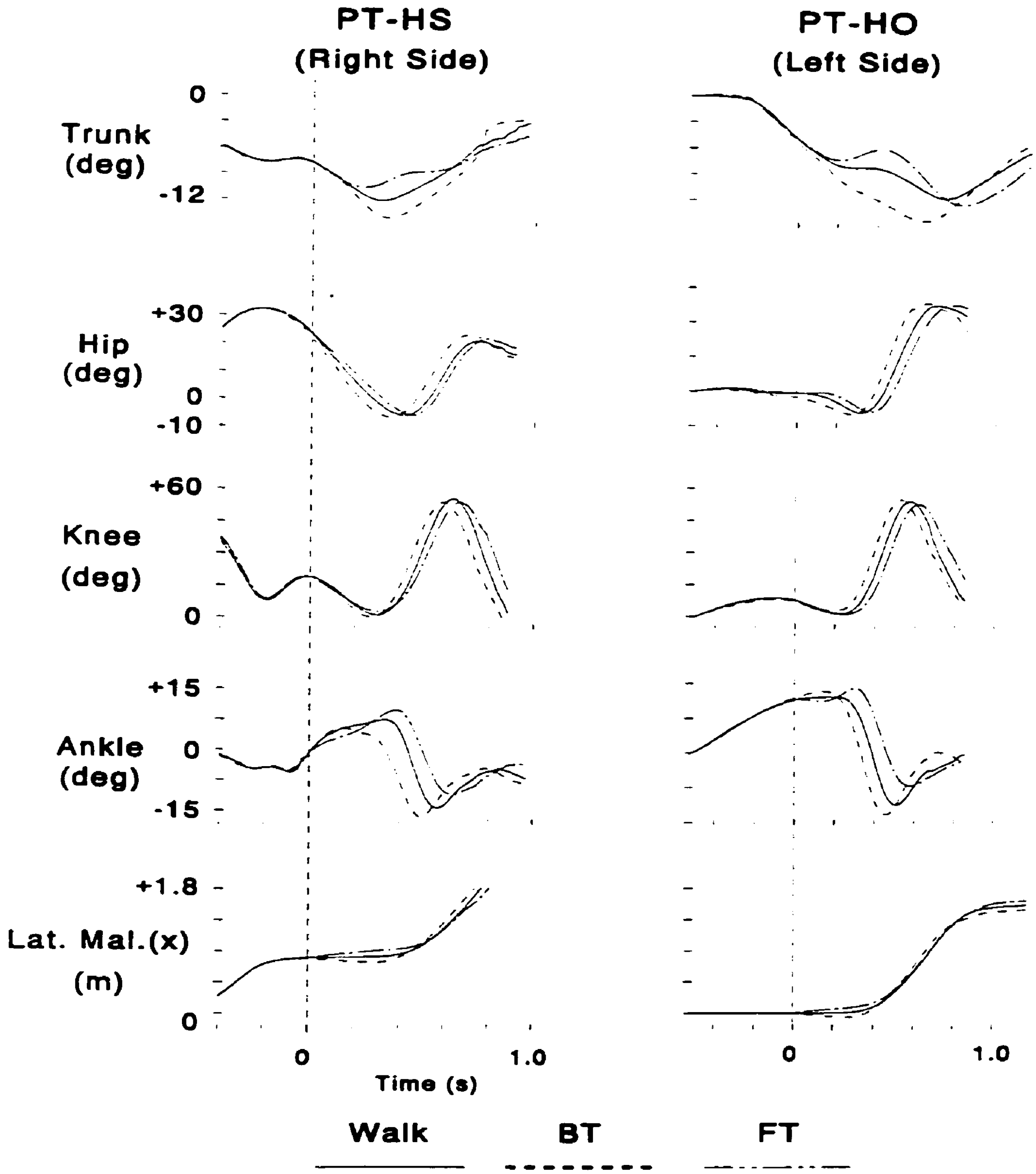

Figure 6. Group average plots of linear displacement of lateral malleolus marker (x-coordinate, meters), angular displacement at the ankle, knee and hip(degrees), and trunk segment rotation (degrees) for NPT, BT, and FT conditions. The side of body corresponding to the stance phase, i.e. sagital left plane during right heel-off trials (PT-HO) and sagital right plane during right heel strike trials (PT-HS) For y-axis, zero represents standing still baseline position, before right heel off and onset of platform translation. Vertical dashed line at time zero is the onset of platform translation. For angular displacement, 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.

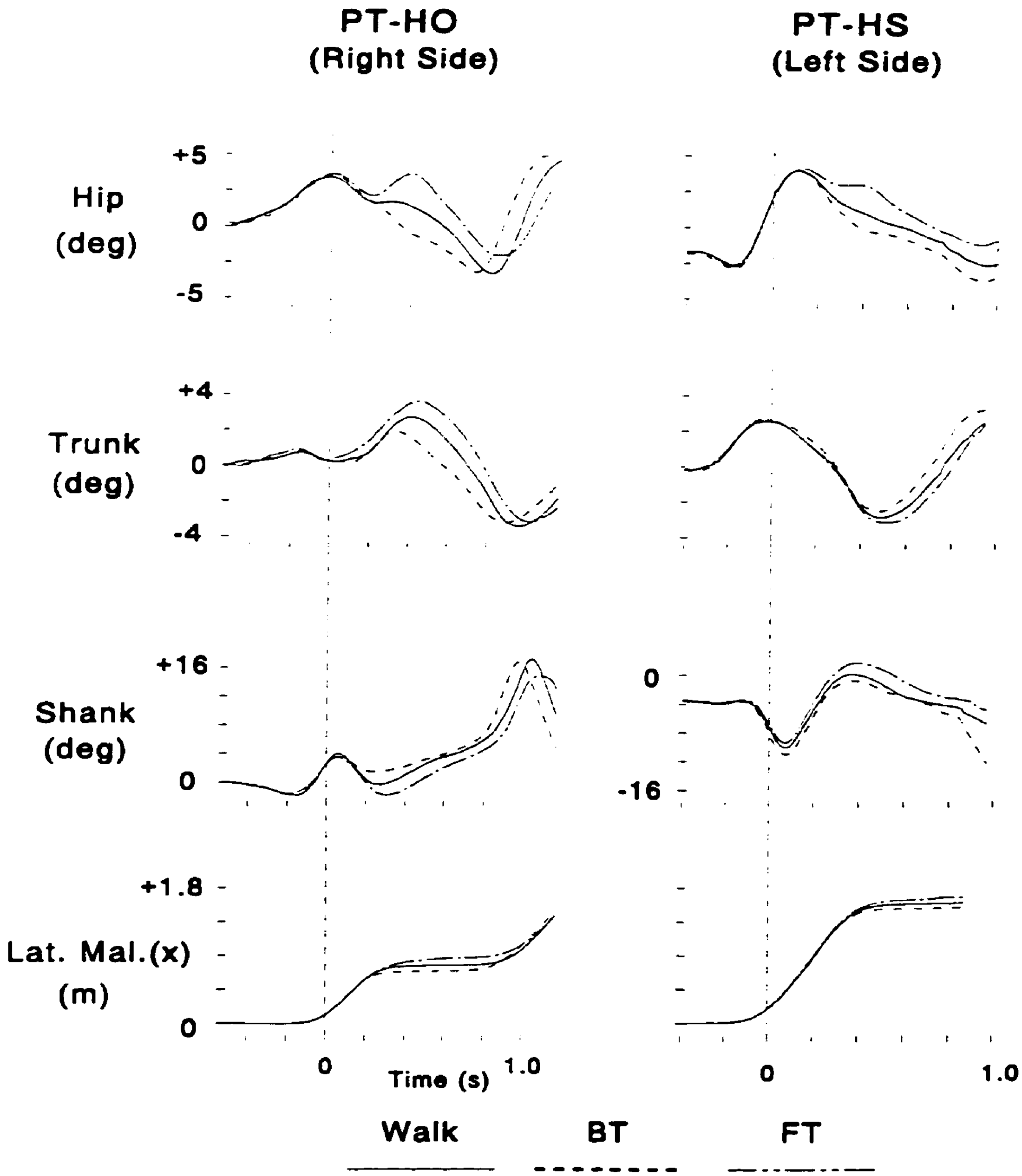

Figure 7. Group average plots of linear displacement of lateral malleolus marker (x-coordinate, meters) in sagital plane, hip angular displacement, and trunk and shank segment rotations in frontal planes for NPT, BT, and FT conditions. The side of body corresponding to the swing limb is presented, i.e. sagital right plane during right heel-off trials (PT-HO) and sagital left plane during right heel strike trials (PT-HS). For y-axis, zero represents standing still baseline position, before right heel off and onset of platform translation. Vertical dashed line at time zero is the onset of platform translation. For right hip angular displacement, positive values are adduction and negative values are abduction. The direction is opposite in the left hip. For right side of trunk segment, positive values represent lateral rotations and negative values are medial rotations. The direction is opposite in the left side of trunk segment. For shank segment, positive values represent counter-clockwise rotations and negative values are clockwise rotations.

4.2. CORRECTIVE BALANCE RESPONSES

4.2.1. General Gait Cycle Parameters—Swing/Stance Duration and Step Length

Table 2 presents the group means and SD of swing/stance duration and step lengths, and the results of one-way repeated ANOVA, and further paired t-test. The table includes NPT, BT, and FT condition for PT-HO and PT-HS. The plots of linear displacement of lateral malleolus marker x-coordinate in Figures 2 and 3 provide a visual presentation of changes in swing/stance duration and step length between BT, FT, and NPT conditions. As shown in Figure 1, for swing leg, PT-HO and PT-HS had similar effects on duration of the initial swing phase and step length. For both PT-HO and PT-HS, swing duration in the first half cycle was significantly decreased for BT and significantly increased for FT as compared to NPT ($p<0.001$). Step length in the first half cycle was also significantly decreased for BT and increased for FT as compared to NPT ($P<0.001$). As evident in Figure 3, similar trail effects on duration of the stance phase were observed during PT-HO and PT-HS. The duration of stance phase was significantly decreased for BT and significantly increased for FT, relative to NPT condition ($p<0.001$).

For PT-HO there was no trials effect on the duration or step length of the second half cycle; right stance and left swing phases. Similarly, there was no significant trial effect on right swing step length or left stance duration during PT-HS. However, there was a trail effect on the second half cycle right swing duration ($p<0.03$) during PT-HS. Here swing duration for BT and FT was less than NPT. There was no difference in right swing step length between BT, FT, and NPT. During the second half cycle in left side in PT-HO, there were no trial effects on left swing duration and step length.

Table 2 Group means (SD) of swing/stance duration (seconds) and step length (meters), and results (p-values) of One-way Repeated ANOVA, and further Paired t-test. The table includes NPT, BT, and FT condition for PT-HO and PT-HS.

					ANOVA	Paired t-test		
PT-HO		NPT	BT	FT	p-value	NPT Vs BT	NPT Vs FT	BT Vs FT
R Swing	Duration	0.53 (0.04)	0.50 (0.04)	0.58 (0.05)	0.001	0.002	0.002	0.001
R Swing	Step Length	0.77 (0.10)	0.64 (0.11)	0.81 (0.08)	0.001	0.006	0.002	0.04
R Stance	Duration	0.43 (0.04)	0.41 (0.07)	0.44 (0.05)	NS			
L Stance	Duration	0.47 (0.05)	0.41 (0.08)	0.53 (0.06)	0.001	0.002	0.001	0.001
L Swing	Duration	0.59 (0.05)	0.58 (0.03)	0.59 (0.07)	NS			
L Swing	Step Length	1.50 (0.12)	1.46 (0.17)	1.49 (0.12)	NS			
PT-HS								
R Swing	Duration	0.54 (0.03)	0.57 (0.04)	0.55 (0.03)	NS			
R Swing	Step Length	0.79 (0.07)	0.77 (0.08)	0.79 (0.08)	NS			
R Stance	Duration	0.41 (0.04)	0.36 (0.09)	0.47 (0.05)	0.001	0.028	0.001	0.001
2nd R Swing	Duration	0.66 (0.05)	0.64 (0.06)	0.61 (0.06)	0.025	0.375	0.003	0.087
2nd R Swing	Step Length	1.47 (0.08)	1.37 (0.31)	1.38 (0.13)	NS			
L Stance	Duration	0.47 (0.03)	0.47 (0.03)	0.46 (0.01)	NS			
L Swing	Duration	0.60 (0.05)	0.56 (0.03)	0.58 (0.03)	0.003	0.018	0.052	0.012
L Swing	Step Length	1.55 (0.08)	1.49 (0.09)	1.59 (0.11)	0.001	0.001	0.181	0.003

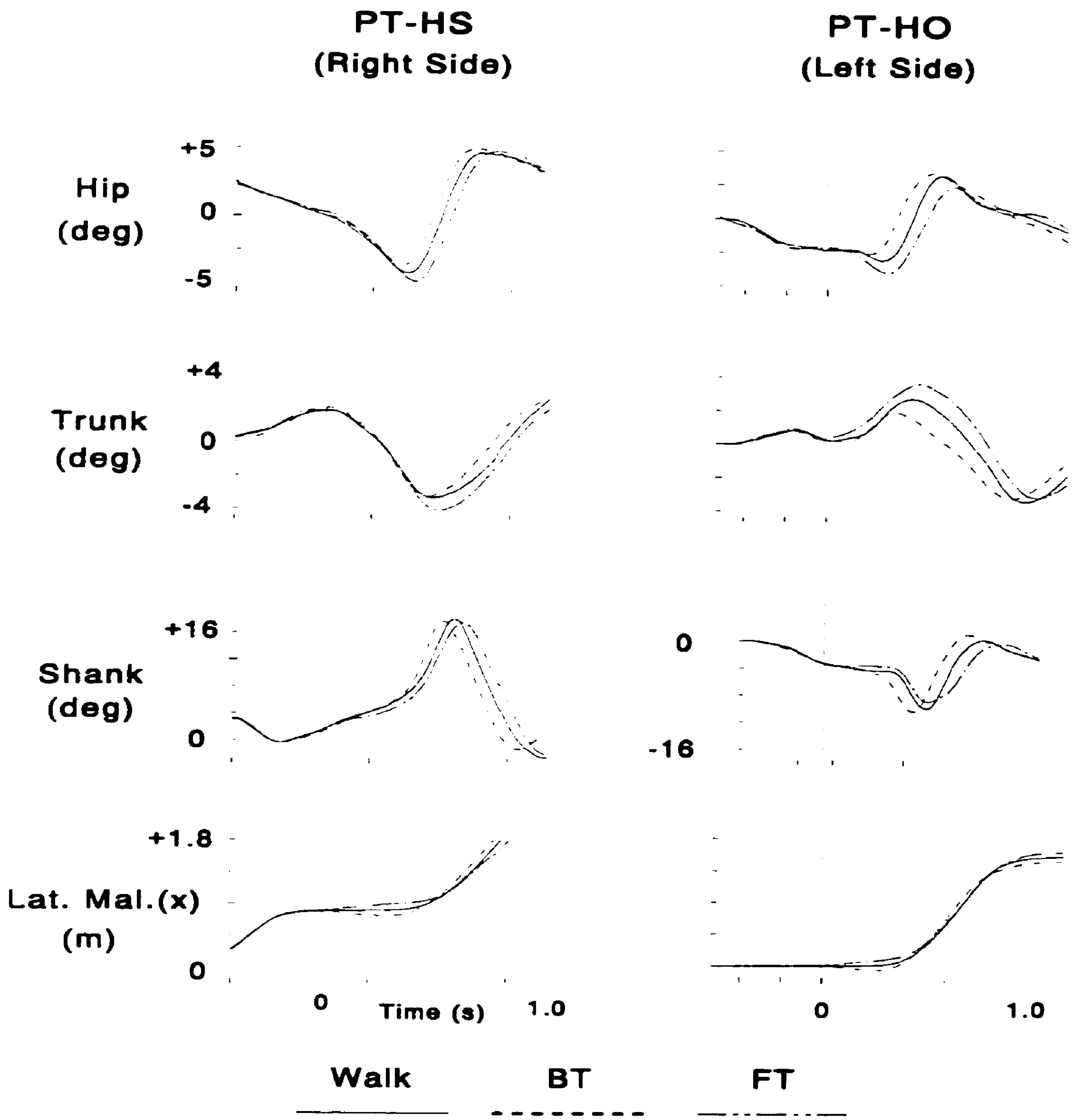

Figure 8. Group average plots of linear displacement of lateral malleolus marker (x-coordinate, meters) in sagital plane, hip angular displacement, and trunk and shank segment rotations in frontal planes for NPT, BT, and FT conditions. The side of body corresponding to the stance limb is presented, i.e. sagital left plane during right heel-off trials (PT-HO) and sagital right plane during right heel strike trials (PT-HS). For y-axis, zero represents standing still baseline position, before right heel off and onset of platform translation. Vertical dashed line at time zero is the onset of platform translation. For right hip angular displacement, positive values are adduction and negative values are abduction. The direction is opposite in the left hip. For right side of trunk segment, positive values represent lateral rotations and negative values are medial rotations. The direction is opposite in the left side of trunk segment. For shank segment, positive values represent counter-clockwise rotations and negative values are clockwise rotations.

4.2.2. CM(S) Displacement

Figure 2 presents the group average (ten subjects, eight trials per subject) plots of CM(S) displacement for swing side of the body; sagittal right during PT-HO and sagittal left side during PT-HS. Figure 3 presents those for stance side of body; sagittal right during PT-HS and sagittal left during PT-HO condition. Statistical analysis revealed no significant difference in trajectories of CM(S) displacement between NPT, FT, and BT during PT-HO or PT-HS in the sagittal plane.

Figure 8 presents the group average (ten subjects, eight trials per subject) plots of CM(S) displacement for NPT, BT, and FT condition in frontal plane during PT-HO and PT-HS. Significant differences between NPT, BT and FT were observed in CM(S) displacement during both PT-HO and PT-HS. Table 1 presents the time period within which the p-value of the results of paired t-test of each time point in CM(S) displacement waveforms was less than 0.05. As evident in Table 1, the time of significant divergence in CM(S) displacement between BT, FT and NPT occurred 300-400 ms from onset of platform motion. For PT-HO, the divergence occurred at the time of early right stance phase when the CM is moving towards the right side of the body. The effect of BT was a significant decrease in duration and magnitude of the right shift in CM(S) displacement and consequently an earlier time to start of the leftward shift in CM that would occur during right mid-swing. The opposite effect was observed for FT as compared to NPT, a delay in time to peak CM(S) displacement to the right with an increased amplitude of right CM displacement, and thus a delay in leftward shift of CM(S) displacement.

The time of divergence in CM(S) displacement during PT-HS occurred at the end of right stance. The direction of divergence during PT-HS was opposite to the one during

PT-HO. At the end of right stance phase, CM(S) was moving towards the right side for BT, and towards left side for FT, relative to NPT. The time to peak CM(S) displacement was earlier in BT, later in FT, relative to NPT condition. There was no significant magnitude difference between BT and NPT, but there was a 2 cm difference between FT and NPT.

4.2.3. CM(F) Displacement (Sagittal Plane)

Ensemble group average (ten subjects, eight trials per subject) plots of CM(F) displacement for left and right sides of body during PT-HO and PT-HS are presented in Figure 2 and 3.

Stance Side: During both PT-HO and PT-HS, there was a reverse in direction of CM(F) displacement after the initial passive response phase at approximately 300-400 ms after onset of platform motion. For both PT-HO and PT-HS, the time to this initial peak CM (F) displacement was significantly earlier in BT, and significantly later in FT as compared to NPT condition ($p<0.001$). For BT, CM was behind the foot than the NPT condition; while for FT, CM was ahead of foot relative to NPT condition. Table 3 shows the group means and SD of time to peak CM(F) displacement. Following the initial peak displacement, there was a phase shift in trajectories of CM(F) displacement for NPT, BT, and FT. As evident in Figure 3 and Table 3, for PT-HO, the time to the second peak CM(F) displacement was significantly earlier in BT, and significantly later in FT as compared to NPT condition ($p<0.001$). There was a trend for trial effect on magnitude to initial peak CM(F) displacement, but a significant trial effect was observed only during

Table 3 Group means (SD) of magnitude of initial and the second peak CM(F) displacement (meters) and time to peak CM(F) displacement (seconds) for sagittal right and left sides, results (p-values) of One-way Repeated ANOVA, and further Paired t-test. The table includes NPT, BT and FT for PT-HO and PT-HS.

Magnitude of peak CM(F) displacement:

				ANOVA	Paired t-test		
PT-HO	NPT	BT	FT	p-value	NPT Vs BT	NPT Vs FT	BT Vs FT
R initial peak	0.44 (0.05)	0.41 (0.05)	0.47 (0.05)	0.001	0.009	0.027	0.001
R second peak	0.47 (0.02)	0.43 (0.02)	0.44 (0.02)	NS			
L initial peak	0.27 (0.03)	0.31 (0.03)	0.23 (0.03)	0.001	0.001	0.001	0.001
L second peak	0.89 (0.07)	0.92 (0.09)	0.88 (0.08)	0.001	0.022	0.112	0.002
PT-HS							
R initial peak	0.49 (0.06)	0.55 (0.10)	0.46 (0.06)	NS			
L initial peak	0.92 (0.07)	0.88 (0.07)	0.95 (0.07)	0.001	0.001	0.003	0.001

Time to peak CM(F) displacement:

PT-HO							
R initial peak	0.27 (0.03)	0.24 (0.05)	0.30 (0.05)	0.001	0.123	0.033	0.001
R second peak	0.95 (0.10)	0.88 (0.13)	0.98 (0.11)	0.001	0.001	0.022	0.001
L initial peak	0.39 (0.06)	0.35 (0.05)	0.44 (0.06)	0.001	0.003	0.001	0.001
L second peak	0.87 (0.09)	0.82 (0.11)	0.91 (0.10)	0.001	0.002	0.003	0.001
PT-HS							
R initial peak	0.47 (0.05)	0.43 (0.06)	0.51 (0.06)	0.001	0.001	0.004	0.001
L initial peak	0.41 (0.04)	0.39 (0.05)	0.42 (0.05)	0.001	0.008	0.023	0.001

PT-HO ($p<0.001$) not PT-HS. The statistical significant difference was also observed in the second peak magnitude of CM(F) displacement during PT-HO ($p<0.001$) (Table 3).

Swing side: As with the stance side, there was a phase shift in trajectories of CM(F) displacement for the side of the body in swing during platform translation. This was the case during PT-HO and PT-HS. As evident in Figure 2 and presented in Table 1, a significant divergence of CM(F) displacement during FT and BT relative to NPT condition did not occur until about 250 ms to 350 ms after the onset of platform translation. BT peaked and changed direction earlier relative to NPT, and the opposite occurred for FT, a delay to peak and a later time to change in direction. There was a significant trial effect on the time to initial peak CM(F) displacement ($p<0.001$). Relative to NPT condition, for BT the CM(F) was further behind the foot, and for FT the CM(F) was ahead of foot. As seen from Figure 2 and Table 3, there was a significant trial effect on time to second peak CM(F) displacement for PT-HO which occurred at about 800 to 900 ms after onset of platform motion. There was a significant trial effect on magnitude of initial peak CM(F) displacement during both PT-HO and PT-HS ($p<0.001$) (see Table 3 for group means and SD of peak magnitudes of CM(F) displacement). There was no significant trials effect on the magnitude of second peak CM(F) displacement during PT-HO.

4.2.4. Angular Displacement

Figure 4 presents group ensemble averages (ten subjects, eight trials per subject) of angular displacement of ankle, knee, hip, and trunk segment rotation for BT, FT, and NPT condition in sagittal right plane during PT-HO and sagittal left during PT-HS.

Figure 5 presents those in sagittal right during PT-HS and sagittal left during PT-HO. Figure 6 presents group ensemble average of hip angular displacement, and trunk and shank segment rotation for BT, FT, and NPT condition in the frontal plane in right side during PT-HO and left side during PT-HS. Figure 8 presents those in frontal right side during PT-HS and left side during PT-HO.

Sagittal Plane

As with the results of CM(F) displacement, the corrective responses observed in angular displacement on the stance side were similar during PT-HO and PT-HS. This was also the case for corrective responses observed on the swing side during PT-HO and PT-HS.

Stance side: As evidence from Figure 5, following the passive responses there was a phase shift in the trajectories of ankle angular displacement between NPT, BT, and FT. The time to the initial peak dorsiflexion following onset of platform motion was earlier significantly in BT, and later significantly in FT, as compared to NPT condition. Similarly, there was a significant trials effect on the time to peak plantarflexion of the next half gait cycle during PT-HO. The time to peak plantarflexion was earlier for BT and delayed for FT relative to NPT during PT-HO. The trials effect on time to peak dorsiflexion and plantarflexion was observed for both PT-HO and PT-HS. These results would confirm a phase shift in the trajectories of ankle angular displacement. See Table 6 and 7 for group means and SD of times to peak dorsiflexion and plantarflexion, and p-values from ANOVA with repeated measures. As evident in Figure 5 and Table 4, 5, there was a trials effect on magnitude to initial dorsiflexion during PT-HS, but not PT-HO. There was no trials effect on magnitude to peak plantarflexion during PT-HS or PT-HO. See Table 4 and 5 for the group means and SD of magnitude to peak dorsiflexion and plantarflexion, and p-values from ANOVA with repeated measures.

A phase shift in the trajectories of knee angular displacement between BT, FT, and NPT during PT-HO and PT-HS is evident also (Figure 4 and 5). The divergence in knee angular displacement between BT and NPT occurred earlier than observed between FT and NPT, 140-250 ms versus 420-600 ms during PT-HS. The divergence of waveforms occurred at 270-300 ms of onset of platform motion in knee angular displacement for BT, FT, and NPT conditions during PT-HO. Table 1 for times at which divergence in waveforms of NPT, BT, and FT were statistically significant. As evident in Figure 5, following the onset of platform translation there was no difference in the time to peak knee extension. Knee flexion was earlier in BT, and later in FT, relative to NPT condition. There was a significant trials effect on time to peak knee flexion for PT-HO and PT-HS. See Table 6 and 7 for group means and SD of time to peak knee extension and flexion, and p-values. The statistical analysis showed no trials effect on peak magnitude to knee extension or the following knee flexion phase. See Table 4 and 5 for group means and SD of peak magnitude of knee extension and flexion, and p-values from ANOVA with repeated measures.

As evident in Figure 5, following the passive responses there was also a phase shift in trajectories of hip angular displacement between NPT, BT, and FT during PT-HO and PT-HS. The time to initial peak hip extension and the following hip flexion phases were significantly earlier in BT, and significantly later in FT as compared to NPT trials.

Table 4 Group means (SD) of magnitude of angular displacements (degrees) for the sagittal (right and left) and frontal (right and left) planes, and results (p-value)of One-way Repeated ANOVA, and further Paired t-test. The table includes NPT, BT and FT for PT-HO.

PT-HO					ANOVA	Paired t-test		
Sagital Right		NPT	BT	FT	p-value	NPT Vs BT	NPT Vs FT	BT Vs FT
Right Swing	peak plantarflexion	15 (3.7)	17 (4.8)	15 (2.8)	NS			
	peak knee extension	38 (13.2)	30 (8.5)	34 (9.0)	NS			
	peak hip flexion	30 (4.4)	28 (4.0)	29 (4.5)	NS			
Right Stance	peak dorsiflexion	14 (2.1)	14 (3.4)	15 (3.9)	NS			
	peak knee flexion	14 (5.7)	11 (4.8)	13 (4.5)	NS			
	peak knee extension	13 (8.4)	17 (7.6)	12 (6.0)	NS			
	peak hip extension	37 (6.6)	38 (5.8)	33 (5.6)	NS			
Sagital Left								
Left Stance	peak dorsiflexion	14 (3.7)	14 (3.3)	14 (3.6)	NS			
	peak hip extension	13 (2.8)	13 (3.5)	14 (3.0)	NS			
	peak knee extension	9 (1.5)	8 (2.0)	9 (1.7)	NS			
Left Swing	peak plantarflexion	30 (4.4)	29 (8.2)	30 (4.8)	NS			
	peak dorsiflexion	10 (5.8)	8 (5.9)	8 (6.9)	NS			
	peak knee flexion	56 (5.2)	52 (5.2)	51 (5.7)	NS	0.022	0.029	0.449
	peak hip flexion	32 (3.9)	33 (4.8)	29 (4.1)	0.037	0.071	0.9	0.077
Frontal Right								
Right Stance	peak hip abduction	11 (1.9)	12 (5.0)	11 (1.9)	NS			
	peak trunk lat.	3 (1.4)	2 (1.3)	3 (1.7)	NS			
	peak trunk med.	6 (2.2)	5 (1.5)	7 (1.8)	0.016	0.369	0.012	0.048
	peak shank ccw.	20 (5.0)	18 (8.3)	21 (10.0)	NS			
Frontal Left								
Left Stance	peak hip adduction	4 (2.8)	3 (2.3)	4 (2.3)	NS			
Left Swing	peak hip abduction	6 (3.9)	6 (4.5)	6 (4.0)	NS			
	peak trunk med.	3 (1.4)	2 (1.3)	3 (1.7)	NS			
	peak trunk lat.	6 (2.2)	5 (1.5)	7 (1.8)	0.016	0.369	0.012	0.048
	peak shank cw.	10 (4.1)	11 (4.2)	9 (5.1)	NS			
	peak shank ccw.	13 (4.7)	14 (4.1)	11 (5.3)	0.002	0.001	0.082	0.079

Table 5 Same as Table 4, except for PT-HS.

PT-HS		NPT	BT	FT	ANOVA p-value	Paired t-test NPT Vs BT	NPT Vs FT	BT Vs FT
Sagital Right								
Right Stance	peak dorsiflexion	13 (4.1)	11 (3.7)	15 (4.3)	0.001	0.134	0.001	0.001
	peak knee flexion	14 (3.6)	14 (3.6)	14 (3.9)	NS			
	peak plantarflexion	22 (14.4)	22 (7.9)	25 (11.6)	NS			
	peak knee extension	21 (5.9)	21 (6.0)	18 (8.5)	NS			
	peak hip extension	47 (7.4)	53 (16.9)	47 (8.3)	NS			
Right Swing	peak knee flexion	55 (9.6)	54 (6.6)	54 (8.3)	NS			
	peak knee extension	66 (6.6)	68 (16.2)	69 (5.8)	NS			
	peak hip flexion	26 (4.5)	29 (5.3)	27 (5.8)	NS	0.028	0.127	0.106
Sagital Left								
Left Swing	peak plantarflexion	28 (6.1)	28 (5.9)	25 (5.9)	NS			
	peak knee flexion	59 (4.9)	60 (4.8)	59 (5.1)	NS			
	peak knee extension	58 (3.5)	55 (4.9)	57 (6.1)	NS			
	peak hip flexion	35 (8.0)	36 (8.7)	38 (7.0)	NS			
Left Stance	peak knee flexion	15 (3.9)	17 (7.7)	20 (5.8)	0.042	0.317	0.001	0.224
	peak hip extension	32 (9.8)	28 (12.1)	32 (9.0)	NS			
Frontal Right								
Right Stance	peak hip abduction	7 (2.5)	7 (2.6)	7 (2.3)	NS			
	peak trunk med.	7 (2.0)	7 (2.4)	7 (1.8)	NS			
	peak shank ccw.	21 (7.7)	21 (7.2)	24 (10.5)	NS			
Right Swing	peak hip adduction	11 (2.1)	10 (2.2)	14 (7.5)	NS			
	peak trunk lat.	6 (0.9)	6 (1.6)	6 (2.0)	NS			
	peak shank cw.	23 (3.8)	19 (3.7)	21 (4.6)	NS			
Frontal Left								
Left Swing	peak hip abduction	7 (4.5)	7 (3.8)	7 (4.6)	NS			
	peak hip adduction	6 (2.6)	7 (4.5)	7 (3.1)	NS			
	peak trunk lat.	6 (1.3)	6 (1.3)	6 (1.3)	NS			
	peak shank cw.	10 (4.2)	11 (5.2)	10 (4.8)	NS			
	peak shank ccw.	10 (5.4)	11 (5.4)	11 (5.0)	0.02	0.083	0.031	0.134

Table 6 Group means (SD) of time to peak angular displacements (seconds) for the sagittal (right and left) and frontal (right, and left). planes, and results (p-values) of One-way Repeated ANOVA, and further Paired t-test. The table includes NPT, BT and FT for PT-HO.

PT-HO					ANOVA	Paired t-test		
Sagital Right		NPT	BT	FT	p-value	NPT Vs BT	NPT Vs FT	BT Vs FT
Right Swing	time peak plantarflexion	0.02 (0.02)	0.03 (0.04)	0.01 (0.02)	NS			
	time peak knee extension	0.27 (0.07)	0.24 (0.08)	0.30 (0.06)	0.001	0.005	0.01	0.001
	time peak hip flexion	0.22 (0.07)	0.20 (0.06)	0.24 (0.09)	NS			
Right Stance	time peak dorsiflexion	0.84 (0.12)	0.77 (0.18)	0.86 (0.11)	0.014	0.026	0.354	0.033
	time peak knee flexion	0.52 (0.08)	0.47 (0.08)	0.55 (0.07)	NS			
	time peak knee extension	0.79 (0.14)	0.82 (0.08)	0.84 (0.11)	NS			
	time peak hip extension	0.93 (0.08)	0.85 (0.13)	0.94 (0.11)	0.034	0.096	0.788	0.003
Sagital Left								
Left Stance	time peak dorsiflexion	0.18 (0.10)	0.15 (0.08)	0.25 (0.11)	0.003	0.304	0.025	0.005
	time peak hip extension	0.35 (0.04)	0.29 (0.06)	0.41 (0.06)	0.001	0.005	0.01	0.001
	time peak knee extension	0.33 (0.06)	0.20 (0.07)	0.36 (0.08)	NS			
Left Swing	time peak plantarflexion	0.50 (0.06)	0.36 (0.07)	0.56 (0.07)	0.001	0.001	0.001	0.001
	time peak dorsiflexion	0.79 (0.08)	0.74 (0.11)	0.88 (0.14)	0.001	0.016	0.027	0.004
	time peak knee flexion	0.55 (0.05)	0.48 (0.06)	0.59 (0.06)	0.001	0.001	0.015	0.001
	time peak hip flexion	0.70 (0.09)	0.68 (0.11)	0.74 (0.10)	0.037	0.398	0.01	0.042
Frontal Right								
Right Stance	time peak hip abduction	0.83 (0.09)	0.75 (0.14)	0.87 (0.11)	0.001	0.009	0.012	0.001
	time peak trunk lat.	0.44 (0.12)	0.39 (0.12)	0.47 (0.08)	0.001	0.018	0.081	0.004
	time peak trunk med.	0.99 (0.08)	0.92 (0.10)	1.04 (0.11)	0.003	0.004	0.669	0.007
	time peak shank cw.	0.28 (0.11)	0.19 (0.12)	0.32 (0.92)	0.003	0.007	0.648	0.001
	time peak shank ccw.	1.07 (0.12)	0.98 (0.24)	1.10 (0.04)	0.001	0.018	0.081	0.004
Frontal Left								
Left Stance	time peak hip adduction	0.28 (0.09)	0.22 (0.10)	0.30 (0.10)	0.024	0.018	0.922	0.065
Left Swing	time peak hip abduction	0.57 (0.05)	0.54 (0.08)	0.64 (0.07)	0.001	0.033	0.01	0.002
	time peak trunk med.	0.44 (0.12)	0.39 (0.12)	0.47 (0.08)	0.001	0.018	0.081	0.004
	time peak trunk lat.	0.99 (0.08)	0.92 (0.10)	1.04 (0.11)	0.003	0.004	0.669	0.007
	timepeak shank cw.	0.58 (0.13)	0.51 (0.10)	0.58 (0.06)	0.001	0.002	0.009	0.001
	time peak shank ccw.	0.84 (0.10)	0.81 (0.08)	0.89 (0.11)	0.03	0.214	0.242	0.001

Table 7 Same as Table 6, except for PT-HS.

PT-HS					ANOVA	Paired t-test		
Sagital Right		NPT	BT	FT	p-value	NPT Vs BT	NPT Vs FT	BT Vs FT
Right Stance	time peak dorsiflexion	0.33 (0.08)	0.23 (0.08)	0.40 (0.08)	0.001	0.001	0.001	0.001
	time peak knee flexion	-0.04 (0.08)	-0.04 (0.08)	-0.05 (0.09)	NS			
	time peak plantarflexion	0.56 (0.08)	0.52 (0.07)	0.58 (0.21)	NS			
	time peak knee extension	0.31 (0.04)	0.31 (0.04)	0.34 (0.05)	NS			
	time peak hip extension	0.66 (0.05)	0.61 (0.05)	0.69 (0.07)	0.001	0.001	0.032	0.001
Right Swing	time peak knee flexion	0.42 (0.05)	0.37 (0.05)	0.48 (0.06)	0.001	0.001	0.001	0.001
	time peak knee extension	0.87 (0.03)	0.86 (0.07)	0.90 (0.05)	0.001	0.001	0.001	0.001
	time peak hip flexion	0.90 (0.08)	0.83 (0.07)	0.91 (0.09)	0.002	0.008	0.12	0.008
Sagital Left								
Left Swing	time peak plantarflexion	0.04 (0.03)	0.02 (0.08)	0.03 (0.08)	NS			
	time peak knee flexion	0.12 (0.04)	0.12 (0.02)	0.10 (0.04)	NS			
	time peak knee extension	0.42 (0.04)	0.40 (0.04)	0.42 (0.04)	0.003	0.218	0.018	0.006
	time peak hip flexion	0.24 (0.04)	0.24 (0.04)	0.24 (0.05)	NS			
Left Stance 2	time peak knee flexion	0.64 (0.08)	0.60 (0.08)	0.65 (0.07)	NS			
	time peak hip extension	0.93 (0.07)	0.90 (0.08)	0.94 (0.07)	0.038	0.017	0.529	0.051
Frontal Right								
Right Stance	peak hip abduction	0.35 (0.03)	0.32 (0.01)	0.39 (0.07)	0.001	0.001	0.001	0.001
	peak trunk med.	0.49 (0.06)	0.48 (0.07)	0.50 (0.05)	NS			
	peak shank ccw.	0.60 (0.06)	0.56 (0.06)	0.64 (0.07)	0.001	0.001	0.008	0.001
Right Swing	peak hip adduction	0.68 (0.08)	0.65 (0.07)	0.87 (0.06)	0.001	0.003	0.119	0.004
	peak trunk lat.	0.97 (0.07)	1.00 (0.09)	1.00 (0.08)	NS			
	peak shank cw.	0.95 (0.02)	0.94 (0.03)	0.97 (0.02)	0.001	0.018	0.081	0.004
Frontal Left								
Left Swing	peak hip abduction	0.06 (0.08)	0.07 (0.05)	0.06 (0.14)	NS			
	peak hip adduction	0.87 (0.10)	0.84 (0.10)	0.92 (0.10)	0.001	0.161	0.001	0.001
	peak trunk lat.	0.51 (0.06)	0.49 (0.06)	0.52 (0.04)	NS			
	peak shank cw.	0.10 (0.18)	0.14 (0.19)	0.14 (0.16)	NS			
	peak shank ccw.	0.43 (0.11)	0.44 (0.13)	0.43 (0.09)	NS			

This was the case for both PT-HO and PT-HS. See Table 6 and 7 for group means and SD of times to peak hip extension and flexion, and results of statistical analysis. These results would confirm a phase shift in the trajectories of hip angular displacement. There was no trials effect on magnitude to peak hip extension or the following right hip flexion phase during PT-HS condition. There was no trials effect on the magnitude to peak hip extension during the left stance phase in PT-HO condition. But there was a trials effect on the magnitude of following peak hip flexion ($p < 0.04$). BT resulted in increase hip flexion, while FT resulted in decrease hip flexion. The largest hip flexion was observed in BT (33 degrees), the smallest in FT (29 degrees) relative to NPT (32 degrees) condition.

Swing side: Similar corrective responses in angular displacement were observed in right ankle in PT-HO and left ankle in PT-HS conditions. The divergence in trajectories of ankle angular displacement is evident in Figure 4. A significant divergence in trajectories between BT, FT, and NPT was only observed during the PT-HO condition. The time at which the divergence was statistically significant was 840 ms for BT, and 900 ms for FT (See Table 1). As evident in Figure 4, Table 6 and 7, there was no difference in time to initial peak ankle plantarflexion between BT, FT, and NPT during both PT-HO and PT-HS. Following the initial peak ankle plantarflexion, there was a phase shift in trajectories of ankle angular displacement for NPT, BT, and FT. The time to the following ankle dorsiflexion was earlier significantly in BT, and later significantly in FT, relative to NPT condition. See Table 6 and 7 for group means and SD of times to peak ankle plantarflexion and dorsiflexion, and results of statistical analysis. There was no trials effect on magnitude to peak ankle plantarflexion during both PT-HO and PT-HS, or dorsiflexion during PT-HO condition. See Table 4 and 5 for group means and SD of peak magnitude of ankle plantarflexion and dorsiflexion phases.

Significant divergence in trajectories of knee angular displacement between BT, FT, and NPT were observed during PT-HO and PT-HS. As presented in Table 1, the time period of significant divergence in trajectories began earlier during PT-HO (200-450 ms) as compared to PT-HS (360-570 ms). As evident in Figure 4, for both PT-HO and PT-HS, the time to initial peak knee extension and the following knee flexion were earlier significantly in BT, later in FT, as compared to NPT condition. See Table 6 and 7 for group means and SD of times to initial peak knee extension and the following peak knee flexion, and results of statistical analysis. There was no trials effect on magnitude to peak knee extension or the following peak flexion during PT-HO and PT-HS conditions. See Table 4 and 5 for group means and SD of peak magnitude of knee extension and flexion phases.

Significant divergence in the trajectories of hip angular displacement between BT, FT, and NPT were also observed during PT-HO and PT-HS. As presented in Table 1, the time period of significant divergence in trajectories began earlier during PT-HS (240-270 ms) as compared to PT-HO (340-600 ms). As illustrated in Figure 4, there was no trials effect on time to peak hip flexion, but the time to following peak hip extension was earlier significantly in BT and later in FT relative to NPT. This was the case for both PT-HO and PT-HS. See Table 6 and 7 for means and SD of time to peak magnitude of hip flexion and extension. There was no trials effect on magnitudes to peak hip flexion and extension during PT-HO and PT-HS. See Table 4 and 5 for means and SD of magnitudes to peak hip flexion and extension.

Significant divergence in the trajectories of trunk segment rotation between BT, FT, and NPT was observed (See Figure 4). As presented in Table 1, the time period of significant divergence in trajectories was similar for PT-HO (190-240 ms) and PT-HS (200-290 ms). As evident in Figure 4, a phase shift was observed in the trajectories of trunk segment rotation displacement for NPT, BT, and FT. The direction of divergence was the same for PT-HO and PT-HS. It showed an increase in magnitude of forward trunk segment rotation in BT condition, while a decrease in magnitude of forward trunk segment rotation in FT condition, as compared to NPT condition. This was the case for both PT-HO and PT-HS.

Frontal Plane

Stance side: As evident in Figure 7, similar responses were observed for hip joint displacement, shank and trunk segment rotation during PT-HS and PT-HO.

As evident in Figure 7 and presented in Table 1, significant divergence in trajectories of hip angular displacement between BT, FT, and NPT was observed. The time period of significant divergence in the trajectories was slightly earlier for PT-HO (200-290 ms) as compared to PT-HS (350-390 ms). For PT-HO, the time to initial peak hip adduction was earlier significantly in BT, later in FT, relative to NPT. The time to next peak hip abduction was earlier in BT, later in FT, relative to NPT. For PT-HS, the time to initial peak hip abduction and the following peak hip adduction were earlier significantly in BT, later in FT, relative to NPT. See Table 6 and 7 for group means and SD of time to peak hip adduction and abduction angular displacements, and results of statistical analysis. As presented in Table 4 and 5, there was no trials effect on magnitude to peak hip adduction and abduction during stance and subsequent swing phases. This was the case for both PT-HO and PT-HS.

As illustrated in Figure 7, significant divergence in trajectories of trunk segment rotation between BT, FT, and NPT were observed. The time period of significant divergence in trajectories was almost the same for PT-HO (100-440 ms) and for PT-HS (100-350 ms). For PT-HO, the time to initial peak trunk lateral rotation and the following medial rotation was significantly earlier in BT, later in FT, as compared to NPT condition. See Table 6 for group means and SD of time to peak trunk lateral and peak medial rotations, and results of statistical analysis. There was no trials effect on magnitude to initial peak trunk segment rotations during PT-HO and PT-HS. A significant difference was only observed in the second peak magnitude of trunk lateral rotation during PT-HO. See Table 4 and 5 for group means and SD of magnitude of trunk segment rotation, and results of statistical analysis.

Although the trajectories of shank rotation were different on the stance side during PT-HO and PT-HS, the corrective responses to the walking disturbance was the same essentially. As evident in Figure 7 and presented in Table 1, significant divergence in trajectories of shank segment rotation between BT, FT, and NPT were observed. The time period of significant divergence in trajectories was earlier during PT-HO (350-670 ms), as compared to PT-HS (140-400 ms).

For PT-HS, the time to initial peak counter-clockwise (ccw) and the following clockwise (cw) shank rotation was significantly earlier in BT, and later in FT, relative to NPT ($p<0.001$) (Table 7). For PT-HO, the time to initial peak cw shank rotation and the following ccw was significantly earlier in BT, later in FT, relative to NPT. There was no trials effect on magnitude to initial peak shank segment rotations during PT-HO and PT-HS. See Table 4 and 5 for group means and SD of magnitude of shank segment rotation.

Swing side: As with the stance side similar responses were observed for hip joint displacement, shank and trunk segment rotation during PT-HS and PT-HO.

As evident in Figure 6 and presented in Table 1, a significant divergence in trajectories of hip angular displacement between BT, FT, and NPT was observed. The time to significant divergence in trajectories began approximately 250 ms after onset of platform motion. The time to initial peak hip joint displacement (hip abduction for PT-HO, hip adduction for PT-HS) was significantly earlier in BT, later in FT, relative to NPT. See Table 6 and 7 for group means and SD of time to peak hip adduction and abduction angular displacements, and results of statistical analysis. As presented in Table 4 and 5, there was no trials effect on magnitude to peak hip abduction during PT-HO and hip adduction during PT-HS.

As evident in Figure 6 and presented in Table 1, significant divergence in trajectories of trunk segment displacement between BT, FT, and NPT was observed. The start time of significant divergence of the trunk segment displacement trajectories ranged from 200 to 500 ms. A statistical significant trials effect on time to initial peak trunk rotation was only observed during PT-HO not PT-HS. The time to initial peak trunk lateral rotation and the following trunk medial rotation was earlier significantly in BT, later in FT, relative to NPT. See Table 6 and 7 for group means and SD of time to peak trunk segment rotation. As presented in Table 4 and 5, there was no trials effect on magnitude to initial peak trunk segment rotation during PT-HO and PT-HS.

Similar phase shift in the trajectories of shank segment rotation between BT, FT, and NPT was observed. As evident in Figure 6 and presented in Table 1, significant divergence in trajectories of shank segment rotation displacement between BT, FT, and NPT was observed. The time period of significant divergence in shank segment rotation trajectories was slightly earlier during PT-HS (120-140 ms) than that during PT-HO (190-200 ms). Table 6 and 7 present the group means and SD of time to peak shank segment rotation. Similar to the results of trunk segment rotation, the statistical significant trials effect on the time to initial peak shank segment rotation was only observed during PT-HO not PT-HS. The time to initial peak shank segment cw rotation and the following shank segment ccw rotation was significantly earlier in BT, later in FT, relative to NPT. As presented in Table 4 and 5, there was no trials effect on magnitude to peak shank segment ccw rotation during PT-HO.

4.2.5. Ground Reaction Forces:

Figure 9 presents the group average plots of ground reaction forces in mediolateral (Fx), vertical (Fz) ground reaction for NPT, BT, and FT during PT-HO and PT-HS.

Statistical analysis did reveal a significant difference in the trajectories of both Fx and Fz between BT, FT, and NPT. The time period of significant divergence in Fz were similar during PT-HO (330-570 ms) and PT-HS (320-400 ms). As with the findings of swing/stance duration describe above, the plots of Fz showed earlier foot off for BT and delayed for FT, relative to NPT. The time period of significant divergence in Fx trajectories was also around 300-500 ms during PT-HO and PT-HS. As evident from Fig 9

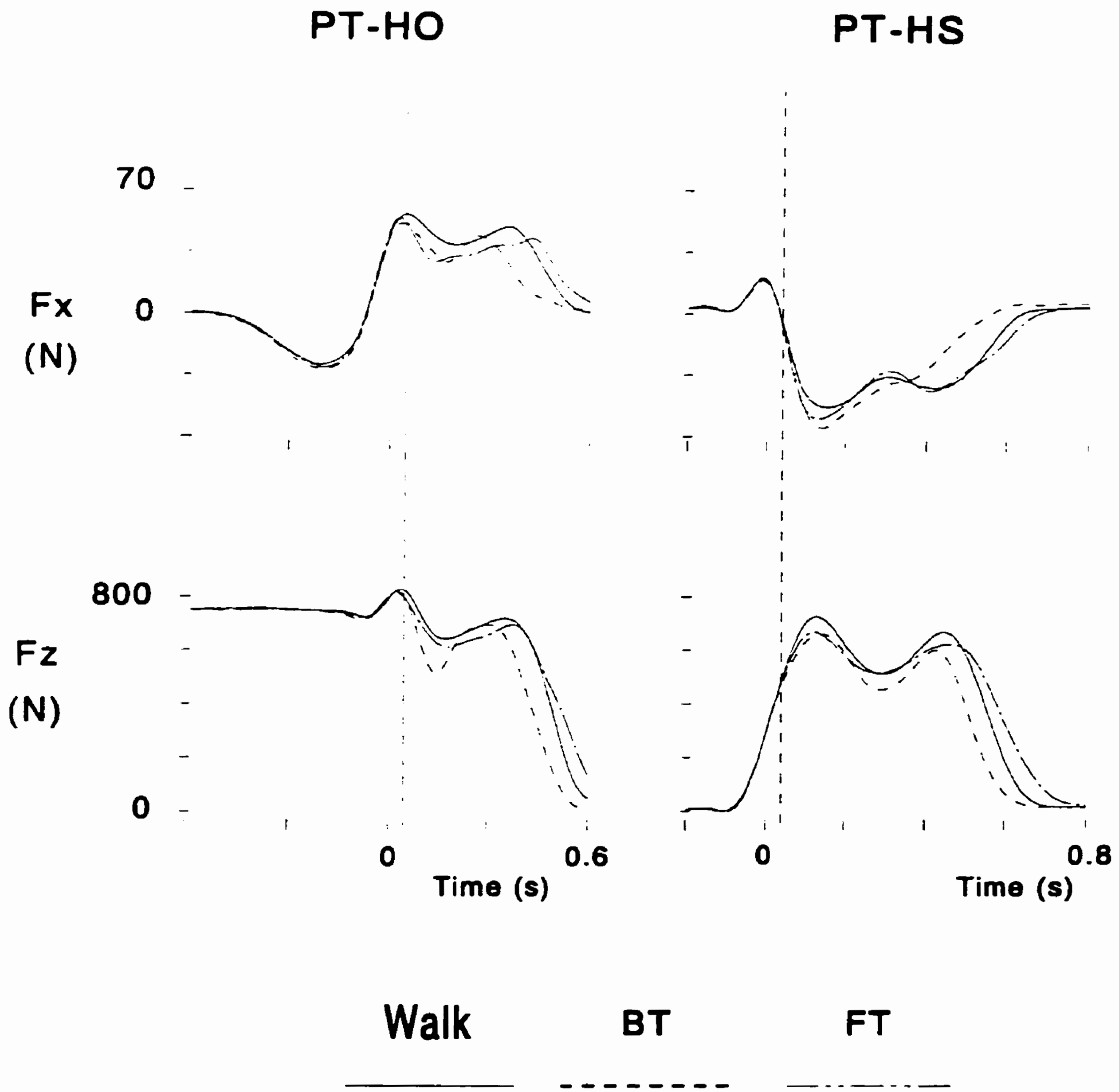

Figure 9. Group average plots of ground reaction forces in medio-lateral (Fx), vertical (Fz) direction for NPT, BT, and FT conditions during right heel-off trials (PT-HO) and right heel strike trials (PT-HS). For y-axis, zero represents standing still baseline position, before right heel off and onset of platform translation. Vertical dashed lines at time zero is the onset of platform translation. For Fx, positive values represent the leftward (medial) direction, negative values are toward right side (lateral).

Figure 9, there was earlier push-off (rightward directed Fx for PT-HO and leftward for PT-HS) for BT and later for FT, relative to NPT.

4.2.6. EMG

Difference EMG waveforms of individual trials for one representative subject are presented in Figure 10-13. Table 8 presents the group means and SD of onset latencies, the frequency of the presence of a response, and frequency of corrective responses which were excitatory for bilateral thigh and leg muscles during backward platform translation. Excitatory responses were increased levels of muscle activity relative to average NPT trials, and inhibitory responses were decreased levels of activity relative to averaged NPT trials. AB and RF means include nine trials per subject for nine subjects, AD and HA means include six trials per subject for nine subjects, and TA and GS means include three trials per subject. Table 9 presents the same information but for forward translation.

Early corrective responses were observed in all muscles during PT-HO and PT-HS for both BT and FT. But the onset latencies and response frequencies were quite different.

Backward Translation

The range of mean onset latencies was from 61ms to 328ms of the platform motion for swing limb and stance limb. The range of response frequency was from 59% to 100%. The range of frequency of excitatory was from 24% to 100%.

Stance limb: Right side limb during PT-HS and left side limb during PT-HO were the stance limbs during the time of platform translation. The patterns of corrective

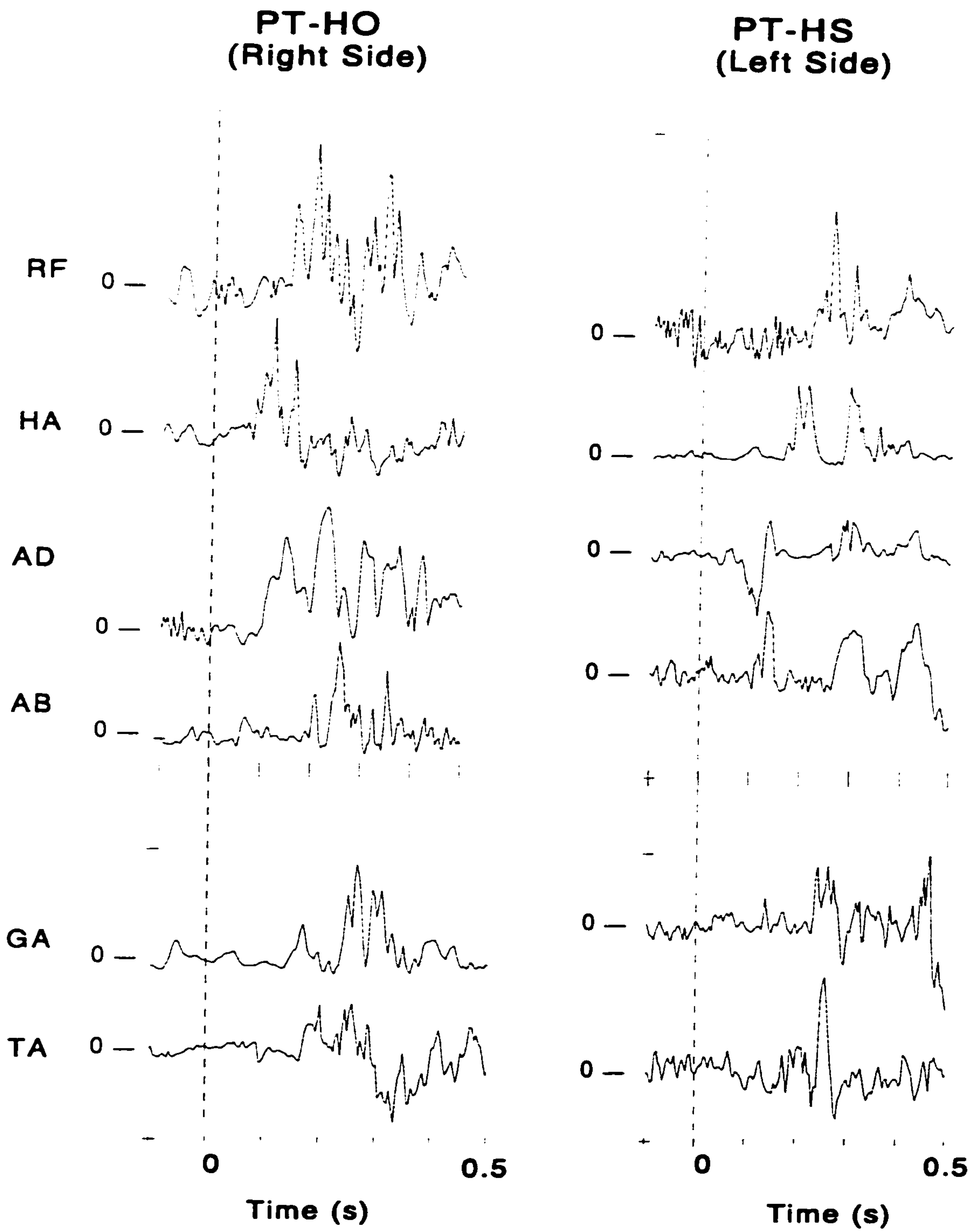

Figure 10 Difference EMG waveforms from one subject of AB, AD, HA, RF, GA, and TA for swing limb during Backward Translation; PT-HO (right side) and PT-HS (left side). For x-axis, time zero is the onset of platform translation.

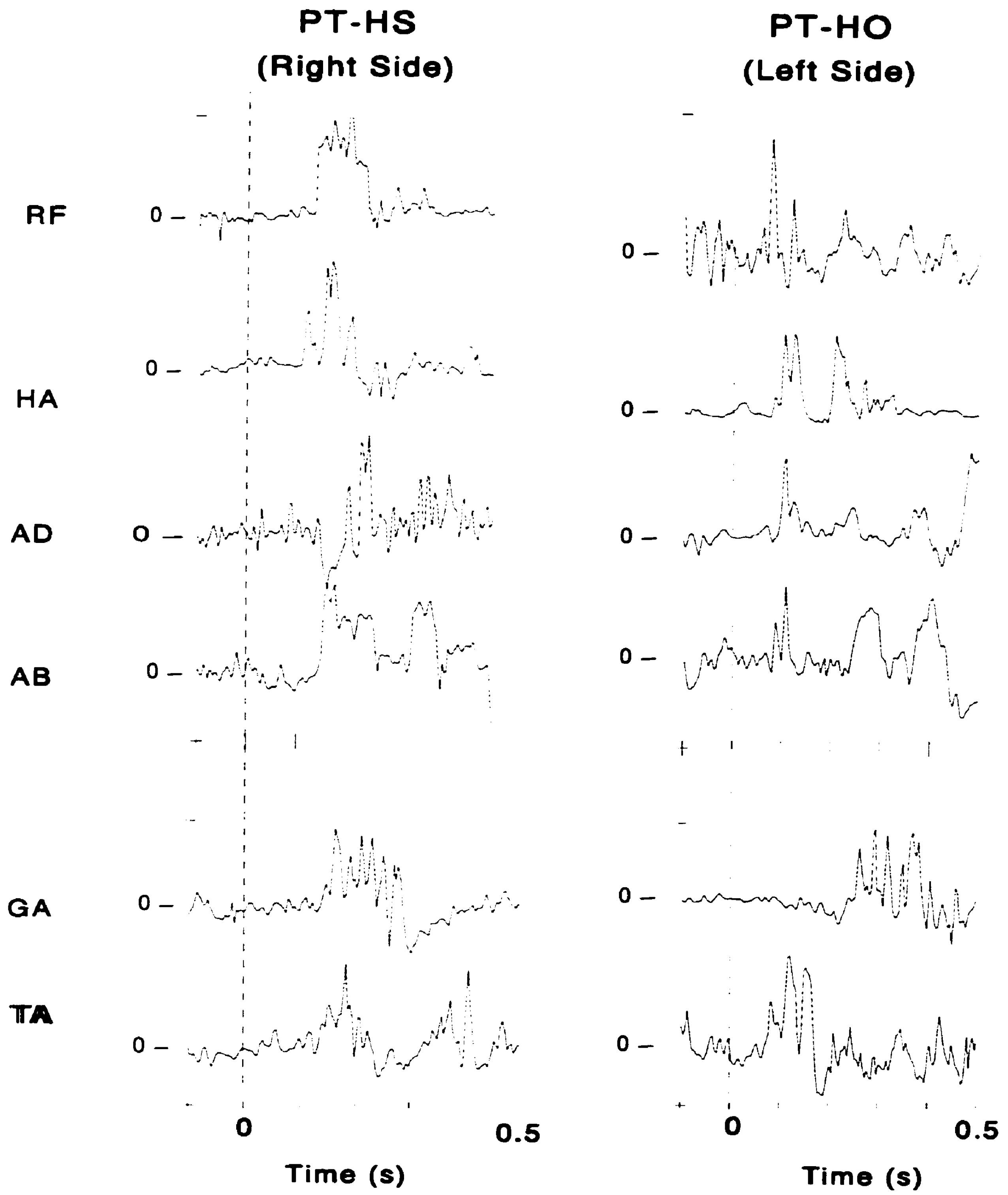

Figure 11 Same as Figure 10, except for stance limb; right side during PT-HS, and left side during PT-HO.

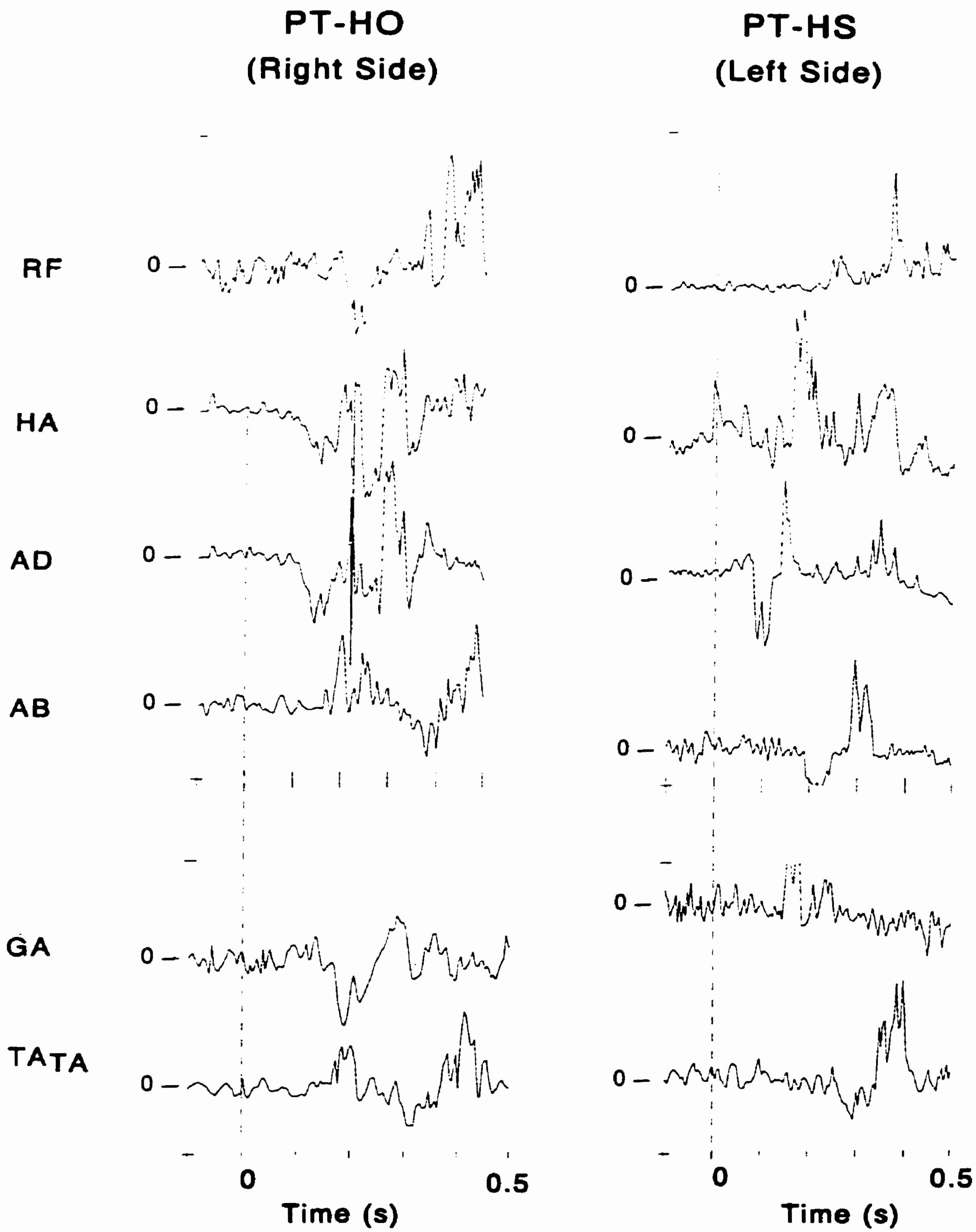

Figure 12 **Difference EMG waveforms from one subject of AB, AD, HA, RF, GA, and TA for swing limb during Forward Translations; PT-HO (right side) and PT-HS (left side). For x-axis, time zero is the onset of platform translation.**

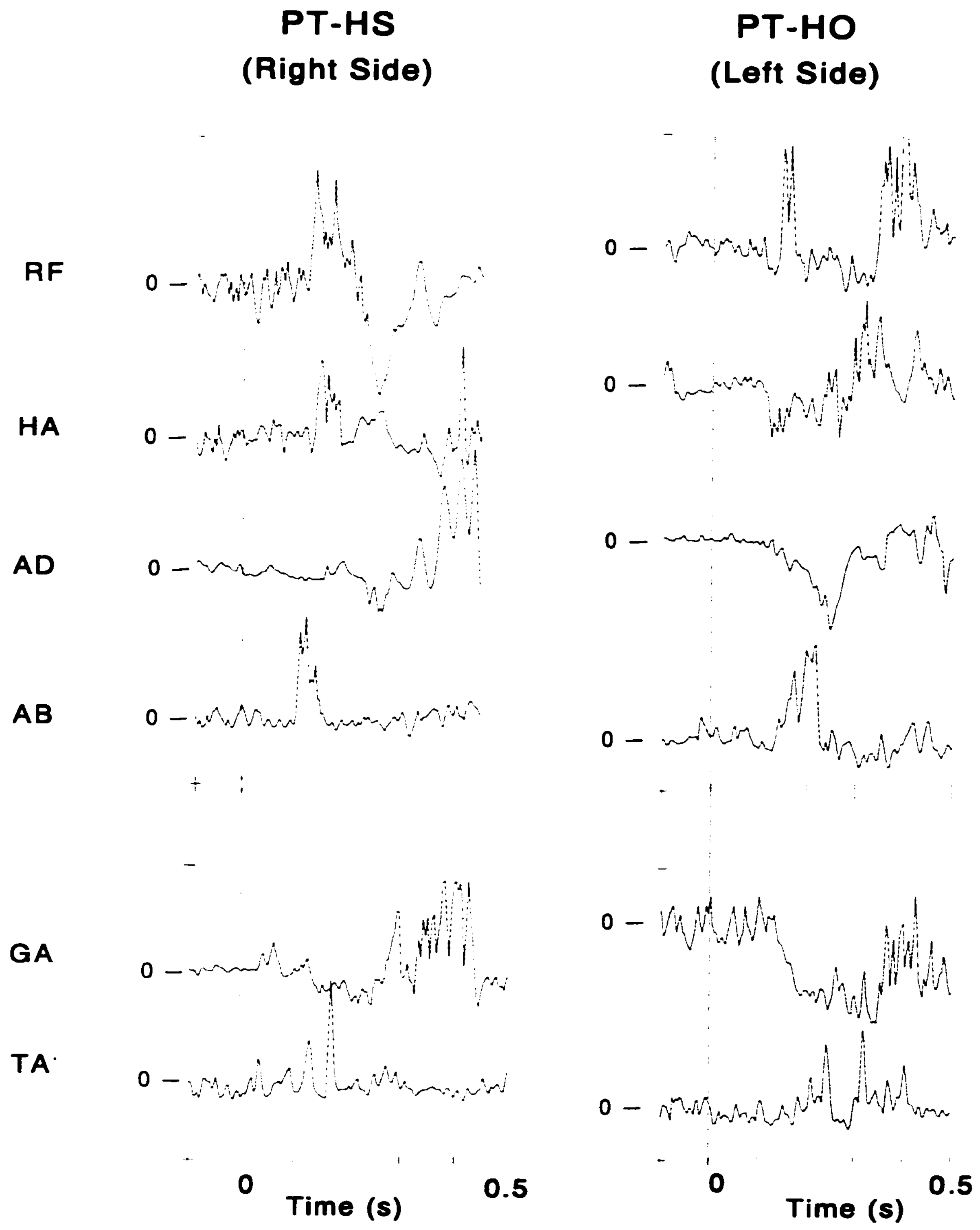

Figure 13 Same as Figure 12, except for stance limb; right side during PT-HS, and left side during PT-HO .

Table 8 Group means (SD) of onset latencie and response frequency for initial corrective responss in respective muscles during Backward Translations. "R" for right side, "L" for left side.

Backward translation

Swing Side		Onset	SD	Response Frequency (% of total)	Frequency of Excitation (% of total)
PT-HO					
	R-AB	201	79	77	100
	R-AD	120	40	96	94
	R-HA	66	17	100	80
	R-RF	142	36	99	83
	R-TA	179	42	92	96
	R-GA	155	94	100	78
PT-HS					
	L-AB	168	83	92	65
	L-AD	73	32	92	24
	L-HA	176	81	98	68
	L-RF	212	110	90	84
	L-TA	320	73	67	100
	L-GA	328	136	71	100
Stance Side					
PT-HS					
	R-AB	160	63	59	92
	R-AD	166	109	67	56
	R-HA	100	27	100	98
	R-RF	146	63	95	74
	R-TA	138	69	62	93
	R-GA	138	102	88	90
PT-HO					
	L-AB	119	60	99	74
	L-AD	106	66	96	28
	L-HA	113	35	95	88
	L-RF	122	52	88	94
	L-TA	252	106	72	100
	L-GA	61	9	83	100

- response frequency = % of total trials with a response within 250 ms
- frequency of excitation = % of trials with excitatory corrective responses
- response time is relative to onset of platform motion

Table 9 Same as Table 8, except for forward translation.

		Onset	SD	Response Frequency (% of total)	Frequency of Excitation (% of total)
Swing Side					
PT-HO					
	R-AB	207	97	99	29
	R-AD	123	75	100	15
	R-HA	147	49	96	43
	R-RF	158	40	99	60
	R-TA	172	84	100	54
	R-GA	160	90	88	46
PT-HS					
	L-AB	218	90	89	36
	L-AD	75	15	98	19
	L-HA	132	93	93	47
	L-RF	223	158	57	88
	L-TA	282	150	42	37
	L-GA	151	139	43	63
Stance Side					
PT-HS					
	R-AB	96	41	60	88
	R-AD	254	103	54	43
	R-HA	144	96	98	81
	R-RF	113	44	92	64
	R-TA	136	78	100	71
	R-GA	126	51	100	25
PT-HO					
	L-AB	135	62	93	63
	L-AD	127	90	100	4
	L-HA	109	30	98	30
	L-RF	123	53	93	87
	L-TA	219	91	41	79
	L-GA	116	33	100	5

- response frequency = % of total trials with a response within 250 ms
- frequency of excitation = % of trials with excitatory corrective responses
- response time is relative to onset of platform motion

responses were different when the disturbance was presented in the initial step as compared to the second step.

For example, onset latency of GA response during initial step (PT-HO) was 61 ms as compared to 133 ms during second step (PT-HS). This was similar for AB and AD. AB and AD for the initial step also have higher response frequencies than during the second step. Such as, the response frequency of AB was 99% during the initial step, as compared to 59% during the second step. While TA for the initial step showed much greater onset latency than during the second step, 252 ms versus 138 ms. There was no difference of response frequency between the initial step and the second step for GA and TA. There was no difference of onset latencies and response frequencies between the initial step and the second for HA and RF.

Four of six muscles showed the excitatory frequency for initial step was not less than that for the second step. The frequencies of excitatory responses for HA, TA, and GA were similar between the initial (PT-HO) and second step (PT-HS). Such as, for GA, the frequency of excitatory response was 100% for the initial step, while was 90% for the second step. Frequencies of excitatory responses for RF were greater during initial step than during second step, 94% versus 74%.

Swing limb: Right side limb during PT-HO and left side limb during PT-HS were the swing limbs during the time of platform translation. As for the stance limb described above, the patterns of corrective responses were different when the disturbance was presented in the initial step as compared to the second step. Earlier onset latency and higher response frequency were observed for HA, RF, TA, and GA during the initial step

(Right side for PT-HO) as compared to the second step (Left side for PT-HS). Such as, the onset latency and response frequency of HA during initial step were 60 ms and 100%, as compared to 176 ms and 98% during the second step. These were opposite for AB and AD. For example, the onset latency and response frequency of AD during initial step were 120 ms and 96%, while were 73 ms and 92% during the second step. The frequency of excitatory was over 78% for all the right side limb muscles during PT-HO. The frequencies of excitatory for RF and TA were similar for initial and second step. Frequency of excitatory responses for AB, AD, and HA were greater during initial step than during second step. For example, the frequency of excitatory response for AB was 100% during the initial step, while was 45% during the second step. GA showed the opposite result; the frequency of excitatory response was 78% during the initial step, while was 100% during the second step

Forward Translation

The range of onset latency was 75ms to 282ms. The response frequency was from 41% to 100%. As evidence from the Figure 12, 13, and Table 9, large percentage of the responses exhibited an initial inhibitory burst followed by an excitatory burst. In general the frequency of excitatory responses was higher during BT than during FT.

For the stance side, the mean frequency of excitatory for the initial step was 83.8% during BT, while 62% during FT; for the second step was 80.7% during BT, while 44.8% during FT. Such as, the frequencies of excitatory for the initial step for TA and GA were 93% and 90% during BT, while 71% and 25% during FT. The frequencies of excitatory for the second step for TA and GA were both 100% during BT, while were only 79% and 5% during FT.

For swing side, the mean frequency of excitatory for the initial step was 88.5% during BT, while was 41.2% during FT; for the second step was 68.5% during BT, and was 48.3% during FT. For example, the frequencies of excitatory for the initial step for TA and GA were 96% and 78% during BT; while were 54% and 46% during FT. For the second step, the frequencies of excitatory for TA and GA were both 100% during BT, while were 37% and 63% during FT.

In general the onset latencies and the response frequencies were similar between BT and FT. For example, for the swing side, the onset latency and the frequency of response for the initial step for AD were 120 ms and 96% during BT, while were 123 ms and 100% during FT. The onset latency and the frequency of the response for the second step for AD were 73 ms and 92% during BT, and were 75ms and 98%. For the stance side, the onset latency and the frequency of response for the initial step for HA were 100 ms and 100% during BT, while were 144 ms and 98% during FT. The onset latency and the frequency of the response for the second step for HA were 113 ms and 95% during BT, and were 109 ms and 98%.

As with corrective responses to BT, the patterns of corrective responses to FT were different when the disturbance was presented in the initial step as compared to the second step in FT.

Stance limb: AD and HA exhibited earlier onset latencies during the initial step (PT-HO) than during the second step (PT-HS). For example, the onset latency of AD for the initial step was 127 ms, and was 254 ms for the second step. There was a higher response frequency for AD during the initial step than during the second step; 100% versus 54%. TA showed greater onset latency and lower response frequency during the initial step than during the second step. Onset latency and response frequency of TA during the initial step were 219 ms and 41%, as compared to 136 ms and 100% during the second step. This was similar for AB, except AB for the initial step had a higher response frequency than during the second step. There was no difference of onset latencies or response frequencies between the initial step and the second step for RF and GA.

Swing limb: RF and TA exhibited earlier onset latencies and higher response frequencies during the initial step as compared to the second step. Such as, the onset latency and response frequency of RF for the initial step were 158 ms and 99%, and were 223 ms and 57% for the second step. AD for the initial step showed a greater onset latency than during the second step, 123 ms versus 75 ms. There was no difference in response frequency between the initial step and the second step for AD. Onset latencies and response frequencies of AB and HA were not different between the initial step and the second step. These were similar for GA, except that response frequency was higher for the initial step than during the second step. The response frequency for the initial step for GA was 88%, while was 43% for the second step.

This study investigated corrective responses of healthy adults in both the sagittal and frontal planes during unexpected disturbances while walking. The intent was to provide a better understanding of the corrective balance responses required during unexpected disturbances at different periods in gait cycle, and to examine corrective responses in M-L plane. Walking was perturbed using a movable platform installed in the middle of a walkway. The types of balance disturbances included backward translation (BT) and forward translation (FT) in the sagittal plane presented at different points in the gait cycle, in particular during the single support phase, just after right heel off (PT-HO), and just after heel strike (PT-HS).

5.1. MAIN FINDINGS

Sudden platform movement in either a forward or backward direction induced passive mechanical effects within 100ms of the onset of platform motion. These were observed in the trajectories of lateral malleolus marker displacement, CM(F) displacement and angular displacement at the ankle and hip in the stance side of the body in the sagittal plane. No passive effects were observed in the trajectories of lateral malleolus marker displacement, CM(F) displacement and angular displacement at the hip in the frontal plane.

The trajectories of CM(S) displacement help us to understand the task performance levels between the perturbed walking and the unperturbed walking condition, since the goal of a walking task is to maintain the rate of forward progression of the body. There was no significant difference between NPT, FT and BT in forward progression as evidenced from magnitude of CM(S) displacement. However, in the frontal plane there were significant differences in the trajectory of CM(S) displacement between BT, FT, and NPT during both PT-HO and PT-HS conditions. The effect of BT was a significant decrease in duration and magnitude of the right shift in CM(S) displacement during PT-HO. The opposite effect was observed for FT. During PT-HS, the effect of BT was decrease in duration and magnitude of the left shift in CM(S) displacement, the opposite effect was observed for FT.

Corrective muscle responses were observed in all muscles; onset latencies between 60-320 ms. Some muscles showed increased activity relative to NPT, and others a decrease. Onset latency and sign of muscle responses were specific to the direction of disturbance and where in the gait cycle the disturbance was presented, i.e. in the initial step (PT-HO) or during the second step (PT-HS). One primary EMG findings of this study was the definite trend for EMG responses to BT to be excitatory while those for FT to be inhibitory.

During BT and FT, there was a change in swing/stance durations and swing distances as compared to NPT condition. There was shorter duration and distance for BT, and longer for FT. In sagittal plane, the change in trajectories of angular displacements, trunk segment rotation, and CM(F) displacement that occurred in response to FT and BT relative to NPT was a phase shift with no significant change in pattern or magnitude. For BT where the foot was displaced backward relative to CM, we observed an initial increase in rate of angular displacement at the ankle, knee and hip, which resulted in a phase lead in the trajectories. For FT where the foot was displaced forward relative to

CM, a delay or decrease in the rate of angular displacements was observed, which resulted in a phase lag in the trajectories. This phase shift was evident for perturbations during the initial step (PT-HO) or during the second step (PT-HS). In the frontal plane, the change in the trajectories of hip angular displacements, trunk and shank segment rotations which occurred in response to BT and FT relative to NPT was a phase shift, with no significant change in magnitude. For BT, an initial increase in rate of angular displacement at hip and segment rotation at the trunk and shank was observed, which resulted in a phase lead in the trajectories. For FT, a delay or decrease in the rate of angular displacements at hip and segment rotation at trunk and shank was observed, which resulted in a phase lag in the trajectories. This phase shift was evident for perturbations during the initial step (PT-HO) or during the second step (PT-HS).

These findings were highly consistent with the view that corrective muscle responses are organized into locomotor like patterns, and that the process organizing balance adjustments and locomotor activities are integrated.

5.2. CORRECTIVE RESPONSES IN SAGITTAL PLANE:

The results of this study were highly consistent with the previous study of Riediger (1999), where the change in trajectory of angular displacement, trunk segment rotation, and CM(F) displacement that occurred in responses to BT and FT relative to NPT was a phase shift with no significant changes in pattern or magnitude. This phase shift was evident during perturbations at heel-off and at mid-swing of the initial step. The present study extended these results to include start of single support phase of the second step. Similarly, the same effects of sudden platform displacements on the trajectories of angular displacement were observed in the present study as were the changes to trunk segment rotation and CM(F) displacement during single support phase of the second step. Interestedly, the trajectories of CM(F) also showed the phase shift in response to BT and FT relative to NPT. The phase shift in CM(F) was observed at both PT-HO and PT-HS. This was true for both BT and FT. These findings suggest that the motor strategy used to restore balance is not dependent on the point in the gait cycle where the disturbance occurred. In the present study the type of disturbance was the same in PT-HO and PT-HS. Taken together, this would be consistent with the view that the type of disturbance, not the period of gait cycle determines the type of the motor strategy which is used to restore the balance.

A number of studies on the healthy young adults have examined the corrective balance responses to perturbations during walking in the sagittal plane (Nashner, 1980; Dietz et al., 1986; Eng, Winter & Patla, 1994, 1997; Tang et al., 1998). With exception of one of the study of Eng et al. (1994, 1997), these studies have focused on muscle EMG analysis. All these studies showed a change in pattern of EMG responses when disturbances are presented at different periods of the gait cycle. In particular, duration and the sign of excitatory or inhibitory responses were specific to the period of gait in which the disturbance was presented. However, the relative segment position or body posture is different in different periods of the gait cycle, velocity of segment motion is different in different periods, and in some cases direction of rotation is different in different periods. For example, the joint angular position and segment position in the current study were different during PT-HO and PT-HS. If the body is in a different position and moving at a different rate or direction, the muscle forces required to restore stability would be different.

Previous studies on corrective balance responses to perturbed walking (Nashner, 1980; Dietz et al., 1986) only examined distal leg muscles. The activation of these muscles are required to restore the relationship of CM(S). In addition to the early activation of TA and BA, observed by Nashner (1980), and Dietz et al., (1986). Tang et al. (1998) showed consistent and early responses in some thigh muscles, such as RF and BF. These corrective responses would be required to restore balance during disturbed walking. The results showed that the onset latencies of proximal muscles (gluteus medius and rectus abdominus) were much longer than distal muscles and response frequencies of the proximal muscles were much less than distal muscles (tibialis anterior and rectus femoris). So, their main conclusion was that proximal muscles are not as important to the compensatory response than distal muscles.

Our findings are not consistent with this view. All muscles showed early corrective responses including proximal muscles. Proximal muscles operating in sagittal and frontal plane showed relatively early onsets and had high occurrence frequencies. Although there was no passive effect in frontal plane to sagittal directed disturbances, the corrections to restore balance required significant changes in trajectories of CM(S), hip angular displacement, and trunk and shank segments rotation in the frontal plane. During BT and FT, short latency corrective responses were observed consistently for AB and AD on the stance limb. These responses would act to assist in the producing the necessary changes in hip angle to compensate for changes in lateral acceleration of CM(S). These showed that the proximal muscles are important especially ones controlling M-L forces set-up by A-P corrective responses. The significant changes in the trajectories of hip, knee angular displacement, trunk segment rotation in the sagittal plane provided also the evidence of the important role of these proximal muscles in sagittal plane. In our study, corrective responses not only at the ankle, but also at the knee, hip, and trunk segment on both the stance and swing limbs in both sagittal and frontal planes were observed.

The early responses in the kinematic trajectories of hip, knee angular displacements and trunk segment rotations showed the important role of proximal muscles. These early kinematic changes would require early activation of proximal muscles. Thus, all muscles are important for the control of whole body balance to modulate disturbances.

Tang et al. (1998)'s study presented disturbance in double support phase of walking, such as early stance, mid-stance, and late stance, not single support phase, similar to the present data. However, they only translated one side of the body. Thus it is likely that theirs is a small or weaker disturbance than ours that it does not require proximal muscles for regulation or restoration. Besides, Tang et al. (1998)'s study was based on EMG analysis alone. Without the kinematic data, the passive mechanical effects on all angular displacements, CM(S), or CM(F) can not be determined.

The other main conclusion in Nashner (1980) and Tang et al. (1998)'s study was that primary sensory input for triggering the corrective balance reaction is muscle inputs from changes in ankle joint displacement (i.e. BT versus NPT condition). This statement was very limited. As we describe before, no one single source of sensory information can define the type and direction of the balance disturbance. Multiple sources of sensory inputs (including proprioceptors spanning all body segments) are required to determine position and motion of CM relative to the foot (the base of support) and consequently to define the type and direction of the balance disturbance.

Nashner (1980) proposed a model to account for regulation of unexpected disturbances during walking. The model stated that the phasic activation of the movement generator for locomotion, via feedback signals from peripheral afferents could produce parametric changes in stepping movement to restore total body stability and maintain the forward progression of the body. He hypothesized that the processes organizing balance adjustments and locomotor activities were closely related or integrated. He also speculated that together with the biomechanical properties of the limb segments, the short latency corrective muscle contractions indicated which parameters of the stepping movement would be modulated during unexpected disturbance. There are a number of modifiable parameters within the stepping movement generator, i.e. rate, duration and magnitude.

The present results and those of the previous study (Riediger, 1999) demonstrated that the change in trajectories of angular displacements, trunk segment rotation, and CM(F) displacement which occurred in response to FT and BT relative to NPT was a phase shift with no significant change in pattern or magnitude. This phase shift was evident for perturbations during heel-off and midswing of the initial step and just after heel off in the second step. For BT where the foot was displaced backward relative to CM, we observed an initial increase in rate of angular displacement at the ankle, knee and hip, which resulted in a phase lead in the trajectories. For FT where the foot was displaced forward relative to CM, we observed a delay or decrease in the rate of angular displacements, which resulted in a phase lag in the trajectories. The corrective response was a phase shift with no change in pattern or quality of the kinematic trajectories. These responses occurred on both sides of the body.

Because of the passive component observed on the stance side, it was difficult to determine the timing of the kinematic changes of the real corrective responses on this side. But the timing of the earliest changes in the trajectories of angular displacement or segment rotation could be determined. A paired T-test to examine the magnitude difference of each time point in the waveforms of NPT, BT and FT (Table 1) was performed. From this analysis, we knew if and where a significant divergence in the trajectories of the kinematic waveforms between BT, FT, and NPT occurred. Some of the ranges were earlier than 200 ms which is a very short time to get a response. For example, during PT-HS, the divergence in right knee (stance side) angular displacement between BT and NPT occurred at 140-250ms, between FT and NPT occurred at 420-600ms. The knee flexion was earlier in BT, later in FT relative to NPT condition. For BT, early onset latency of HA (100ms) may assist the early knee flexion compared to NPT condition. Relatively greater onset latency of HA (140ms) may result in the delay knee flexion for FT. The muscle activity seems to be organized into locomotor like patterns. The onset latency of corrective responses and kinematic changes are relatively short latency. These findings were highly consistent with the statement that the early muscle responses are organized into locomotor like patterns, and that the process organizing balance adjustments and locomotor activities are integrated.

5.3. CORRECTIVE RESPONSES IN FRONTAL PLANE:

Winter et al. (1996) proposed two different postural control mechanisms for sagittal and frontal plane. Sagittal or A/P balance is maintained by ankle (plantar/dorsiflexor) control, whereas frontal or M/L balance is maintained by hip (abductor/adductor) control. Their study pertained to normal gait. It based on the steady state over ground walking with no perturbation.

The present study examined trajectories of CM(S) in ML plane for the sagittal directed perturbations in non-stead state locomotion. The differences in trajectories of CM(S) displacement between BT, FT, and NPT conditions were significant. During PT-HO, the divergence of CM(S) trajectories occurred at the time of early right stance phase when the CM was moving towards the right side of the body. During PT-HS, the time to divergence in CM(S) displacement occurred at the end of right stance when the CM was moving towards the left side of the body. The direction of divergence was opposite for BT and FT relative to NPT. For PT-HO, the effect of BT was a significant decrease in duration and magnitude of the right shift in CM(S) displacement and consequently an earlier time to start of the leftward shift in CM that would occur during right mid-swing. The opposite effect was observed for FT. During PT-HS, the effect of BT was decrease in duration and magnitude of the left shift in CM(S) displacement, the opposite effect was observed for FT. For PT-HS, there was no significant magnitude difference between BT and NPT condition, but there was a 2 cm difference between FT and NPT. The significant early changes in trajectories of CM(S) displacement in the frontal plane were required, during the corrective balance reaction to AP directed disturbances.

A significant divergence of trunk segment rotation trajectories between BT, FT, and NPT occurred, which began 100-200 ms after the onset of platform motion. During PT-HO, the effect of BT was earlier trunk lateral to right side rotation and earlier following medial to left side rotation. The effect of FT was later lateral to right side rotation and later following medial to left side rotation. There was no significant magnitude difference between BT, FT and NPT. These early corrective trunk segment rotations would contribute significantly to control of CM(S) displacement in the frontal plane. For BT, the earlier lateral trunk rotation in the stance phase and following earlier medial rotation resulted in the decrease in duration and magnitude of the right shift in CM(S) displacement and consequently an earlier time to start of the leftward shift in CM that would occur during right mid-swing. During PT-HS, there was no trials effect on time or magnitude to peak trunk rotation.

As with flexion/extension movements in sagittal plane, a similar phase shift in the trajectories of hip joint abduction and adduction between BT, FT, and NPT was observed. The time of initial significant divergence in trajectories of hip angular displacement was about 200-250 ms after the onset of platform motion. The effect of BT was a significant lead in the trajectories of hip angular displacement relative to NPT condition. The opposite effect was observed for FT. We did not observe the magnitude significant changes in these trajectories. These differences would related to the control requirements of CM(S) displacement. For PT-HO, the effect of BT was a significant decrease in duration and magnitude of the right shift in CM(S) displacement and consequently an earlier time to start of the leftward shift in CM(S) that would occur during right mid-swing. The opposite effect was observed for FT. For PT-HO, the effect of BT in left side hip was significant earlier adduction while in left stance phase and in right side hip was significant earlier abduction in late right stance phase, relative to NPT condition. For BT, the earlier left hip adduction in left stance phase and right hip abduction in right stance phase resulted in the decrease in duration and magnitude of the right shift in CM(S) displacement, relative to NPT condition. The opposite effect was observed for FT. For PT-HS, the effect of BT was a significant decrease in duration and magnitude of the left shift in CM(S) displacement and consequently an earlier time to start of the rightward shift in CM(S) that would occur during right mid-swing. The opposite effect was observed for FT. For PT-HS, the effect of BT in right side hip was significant earlier abduction in right stance phase and in left side hip was significant earlier adduction while in late left stance phase, relative to NPT condition. For BT, the earlier right hip abduction in right stance phase and left hip adduction in late left stance phase resulted in the decrease in duration and magnitude of the left shift in CM(S) displacement, relative to NPT condition. The opposite effect was observed for FT.

Corrective responses were consistently observed in AB and AD. The responses in these muscles were of the short onset latencies. The range of onset latency was from 73 ms to 254 ms. During PT-HO, a backward translation elicited short latency responses in both left AB and AD in 110 ms. Here we thus see synchronous activity of agonist and antagonist. From the kinematic perspective, we observed earlier left adduction, then right abduction, relative to NPT condition. This finding confirmed that the body segment movement can not be predicted by EMG analysis information alone. We need more information about other forces. As we described above, the stability of body segment or each link is achieved by maintaining a balance of external and internal forces acting at each joint. External forces are gravity-dependent forces, motion-dependent forces, and ground reaction forces. Internal forces are muscular forces, bony contact forces, and visco-elastic forces of tissues spanning a joint (Putnam, 1991). Body segment kinematics arises from at least four different types of forces. Muscular force is certainly one of them, and it contributes to dynamic stability of the system and to the movement. We may assume greater joint stiffness in the stance side when we observed the synchronous activity of agonist and antagonist in this side. The greater joint stiffness might be associated with greater stability for the purpose of dealing with the large disturbing forces produced by sudden support surfaces translation. Synchronous activity of agonist and antagonist was also observed during PT-HS for BT, and during PT-HO for FT.

Another main EMG finding of this study is that we observed more inhibitory responses during FT than during BT. During FT, we observed a phase lag or delay in the trajectories of angular displacements. A decrease in level of muscle activity relative to NPT condition would be required to achieve a delay in rate of angular displacements. Conversely, during BT, we observed a phase lead in the trajectories of angular displacements. An increase in level of muscle activity relative to NPT condition would be required to increase rate of angular displacements.

Mallroy and Maki (1996) examined age-related differences in spatial and temporal characteristics of compensatory stepping during large and fast FT and BT perturbations. The amplitude of platform translations was greater than the current study. The control of lateral stability in stance as measured by variability in lateral center of pressure (CoP) has been demonstrated to be a better predictor of falls in the elderly than A/P stability. Rapid onset and relatively short duration steps were observed in nearly all trials of AP support surface translation. One common feature of steps was that they were directed laterally. This was observed more in elderly than the young subjects. Our results showed early divergence in trajectories of lateral trunk segment rotation and hip abduction/adduction between NPT and the platform translations. This was accompanied by direction-specific changes in trajectories of CM(S) displacement. This is consistent with the plots of Fz and Fx, which showed earlier foot off for BT and delayed for FT, relative to NPT. For BT, the direction of Fx was leftward for BT during PT-HO. All this supports the conclusion that early corrective responses acting in the M-L plane are required A-P directed perturbation to regulate lateral acceleration of CM during walking.

5.4. CONCLUSION:

The present results and those of the previous study (Riediger, 1999) demonstrated that the change in trajectories of angular displacements, trunk segment rotation, and CM(F) displacement which occurred in response to FT and BT relative to NPT was a phase shift with no significant change in pattern or magnitude. This phase shift was evident for perturbations during heel-off and midswing of the initial step and just after heel off in the second step. These responses occurred on both sides of the body. Corrective muscle responses were observed in all muscles. There was a definite trend for EMG responses to BT to be excitatory while those for FT to be inhibitory. In the sagittal plane, the change in trajectories of angular displacements, trunk segment rotation, and CM(F) displacement that occurred in response to FT and BT relative to NPT was a phase shift with no significant change in pattern or magnitude. In frontal plane, the change in the trajectories of hip angular displacements, trunk and shank segment rotations which occurred in response to BT and FT relative to NPT also was a phase shift, with no significant change in magnitude. This phase shift was evident for perturbations during the initial step (PT-HO) or during the second step (PT-HS).

These findings were highly consistent with the statement that the early muscle responses are organized into locomotor like patterns, and that the process organizing balance adjustments and locomotor activities are integrated.

Our current information provides the baseline data for the future studies on age-related corrective responses during unexpected walking in the frontal plane.

5.5. LIMITATION:

We assumed this study represented the true corrective balance responses people might make during unexpected perturbation during their daily life. But, there were a couple of limitations in this study. First, the results were influenced by the subjects' experiences after several trials. Secondly, the sample size was small. They might not represent the true healthy active adult population.

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Appendix A

UNIVERSITY OF MANITOBA

FACULTY COMMITTEE ON THE USE OF HUMAN SUBJECTS IN RESEARCH

NAME: Dr. Tony Szturm

OUR REFERENCE: E94:277

DATE: February 14, 1995

YOUR PROJECT ENTITLED:

Analysis of Sensory and Motor Processes Regulating Postural Stability During Initiation of Locomotion.

HAS BEEN APPROVED BY THE COMMITTEE AT THEIR MEETING OF:

Approved by Dr. Gordon Grahame on behalf of the Committee on February 7, 1995.

COMMITTEE PROVISOS OR LIMITATIONS:

Approved as per your letter dated January 9, 1995.

You may be asked at intervals for a status report. Any significant changes of the protocol should be reported to the Chairman for the Committee's consideration, in advance of implementation of such changes.

****THIS IS FOR THE ETHICS OF HUMAN USE ONLY. FOR THE LOGISTICS OF PERFORMING THE STUDY, APPROVAL SHOULD BE SOUGHT FROM THE RELEVANT INSTITUTION, IF REQUIRED.**

Sincerely yours,

Gordon R. Grahame, M.D.,
Chairman,
Faculty Committee on the Use of
Human Subjects in Research.

GRG/11

TELEPHONE INQUIRIES:
789-3255 - Lorraine Lester