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Postural coordination during self-perturbed standing

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Abstract

The central nervous system employs control mechanisms in response to both external perturbations and self-perturbations in order to maintain stable upright stance. Previous studies on self-perturbation tasks involving arm movements have examined postural patterns and the relationship between arm movement and the body's center of pressure: however, the coupling within and between the components of the lower limb have yet to be investigated. To examine these issues ten healthy subjects swung their arms inphase and antiphase in the anterior-posterior direction to a metronome at frequencies of 0.67, 0.75, 1.0 and 1.5 Hz. Our results show differences in the phase relationships, joint amplitudes and sway patterns between inphase and antiphase swinging. These findings provide insight into the existence of separate control mechanisms for the two swinging modes, and therefore highlight task-specific control mechanisms for the lower limbs during self-perturbations.

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CHAPTER I

Introduction

1.1 Introduction to postural control

The Central Nervous System (CNS) contains billions of neurons and neuronal connections as well as hundreds of muscles and joints, and therefore an infinite number of possible muscle activation patterns that can be utilized to accomplish a task (Turvey, 1990). Thus, all human movements require coordination across many degrees of freedom. In order to simplify the problem of motor control, it has been hypothesized that the CNS organizes degrees of freedom required for a movement into synergies (Kelso, Putnam, & Goodman, 1983). The idea of coordinated structures or synergies (defined as functional groupings of structural elements that are temporarily constrained to act as a single unit) was first introduced in a seminal work by Bernstein (1967).

A major question regarding the function of the human postural control system is how the CNS controls the numerous associated joints and musculature to keep the body standing. The ability to stand and walk is quintessential to human movement and survival; humans are among many mammals that can maintain a normal gait without falling or losing their balance. When forces act on the body, perturbing it from a stable position there exist control mechanisms and recovery strategies coded within the CNS that enable the body to remain stable and prevent an individual from falling.

Posture is defined as the geometric relationship between two or more segments of the body. This relation is expressed in terms of joint angles between segments; for example, the ankle and knee angles describe the posture of the leg (Horak & MacPherson, 1996). In contrast, balance is the equilibrium that results from matching torques either by anticipating or reacting to the effects of postural disturbances (Balasubramaniam & Wing, 2002). Postural sway is typically measured from changes in centre of pressure (CoP), where CoP is defined as the point location of the sum of the vertical reaction forces in the anterior-posterior (AP) and medio-lateral (ML) directions (Balasubramaniam & Wing, 2002). When faced with postural disturbances, adjustments are made by the CNS. Consequently the CNS scales an appropriate response (i.e. muscle force) to counteract the forces acting on the body (Horak, Henry, & Shumway-Cook, 1997). The CNS does this by controlling the CoP to shift in phase with the body's center of mass (CoM), which in turn keeps the CoM within the body's base of support (Winter, Prince, Frank, Powell, & Zabjek, 1996; Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998).

Postural control is a complex perception-action task that involves a culmination of effort from physiological, perceptual, biomechanical and goal related factors acting in conjunction with the body's major muscle groups (Riley, Balasubramaniam, Mitra, & Turvey, 1998). This problem becomes accentuated when the body is perturbed during standing. In fact, much of the perturbations that influence body equilibrium are the product of voluntary movement. Whether reaching for an object or swinging ones arms during walking or cross country skiing, voluntary arm movements can perturb or act to maintain body equilibrium when standing in an upright position. Changing the magnitude of the perturbation produces interacting torques throughout the body, which in turn affects the postural response.

Postural reflexes in response to platform perturbations have been shown to have latencies in the order of 100ms (Nashner & Cordo, 1981). These postural reflexes are more complex than typical spinal reflexes because they adapt to contextual changes and are organized sequentially over groups of muscles (Horak, 1996). Two prominent neuromuscular control strategies for postural control are the ankle strategy and the hip strategy (Nashner & McCollum, 1985). Muscles spanning these joints are organized in synergies to control action by minimizing the number of active muscles and consequently, neural decision making to return the body to equilibrium. These synergies have been shown to be both directionally specific (Winter et al., 1996) and also independently modulated in a task specific manner (Balasubramaniam, Riley, & Turvey, 2000). Other studies have focused on the postural patterns and phase relationships between the hip and ankle that functionally emerge based on constraints of the task (Bardy, Marin, Stoffregen, & Bootsma, 1999)

When studying balance, recordings are typically taken from movements of body segments and ground reaction forces and forces. Balance can be perturbed experimentally in a number of manners: by applying forces in a predictable or an unpredictable way through platform perturbations (i.e. Nashner & McCollum, 1985), by having subjects perturb their own posture by displacing their arms (Balasubramaniam & Wing, 2002), or by having subjects track or look at objects in their visual field (Bardy, Oullier, Lagarde, & Stoffregen, 2007).

Work investigating self-perturbations most often focuses on the postural responses associated with arm-movements. Studies investigating CoP changes in response to different arm swinging frequencies have found evidence that shows CoP displacements under each foot exhibit frequency-related changes; more specifically, at fast frequencies, CoP moves in a direction opposite to the arm movement (Ustinova, Goussev, Balasubramaniam, & Levin, 2004). This suggests that perhaps the CoP shifts backwards with the centre of mass to compensate for (or in advance of) forward arm motion. Other studies have shown that during a self-perturbating arm swinging task, one of two postural patterns between the lower limbs and arm emerge: a hip-shoulder inphase mode at low swinging frequencies and an ankle shoulder inphase mode at higher arm swinging frequencies (Abe & Yamada, 2001). While postural coordination in the lower limbs has been investigated in response to other suprapostural tasks (any task performed in addition to maintaining posture) (i.e. Bardy et al., 1999), the coupling within and between the elements of the lower limbs has not been investigated during continuous, rhythmical arm movements while standing.

The bulk of this section will focus on theoretical approaches and experimental paradigms for the study of postural control, while also highlighting the contribution of the current study to this area of research.

1.2 Theoretical approaches to the study of postural control

There are two main approaches to the study of postural control. The neuromuscular approach (Nashner & McCollum, 1985) focuses on the idea that there exist two discrete strategies used by the nervous systems to control posture. These strategies are a consequence of basic coordination patterns of muscle activity that are controlled by neural activation patterns. More recently Bardy et al., (1999) applied the pattern dynamics approach, which focuses on the idea that multiple coordination modes exist and their emergence is a self-assembling phenomenon.

1.2.1 Neuromuscular approach

Nashner and colleagues (Nashner & McCollum, 1985; Horak & Nashner, 1986) have proposed that the maintenance of stance relies on a small number of postural strategies or organizational schemes, which are rooted in the neuromuscular system. These strategies depend on muscular synergies or patterns of muscle activation about the ankles, hips, and knees. Furthermore, these muscular synergies are activated by the CNS in response to the constraints of a given task. During upright stance, the knees control primarily the vertical position of the centre of mass, whereas the ankles and hips control the horizontal position. The ankle strategy moves the body about the ankle joint although weak hip movements are observed. Conversely, the hip strategy involves rotations about the hips, accompanied by ankle rotations in the opposite direction. The idea of ankle and hip strategies has been widely accepted because it succeeds in reducing the dimensionality of the postural system (Bardy et al., 1999). Ankle and hip strategies compress a high dimensional postural space with many degrees of freedom into a low dimensional space requiring control of only a few degrees of freedom. Controlling only a few degrees of freedom reduces the variability and uncertainty in the system and simplifies the integration of sensory information (c.f. Bernstein, 1967; Turvey, 1990). The hypothesized ankle strategy has served as the basis of many recent postural control models (e.g. Horak, Nashner, & Diener, 1990).

One hypothesis that simplifies the problem of coordinating so many degrees of freedom was proposed by Winter et al. (1998); this hypothesis stipulates that the body acts like an inverted pendulum during quiet standing and is stabilized primarily by active control of the ankle joint with only passive control about other joints. If postural sway is in fact localized about the ankle joint, the position of other locations in the body (i.e. CoM, head) should be trivially related to each other. Directionally speaking, Winter et al. (1996) found that the hip joint's abductors/adductors produce torques that limit sway in the frontal (M-L) plane and the ankle joint's plantar flexors limit sway in the sagittal (A-P) plane.

Furthering these last findings, Balasubramaniam et al. (2000) reported that AP and ML sway could be independently modulated to facilitate the performance of supra-posture precision tasks and is therefore task-specific.

There are problems however, with inverted pendulum models since more recent studies have suggested these models are largely an oversimplification of postural control (Bardy et al. 1999, Bardy, Oullier, Bootsma, & Stoffregen, 2002; Alexandrov, Frolov, Horak, Carlson-Kuhta, & Park, 2005; Creath, Kiemel, Horak, Peterka, & Jeka, 2005). Additionally, studies have shown that there is consistently some degree of rotation about both the hip and ankle joints (Horak & Nashner, 1986; Horak, Nashner & Diener, 1990). Furthermore, posture is rarely controlled solely for its own sake, in the context of quiet standing, but has to be maintained while performing a variety of suprapostural tasks. In wake of these limitations new theoretical paradigms have developed in postural control research.

1.2.2 Pattern Dynamics approach

The pattern dynamics approach to postural coordination is consistent with the dynamical systems approach to movement coordination, which relies on the existence of multiple coordination modes and their self-assembling emergence towards stable patterns (Kelso, 1995).

The idea of a set of oscillators self-assembling and emerging to form stable patterns was first brought to light through experiments by Von Holst (1973). In de-cerebrated fish, Von Holst found that fins oscillated at preferred frequencies in isolation, but at a common frequency while swimming. He termed the tendency for each fin to oscillate at its own preferred frequencies its 'maintenance' frequency and the tendency of a fin to be attracted to another fin's frequency the 'magnet effect'. These effects can also be observed at the level of the fin ray (which composes the fin). Thus coordination is not specific to any particular degree of freedom and is present at multiple levels.

In a series of experiments by Kelso (1984), similar results were observed in human interlimb coordination. Kelso had subjects tap their two index fingers to a metronome beat and observed the patterns between the two fingers. It was found that subjects could stably produce two basic movement patterns with their fingers: inphase (with the fingers moving in unison in time and space) and antiphase (with the fingers moving in an opposite fashion). To calculate the phase relationship between the two fingers, a point estimate of relative phase (i.e. latency of one finger with respect to the other finger based on peak-to-peak displacement) was used. Coordination patterns can therefore be described by phase difference between the two segments or joints of interest (i.e. relative phase between the two index fingers).

$$\Phi\text{rel} = \Phi\text{effector1} - \Phi\text{effector2}, \qquad [1]$$

It was found that as tapping frequency increased, subjects tapping antiphase spontaneously switched to an inphase coordination pattern. This transition is not reversed if finger-oscillating frequency is decreased. Conversely, when subjects start tapping with the fingers inphase, no frequency dependent transitions to an antiphase coordination pattern were observed.

These experimental observations on phase transitions during finger tapping led to the development of theoretical explanations to account for this behaviour. Haken, Kelso, and Bunz (1985) developed a mathematical model (HKB) that describes the stability of both inphase and antiphase movement patterns. They utilize a potential function, with attractors at 0° and 180°, corresponding to the stable movement patterns previously identified:

$$V(ø)= -a\cos(ø) -b\cos(2ø), \qquad [2]$$

where ø is the relative phase, and *a* and *b* are coupling parameters (i.e. *b/a* is a coupling ratio between effector frequencies) (Haken et al, 1985). Since relative phase describes the spatial order of the system, it has been identified as the order parameter, (Turvey, 1990). Relative phase is also a cyclic variable and therefore any equation of motion has to be written in terms of periodic functions (i.e. sines and cosines). Furthermore, changing the tapping frequency as dictated by a metronome has been identified as the control parameter (Turvey, 1990). By changing the coupling ratio, which is inversely related to movement frequency, changes in the landscape are observed. The landscape shifts from a bi-stable landscape with attractors at ±180° and 0°, to a mono-stable landscape with only one attractor about 0° present (Swinnen, 2002). Once in this 0° inphase mode, movements are very stable and changes in the landscape will not cause further transitions.

The HKB model was successful in providing an accurate description of the experimental observations observed by Kelso (1984). This dynamical systems approach suggests that the components of a system self-organize or emerge based on functional units of action (Turvey, 1990) and aren't based on relationships among mechanical, muscular, or neural activity (Bardy et al., 1999).

The idea of using relative phase to investigate the presence of phase transitions during multi-segmental postural coordination was first examined by Bardy and colleagues in 1999. During multi-segmental postural control most movements occur about the hip and ankle with both exhibiting an infinite number of positions. Therefore, the dimensionality of the system can be reduced using relative phase as a composite variable to incorporate positions and velocities of both ankle and hip. The relative phase between the two joints is able to compress four variables (i.e. position and velocity of both joints) into one collective variable. During a suprapostural tracking task where subjects had to follow periodic movements of a display target with the head, this group found that as target amplitude increased, the relative phase between the ankle and the hip shifts from an inphase (more stable) to an antiphase (less stable) pattern. These patterns of coordination emerge spontaneously in suprapostural tasks depending on task and environmental constrains. Therefore emergent postural coordination modes can potentially describe coordinated movements of the segments of the postural control system better than discrete neural control strategies.

The inphase and antiphase patterns described by Bardy et al. (1999) differ slightly from those of Kelso (1995). Whereas Kelso found an inphase pattern about 0°, Bardy observed inphase coordination in the postural system to occur close to 20° with the hip joint lagging the ankle joint. They argued, that while it is easier to couple two fingers from different hands because they are unconstrained, the postural control system is highly concatenated, and therefore, movements of the trunk and hip work in opposition to maintain posture.

Results similar to that of Bardy et al. (1999) were also found when the tracking task was replaced by a simple task whereby participants watched a visual display (Oullier, Bardy, Stoffregen, & Bootsma, 2002). This result is of importance because it highlights the presence of adaptive postural coordination patterns simply during looking.

A dynamical systems approach to postural control has continued with studies by Creath et al. (2005) who used spectral coherence to reveal two modes of coupling between variations of leg and trunk segment angles. In-phase coupling was found below 1 Hz, whereas a shift to anti-phase coupling occurred above that value. This result suggests that ankle and hip strategies may exist simultaneously as excitable modes of coordination between the leg and trunk segments during quiet stance and refutes the argument that they are two strategies at the extremes of a continuum. These results support previous data in perturbation experiments (Alexandrov et al., 2005) that have found that postural strategies co-exist with varying power depending upon task, environmental and biomechanical constraints. This study revealed that the ankle, knee, and hip joints are coordinated as different eigenmodes, or as independent combinations of all three joints defined by a principal component analysis. Furthermore, the ankle eigenmode involves predominately ankle motion, with motion of the knee and hip making important contributions to the control of upright stance as well.

More recently Martin, Cahouet, Ferry, and Fouque (2006) found that an inphase mode of coordination is the less expensive strategy in terms of minimizing variability of joint torque, while antiphase was more effective to maintain balance in suprapostural tasks. Authors have also begun to investigate the causes of the phase transitions during postural tracking tasks (Ferry, Cahouet, & Martin, 2007). A kinetic analysis showed that CoP position is a crucial parameter in determining the adopted coordination mode: AP displacements of CoP were near the edge of the supporting area just before transition. They also showed that CoP displacement was directly linked to the moment of force about the ankle and an antiphase mode of coordination presents a better distribution of muscular moment between the hip and ankle joints. Even as transition occurred the moment about the ankle was still greater than the moment about the hip. An antiphase mode of coordination is also better suited in keeping the feet flat on the ground while requiring less neural effort for a given amount of CoM acceleration and achieving high frequency oscillations with limited CoP displacement.

1.3 Postural Reponses to Perturbations

Postural perturbations can be defined as an abrupt change in conditions that displace the body position away from equilibrium (Horak et al., 1997). Perturbations can be sensory, vestibular, visual or somatosensory, leading to responses based on perceived instability, or they can be mechanical in nature, in which body position is actually displaced (Horak et al., 1997). A brief overview outlining methods and results from mechanical and self-perturbations is now introduced.

1.3.1 Mechanical Perturbations

The theoretical and experimental observations of Nashner and McCollum (1985) led to subsequent studies using a moving platform as a means to perturb standing subjects. Mechanical perturbations serve to physically reduce stability and cause displacement of the body's centre of mass toward the boundaries of the base of support. The most common way of doing so is to perturb the supporting surface, which displaces the base of support beneath the body's CoM.

Manipulating the frequency of the perturbing surface leads to different control strategies. The ankle strategy is observed during slow perturbations (i.e. frequencies under 0.2 Hz, Nashner, Shupert, Horak, & Black, 1989) and at sway amplitudes below 20° (McCollum & Leen, 1989). Conversely the hip strategy involves a combination of ankle extension and hip flexion or vice versa and is used in response to faster perturbations and greater sway amplitudes. In addition to manipulating the frequency of the support surface, manipulations to the support surface itself have been made and have also been found to isolate different control strategies. The ankle strategy is found when the support surface is firm, whereas the hip strategy has been found when the support structure is compliant or smaller than the feet, like the balancing beam in gymnastics (Horak et al., 1990). Furthermore, an additional mechanism exits when the perturbation is too large to overcome with the feet in place. Under these circumstances the CoM is projected outside the base of support provided by the feet, and participants often take a step to recover balance (Horak & Nashner, 1986). These control mechanisms are also adaptive, as studies have shown that with practice, postural strategies become more efficient in their responses to a repeatedly destabilizing stimulus (Horak, Diener, & Nashner, 1989).

Other studies have investigated the possibility of phase transitions when faced with a mechanical perturbation. Buchanan and Horak (2001) had participants stand on a displacing platform that increased in oscillating frequency. They found that the postural pattern in the lower limbs showed gradual transitions, as opposed to abrupt phase transitions observed in both tapping (Kelso, 1984) and other suprapostural tasks (Bardy et al., 1999). This shows that the CNS is able to modulate joint amplitude in a continuous manner, which in turn allows for gradual modulation of CoM amplitude regardless of the postural pattern produced.

1.3.2 Self Perturbations

Another way of inducing perturbations is to have subjects perform arm movements, such as arm swinging while walking or voluntarily reaching for on object, to displace the

CoM. As expected, these every day voluntary arm movements require compensatory adjustments in body equilibrium to keep the body in an upright posture. When body dynamics change due to arm movement, the posture control system in the lower limbs, hips, trunk and pelvis compensates in an anticipatory manner to ensure stable upright posture is maintained (Hodges & Richardson, 1999). This occurs because upper limb movements produce dynamic predictive and reactive forces acting on other joints of the body. Abe and Yamada (2001) examined lower limb coordination patterns in response to varying arm swinging frequencies. They found that an increase in arm swinging frequency causes a change in biomechanical properties of lower limb segments. More specifically, they found that at lower frequencies (below 40% max) postural coordination is controlled by the hip, hence a hip-shoulder inphase mode, while at higher frequencies (above 45% max) postural coordination is controlled by the ankle in an ankle-shoulder inphase mode.

In a similar arm-swinging task, Ustinova, et al., (2004) examined the relationship between arm trunk and CoP displacements in healthy and hemi-paretic subjects. The subjects performed both inphase and antiphase arm-swinging to investigate differences between the two modalities. They found that CoP displacements under each foot were not related to arm swinging during in-phase swinging at the preferred speed in healthy subjects but when speed of arm swinging was increased, the CoP moved in a direction opposite to the arm movement. It was thought that in healthy subjects, the trunk counterbalanced antiphase arm movements, while in hemi-paretics the trunk moved with the affected arm.

A more recent study investigated the effects of a complex bimanual multi-joint coordination task on posture (Forner-Cordero, Levin, Li, & Swinnen, 2007). In this study participants had to produce different inter and intra-limb coordination patterns about the arms and elbows at increasing frequencies. Results from this study show that increasing the complexity of coordination, more specifically producing an antiphase pattern between the elbow and the wrist across all frequencies leads to adverse effects on postural sway (increase in CoP amplitude) (Forner-Cordero et al., 2007).

Important findings have been made when examining the differences in the relationship between upper limb movement speed and reaction time for trunk and lower limb postural muscles. It is known that preprogrammed trunk muscle activity is associated with slower, natural movements and rapid upper limb movement (Hodges & Richardson, 1999). Upper limb movement parameters have a limited influence on the reaction time of the associated trunk muscle response indicating that these muscles control a different component of the postural disturbance than the lower limb postural muscles (Hodges & Richardson, 1999). Furthermore, when the reaction time of trunk and lower limb muscles was analyzed over a variety of speeds, the reaction time of the lower limb postural muscles was correlated with upper limb acceleration only at speeds greater than 70 – 90 degrees per second (Lee, Buchanan, & Rogers, 1987).

There has also been reported high variability with movements at slow speeds because the initiation of slow movement occurs with a slow increase in EMG amplitude compared to the large burst associated with rapid movement perhaps indicating an absence of anticipatory responses with slow movement (Horak, Esselman, Anderson, & Lynch, 1984; Lee et al., 1987). It has been suggested that passive mechanical forces from inertia and relaxed trunk muscles may be sufficient to resist the low forces associated with the slow movement condition (Lee et al., 1987).

1.4 How is posture controlled?

Postural reflexes, often termed 'automatic postural responses' (Horak et al., 1997), are longer than spinal stretch reflexes (40-50 ms), but shorter than voluntary responses (180-250ms) (Nashner & Cordo, 1981). These responses occur at a latency of 70–110 ms in the lower extremities (Nashner, 1977). Platform perturbation studies have shown that muscles closest to the spinal cord are not always activated first and in some responses the ankle is activated before trunk, indicating centrally programmed synergies can delay activation of proximal muscles to maintain a functioned spatio-temporal pattern (McCollum, Horak & Nashner, 1984).

1.4.1 Predicting Postural Disturbances

The ability to predict and anticipate perturbations in advance is essential for remaining in a stable upright stance. Postural responses counteracting the effects of perturbing forces have been found to be initiated prior to the onset of contraction of the muscles responsible for limb movement. Bouisset and Zattara, (1981) had subjects accelerate their arms in the vertical direction while recording signals from accelerometers on the lower and upper body muscles. They found presence of specific patterns of movements in the lower limbs and trunk in advance of arm raising. They identified these patterns of activation as anticipatory postural adjustments (APA). These authors hypothesized that during the arms movements, predictive and reactive forces of the movement of the shoulder cause the body's centre of mass to be displaced and accelerate backwards and downwards, and that APA's counter this force by shifting the CoM in the opposite direction. This was confirmed by further accelerometric studies (Bouisset and Zattara 1987a, 1987b) in which the forces at the shoulder level were calculated and the forces and moments generated by the anticipatory postural adjustments prior to the movement onset were measured. It was found that anticipatory forces were directed in the direction opposite to the reaction forces associated with the arm movement and therefore served to minimize the postural disturbance caused by the movement. Furthermore, the dynamic predictive and reactive forces resulting from upper limb movement, in addition to acting in the opposing, direction, are actually equal in magnitude to the forces producing the movement. It has been further confirmed that as the body, along with its CoM move during an action, dynamic reactive and predictive forces are exerted on body, acting in the opposite direction to the movement (Lee et al. 1987). Studies using EMG techniques (Horak et al. 1984) have also found that the magnitude of APAs scale to the magnitude of the perturbation (i.e. the degree to which the arm movement disturbs posture).

Predictably, subjects are also able to make appropriate adjustments when the characteristics of a platform displacement are known. In line with this argument, subjects are able to shift CoM (Cordo & Nashner, 1982), change ankle moment responses (Horak, 1989), change the gain of postural muscle reflexes and decrease body displacement in a predictive manner. Further studies have shown that in order to maintain balance, all subjects exhibited early postural muscle activation, and tighter coupling of the center of pressure (CoP) to platform movements, and overall smaller CoP excursions (Bugnariu & Sveistrup, 2006). There also exists the possibility that reactive postural adjustments are made in response to available feedback, and occur later in the motor response (Massion, 1992). These findings highlight what is essential to maintaining upright balance: if we move our arms forward our body compensates by shifting our CoM backwards and imply that the CNS is able to predict the magnitude of the upcoming destabilizing arm movements and the destabilizing forces and is able to compensate for them in advance (Friedli, Cohen, Hallet, Stanhope, & Simon, 1988). Despite what appears to be quite a precise control over balance, there have been occasions where postural responses in the lower limbs and trunk have been found to over compensate for arm raising movements causing potential postural disturbances (Bouisset & Zattara, 1987b).

When investigating the anticipatory response of the trunk muscles associated with movement of the upper limb, studies show contraction of the erector spinae (ES) prior to upper limb flexion or contraction of the rectus abdominis (RA) preceding upper limb extension (Aruin & Latash, 1995). Regardless of the direction of the movement, contraction of transversus abdominis (TrA) has been found to be the first trunk muscle active, and even indicating contraction of TrA prior to the prime mover of shoulder flexion (Hodges & Richardson, 1999). These anticipatory movements serve to reduce postural disturbance due to a forthcoming movement and contribute to the general dynamic organization of balance.

1.4.2 The role of the cerebellum

One of the first experiments characterizing deficits of patients with cerebellar lesions was performed by Babinksi in 1899. He observed the movements of the patients, when compared to normal subjects, were absent of opposing hip and knee movements required to maintain equilibrium in response to upper trunk movements.

A critical function of the cerebellum is to combine actions of muscles to function synergistically in adaptive and coordinative movement control (Thach, 1998). Thach, Goodking and Keating (1992) also proposed that each of the three deep cerebellar nuclei have representative body maps, each coding for difference movement types and contexts, while the cerebellar cortex functioned to combine simple elements of movements into complex synergies.

Some of the most distinctive signs of cerebellar damage are loss of coordination of motor movement and disturbances to balance and gait including increased postural sway (as measured by CoP) and irregular responses to perturbations (Mauritz, Dichgans, & Hufschmidt, 1979; Dichgans & Mauritz, 1983; Earhart & Bastian, 2001). Lesions localized to the medial zone and flocculonodular lobe appear to have the most detrimental affects in postural tasks.

Lesions in these areas in cats and monkeys lead to changes in upright postural tone, difficulty walking and impairments in sitting and standing balance (Sprague & Chambers, 1953; Thach et al., 1992). Furthermore, these lesioned animals fall frequently backward and toward the side of the lesion (Chambers & Sprague, 1955a, 1955b) and demonstrate abnormal timing of relative limb movements during locomotion (Yu & Eidelberg, 1983). Damage to the anterior lobe is characterized by increased sway in the AP direction (high frequency, low amplitude) and increased inter-segmental movements (Dichgans & Mauritz, 1983). In response to backward platform displacements cerebellar patients showed hypermetric postural responses to surface displacements and impaired ability to adapt to predictable perturbations (Horak & Diener, 1994).

Cerebellar damage can also lead to the inability to predict and compensate for interaction torques (Bastian, Martin, Keating, & Thach, 1996). When faced with perturbations the role of the cerebellum is to coordinate movements about mutliple joints by controling agonist-antagonist muscle couplings thereby restoring equilibrium (Thach, 1998).

The cerebellum also sends predictive signals to prevent errors from occuring that would otherwise result from uncontrolled interaction torques (Thach, 1998). In order to achieve these predictions, internal forward models have been hypothesized to use an efference copy of the motor command to predict of the sensory consequences of the movement postulated (Wolpert, Ghahramani, & Jordan, 1995; Wolpert, 1997). The sensory prediction is then compared with actual sensory feedback of the movement and, can be used to filter sensory information. The cerebellum is involved in signaling the sensory discrepancy resulting from the forward model's comparison between the predicted and actual sensory feedback from movements (Blakemore, Frith, & Wolpert, 2001).

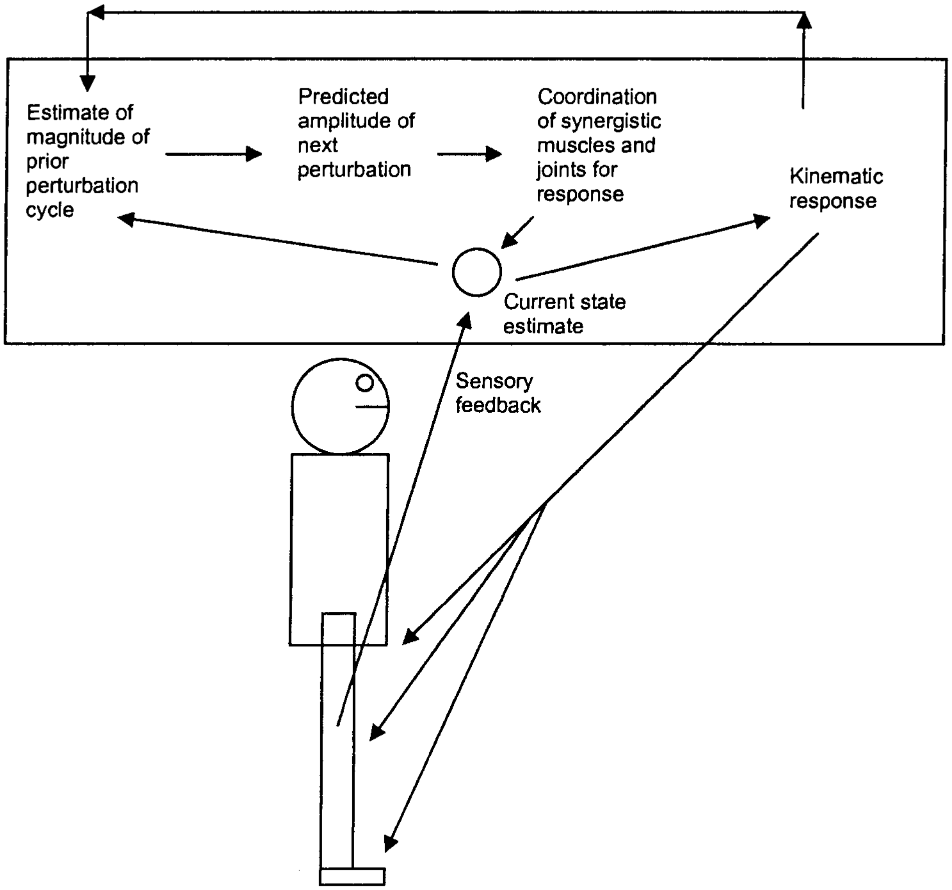

Figure 1. Schematic of a theoretical model that shows how compensation of CoM displacements occurs during self perturbations. The current state is constructed from the previous state estimate which represents a number of possible lower limb configurations. This estimate is then refined by using it to predict the current sensory feedback. The error between this prediction and the actual sensory feedback is used to correct the current estimate. Signaling the sensory discrepancy resulting from the forward model's comparison between the predicted and actual sensory feedback from movements involves the cerebellum (Blakemore et al., 2001). Additionally (although not depicted), in other models for tasks such as reaching, a Kalman gain filter uses sensory feedback and forward models to estimate the current state while compensating for sensorimotor delays to reduce error in state estimate (Wolpert & Gharamani, 2000).

1.5 Current Study

There has been little research investigating how the CNS compensates for or adapts to perturbations that are self-generated, as in the case of arm swinging. Here we look at coupling in lower limb joints and sway patters in response to voluntary and repetitive anterior-posterior inphase and antiphase upper limb perturbation at various frequencies. The present study is concerned with:

1) Determining whether increasing the frequency of the upper limb perturbation causes significant disruption in coordination of the lower limbs
2) Determining whether a different coordination pattern in the lower limbs exists for inphase and antiphase arm movements

Given the changes in the postural pattern observed in previous studies (i.e., Abe & Yamada, 2001; Bardy et al., 1999), we hypothesize that increasing arm swinging frequency will result in the emergence of different coordination patterns observed in the lower limbs. More specifically as swinging frequency is increased, increases in mean relative phase, relative phase variability and joint amplitudes will be observed. We also hypothesize that inphase and antiphase swinging present different problems for the CNS and therefore may require different patterns for postural coordination, with inphase swinging exhibiting a more coordinated and stable pattern with mean relative phase values and relative phase variability being closer to 0°. To investigate these issues we will examine sway and joint amplitudes as well as the relative phase between joints of the lower limbs. We will also apply a principal component analysis (PCA) to examine the relationship between these variables (lower limb joints). While relative phase has been a traditional measure of coordination between joints, PCA has recently been shown to have practical applications in human movement and posture by reducing the dimensionality of the data series, based on the mutual information between variables (Daffertshofer, Lamoth, Meijer, & Beek, 2004; Forner-Cordero, Levin, & Swinnen, 2005). This provides a more global measure of performance in a given task. Here we also apply a directional coupling analysis to the postural perturbation task in order to determine the leading oscillator between a pair of joints in the lower limbs.

CHAPTER II

Methods

2.1 Subjects

Ten healthy participants (5 male and 5 female) aged 20-23 years with no history of neurological or musculo-skeletal problems were tested. Subjects had a mean height of 166.48 ± 7.0 cm, and weight of 63.64 ±11.80 kg. All subjects gave their informed consent and the experiment was approved by the University of Ottawa ethical committee on testing human subjects.

2.2 Apparatus

Subjects wore minimally invasive light sensors (passive reflective spheres), which were comfortably affixed to bony landmarks on the left and right hip, knee and ankle as well as the index finger. The subject's clothing was affixed by tape inferiorly and superiorly to the marker to ensure the marker remained fixed on the desired landmark. An eight camera MX-40 Vicon™ motion capture system recorded movement kinematics from these sensors at 200Hz (frames/second) and two force plates (AMTI-6df 2000, AMTI-6df 1000) measured ground reaction forces which were used to calculate CoP dynamics under each foot. The movement frequency was dictated by an electronic metronome that produced a 1 kHz tone for 20ms.

2.3 Experimental conditions

Subjects stood shoeless in a comfortable upright stance with the feet a shoulder width's apart and with one foot on each force plate. Subjects were instructed to swing their arms forwards and backwards to an auditory cue dictated by a metronome, such that the beat occurred at the same point in their swing cycle. Subjects performed the arm swinging under two experimental conditions: inphase and antiphase (Figure 2). Each condition consisted of a very slow (0.67 Hz), slow (0.75 Hz), medium (1.0 Hz) and fast (1.5Hz) paced arm swinging frequencies, with five 30 second trials under each condition. Subjects were instructed to perform the movements, with an amplitude that would allow them to maintain swinging cadence.

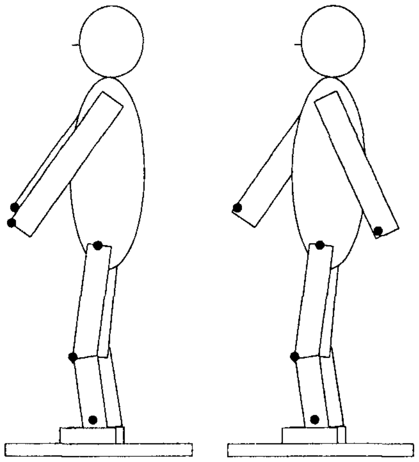

Figure 2. Experimental setup showing inphase movement (condition 1, left) and antiphase movement (condition 2, right), with marker placement.

Condition 1

Subjects swung their arms upwards and backwards at the same time so their arms were in phase with each other.

Condition 2

Subjects swung their arms upwards 180 degrees out of phase with each other such that as one arm was swinging forwards, the other was swinging backwards.

2.3 Data reduction and analysis

The 3-D kinematic data (from the sagittal, frontal and horizontal directions) and corresponding analog centre of pressure data from the force plates and metronome data was stored onto a conventional PC for reduction and analysis. Three dimensional marker positions for each frame and analog data were reduced in Excel.

Movement coordination was evaluated using the following dependent measures: mean relative phase and relative phase variability between joints of the lower limbs, and relative phase between the arm and CoP, joint, CoM and CoP amplitudes, directional coupling between joints of the lower limbs, and a principal component analysis (PCA) of the joints of the lower limbs.

The raw kinematic and force plate data were filtered using a 4th order bidirectional Butterworth filter with bandpass frequencies of 0.5Hz and 4Hz. The Matlab Hilbert transform function was used to compute mean relative phase values, relative phase variability and amplitudes of the lower limb joints (Rosenblum & Kurths, 1998). The Hilbert transform uses a Fast-Fourier transform and is used in calculating instantaneous attributes of a time series such as the amplitude and frequency (instantaneous frequency is the time rate of change of the instantaneous phase angle). Relative phase was computed between the joints of the lower limb (i.e. between the hip-ankle, ankle-knee, hip-knee of each leg), and between the arm and the CoP under the ipsilateral foot (arm-CoP) for movements in the AP direction, and were averaged across the left and right side. Joint amplitudes were defined as the amplitude of excursion of each marker in the AP direction and were averaged across the left and right side. The instantaneous amplitude of each of the lower limb joints are the amplitudes of the complex Hilbert transform. The amplitudes of the arms and CoP were also calculated via the Hilbert transform:

$$H(t) = \frac{1}{\pi} \int_{-\infty}^{\infty} \frac{x(\tau)}{t-\tau} d\tau , \qquad [3]$$

Where *H(t)* is the Hilbert variable, *t* is time and τ is a delayed time component of the Hilbert transform. The relative phase, or phase difference between two signals x(t) and y(t) can then be obtained by:

$$\mathrm{f}(\phi) = \arctan \frac{\mathrm{H_y(t)x(t) - H_x(t)y(t)}}{\mathrm{x(t)y(t) + H_x(t)H_y(t)}}, \qquad [4]$$

The circular mean and circular variance of the relative phase was then computed for each trial using CircStat, a circular statistics toolbox developed for Matlab (Berens, 2006). The absolute value of the mean relative phase was then computed.

In order to identify interdependencies of the lower limb joints a directional coupling analysis was used to investigate the coupling and the intensity of the interaction (see Rosenblum, Cimponeriu, Bezerianos, Patzak, & Mrowka, 2002 for a review). For example, if joint x is driven by joint y, then the phase angle of *x* depends on *y*. Predicting x from previous values of y is possible by taking into account the history of *y* (Rosenblum et al., 2002). Cross dependences, c, of phase dynamics of two systems, Φ_1 and Φ_2 are given by the following equation:

$$c_{1,2}^2 = \int \int_0^{2\pi} \left(\frac{\partial F_{1,2}}{\partial \Phi_{2,1}}\right)^2 d\Phi_1 d\Phi_2, \qquad [5]$$

where F represents periodic components or the Fourier series. The directionality index, d, is then given by:

$$d^{(1,2)} = \frac{c_2 - c_1}{c_1 + c_2}, \qquad [6]$$

A principal component analysis was used to reduce dimensionality of the data series based on the mutual information between variables being analyzed. The correlation matrix was composed of the positions of the six joints of the lower limb (left and right hip, knee and ankle) in the x, y and z direction. In the analysis of each condition (phase x frequency), the first eigenvalue measures the quality of the performance (Forner-Cordero et al., 2005), while the components of the first and second eigenvectors, or coefficient vectors, measure the importance and contribution of each joint on coordination performance (Forner-Cordero et al., 2005). Since the eigenvalues measure the percentage of variance of the data series explained by each principal component (PC) and coordination in the lower limbs was, for the most part in phase, the correlation matrix of the lower limb joints should have only one large eigenvalue to explain the global coordination performance. Additional eigenvalues indicate deviation from the coordination pattern and the noise components of the signals or the presence of a more complex pattern that simply cannot be prescribed with one eigenvalue (Forner-Cordero, 2007). For the CoP spectral analysis, directional coupling, principal component analysis the raw kinematic data was filtered with a 768 order FIR1 bidirectional filter with bandpass frequencies of 0.1 and 10.0 Hz.

Centre of Mass (more precisely, the horizontal projection of centre of gravity, CoG_h) in the AP and ML direction was estimated from CoP data collected from the force plates (Rougier, 2004). To calculate CoG_h, the CoP displacement time-series in the AP and ML directions are processed through a fast Fourier transform (FFT) to get a CoP spectrum. The obtained CoP spectrum is then multiplied through a CoG_h/CoP filter to give CoG_h spectrum. CoG_h/CoP filter is a mathematical low pass filter that expresses an amplitude ratio between CoG_h and CoP as a function of the movement frequency. An inverse FFT then returns temporal domain to CoG_h and amplitude can be measured.

To calculate net CoP amplitudes in the AP direction (used to calculate CoM), data collected from each force plate as well as the vertical reaction force was entered into the following equation (Winter et al., 1996):

$$\text{Net CoP}_{AP} = \frac{\text{LCoP}_{AP} \cdot \text{LF}_Z}{(\text{LF}_Z + \text{RF}_Z)} + \frac{\text{RCoP}_{AP} \cdot \text{RF}_Z}{(\text{LF}_Z + \text{RF}_Z)}, \qquad [7]$$

where LCoP and RCoP are the displacements of the left and right CoP's in the sagittal (anterio-posterior) direction, and LF_Z and RF_Z are the total vertical forces under the left and right foot respectively, as determined by each force plate. The same process was then used to calculate net CoP in the ML direction.

The power spectral densities of the CoP were calculated by processing the CoP signals through a FFT and then by multiplying the FFT signal by the signal of the conjugate FFT.

CHAPTER III

Results

First the dependent measures (mean relative phase, variability of relative phase and joint amplitudes) are shown for the independent measures concerning the lower limbs as an effect of mode, frequency and joint. Secondly CoP and CoM amplitudes are presented, and lastly a PCA and directional coupling analysis of the lower limbs are shown.

3.1 Mean Relative Phase

Mean relative phase between joints within each leg was calculated to give an indication of how tightly coupled or the degree of coordination between two joints. A 2 (mode) × 3 (joint pair) × 4 (frequency) repeated measures ANOVA revealed significant main effects of phase, F (1,49)=121.084, $p<0.001$ (Figure 3). This shows that mean relative phase values between joints in the lower limbs are higher during antiphase (M=34.451) arm swinging than during inphase arm swinging (M=7.687).

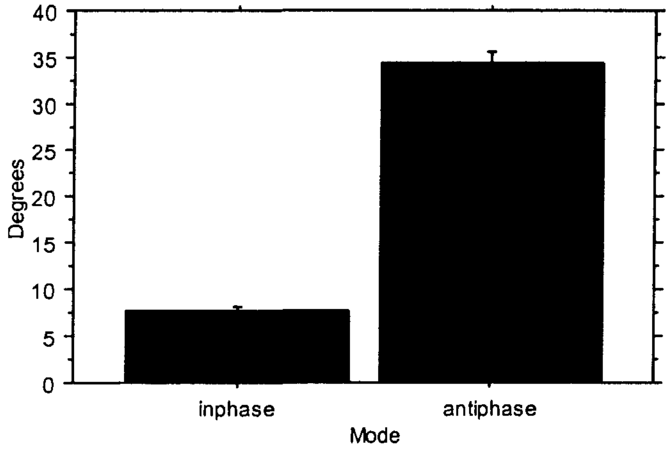

Figure 3: Mean relative phase values between joints of the lower limbs (ankle-hip, knee-hip, ankle-knee) for inphase and antiphase arm swinging. Error bars represent standard error of the mean.

A main effect of frequency, F (1, 3) = 2.850, $p<0.05$ was found when examining relative phase of the lower limb joints. This effect shows that, as a trend, mean relative phase values increase as a function of swinging frequency. However, Figure 4 demonstrates that Bonferroni-Dunn post-hoc testing revealed significant differences only between slow (M=17.733) and fast frequencies (M=24.914) ($p<0.001$).

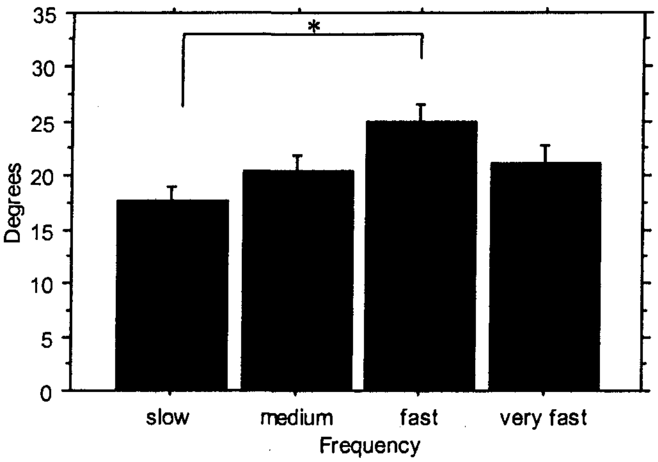

Figure 4: Main effect of mean relative phase values between joints of the lower limbs (ankle-hip, knee-hip, ankle-knee) for slow (0.67Hz), medium (0.75Hz), fast (1.0 Hz) and very fast (1.5 Hz) arm swinging. Error bars represent standard error of the mean. * $p<0.001$.

Lastly, analysis revealed a significant main effect of mean relative phase between pairs of joints F(1,2)=31.065, $p<0.001$. A Bonferroni-Dunn post hoc analysis revealed significant differences between all joint pairs at a level of at least $p<0.001$. This effect reveals that relative phase values were lowest between the knee and the hip and were greatest between the ankle and the hip.

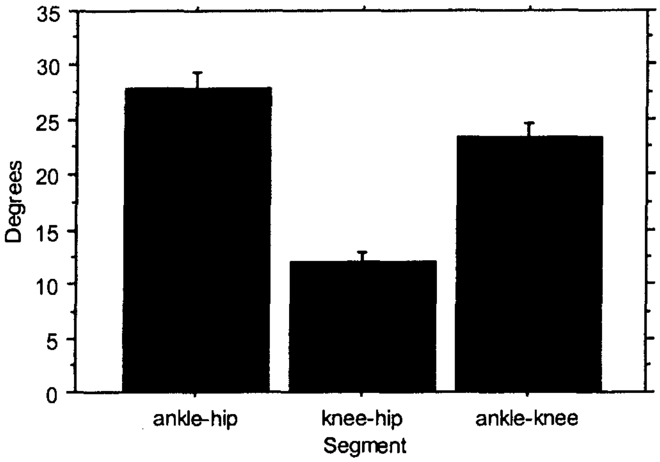

Figure 5: Main effect of mean relative phase values between joints of the lower limbs (ankle-hip, knee-hip, ankle-knee). Error bars represent standard error of the mean.

In addition to main effects, significant two-way interactions between: joint pair × frequency F(1, 6)=7.613, p<0.001, joint pair × mode F(1, 2) = 24.398, p<0.001 were observed, while a significant three-way interaction between joint pair × frequency × mode F(1,6)=7.374, p<0.001 is shown below in Figure 6. This interaction shows that as a function of arm swinging frequency, relative phase values between joint pairs of the lower limbs remain constant during inphase swinging. During antiphase swinging mean relative phase values of the lower limbs joint pairs involving the ankle (i.e. ankle-knee, ankle-hip) increase until 1.0 Hz then decrease at 1.5 Hz while the mean relative phase of the hip-knee joint pair increases at 1.5Hz.

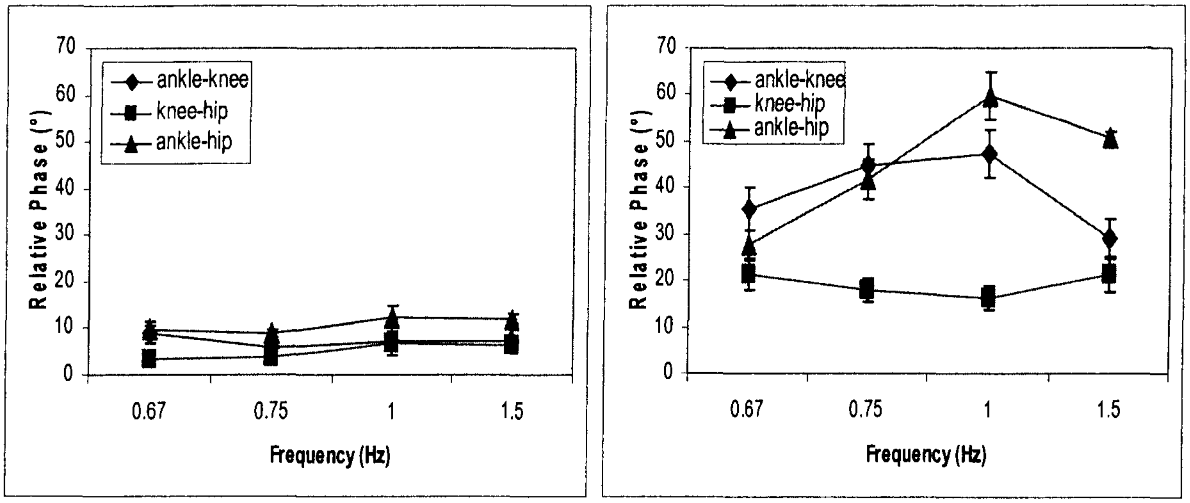

Figure 6: Mean relative phase values between joints of the lower limbs during both inphase (left) and antiphase (right) at slow (0.67Hz), medium (0.75Hz), fast (1.0 Hz) and very fast (1.5 Hz) arm swinging. Error bars represent standard error of the mean.

3.2 Variability of Relative Phase

Variability of the relative phase between lower limb joints was investigated, in order to look at how stable and consistent a subject's movements were. A 2 (mode) x 3 (joint pair) x 4 (frequency) repeated measures ANOVA revealed main effects of mode F(1,1)=233.594, p <0.001 (Figure 7), frequency F(1,3)=5.755, *p*<0.001 (Figure 8) and segment F(1,2)=53.743, *p*<0.001.

A main effect of phase revealed that inphase arm swinging produces less variability in the mean relative phase between joint segments of the lower limbs (M=4.562) compared to antiphase arm swinging (M=17.428) (Figure 7).

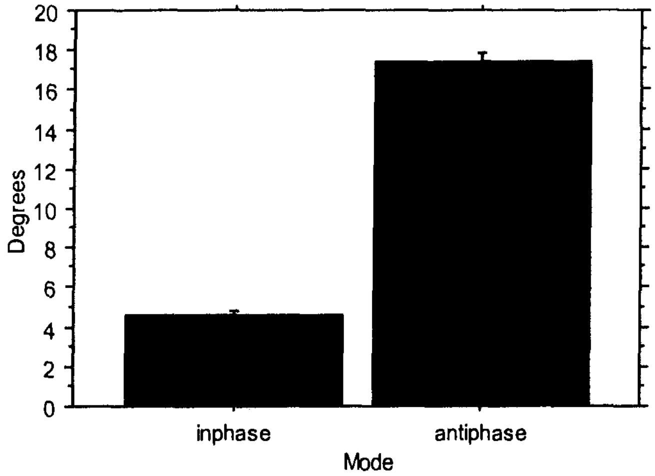

Figure 7: Mean variability of relative phase values between joints of the lower limbs (ankle-hip, knee-hip, ankle-knee) for inphase and antiphase arm swinging. Error bars represent standard error of the mean.

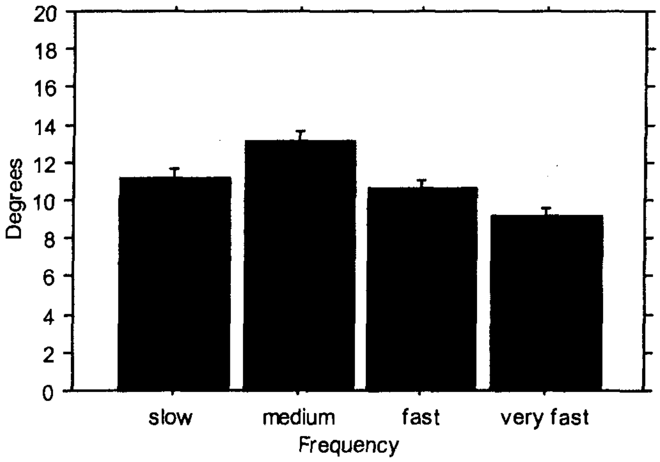

Figure 8: Mean variability of relative phase values between joints of the lower limbs (ankle-hip, knee-hip, ankle-knee) for slow (0.67 Hz), medium (0.75 Hz), fast (1.0 Hz) and very fast (1.5Hz) arm swinging conditions. Error bars represent standard error of the mean.

The main effect of frequency revealed that as arm swinging frequency increased, mean relative phase variability peaked at 0.75 Hz and the decreased at 1.0 Hz and 1.5 Hz.

A Bonferroni-Dunn post-hoc analysis revealed significant differences (at least a level of $p<0.001$) between all frequencies except slow and fast and fast and very fast (Figure 8). Significant two way interactions between joint pair × frequency, $F(1,6)=5.518, p<0.001$, joint pair × mode, $F(1,2)=5.259, p<0.01$ and frequency × mode, $F(1,3)=18.041, p<0.001$ were observed, while a significant three way interaction of joint pair x frequency x mode, $F(1,6)=7.644, p<0.001$ was also observed (Figure 9). This interaction reveals that during inphase arm swinging relative phase variability was not markedly different as a function of frequency, but during antiphase arm swinging relative phase variability decreased linearly after peaking at 0.75 Hz.

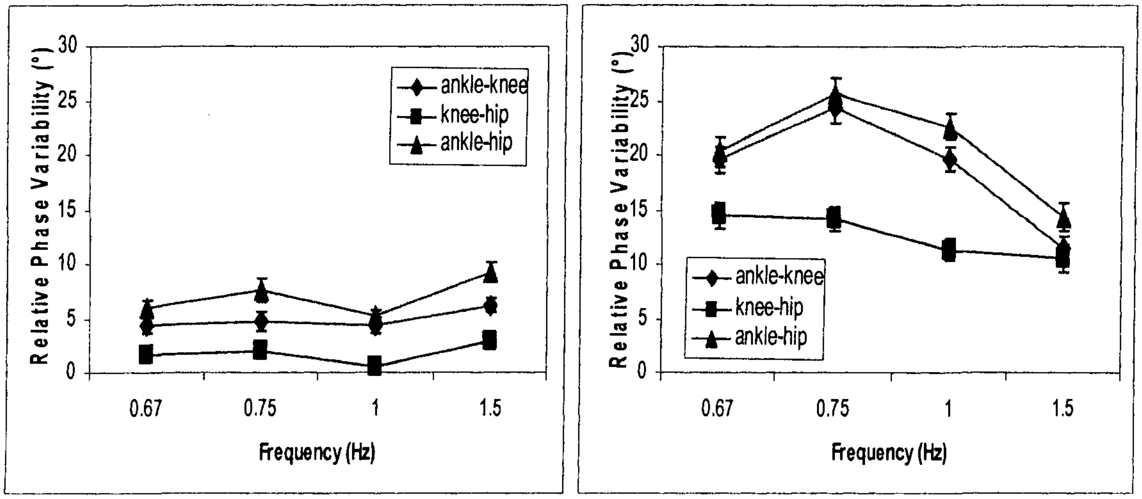

Figure 9: Mean variability of the relative phase between the ankle-knee, knee-hip, ankle-hip during slow (0.67 Hz), medium (0.75 Hz), fast (1.0 Hz) and very fast (1.5Hz) arm swinging. The inphase is depicted on the left while the antiphase condition is depicted on the right. Error bars represent standard error of the mean.

3.3 Joint Amplitudes

Hip, knee and ankle joint excursions were calculated to determine which joints contributed most to the centre of mass displacement. A 2 (phase) × 3 (joint pair) × 4 (frequency) repeated measures ANOVA was performed. Although differences in mean movement amplitude in the lower limb joints between inphase arm swinging (M= 4.370) and antiphase arm swinging (M=3.383) were observed, no significant main effect of mode was found between inphase and antiphase arm swinging, F (1, 49) =2.244, $p>0.05$. However, a main effect of frequency, F(1,3)=3.079, $p<0.05$ was found. Bonferroni-Dunn post-hoc analysis revealed significant differences between slow (0.67Hz; M=3.026) and very fast (1.5Hz; M=5.058), medium (0.75Hz; M=2.916) and very fast, and medium and fast (1.0Hz; M=4.507) (Figure 10), showing in general, a linear increase in joint amplitude as an effect of frequency.

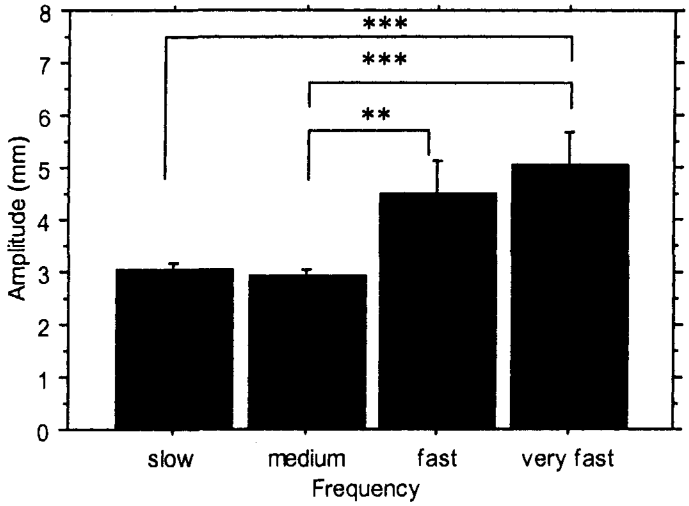

Figure 10: Main effect of lower limb joint amplitude as a function of slow (0.67Hz), medium (0.75Hz), fast (1.0Hz) and very fast (1.5Hz) arm swinging. Error bars represent standard error of the mean. **$p<0.01$, ***$p<0.001$.

A main effect of joint F(1,2)=57.047, p<0.001 was also observed, with Bonferroni-Dunn post-hoc testing revealing significant differences between the amplitudes of the three joints (Figure 11) at a level of p<0.001. This effect reveals that joint amplitudes were greatest in the hip, while very little ankle amplitude was evident.

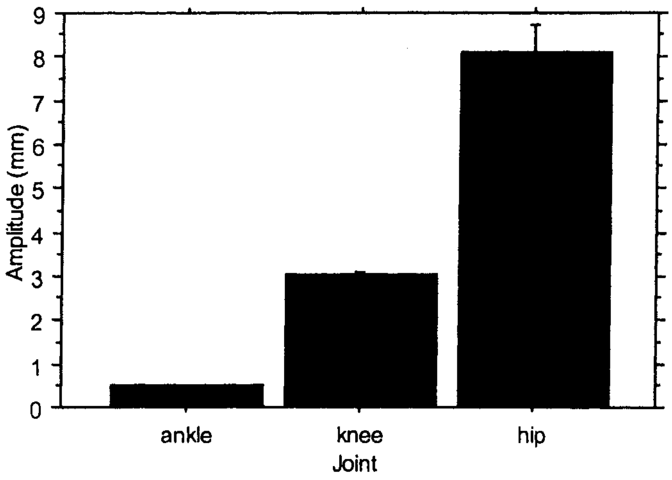

Figure 11: Mean lower limb joint amplitude for the effect of joint. Error bars represent standard error of the mean.

A significant two way interaction was observed between frequency × mode, F(1, 3), *p*<0.001, while Figure 12 shows a significant three way interaction between joint x frequency × mode, F(1,6)=4.972, *p*<0.001. This interaction shows that during inphase movements the ankle and knee joint amplitude remain relatively unchanged as a function of frequency however hip amplitude rises at 1.0Hz and was observed to decrease at 1.5 Hz. During antiphase movements, ankle amplitude remained invariant, knee amplitude appeared to increase linearly and hip amplitude showed an abrupt increase at 1.5 Hz. Except for the abrupt increase in antiphase hip amplitude at 1.5 Hz, joint amplitude tended to be greater in the inphase condition.

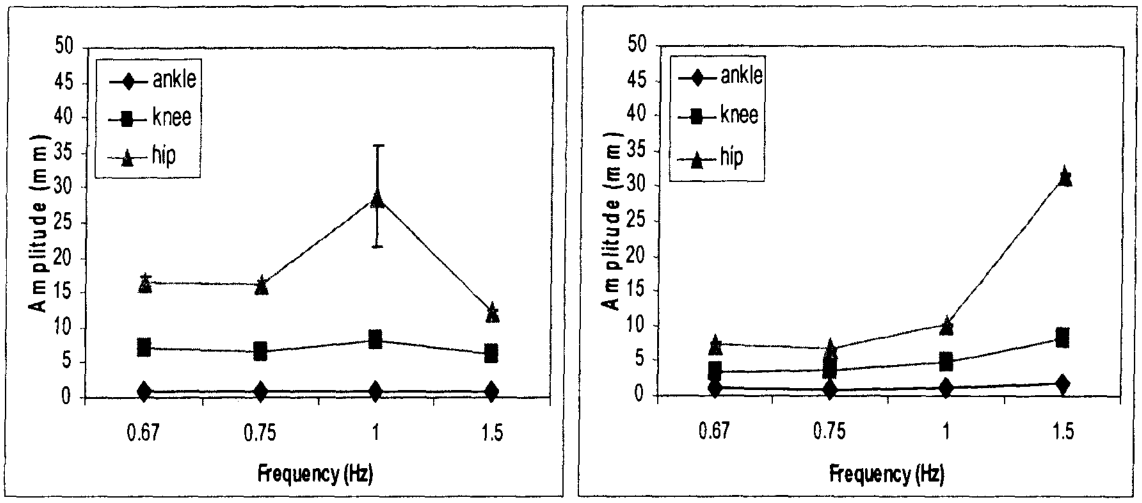

Figure 12: Ankle, knee and hip joint amplitudes averaged across the left and right side at slow (0.67 Hz), medium (0.75 Hz), fast (1.0 Hz) and very fast (1.5Hz) arm swinging frequencies for inphase (left) and antiphase (right) swinging. Error bars represent standard error of the mean.

3.4 Centre of Mass Amplitude

Separate 4 (frequency) × 2 (mode) ANOVAS with repeated measures were performed for both AP and ML centre of mass amplitudes. In the AP direction, significant main effects were found for both frequency F (1,3)=6.235, p<0.001 (Figure 13, left) and mode F(1,1)=29.543, p<0.0001 (Figure 14, left) . The same trend was observed in the ML direction: main effects were found for frequency, F (1,3)=5.995, p<0.001 (Figure 13, right) and for mode F(1,1)=134.634, p<0.001 (Figure 14, right).

As a function of frequency, an increase in mean AP CoM amplitude was observed, while post-hoc analysis revealed significant differences between the slow and very fast (p<0.001), and between medium and very fast arm swinging frequencies (p<0.01). An increase in CoM amplitude as a function of frequency was also observed in the ML direction with post hoc analysis showing significant differences between the slow and very fast (p<0.001), and between medium and very fast (*p*<0.01) arm swinging.

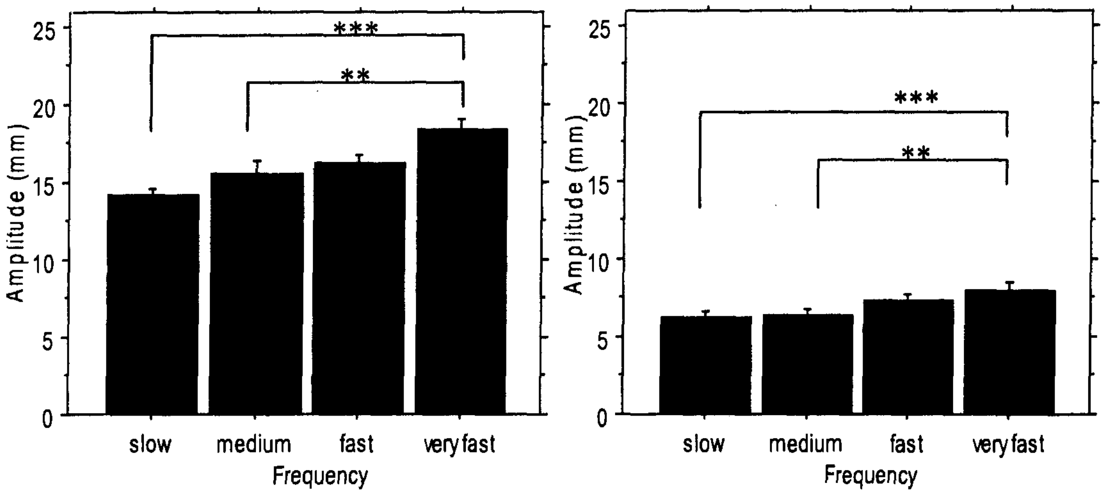

Figure 13: Main effect of frequency of arm swinging for both AP CoM amplitude (left) and ML CoM amplitude (right). Error bars represent standard error of the mean. ***p*<0.01, ****p*<0.001.

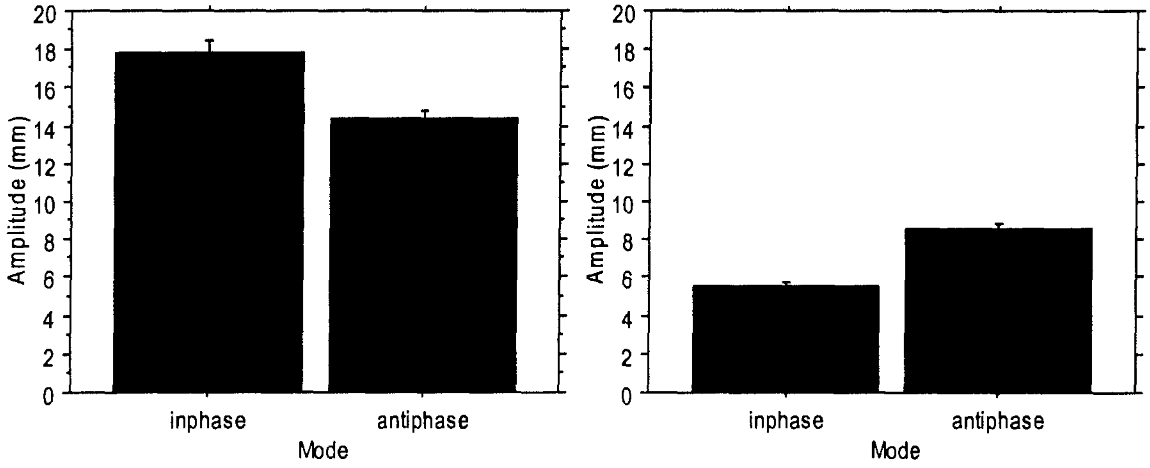

Figure 14: Main effect of mode of arm swinging for CoM amplitudes in both the AP (left) and ML (right) directions. Error bars represent standard error of the mean.

In the AP direction, inphase arm swinging resulted in CoM having greater mean amplitude (M=17.848) than during antiphase swinging (M=14.372), while the opposite trend was evident in the ML direction where antiphase arm swinging resulted in greater CoM amplitude (M=8.558) than inphase arm swinging (M=5.482).

No significant interactions for frequency × mode were observed for either AP or ML CoM amplitude. However, Figure 15 shows a linear increase in mean CoM amplitude as a function of frequency for both inphase and antiphase movements in both the AP (left) and ML (right) directions.

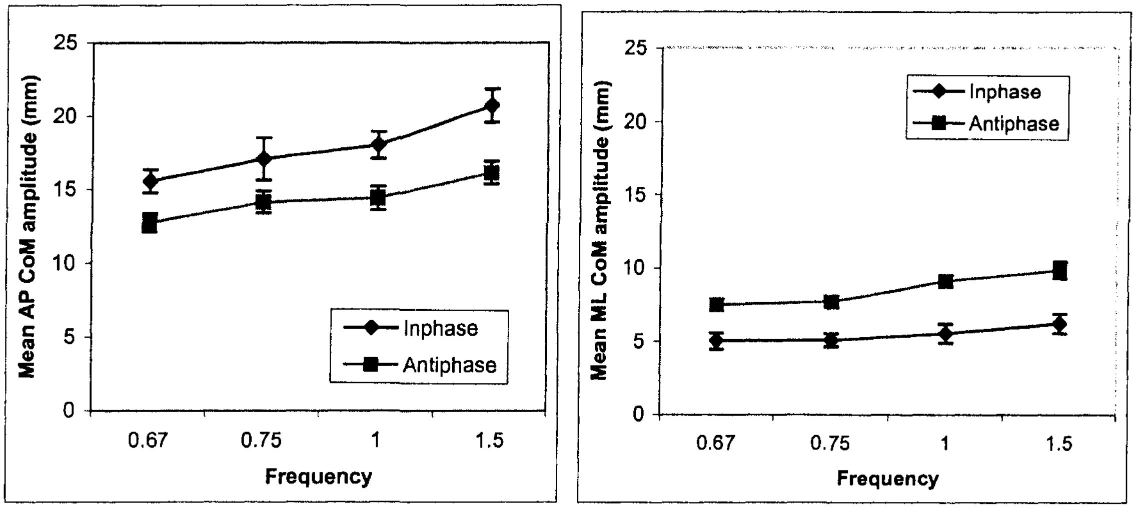

Figure 15: Mean Centre of Mass amplitudes for AP (left) and ML (right) directions for both inphase and antiphase arm swinging at increasing frequencies. Error bars represent standard error of the mean.

3.5 Centre of Pressure Amplitude and Spectral Density

A 4 (frequency) × 2 (mode) ANOVA was performed to investigate changes in mean center of pressure amplitude averaged across the left and right foot in the AP direction. Significant main of effects of both frequency F (1,3)=4.848, $p<0.005$ and phase F(1, 1)=51.642, $p<0.001$ were found. The main effect of phase shows that centre of pressure amplitude was greater during antiphase arm swinging (M=28.801) than inphase arm swinging (M=20,181) (Figure 16).

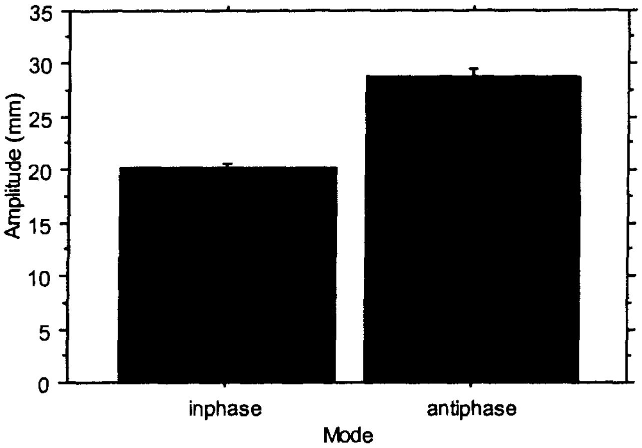

Figure 16: Mean AP CoP amplitude as a function of mode inphase and antiphase arm swinging. Error bars represent standard error of the mean.

To better demonstrate these effects a significant two way interaction of frequency × mode, F(1,3)=8.767 $p<0.001$ is plotted below (Figure 17). Mean AP CoP amplitudes show no real trend as a function of frequency during both inphase and antiphase swinging.

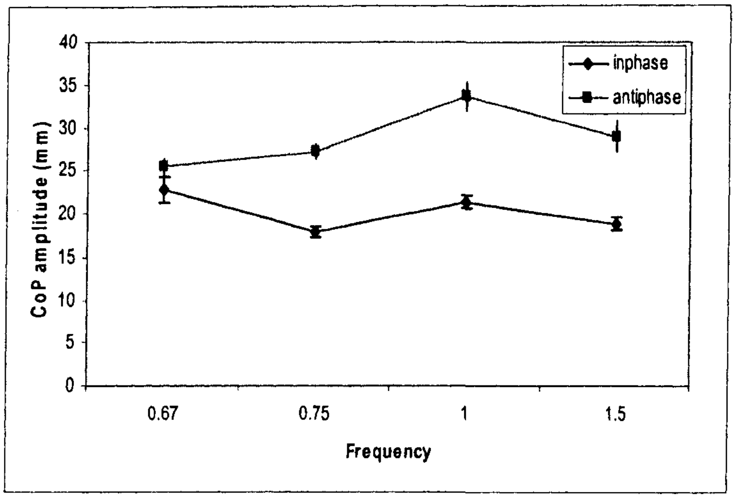

Figure 17. Centre of pressure amplitudes in the anterior-posterior direction at slow (0.67 Hz), medium (0.75 Hz), fast (1.0 Hz) and very fast (1.5Hz) arm swinging frequencies during both inphase and conditions. Error bars represent standard error of the mean.

To gain insight into the properties (or characteristics) of the CoP oscillations in the AP direction, a power spectral density analysis was performed for each trial and the mean frequency was calculated. A 2 (mode) × 4 (frequency) ANOVA with repeated measures was performed. Despite differences in the mean frequency during inphase (M=0.460) and antiphase (M=0.502) swinging, no main effect of mode of swinging was observed, F(1,1)=6.594, $p>0.01$ ($p=0.0133$). As predicted, a main effect of frequency was observed, F(1,3)=26.330, $p<0.001$ and showed an increase in mean frequency as arm swinging frequency increased. Bonferroni-Dunn post-hoc analysis revealed significant differences of mean frequency between all frequencies (except slow and medium) at a level of at least $p<0.01$. A sample spectral plot reveals that as arm swinging frequency increases, density becomes more dispersed, and larger peaks are present around the target arm swinging frequency during inphase conditions.

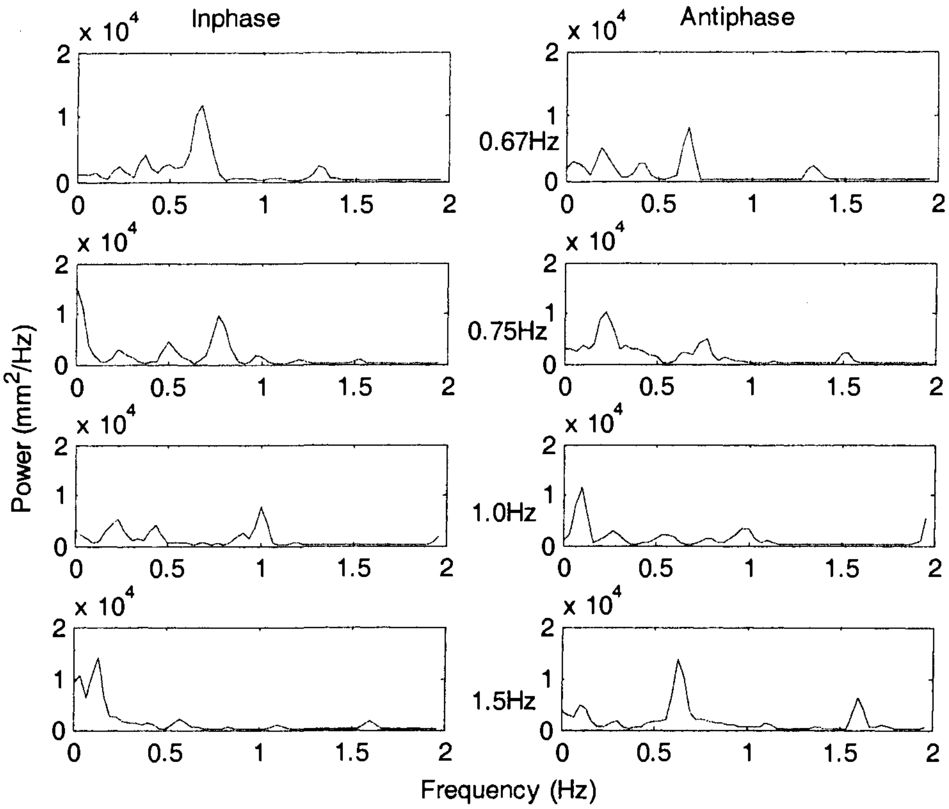

Figure 18: Power spectral density plots of the CoP in the AP direction for inphase and antiphase swinging at slow (0.67 Hz), medium (0.75Hz), fast (1.0Hz) and very fast (1.5Hz) arm swinging for one subject and one trial per condition.

Power spectral density analysis revealed that there were greater oscillations around the target arm swinging frequency defined by the condition (i.e. there were larger peaks at these frequencies). Furthermore, there are larger peaks at the target frequencies during inphase swinging, whereas additional peaks were present during antiphase swinging. These additional peaks during antiphase swinging are likely due to a less coordinated movement pattern in the lower limbs as evidence by greater mean relative phase values between joints of the lower limbs during antiphase swinging (Figure 6).

3.6 Arm Amplitude

To investigate changes in arm amplitude, a 2 (mode) × 4 (frequency) ANOVA with repeated measures was performed. Although differences were found between inphase (M=296.795) and antiphase (M=305.342) arm swinging no main effect F(1,1)=3.544, $p>0.01$ ($p=0.657$) of arm swinging mode was observed (Figure 19). However, a significant main effect of frequency was found, F(1,3)=142.293, $p<0.001$ with a Bonferroni-Dunn post hoc analysis revealing significant differences between all swinging frequencies at a level of at least $p<0.01$. This shows that arm amplitude decreases as a function of swinging frequency.

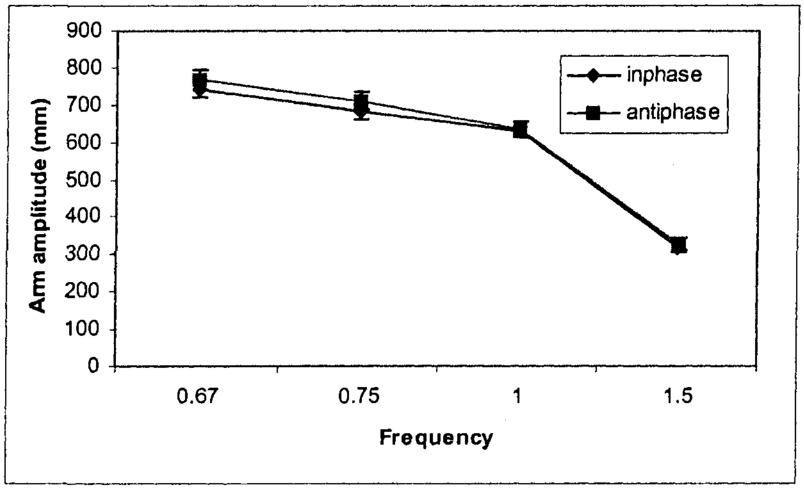

Figure 19: Arm amplitudes for inphase and antiphase arm swinging at slow (0.67 Hz), medium (0.75 Hz), fast (1.0Hz) and very fast (1.5 Hz) arm swinging. Error bars represent standard error of the mean.

3.7 Principal Component Analysis

A principal component analysis (PCA) was performed to reduce the dimensionality of the data set and to give an indication of the contribution of each joint to movement variance in 3D space. Each eigenvalue measures the percentage variance of the movement pattern explained by each PC. The eigenvectors measure the importance of each joint on the principal components and how much each joint contributed to the coordination observed. If two eigenvectors have the same sign, they are inphase with each other, if they had opposite signs they are antiphase to each other. The *x* (ML,) *y* (AP), and *z* (vertical) components of each eigenvector are displayed. Results of the PCA show that one eigenvalue explains over 90% of the variance of the movement pattern during all inphase swinging frequencies, while two eigenvalues are required to explain 90% of the variance during antiphase swinging.

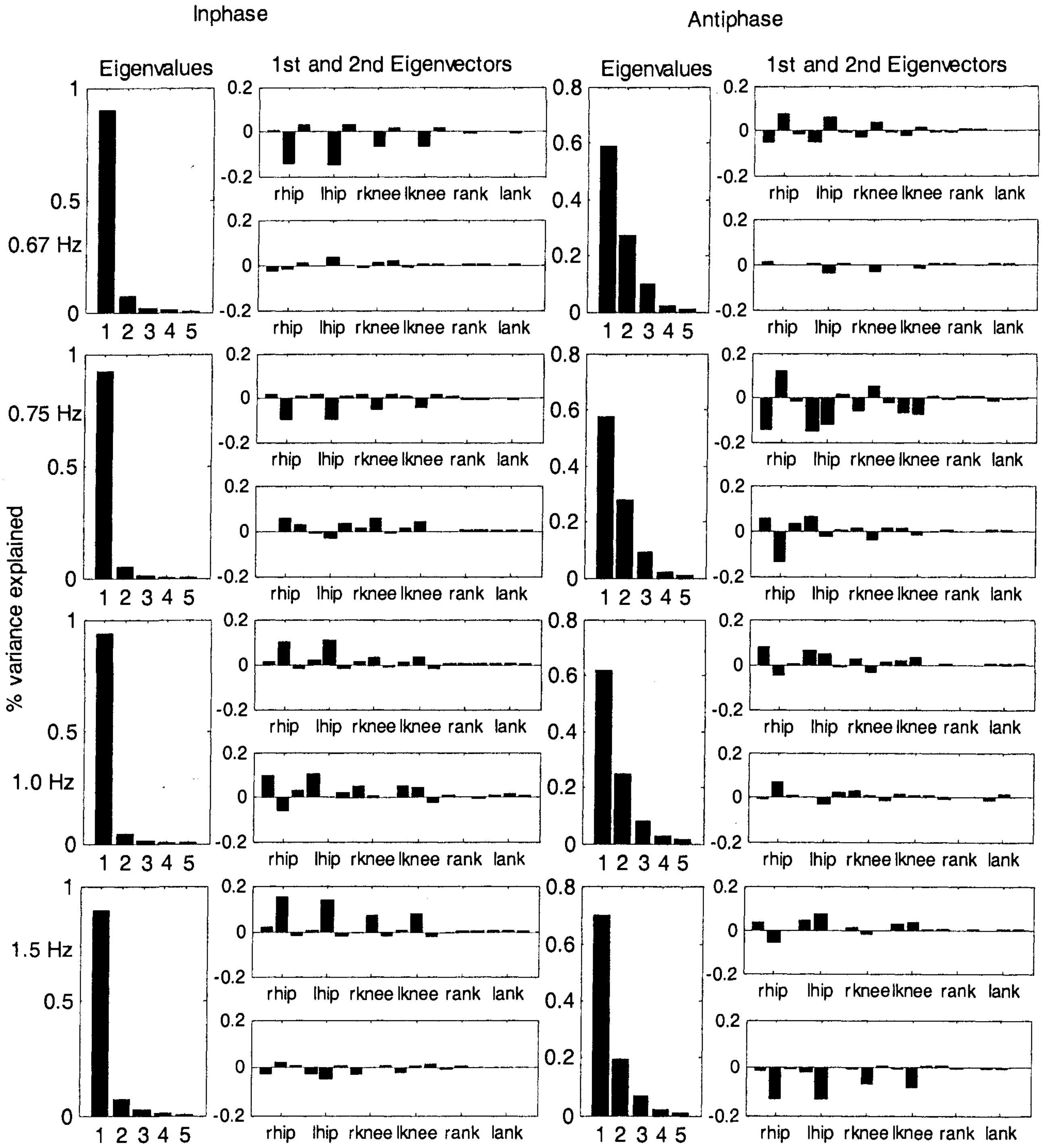

Figure 20: Principal component analysis of the lower limb joints for inphase (left) and antiphase (right) arm movements at increasing frequencies. The top row is slow frequency, second from the top is the medium frequency, third row is fast and bottom row is very fast. The contribution of each joint is shown in the *x* (ML), *y* (AP), and *z* (vertical) direction.

3.8 Directional Coupling

A directional coupling analysis on the lower limb joints was performed in order to determine which joint in each lower limb pairing was the leading oscillator. A 2 (mode) × 3 (joint pair) × 4 (frequency) repeated measures ANOVA revealed, significant main effects of joint pairing, F (1,2)=3.838, $p<0.05$ (Figure 21) and phase F (1,1)=7.635, $p<0.01$ (Figure 22), showing that during inphase arm swinging, there is greater coupling in the joints of the lower limbs (M=0.106) compared to antiphase (M=0.036) swinging. A Bonferroni-Dunn post hoc analysis revealed significant differences between the coupling of the hip-ankle pairing and the knee-ankle pairing ($p<0.01$) and between the hip-ankle and hip-knee ($p<0.001$). Coupling is greatest between the ankle and hip (M=0.100), while coupling in pairings involving the knee (ankle-knee, M=0.62, knee-hip, M=0.052) was lowest.

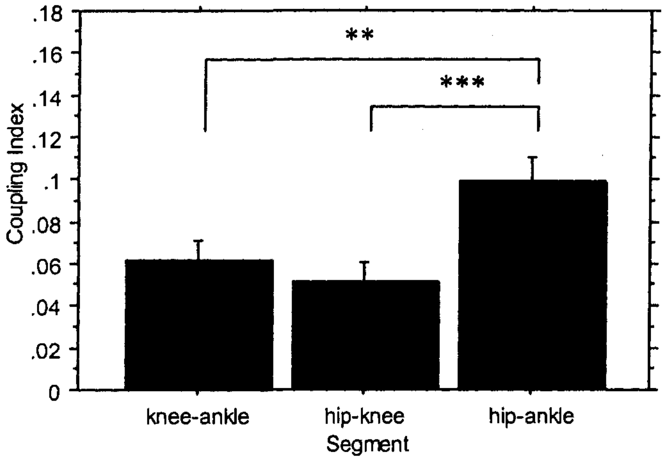

Figure 21: Directional coupling between joints of the lower limbs. A positive value indicates that the first joint in the pair being examined is the leading oscillator. Error bars represent standard error of the mean. ** $p<0.01$, *** $p<0.001$.

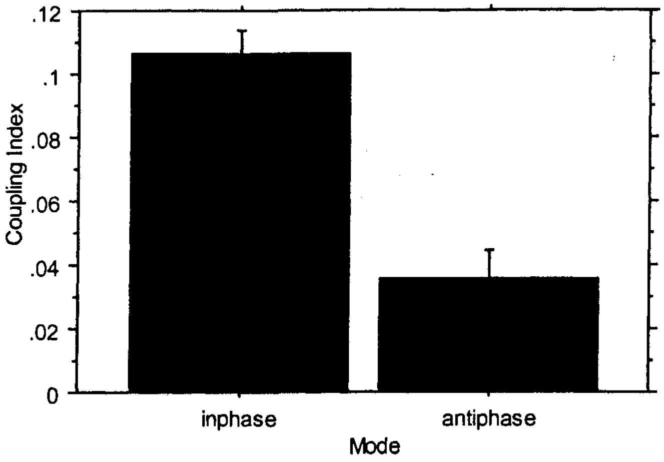

Figure 22: Directional coupling as an effect of inphase and antiphase arm swinging. Error bars represent standard error of the mean.

Significant two way interactions were observed for joint pairing × frequency F(1,6)=2.99, $p<0.01$, joint pair × mode F(1,2)=7.936, $p<0.001$ and frequency × mode F(1,3)=19.232, $p<0.001$. A three way interaction of joint pairing × frequency × mode F(1,6)=4.939, $p<0.001$ (Figure 23) was also found to be significant. This interaction shows that during antiphase movements the ankle was the leading oscillator at low frequencies, with knee and hip leading at faster frequencies. During all inphase arm-swinging frequencies, the hip leads both the knee and ankle, while the knee was also responsible for leading the ankle. The magnitude of the coupling strength is also greater during inphase swinging.

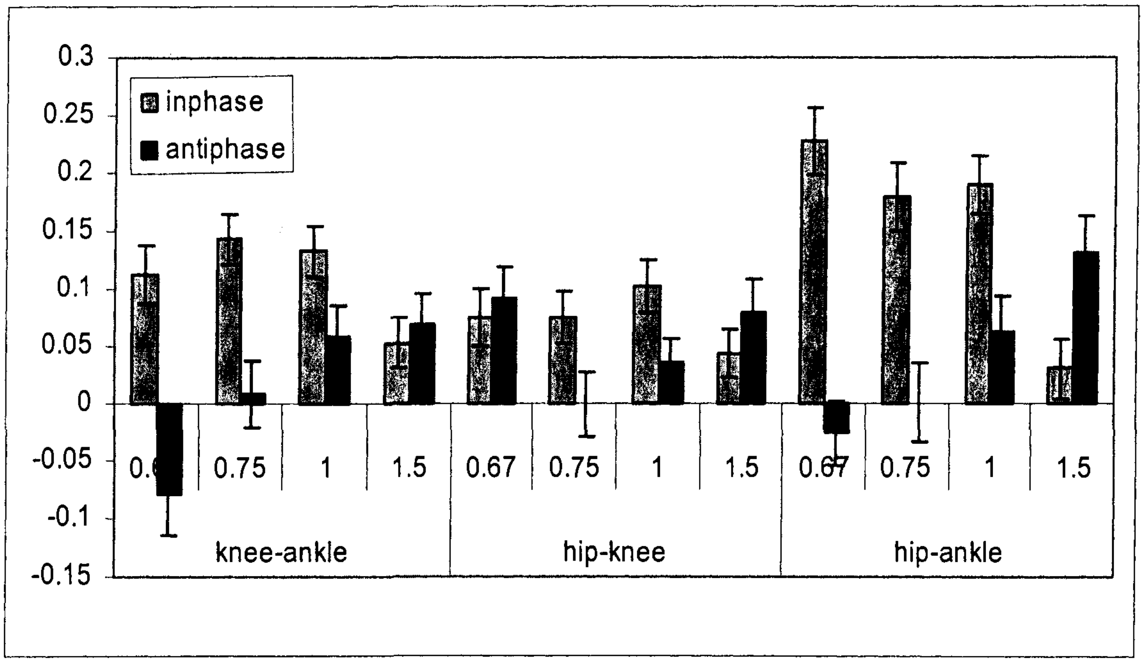

Figure 23. Directional coupling analysis between the knee-ankle, hip-knee and hip-ankle at slow (0.67 Hz), medium (0.75 Hz), fast (1.0 Hz) and very fast (1.5Hz) arm-swinging frequencies for both inphase and antiphase arm swinging. A positive value indicates the first oscillator is leading the second, while a negative value indicates the opposite. Error bars represent standard error of the mean.

3.9 Arm-CoP Relative Phase

Relative phase between CoP under each foot and ipsilateral arm movement was calculated to see how CoP was displaced as a result of the swinging arm. A 4 (frequency) × 2 (mode) repeated measures ANOVA was performed, revealing significant main effects of both frequency F(1,3)=57.139, $p<0.001$ and mode F(1,1)=36.501, $p<0.001$. For the effect of frequency, it was found that the relative phase increased linearly as arm swinging frequency increased (Figure 24). A Bonferroni-Dunn post hoc analysis revealed significant differences between all frequencies except for fast and very fast arm swinging with a significance level of at least $p<0.05$ between all frequencies. For the effect of mode (Figure

25) results show that inphase movements had greater arm-CoP mean relative phase than did antiphase.

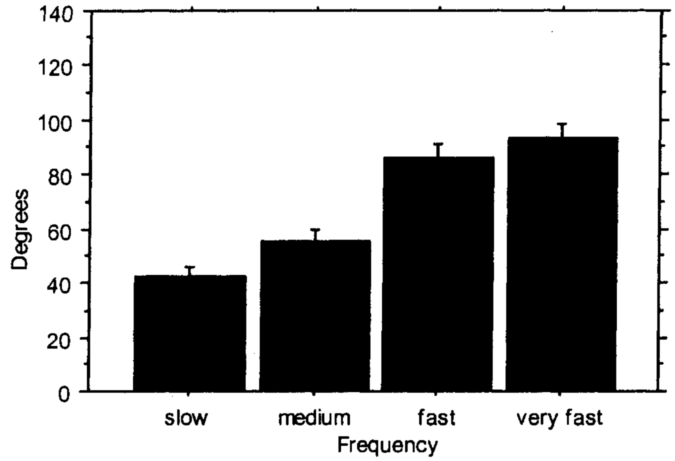

Figure 24: Main effect of frequency for the mean relative phase between AP CoP and the ipsilateral arm at slow (0.67 Hz), medium (0.75 Hz), fast (1.0 Hz) and very fast (1.5 Hz) arm swinging. Error bars represent standard error of the mean.

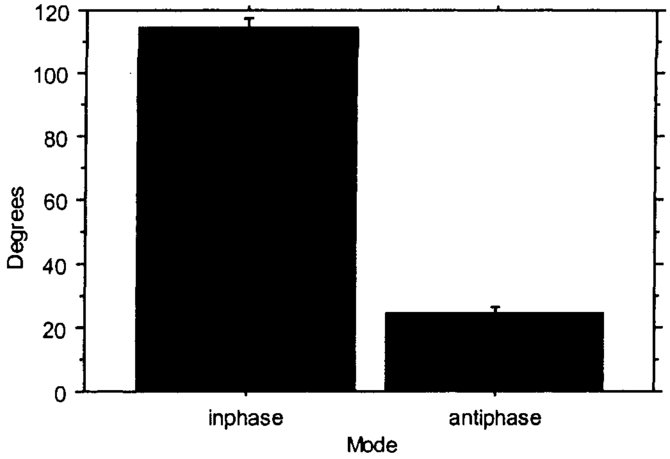

Figure 25: Main effect of mode for the mean relative phase between AP CoP and the ipsilateral arm for inphase and antiphase arm movements. Error bars represent standard error of the mean.

An interaction of frequency × mode F(1,3)=20.968, p<0.001 was also observed to be significant (Figure 26). During inphase swinging this interaction shows that the arm-CoP relationship becomes more antiphase as swinging frequency is increased. However, during antiphase swinging, CoP under each foot moves in the same direction and therefore inphase with the ipsilateral arm.

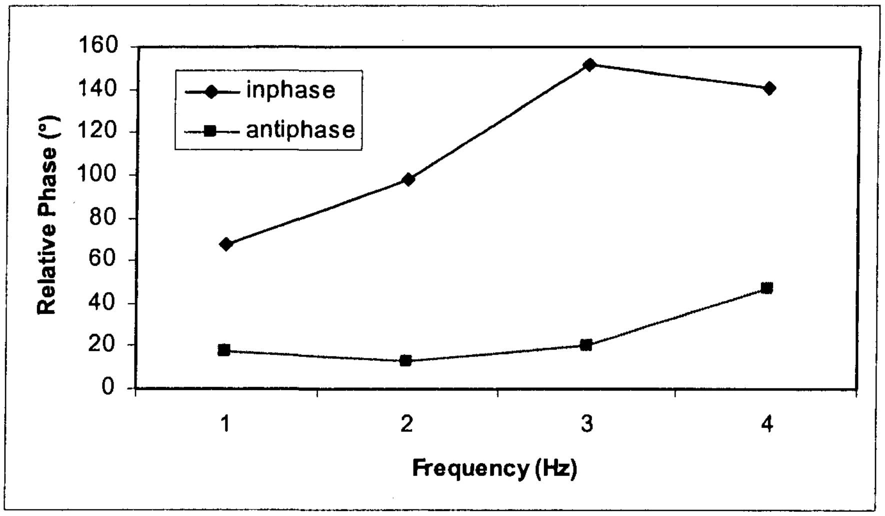

Figure 26. Relative phase between the arm and ipsilateral CoP during inphase and antiphase swinging at increasing swinging frequencies. Error bars represent standard error of the mean.

CHAPTER IV

Discussion

The primary hypothesis of this study was that the coordination in the lower limbs would exhibit different patterns for inphase and antiphase movements and would become less coordinated and more variable as a function of frequency. Specifically it was thought that inphase arm swinging would yield a coordination pattern closer to 0° and that at increased frequencies different coordination patterns may emerge. Results support these hypotheses, as relative phase values were closer to a 0° coordination mode during inphase swinging, while frequency related changes in the role of the hip and ankle were observed in both the relative phase and directional coupling data. Antiphase swinging greater mean relative phase and relative phase variability, and also resulted in greater ML CoM amplitude, with a PCA confirming the role of the hip in the ML direction during antiphase swinging. Inphase swinging results in greater displacement of CoM in the AP direction and as a result CoP shifts antiphase to the arms to compensate for the perturbation. The results are discussed in detail below.

4.1 Mean Relative Phase

The most obvious result of this study is that lower limb synergies during inphase and antiphase arm swinging appear to be organized differently based on the task constraints specific to each condition. Analysis of mean relative phase showed that there are larger angular differences between the movements of joints of the lower limbs while performing antiphase arm movements than during inphase movements, indicating that inphase movements were performed with greater coordination (i.e. closer to a 0° inphase pattern). Previous studies (i.e. Bardy et al 1999, 2002) have shown that during a visual tracking postural coordination task, an inphase coordination pattern between the hips and ankles exhibits relative phase values around 20°. Our results for the inphase condition were approximately 10 ° and support these findings.

As previously hypothesized (Oullier et al., 2006), during postural control there are mechanical constraints that prevent the joints of the lower limbs from reaching a near perfect 0° phase relationship. While it is easy to couple two fingers from different hands during bimanual finger oscillations, the movements of the trunk and hip work in opposition to each other to compensate for the postural perturbation. As a result 0° coordination pattern is difficult to achieve (Bardy et al. 1999, 2002).

During antiphase arm swinging, relative phase values were significantly higher than during inphase swinging, and also much higher than the 20° inphase pattern. The increase in mean relative phase values are likely due to added mechanical constraints imposed upon the postural system during antiphase swinging. It has previously been shown that during antiphase arm swinging there is far more trunk rotation than during inphase swinging (Ustinova et al., 2004). It is hypothesized that these increased trunk rotations cause the hip to be pulled out of phase from the knee and ankle, which in turn could also draw the knee more out of phase with the ankle. This idea is supported by previous studies in which no abrupt transition between inphase and antiphase pattern was observed (Buchanan & Horak, 2001). This observation has been attributed to gradual recruitment/suppression of lower limb degrees of freedom. These ideas are further discussed below.

Previous studies have shown frequency dependent changes in postural coordination patterns in the lower limbs during both arm swinging (Abe & Yamada, 2001) and other suprapostural tasks (Bardy et al., 1999). A secondary objective of this study was to examine coordination in the lower limbs as an effect of frequency. Analysis of mean relative phase values between joints of the lower limbs during inphase swinging revealed no increase in mean relative phase from the initial inphase pattern as a function of frequency. However, during antiphase swinging mean relative phase values gradually increased from slow (0.67 Hz) to fast (1.0 Hz) in both the ankle-knee and the ankle-hip pairings. This gradual change or transition in the postural pattern has previously been documented in suprapostural tasks (Buchanan & Horak, 2001). The exception to this linear trend during antiphase swinging was the very fast (1.5Hz) frequency, where the mean relative phase values between these joints decreased, implying that there is increased activity about the ankle joint at slower frequencies, with activity increasing to a critical point. This finding supports previous claims that the ankle plays a critical role in slower and smaller magnitude perturbations (i.e. McCollum & Leen, 1989).

Contrarily, as arm swinging frequency increased, the relative phase between the hip and the knee increased, implying that there is increased activity about the hip joint at faster swinging frequencies. This finding is in line with previous research (Ferry et al., 2007, Martin et al. 2006; McCollum & Leen, 1989) that has shown increased role of the hip at increased perturbation frequencies. Thus, at this very fast frequency it is hypothesized that the ankle joints stiffened, while rotations about the hip joints increased.

4.2 Relative Phase Variability

To gain an idea of the stability of the phase relationships, we also examined the variance in the relative phases of the lower limb joints during in phase and antiphase movements. During antiphase arm swinging the variability of the relative phase, or of the coordination pattern of the lower limbs, was also greater compared to inphase movements.

Previous studies have shown greater relative phase variability during antiphase movements when compared to inphase movements (between the effectors performing the movements) (ie Kelso, 1984, Haken et al. 1985). It is likely that during antiphase arm swinging, the increased variability in the coordination between joints of the lower limbs occurs as a result of increased variability in the swinging arms. Previous studies have demonstrated that phase transitions are more likely to occur as the variability of the given coordination pattern increases (Bardy et al., 1999), however here we observed no phase transitions as a result of increased relative phase variability, since all the movements were performed at a constant frequency within trials.

While performing antiphase swinging, the variability of the phase relationships between the joints of the lower limbs started to decrease during the faster arm swinging frequencies. During both fast and very fast arm swinging frequencies the relative phase variability between all joints of the lower limb decreased, showing a trend similar to that observed for mean relative phase. This most likely occurs due to increased demands of the task: at lower frequencies the mechanical task demands are low and subsequently movements are more variable, but as task demands become more exacting, movements become more restricted to the required coordination pattern.

4.3 Joint Amplitudes

Joint amplitudes were not significantly different between inphase and antiphase swinging. However we can see (Figure 12) that as a trend, joint amplitudes were greater during inphase arm swinging. The increased joint amplitude during inphase swinging occurs because the body's centre of mass shifts backwards and forwards with greater amplitude compared to antiphase movements. During inphase movements, the arms are acting together to pull the centre of mass in the fore-aft direction, whereas during antiphase movements, the body's centre of mass is displaced less because as one arm is in front of the body, the other is behind imposing a counteracting effect on the location of the centre of mass. Joints of the lower limb, especially the hip joint given its change in amplitude, are likely responsible for the shifting of the centre of mass.

In general, joint amplitudes increased as a function of arm swinging frequency in the hip and knee, however, no changes were observed in ankle amplitude. Ankle amplitudes remained low (relative to the hip and knee), since the feet remained in contact with the force plates, limiting angular displacements and amplitude about the joint. An exception to this trend lies between the fast (1.0Hz) and very fast (1.5 Hz) swinging frequencies. During inphase swinging the hip joint showed a marked drop in amplitude between the two frequencies. However, during antiphase movements, hip joint increased in amplitude.

The drop in hip amplitude is possibly related to the sharp decrease in arm amplitude observed at this very fast swinging frequency. Friedli et al., (1988) showed that the hip joint plays a role in absorbing reactive torques from the shoulder joint during rapid arms movements. Thus it is conceivable that less hip amplitude is required for compensation at this frequency. These speed related biomechanical changes observed in joint amplitudes during inphase swinging are similar to findings of Abe and Yamada (2001) who found that while hip angular displacements were greatest at slower frequencies, they decreased at faster frequencies.

During antiphase arm swinging the sharp increase in hip amplitude can be attributed to increased hip rotations at this frequency. Previous work (Ustinova et al., 2004) has demonstrated that the magnitude of hip rotations were greater during antiphase swinging than during inphase swinging and increased as a function of swinging frequency. Thus, as swinging frequency increased, the rotations about the hip increased as well. This is further supported by CoM amplitude data, which shows that there is greater ML amplitude of CoM in antiphase conditions and at the faster swinging frequencies.

4.4 Centre of Mass Amplitudes

As previously mentioned, CoM amplitudes were analyzed to help explain the anomalous hip amplitude during 1.5 Hz swinging (sharp decrease in hip amplitude during inphase swinging, while there was a sharp increase in hip amplitude during antiphase swinging). In general, CoM amplitude showed increases across all frequencies for both inphase and antiphase movements in both the AP and ML directions.

During inphase swinging at 1.5 Hz, CoM amplitude continued to rise in the AP direction despite the decrease in hip amplitude. A potential explanation for this finding lies in the movements of the upper body. At this frequency it is possible that the body is behaving as a single link inverted pendulum with movements of the trunk, upper body and head increasing about a more stationary hip (Forner-Cordero, Koopman, & van der Helm, 2004). Forner-Cordero et al (2004) also showed that the trunk could be controlled independently of body center of mass acceleration, which would also explain the linear increase CoM at this frequency, despite perhaps nonlinear increases in trunk and upper body movements. These findings confirm those by Ustinova et al. (2004) who found that at increased frequencies there was increase in trunk amplitude for inphase movements while, no increase in trunk amplitude during antiphase movements. They also found that the trunk was responsible for counterbalancing the perturbations induced by the arm movements during inphase arm swinging at fast frequencies.

During antiphase swinging CoM amplitude increased linearly in both the AP and ML direction, however there was disproportional increase in hip amplitude when compared to CoM amplitude. Since there is greater ML CoM amplitude during antiphase movements when compared to inphase movements, it is likely that hip rotations increase due to the mechanical properties of the swinging arms. The hip rotations occurring in the ML direction about the vertical axes of the body are likely the reason why the ML and AP CoM do not show increases in amplitude proportional to the hip.

4.5 Centre of Pressure Amplitudes

CoP under each foot was measured in the AP direction to quantify postural sway during each experimental condition. It was found that greater CoP amplitude was observed during antiphase swinging when compared to inphase swinging. Since forces generated by the arm movements result in reactive torques at the joints of the lower limbs in order to stabilize the body (which in turn displace the CoP), it is inferred that more reactive torques in the lower limbs are generated during antiphase swinging. It is possible that since there was greater variability in the phase relationship between joints of the lower limbs that there were increased oscillations at these joints during antiphase swinging causing a greater amount of CoP amplitude. Arm amplitudes were also greater during antiphase swinging (albeit not significant), which could also account for the greater amount of CoP amplitude during antiphase swinging. Alternatively, during inphase swinging the CoP works to control CoM very tightly to due greater CoM amplitude, while during antiphase swinging CoM amplitude is of a lesser magnitude and therefore controlled less tightly.

The greater CoP values and increased mechanical constraints during antiphase swinging could also be a cause of the higher relative phase values between the joints of the lower limbs which were shown to deviate from the previously identified inphase coordination pattern. Ferry et al., (2007) showed that CoP position is a crucial parameter in determining the adopted coordination mode, and that prior to phase transitions in the lower limbs, AP amplitude of CoP were near the edge of the supporting area.

4.6 Arm Amplitudes

Arm amplitude was calculated to gain insight of their possible effects on other movement parameters. Despite greater arm amplitude during antiphase arm swinging, no main effect was observed. However, significant decreases in amplitude are observed as swinging frequency increased (Figure 25). This inverse relationship between frequency and amplitudes has previously been found in tapping studies (Kay, Kelso, Saltzman, & Schoner, 1987).

4.7 Principal Component Analysis

A PCA was performed to determine the magnitude and directional contributions of each of the lower limb joints to movement coordination and also served to confirm results obtained from relative phase data. Furthermore, since our relative phase and joint amplitude analysis were only performed for movements in the AP direction, the PCA was useful in providing a three dimensional analysis of the movements in the lower limb joints.

Consistent with the relative phase data, the PCA revealed differences between the organization of the lower limbs in response to inphase and antiphase arm swinging. For inphase swinging one large eigenvalue that explains over 90% of the movement variance was present, indicating that there was largely one major coordination mode (Forner-

Cordero et al., 2007). This result is consistent with the relative phase data, which shows that there is an inphase relationship between all joints of the lower limbs during inphase arm swinging. During antiphase swinging two eigenvalues were required to explain 90% of the variance, indicating that there is more variability in the lower limbs during antiphase swinging. This also confirmed results of the relative phase data, which showed that during antiphase arm swinging the coordination pattern between joints of the lower limbs was more variable and deviated from a more stable inphase coordination pattern.

During inphase swinging the first eigenvector set reveals that the hip (both left and right) in the AP direction (direction of the movement) is contributing most to the variance, with smaller contributions of the knee in the AP direction. During antiphase movements the primary eigenvector reveals contributions from the hip in both the AP and ML direction as well as the knee in both AP and ML directions. Hip and knee contributions in AP direction confirmed what was observed in joint amplitudes, while the contributions of the hip and knee in the ML direction supports our hypothesis that there is increased rotation in these joints due to mechanical constraints of the arms. The secondary eigenvector shows mainly contribution from the hip and knee joints, with very little contributions of the ankle in the AP direction.

4.8 Directional Coupling

A directional coupling analysis was performed to identify the leading oscillator within a pair of signals. This analysis revealed that there was positive coupling between the hip and knee (indicating that the hip is leading the knee). There was also positive coupling between the knee and ankle and the hip and ankle. This brings us again to the conjecture that the arm movements produce reactive torques and forces in the lower limb, and that these reactive torques are produced in a sequential manner (i.e. first to the hip, then to the knee then the ankle). This sequence of reactive torques being translated from the hip joint to the ankle joint has previously been shown during arm swinging (Eng, Winter, MacKinnon, & Patla, 1992).

Coupling was also found to be greater during inphase arm swinging than during antiphase swinging. This result is similar to that of the mean relative phase, which showed lower relative phase values during inphase swinging and hence greater coupling. This is most likely due to mechanical constraints; during inphase swinging the legs move together, but during antiphase swinging there is a reciprocal relationship between the two legs, and more joint rotations which in turn lead to decreased interactions and interdependencies as well as de-coupling between the joints.

It was also found that coupling was greatest in joint pairings involving the ankle, with the greatest coupling observed between the hip and ankle joints. During antiphase swinging at the slow 0.67 Hz frequency, the ankle is leading oscillator. As swinging frequency increased there was a switch over (to both the knee and the hip). It is likely that at 0.67 Hz the perturbation and CoM amplitudes are small enough that the body resembles an inverted pendulum with rotations about the ankle being sufficient to stabilize the body. This is consistent with the relative phase data, which showed increased relative phase in lower limb pairings involving the hip at slower frequencies (Figure 6).

4.9 Arm-CoP Interactions

Interaction between the swinging arm and the CoP under the ipsilateral foot was investigated. Results of the current study show that as inphase arm swinging frequency increases, the arm and CoP move in opposite direction to each other. This confirms findings of previous work (Ustinova et al. 2004) which showed that the pattern of interaction between arm and CoP displacement changed with increasing frequency; oscillations of the CoP were not related to arm movements made at a slower frequency, but increasing arm movement frequency resulted in a shift of the CoP in the opposite direction to the arm amplitude.

Ustinova et al. (2004) speculated that it is possible that the counter-phase relationship between arm and CoP displacement during rapid movement resulted from reactive forces acting on the body following the motion of the arm and that the reactive force remained and resulted in the shift in the CoP in the direction opposite to the motion of the arm.

Conversely, during antiphase swinging it was found that arm and CoP under the ipsilateral foot remained relatively in phase as frequency increased (therefore contralateral CoP moved in the opposite direction with the opposite arm). This likely occurred since the effect of one arm is being counteracted by the other, keeping CoM displacement minimal.

4.10 General Discussion

Coordination of the human postural system and human movements in general, involves task specific assembly of several joints, limbs and muscles (Bernstein, 1967). Studies by Bardy et al. (1999; 2002) revealed that postural coordination transitioned spontaneously from an antiphase mode to an inphase mode at a critical oscillation frequency. However, there are other possible solutions that functionally emerge based on task demands (i.e. Abe & Yamada, 2001). Furthermore, complex biological systems have the ability to engage/disengage appropriate components according to their functional (contextual) requirement and to enhance efficiency and reduce energy costs (Bernstein, 1967, Kelso, 1995). Therefore, as a movement parameter is varied the degrees of freedom that are controlled within a coordinative structure can be spontaneously recruited and annihilated Bernstein (1967). For example, Buchanan and Horak (2001) showed that increasing the magnitude of a postural perturbation could lead to a gradual shift in the postural pattern. As the postural system undergoes a transition, it is recruiting (or suppressing) motion about individual joints. This gradual recruitment of lower body degrees of freedom allows for gradual transition in postural states without a loss of stability. In the case of inphase and antiphase arm swinging, the postural perturbation is self-generated which produces a complex pattern of predictive and reactive torques in the lower limbs. These additional constraints to the postural control system must also be compensated for and subsequently, additional degrees of freedom (i.e. hip rotations during antiphase, and upper body and trunk movements during inphase) are recruited.

4.11 Conclusion

Our results show that during a self-perturbation task the joints of the lower limbs may not be restricted to specific coordination modes. This further elucidates the idea that postural coordination is a complex process that is highly dependent on context specific demands.

During inphase arm swinging, joints of the lower limbs remained close to a zero degree phase relationship, with the hip largely contributing to shifting the center of mass to compensate for the perturbing arms. During the very fast frequency we suggest that degrees of freedom about the hip are annihilated and more degrees of freedom are recruited in the upper body (i.e. the trunk) to keep the center of mass within its base of support. It is likely that in antiphase arm swinging, more complex interactions are at play. Relative phase values and directional coupling analyses suggested that at lower frequencies there is more ankle involvement while there is an increased role of the hip during the very fast frequency. At higher arm swinging frequencies CoM is also displaced more in the ML direction due to increased rotations at the hips

In this study, subjects were able to predict the sensory consequences of their swinging arms, develop appropriated compensatory (and most likely anticipatory) kinematic responses, and when CoM amplitude increase during inphase swinging, subjects shift their CoP in the direction opposite to the arms. Being able to predict this consequence is essential for maintaining upright balance in the wake of perturbations (Earhart & Bastian, 2001). This study focused on young healthy subjects, but with aging, different control strategies may be implemented (Bugnariu & Sveistrup, 2006). Further studies should investigate age-related changes, and deficits exhibited by patients with balance related impairments, such as cerebellar patients. Bastian et al., (1996) report that individuals with damage to the cerebellum exhibit, increased postural sway, atypical responses to perturbations, and reduced control of equilibrium during movements of other body parts. These deficits occur as a result of a failure to make predictions about body position, and sensory consequences of their actions. Furthermore, in the case of patients with unilateral cerebellar damage, coordinating the responses about individual joints and joint segments could also become compromised (White & Balasubramaniam, 2008).

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Appendix I: Ethics Certificate

January 20, 2006

Dr. Ramesh Balasubramaniam
School of Human Kinetics
University of Ottawa
125 Université
Room 336
Ottawa, ON K1N 6N5

Mr. David White

Object: Interaction Between Standing Balance and Upper Limb Rhythmic Activity (file H 10-05-08)

Dear Researchers,

You will find enclosed the Health Sciences and Science REB ethical clearance for the abovementioned study.

During the course of the study, any modifications to the protocol or forms may not be initiated without prior written approval from the REB. You must also promptly notify the REB of any adverse events that may occur.

This certificate of ethical clearance is valid until January 20, 2007. Please submit an annual status report to the Protocol Officer in January 2007 to either close the file or request a renewal of ethics approval. This document can be found at: http://web9.uottawa.ca/services/rgessrd/ethics/application_dwn.asp

A copy of this approval will be sent to research services, if necessary.

If you have any questions, you may contact the undersigned at the number 562-5387.

Sincerely yours,

Rita D'Alessandro
Protocol Officer for Ethics in Research
For Dr. Daniel Lagarec, Chair of the Health Sciences and Science REB

HEALTH SCIENCES AND SCIENCE RESEARCH ETHICS BOARD

CERTIFICATE OF ETHICAL APPROVAL

This is to certify that the University of Ottawa Health Sciences and Science Research Ethics Board has examined the application for ethical approval of the research project entitled **Interaction Between Standing Balance and Upper Limb Rhythmic Activity (file H 10-05-08)** submitted by Dr. Ramesh Balasubramaniam (professor's research) who is supervising Mr. David White (4th year student) for this same project, both of the School of Human Kinetics, University of Ottawa. The Board found that this research project met appropriate ethical standards as outlined in the Tri-Council Policy Statement and in the Procedures of the University of Ottawa Research Ethics Boards, and accordingly gave it a Category 1a (approval). This certification is valid one year from the date indicated below.

________________________________	January 20, 2006
Rita D'Alessandro	Date

Protocol Officer for Ethics in Research
For Dr. Daniel Lagarec, Chair of the
Health Sciences and Science REB