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THE INTER- AND INTRA-FOOT COORDINATION DYNAMICS OF QUIET STANDING POSTURES

A Dissertation in

Kinesiology

by

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ABSTRACT

Human postural stance is inherently unstable and is a complex task involving the coordination and control of redundant degrees of freedom at each level of analysis (e.g., limbs, joints, muscles, etc.). In numerous investigations of the postural control system, it has been found that, the coordination and control of posture and balance is simplified by employing different strategies for postural stability. For example, strategies may utilize the ankle joint alone (i.e., inverted pendulum model), ankle-hip coordination (i.e., double inverted pendulum model) or multi-joint synergy (i.e., multi-linkage model). Even though the postural control mechanism has been comprehensively documented by investigations of EMG, kinematic and kinetic data, there are limited data on the investigation of the inter- and intra-foot coordination dynamics in quiet stance.

The primary focus of this dissertation is on the center of pressure (COP) foot coupling under the varying mechanical and task constraints induced by different stances. We investigated: 1) how foot position and asymmetrical body weight distribution interact to influence postural control and inter-foot coordination; 2) the effect of foot position and orientation on the body weight distribution and the foot coordination dynamics of standing postures; and 3) the influence of a shortened support surface together with its orientation on the COP coordination in quiet stance.

Three experiments were conducted that manipulated foot position, foot orientation and the base of support properties in quiet standing. The following conclusions have been reached from the analyses of the individual foot COPs and the coupling between COPs: 1) The time evolutionary properties of the inter-foot coordination dynamics revealed patterns of “stable” but “flexible” control of the postural

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system as a function of foot position and orientation. In particular, the staggered stance represents a “hybrid” blend of the properties of the side-by-side and tandem foot position in both linear and nonlinear analyses. 2) Foot position played a more important role than its orientation in channeling the inter- and intra-foot coordination dynamics. When the postural stance was challenged by the limitation of the base of support (side-by-side and tandem stances), the COPs in the unstable plane (inter-foot coordination) were predominantly involved in postural control. In contrast, when standing posture was not challenged by the support boundary (staggered stance), the COPs of the more loaded foot (intra-foot coordination) dominated postural stability. 3) When the shortened base of support was oriented along the horizontal axis, the four COP time series revealed a parallel contribution indicating an inter-dependence of the inter- and intra-foot coordination. When the shortened support area was positioned along the longitudinal axis, the COPs in the sagittal plane (inter-foot coordination) displayed a more significant contribution to postural stability.

Multistability exists at different levels of the motor control hierarchy (Braun and Mattia 2010; Kelso 2012). The “phase wandering” of the inter-foot coordination dynamics presented in Experiment 1 reflects the multistability properties of the postural control system. The phase synchronization of the COP_L—COP_R coupling characterizes the capability of the self-organizing system to retain its intrinsic attractive states (i.e., stability) whereas the phase transition specifies the escape or de-affiliation tendencies of the system from an attractive steady state. Overall, our findings support the proposition that postural control is achieved through the foot coordination dynamics and the asymmetrical body weight distribution.
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CHAPTER 1: INTRODUCTION

Background

Bipedal standing posture is inherently unstable because approximately two-thirds of our body mass is located two-thirds of the body height above the ground (Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998). To preserve postural stability, individuals need to actively align their trunk and head position, in addition, to maintaining the projection of the center of mass (COM) within the base of support boundaries (Massion, 1992; Horak & McPherson, 1996). Given the fact that there are multiple degrees of freedom at different level of analysis (e.g., limb segments, joints, muscles, etc.) along the axis of the body, muscle activities and joint motions have to be coordinated in certain ways to maintain postural orientation and equilibrium. The question of how to achieve postural stability in a redundant system, although it has been studied extensively in the past decades (Massion, 1994; Horak & Kuo, 2000; Alexandrov, Frolov, Horak, Carlson-Kuhta, & Park, 2005), is still unresolved.

The inverted pendulum model has dominated the past literature for a long period of time due to the fact that it simplified the control system to only one degree of freedom—the ankle joint (Winter et al., 1998; Winter, 1995; Loram & Lakie, 2002). However, debate still exists on whether the anti-gravity torque is generated through a passive or active process and whether the process involves continuous or intermittent control (van der Kooij & de Vlugt, 2007; Park, Horak, & Kuo, 2004; Peterka, 2002;

The double inverted pendulum model holds that postural control is realized by the coordination of the ankle and hip joints in the sagittal plane (Nashner & McCollum, 1985; Horak & Nashner, 1986). From the central programming perspectives, the ankle and hip strategies are the two dominant effector contributions to postural control. However, typical upright standing is maintained through a mixture of both strategies—the “hybrids” (Horak & Moore, 1993). The dynamic system approach has led to the proposal that the ankle—hip strategy is the order parameter or collective variable of the postural control system, that is driven by the amplitudes or the frequencies of the participants’ self-initiated perturbations (Bardy, Marin, Stoffregen, & Bootsma, 1999; Bardy, Oullier, Bootsma, & Stoffregen, 2002; Bardy, 2004).

Both the inverted and double inverted pendulum models tend to simplify the control system by reducing the number of the controlled elemental variables or individual joint degrees of freedom. Therefore, they cannot adequately deal with the multiple degrees of freedom problem of motor skill acquisition raised by Bernstein (1967). The multi-linkage model (Scholz & Schöner, 1999; Hsu, Scholz, Schöner, Jeka, & Kiemel, 2007; Scholz, Schöner, Hsu, Jeka, Horak, & Martin, 2007), on the other hand, makes use of the system’s motor redundancy instead of treating it as a problem (Bernstein, 1996; Latash, 2008; Latash, 2012). The coordination and coupling of the redundant degrees of freedom...
freedom facilitate postural orientation and equilibrium based on the demands of the task and environmental constraints (Latash, 2008; Newell, 1986; Scholz & Schöner, 1999).

In the past literature, the net center of pressure (COP_{Net}) is one of the most commonly analyzed parameters that indirectly represents postural sway and the control of human upright standing. The motion of the COP_{Net}, collected from a single force platform, is a controlled variable determined by the collective average of the COP of each foot (COP_L and COP_R), the body weight loading of each foot and mediated by the availability of the visual, vestibular and somatosensory information (Winter, Prince, Stergiou, & Powell, 1993; Riccio, 1993; Massion, 1994). The effects of the body weight loading and the inter- and intra-foot coordination dynamics can be revealed by requiring the participants to stand with each foot on a separate force platform. However, investigation of the foot COP coordination and its influence to the COP_{Net} has not been fully exploited. More specifically, we investigated the proposition that mechanical factors (e.g., foot position, foot orientation, reducing the size of the support area) play significant roles channeling the body weight loading, the coordination of COP_L-COP_R and thus the COP_{Net} pattern in the control of standing posture.

Focus of the Dissertation

The functional influence of the mechanical factors (e.g., asymmetrical foot position, body weight loading, and the area of the base of support) and their interaction, on the coordination dynamics of postural control has not been fully explored to date. In particular, the asymmetrical body weight loading serves not only as an independent
mechanical factor in postural control but also covaries with other factors such as foot position and orientation (Jonsson, Seiger, & Hirschfeld, 2005; Genthon & Rougier, 2005; Anker, Weerdesteyn, van Nes, Nienhuis, Straatman, & Geurts, 2008; Wang & Newell, 2012a, b). Although independent control of limb position and contact forces has been found in a series of animal postural control studies, the control mechanisms for foot position and weight loading on human inter- and intra-foot coordination dynamics have not been established (Lacquaniti, Taillanter, Lopiano, & Maioli, 1990; Lacquaniti & Maioli, 1992, 1994a, 1994b). Therefore, the main and interactive effects of these mechanical constraints on the foot coupling dynamics are the focus of this investigation.

These issues are investigated here by testing the hypotheses that: 1) there are different foot coordination dynamic structures for the upright postural stances; and 2) there are different qualitative inter- and intra-foot coordination patterns with the manipulation of the mechanical factors. In our previous studies, preliminary findings indirectly suggested the existence of the compensatory effect between the inter-foot coordination in the AP and ML directions (Wang, Jordan, & Newell, 2012; King, Wang, & Newell, 2012). In this dissertation, we focused on the examination of the inter- and intra-foot coordination dynamics in a direct way (Experiments 2 and 3).

The first question examined by Experiment 1 is whether foot position and intentionally induced asymmetrical body weight loading influence the inter-foot coordination in parallel or interdependently. It is expected that foot position and asymmetrical body weight loading main and interaction effects can be observed in both linear and nonlinear analysis reflecting that the covariation of these two mechanical constraints channels the postural stability and the postural control mechanism.
The 2nd experiment investigated the question of whether the inter- and intra-foot coordination dynamics and the body weight loading change qualitatively as a function of foot position and orientation. It is hypothesized that foot position plays a more important role than foot orientation in channeling the foot coordination dynamics due to the fact that it constrains both the area of the base of support and the loading of the feet.

The question investigated in the 3rd experiment is whether the width and orientation of the base of support influence the inter- and intra-foot coordination dynamics in the side-by-side foot position. It is hypothesized that the inter- and intra-foot coordination will increase as the postural instability increases through the change of the unstable board width and orientation. In particular, increased foot coordination dynamics is expected when the unstable board is oriented along the horizontal axis of the feet due to the fact that posture is less stable in the sagittal plane in the side-by-side stance.

To sum up, the COPNET pattern is mediated by the body weight distribution and the coupling of the individual COP of the feet. In other words, information on the body weight distribution and the COPL—COPR coordination cannot be separated apart through the investigation of the COPNET pattern. In particular, when humans adapt to different foot positions and foot orientations while standing, loading over the feet and the COPL and COPR patterns might not be similar. Therefore, understanding postural control requires an examination of the inter- and intra-foot coordination dynamics along with the body weight loading in different quiet stances.
CHAPTER 2: LITERATURE REVIEW

Even though postural control for healthy individuals is typically conducted subconsciously and requires limited effort or attention, human bipedal stance is inherently unstable given that approximately two-thirds of our body mass is located two-thirds of the body height above the ground (Winter et al., 1998). The two main functional goals of postural control are postural orientation and postural equilibrium (Nashner & McCollum, 1985; Massion, 1992; Riccio, 1993; Horak & McPherson, 1996). In other words, postural control includes not only the antigravity control under the static conditions but also the interface between perception and action via postural adjustments associated with supra-postural activities.

Postural orientation involves the active alignment of the trunk and head with respect to gravity, support surface, the visual surround and internal references. Postural orientation is established by somatosensory, vestibular and visual sensory information integration and re-weighting depending on task demands and the environmental context. Studies have shown that perception of verticality has multiple neural representations (Karnath, Ferber, & Dichgans, 2000). A healthy individual can adjust his postural orientation according to the internal and external representation of gravity and also the task constraints. For example, a person may orient the body perpendicular to the ground until the support surface tilts, and then they orient their postural to gravity.

Postural equilibrium, on the other hand, involves the coordination of movement strategies to stabilize the body center of mass against gravity during both self-imitated and externally triggered perturbations. Note that postural equilibrium in our context does not represent a motionless body configuration or a fixed steady state (fixed point) as the
traditional literature held (Murray, Seireg, & Sepic, 1975; Goldie, Bach, & Evans, 1989; Chaudhry, Findley, Quigley, Ji, Maney, & Sims, 2005). Instead, it is a set of dynamically and instantaneously varied equilibrium states migrating within the boundary of the support area [e.g., “stability cone” McCollum & Leen, 1989; Riccio, 1993; “instant equilibrium” Zatsiorsky & Duarte, 1999; “stable manifold” Morasso & Sanguineti, 2002].

Given infinite number of degrees of freedom, originated within the human body, and their combinations at each level of analysis (e.g, limbs, joints, muscles, motor neurons etc) that are available, the question is how the postural control system chooses one solution or a set of solutions out of the infinite to sustain postural equilibrium. This question can be viewed as a challenge of the broader motor redundancy issue introduced by Bernstein (1967): namely that coordination, control and skill in action is a reflection of a system with many redundant degrees of freedom. There are, however, several proposed solutions that can possibly parse the control mechanisms of the redundant postural system.

**Postural Control Strategy and Modeling**

**Inverted Pendulum Model**

The inverted pendulum model is one of the most well-documented models of postural control. It holds that the destabilizing torque due to gravity is counterbalanced by a restorative torque exerted by the ankle joints against the base of support. Since studies in the side-by-side quiet stance are predominantly focused on the control mechanism in the sagittal plane and treat the ankle joints’ motion identical, the inverted
pendulum model simplifies the postural control system to only one degree of freedom at the kinematic joint level (Riach & Starkes, 1994; Winter, 1995). There are, however, two main debates that exist in the current literature on: 1) whether the ankle restorative torque is generated through the action of a feedback control system (e.g., vision, vestibular and somatosensory feedbacks) or muscle tone, muscle stiffness itself; and 2) whether the CNS controls the restorative torque continuously or in an intermittent manner.

According to Peterka (2002), the restorative torque generated through the feedback process an “active torque”, whereas torque generated from the muscle elastic properties without time-delay is called a “passive torque”. So, the first question appears to be centered on whether postural control mainly involves an active process, passive activities or both. Zatsiorsky and Duarte (1999) proposed that, in the side-by-side stance, the CNS defines a set of attractive equilibrium points, deviation from which induces restoring forces in the horizontal plane. Given the fact that body sway is relatively slow with small amplitude motion, the restoring forces can be generated from the intrinsic stiffness of the ankle joint muscles. Winter et al. (1998) reported that the COP displayed an in-phase and 4 ms delayed motion relative to the COM. The limited time delay between the controlling and controlled variables supports the idea that the postural control system cannot afford any feedback or reactive process. Instead, muscle stiffness itself should be the predominant factor for postural correction.

Morasso and Schieppati (1999) argued, however, that the in-phase COP-COM relation is due to physical law whereby the COP-COM time delay cannot represent the delay of the control circuitry. Moreover, studies measuring the ankle stiffness have shown that the restorative torque generated from pure muscle stiffness could counter no
more than 90% of the destabilizing gravitational torque (Casadio, Morasso, & Sauguineit, 2005; Loram & Lakie, 2002). Therefore, postural control cannot be established by pure “passive” torque without the activation of the feedback process (van der Kooij & de Vlugt, 2007; Park et al., 2004; Peterka, 2002; Kiemel et al., 2011; Pasma et al., 2012).

However, a question that emerges is whether the “active torque” is generated by a continuous or intermittent feedback mechanism. This debate was phrased as “brute force” vs. “gentle taps” in the control of the unstable system by Morasso (2011). The simplest modeling taking into account of the feedback process by assuming no time-delay in the control circuitry (Barin, 1989; Park et al., 2004), whereas other models incorporated stochastic noise either at the sensory input level or the system’s error estimation state (van der Kooij & de Vlugt, 2007; Peterka, 2002; Kiemel et al., 2011). The continuous feedback model proposes that the upright stance can only be balanced at the “equilibrium state” (Winter et al., 1998). Any deviation from the state requires active correction whereby the system behaved like an instantaneously manipulated “servo loop”.

It has also been proposed that postural control is confined within a bounded stable manifold around the equilibrium state so that the feedback process can be conducted in an intermittent manner (Eurich & Milton, 1996; Bottaro et al., 2008; Asai et al., 2009; Loram et al., 2010; Morasso, 2011; Suzuki et al., 2013). A little deviation from the equilibrium might not require any feedback control process until the deviation approaches the equilibrium boundaries. This control scheme is event-driven, more metabolically robust and even not noise-sensitive.
Double Inverted Pendulum Model

The moving platform protocol evaluates control mechanisms with respect to the adaptive response of the postural system to external perturbations. A central finding from this approach is that pre-programmed stereotyped ankle and hip strategies are triggered when unexpected external perturbations to the postural system are applied (Nashner & McCollum, 1985; Horak & Nashner, 1986; Horak, Nashner, & Diener, 1990; Allum, Honegger, & Schicks, 1993). For example, exposing participants standing on a normal support surface to brief forward and backward horizontal surface perturbations elicited relatively stereotyped ankle joint muscles activation. The inverse activation pattern from the distal to proximal end of the lower extremity primarily utilizes the ankle muscle torque to compensate perturbation and restore equilibrium (ankle strategy). On the other hand, to successfully maintain balance while standing on a short support surface, the lower extremity muscle activates in a proximal to distal manner with the same activation latency as the ankle strategy, so that equilibrium can be resorted by the shear force generated from hip joints (hip strategy).

The ankle and hip strategies have recently been shown to be independent modes of movement, in which kinematic joint motions and muscle activation patterns have different spatial and temporal characteristics (Alexandrov et al., 2001a, b; Massion et al., 2004; Torres-Oviedo et al., 2007). For example, by translating the participants to multiple spatial directions, Torres-Oviedo and Ting (2007) observed two time varied muscle modes associated with the ankle and hip strategies. Bardy and colleagues (1999, 2002, 2004; Stoffregen, Pagulayan, Bardy, & Hettinger, 2000), based on a dynamic system approach, investigated the ankle-hip coordination dynamics during different
supra-postural tasks. By requiring the participants to track a forward-backward oscillating target displayed from the screen with their head motion, two relative phase patterns between the ankle and hip joint motion were observed—in-phase ($\phi_{rel} \approx 20 - 30^\circ$) and anti-phase ($\phi_{rel} \approx 180^\circ$). Other evidence also exists that the ankle-hip coordination is the variable predominantly involved in postural and supra-postural tasks (Vierordt, 1862; Buchanan & Horak, 1999; Haddad, Rietdyk, Claxton, & Huber, 2013).

The ankle-hip coordination has also been reported in a series of experiments involving quiet stance. Aramaki, Nozaki, Masani, Sato, Nakazawa, & Yano (2001) found a reciprocal relation between the COM acceleration and the angular acceleration of the ankle-hip coupling. They noted that, instead of the COM displacement, the COM acceleration could be the potential controlled variable in quiet stance due to the fact that the COM motion is not induced by a fixed-point attractor (Zatsiorsky & Duarte, 1999; Sasagawa, Ushiyama, Kouzaki, & Kanehisa, 2009). Other studies have been shown that the ankle joint motion plays a significant role when postural sway is less than 1Hz (Aramaki et al., 2001; Creath, Kiemel, Horak, Peterka, & Jeka, 2005; Bonne, Bardy, Fraisse, Ramdani, Lagarde, & Ramdani, 2009). Visual feedback increases the trunk motion and its flexibility; therefore, postural control strategy could shift from the inverted pendulum to the ankle-hip coordination when the participants stand with their eyes open (Accornero, Capozza, Rinalduzzi, & Manfredi, 1997). Suzuki et al. (2013) have furthered the investigation by an intermittent control model scheme.

Even though both inverted pendulum and double-inverted pendulum models uncovered a fruitful knowledge related to the postural control system, postural stability
cannot be realized by only involving one or two degrees of freedom (Bernstein, 1967). Given the fact that human body is a multi-linkage system, joints (e.g., knee, neck, etc.) other than the ankle and hip should also involve maintaining the projection of the COM within the base of support (Hsu et al., 2007). Moreover, these joints must be coordinated and compensated in certain spatial-temporal manner according to the demands of the task, environment and organism constraints (i.e., coordinative structure; synergy).

**Multi-Linkage Model**

Babinski (1899) was the first to observe and demonstrate multiple degrees of freedom coordination regarding postural control during quiet stance on an “asynergie cérébelleuse” patient. He observed the absence of a forward displacement of the hip and knee joints to preserve a fall, when the patient was asked to look upward by tilting his head and trunk backward. The multi-linkage model attributes the motor redundancy as a “blessing” instead of a “burden” to the control system that the system makes use of the degrees of freedom at different levels during the control of upright standing (Gelfand & Tsetlin, 1971; Gelfand & Latash, 1998). “Synergy” can be characterized by sharing, error compensation and adaptation (Scholz & Schöner, 1999; Scholz, Schöner, & Latash, 2000; Latash, 2008).

Postural orientation and equilibrium cannot be realized by the activation of the minimum number of degrees of freedom. Instead, it is the coordination and co-variation of the multiple redundant elements that facilitates postural stability. When one element of the motor system makes an error or fails to perform the task, a redundant element can
compensate for the error or take over the task. Given the fact that there are multiple resources channeling postural stability (e.g., the internal neuromuscular activations induced by the voluntary movements and the external environment), the task goal of maintaining the COM equilibrium cannot be realized unless motor coordination accommodates the task, environmental and organism constraints in an adaptive manner (Newell, 1986; Davids, Button, & Bennett, 2007).

Approaches revealing the synergetic behavior have been quantified via the analysis of variability structure of the elemental variables (e.g., the kinematic joint motion or EMG muscle activations) in relation to the performance variable (i.e., COM, head or trunk position). One method is the uncontrolled manifold (UCM) hypothesis (Latash, Krishnamoorthy, Scholz, & Zatsiorsky, 2005; Krishnamoorthy, Yang, & Scholz, 2005; Freitas, Duarte, & Latash, 2006; Hsu et al., 2007). According to this hypothesis, the variability of multiple joint motions or EMG coordination is partitioned into two subspaces: one subspace assists the stabilization of the effectors position (e.g., sustains the COM motion within the base of support); and a second sub-space induces performance error (e.g., leads the COM deviate from its equilibrium).

Therefore, the variability within the first sub-space is termed as the “good” variability in that co-variation within this sub-space reflects synergetic properties of the elemental variables, which stabilizes the upright posture. The variability within the second subspace is termed as the “bad” variability in that it deteriorates postural equilibrium and orientation. It is hypothesized that the controller does not eliminate redundant degrees-of-freedom but rather uses them to limit the amount of “bad” variability with respect to important performance variables.
Another approach applies the principal component analysis (PCA) to the multivariate kinematic joint motion in postural control (Hong & Newell, 2006; Pinter, van Swigchem, van Soest, & Rozendaal, 2008). The PCA itself is one of the statistical analyses holding the capacity of revealing the patterns and compressing the dimension of the multivariate data sets. The potential assumption of using PCA is that the central nervous system does not control the multiple degrees of freedom at different levels individually. Instead, it controls a set of the elemental variable synergies (e.g., muscle synergy or muscle mode, joint mode). A set of muscle synergies, for example, accounts for a wide range of muscle activation patterns during quiet stance or postural responses to the external perturbations. Within each synergy, elemental variables (e.g., muscle or joint motions) are coordinated and coupled in a performance error compensative way. However, conclusions should be cautiously drawn from the PCA results because, definitely, there is no direct mapping between the principal components with high eigenvalue and the synergies formed by elemental variables (Witte, Ganter, Baumgart, & Peham, 2010).

According to a three-segmental linkage model, Pinter et al. (2008) projected the 1st PC to the plane formed by either two of the ankle, knee and hip joints (e.g., ankle-hip plane, hip-knee plane and etc.). They found that the slopes of the projected principal axis are consistently larger than 0 indicating a multiple inverted pendulum model. Based on the PCA technique, Alexandrov, Frolov, & Massion (2001a, b) and Alexandrov et al. (2005) partitioned the postural linkage to three eigen-modes that each is dominated by the motion of one of the lower limb joints (i.e., ankle, knee and hip). In particular, Park et al.
(2004) showed adaptable capacities of these model gains channeled by biomechanical constraints through an active feedback control process.

Another perspective of the multi-linkage model specifies the spatial-temporal proximity of the COP\textsubscript{NET} to the stability boundary instead of the postural equilibrium (Slobounov, Slobounova, & Newell, 1997; Haibach, Slobounov, Slobounova, & Newell, 2007; Slobounov, Cao, Jaiswal, & Newell, 2009). The time-to-contact (VTC) approach, originated from the organism-environment-task interaction, emphasizes on information available for postural control instead of the departures from the equilibrium region. VTC is determined from the dynamics of the COP over each point in time of the COP time series and is based on the virtual trajectory of the COP should it continue to move with those same dynamics (Slobounov, Moss, Slobounova, & Newell, 1998). The stability boundary of the BOS can be measured both geometrically and functionally. The geometric BOS is usually determined by the area contained within the spatial boundaries of the position of the feet on the surface of support. In contrast, the functional BOS has been defined as the proportion of the AP dimension of the BOS utilized during sustained maximal forward and backward leaning. In young adults, VTC systematically decreases as the base of support in stance is reduced and as the speed of postural swaying movement is increased (Slobounov et al., 1998).

**Focus of the Dissertation**

According to the constraint-on-action theory (Newell, 1986) and the dynamic system approach (Kelso, 1995; Kulger & Turvey, 1987), this dissertation focuses on the
influence of task constraints on the inter-and intra-foot coordination dynamics of postural control. Instead of investigating synergies at the muscular-articular level (Alexandrov, Frolov, & Massion, 1998; Danna-Dos-Santos, Slomka, Zatsiorsky, & Latash, 2007; Pinter et al., 2008; Silva, Sousa, Pinheiro, Feraz, Tavares, & Santos, 2013), this dissertation explores the coordination and coupling of the COPs of each foot. The COP_{Net} (the COP trajectories recorded from a single force platform) has been treated as the controlling variable to maintain the projection of the COM within the base of support. Based on the inverted pendulum model, the motion of the COP_{Net} in the sagittal plane is primarily dominated by the ankle joint (Winter et al., 1998). From the multi-linkage point of view, COP_{Net} is driven by the activation and coordination of the redundant muscular-articular degrees of freedom of the system (Hsu et al., 2007).

Postural control has been extensively studied with the side-by-side upright stance, especially the control mechanisms and synergies in the sagittal plane. However, postural control involves coordination of both the sagittal and the frontal plane (Mochuzuki, Duarte, Zatsiorsky, Amadio, & Latash, 1999, 2006). Given the COP_{Net} is a composition of the COPs of each foot (COP_{L} and COP_{R}) and the body weight loading over the foot (Winter et al., 1993), the inter- and intra-foot COP coupling and the body weight distribution can be channeled by different task constraints (e.g., asymmetrical foot position, body weight loading, and the area of the base of support, and etc.). However, the functional influence of the mechanical factors and their interaction, on the COP coordination dynamics has not been fully explored. As a result, this thesis focuses on testing the following hypotheses: 1) there are different foot coordination dynamic structures for the upright postural stances; and 2) there are different qualitative inter- and
intra-foot coordination patterns with the manipulation of the mechanical factors of foot position and orientation in postural stance.

Experiment 1 investigated the interactive influence of foot position and intentionally induced asymmetrical body weight loading on the inter-foot coordination dynamics. We tested the hypothesis that the covariation of the two mechanical task constraints channels the postural stability and dynamics.

Experiment 2 examined the question of whether the inter- and intra-foot coordination dynamics and the body weight loading changes as a function of foot position and orientation in a qualitative manner. In the experiment, the asymmetrical body weight distribution is not an independent mechanical factor instead it covaries with foot position and orientation.

Experiment 3 investigated the interactive effect of the area and orientation of the base of support on the inter- and intra-foot coordination dynamics in a side-by-side stance. It has been shown that reducing the area of the base of support could effectively induce postural instability and further challenge the upright standing. However, limited investigation has been focused on how the shortened support area channels the inter- and intra-foot COP coupling.
CHAPTER 3: ASYMMETRY OF FOOT POSITION AND WEIGHT DISTRIBUTION CHANNELS THE INTER-LEG COORDINATION DYNAMICS OF STANDING

Abstract

The study of quiet standing has mainly been conducted in the foot side-by-side position with the assumption that the contribution of the lower limbs is structurally and functionally equivalent. The purpose of this study was to examine how the two mechanical factors of foot position and weight distribution interact to influence postural control and inter-leg coordination dynamics. Participants were required, while standing in either a side-by-side, staggered or tandem right foot forward position, to intentionally produce three different levels of weight distribution (50/50, 30/70 and 70/30) over the two feet. Our results showed that the interaction effects of the two mechanical constraints were represented in both linear and nonlinear analyses. The COP mean velocity was predominantly influenced by body weight distribution in the side-by-side stance, whereas foot position was more influential in the tandem stance. The nonlinear analysis showed that the least experienced postural condition (i.e., tandem stance with a 70/30 loading level) had the lowest number and total duration of COP_L–COP_R phase synchronization epochs in the AP direction that were compensated by “stable” coordination dynamics in the ML direction. The findings revealed that the staggered stance represents a “hybrid” blend of the properties of the side-by-side and tandem foot positions. Collectively, foot position and weight distribution interact to determine the stability and flexibility of inter-leg coordination dynamics in postural control.
Introduction

To sustain standing posture the neuromuscular control system must instantaneously support the passively unstable, multi-leveled body components (e.g., head, torso, limb segments, muscles etc.) against gravity to ensure the vertical projection of the center of mass (COM) is balanced within the base of support (Massion, 1994; Riccio, 1993; Nashner & McCollum, 1985). Although stable control of undisturbed standing is well practiced and subconsciously processed for healthy subjects, it may be quite an achievement for individuals with central nervous system or peripheral musculoskeletal pathologies. Patients adapt different foot positions and/or asymmetric body weight-bearing strategies when required to stand upright. But, the general role of foot position and asymmetric body weight distribution in human postural balance is not well established.

Upright standing in most previous studies has primarily been examined in the side-by-side position in which the lower limbs are aligned in parallel about hip or shoulder width apart assuming they are anatomically and functionally identical. Independent control of limb position and contact forces has been found in a series of animal postural studies, but previous experiments with humans have not yet fully investigated the control mechanisms of foot position and weight distribution on upright stance (Lacquaniti et al., 1990, 1994a, 1994b). Winter et al. (1993; Winter, Prince, Frank, Powell, & Zabjek, 1996) required participants to stand with different foot positions by loading their body weight evenly on the lower limbs and found that in side-by-side stance, COP_{NET} motion in the anterior-posterior (AP) direction is predominantly driven by ankle plantar- and dorsi-flexors and that motion in the medio-lateral (ML) direction is mainly
determined by a hip loading/unloading mechanism. On the other hand, the migration of COP\textsubscript{NET} in staggered and tandem stances is due to the combinations of the ankle and hip control mechanisms.

Wang et al. (2012) reported similar findings from an investigation of the coupling dynamics of the COP under each foot (COP\textsubscript{L} and COP\textsubscript{R}) in quiet standing. In side-by-side stance, ML motion was dominated by anti-phase coupling while the AP direction was dominated by in-phase coordination. In staggered stance, however, the in-phase coupling contributed to the coordination dynamics in both directions. While in tandem stance, only the ML direction displayed a consistent in-phase coordination pattern.

By gradually changing foot position in both ML and AP directions together with the angles between the toes and heels, Holbein & Chaffin (1997) and Kirby, Price, & MacLeod (1987) found that increased separation of the feet in a particular direction resulted in larger displacement of the center of gravity (COG) in that direction. In addition, the most COP\textsubscript{NET} (equivalent to the center of pressure collected from a single force platform) variation was in the extreme stances, such as the feet together or 45cm apart in the ML direction in comparison to a position of the feet being 15cm apart. However, in these studies, the body weight distribution was not the primary variable that was controlled. Participants were instructed to either maintain an evenly distributed body weight under the assistance of visual cues or stand in their own comfortable manner.

The fact that body weight-bearing plays an important role in postural control is evident in even short-term self-chosen foot side-by-side stance (Blaszczyk, Prince, Raiche, & Hébert, 2000; Prado, Castanharo, Vilela, & Duarte, 2010). It has been shown that increasing body weight-bearing asymmetry in side-by-side stance induced an
increase in $\text{COP}_{\text{NET}}$ velocity predominantly in the ML direction and the asymmetry control of postural sway velocity in favor of the more loaded foot in both directions (Anker et al., 2008). However, a larger increase of the COP amplitude has been observed for the less loaded compared to the more loaded foot (Genthon & Rougier, 2005). In another example, when instructed to load evenly in a tandem posture, only one out of 58 participants (26 young, 26 elder adults) could maintain that position for 30 s whereas most participants distributed roughly 60% of their body weight on their rear foot (Jonsson et al., 2005).

Patients with Parkinson’s disease (PD) have been reported to stand with an increasingly narrowed support surface with pronounced backward or lateral sway as PD becomes more severe (Selby, 1968; Dimitrova, Horak, & Nutt, 2004; Horak, Dimitrova, & Nutt, 2005) whereas patients with midline cerebellar syndromes are characterized as unable to stand in the Romberg position and perform tandem gait (Nashner & Grimm, 1978). Due to the skeletal alignment changes of the lower extremity, children with cerebral palsy develop crouch stance with on-toe posture, which predominantly relies on limb protraction/retraction mechanism to maintain balance instead of the commonly used ankle strategy in the sagittal plane (Gage, 1991; Burther, Woollacott, & Qualls, 1999; Ferdjallah, Harris, Smith, & Wertsch, 2002). As a result of impaired ankle proprioception and muscle weakness, weight-bearing asymmetry in favor of the non-paretic leg as well as increased spontaneous postural sway in the frontal plane are among the most characteristic consequences of incompletely recovered stroke patients (Eng & Chu, 2002; Geurts, de Haart, van Nes, & Duysens, 2005). Limb loading and COP pattern also change as a function of amputee location (e.g., below or above the knee joint) for
patients after unilateral lower extremity surgery (Summers, Morrison, & Cochrane, 1987; Talis, Grishin, Solopova, Oskanyan, Belenky, & Ivanenko, 2008). In addition, patients with diabetes apply various stance positions and asymmetric weight-bearing due to foot deformities, patterns of plantar pressure distribution, callus formation and ulceration (Bevans, 1992; Greene, Stevens, & Feldman, 1999; van Schie, Vermigli, Carrington, & Boulton, 2004).

The focus of the current experiment is the functional influence of foot position and weight distribution on healthy subject’s postural control and particularly the coordination dynamics of the interacting lower limbs. Our questions are: 1) whether foot position and asymmetric body weight distribution influence postural control in parallel or interactively; and 2) whether inter-leg coordination dynamics are channeled by these two mechanical factors. In order to answer these questions, the first objective of the present study was to investigate the role played by different foot positions and body weight distribution on upright quiet stance. The second objective was to investigate the inter-leg coordination dynamics, the coupling of the COP_L and COP_R, as a function of these two factors. To quantify the COP changes under each foot, two synchronized force platforms were used. From this perspective, we assumed that COP_L and COP_R are two interacting variables driven by the torso and limb motion of the degree of freedoms of the body in order to maintain the projection of the COM within the support surface. In this framework, the motion of the controlled variable (i.e., COM) is driven by the coordination dynamics of these two control COP variables.
Methods

Subjects and Apparatus

Eleven (5 female, 6 male; height 168.95± 2.85 cm, weight 66.79± 3.60 kg) right-footed subjects pre-screened by Waterloo Footedness Questionnaire—Revised (Elias, Bryden, & Bulman-Fleming, 1997), between the age of 25 and 35 years with no diagnosed musculoskeletal and balance pathology, volunteered to participate in this study. All subjects provided informed consent to this project that was approved by the Institutional Review Board (IRB) of Pennsylvania State University. The foot kinetic data were collected by two adjacent AMTI force platforms (Advanced Mechanical Technology Inc., OR6-5-1000, size 46×50 cm) at a sample rate of 100 Hz. The platforms were calibrated and synchronized for data collection.

Tasks and Procedures

Three upright postures were investigated in this study: feet side-by-side standing about hip width apart (SS), staggered (SR) and tandem positions (TR). In the non-side-by-side stances, the participants were instructed to position their right foot forward and in the tandem stance directly in front of the left foot (Figure 3-1). Hence, the coding of left/front foot in this paper indicates the left foot in the side-by-side stance and the front foot in the staggered and tandem stances while right/rear foot indicates the right foot in the side-by-side stance and the rear foot in the other postures.
Figure 3-1. Schematic of the foot position for the 3 quiet postural stances: SS—side-by-side, SR—staggered right foot forward and TR—tandem right foot forward. Foot distances are measured by the length between the forefoot heel and the rear foot’s big toe in the AP direction and the first metatarsophalangeal articulations between the left and right foot in the ML direction (landmarks: dash line; distances: solid line). These distances were (mean ± se): SS: ML 23.39±1.08 (cm); SR: ML 9.61±0.89 (cm), AP 4.67±0.82(cm); TR: AP 6.82±1.16(cm).

Each posture had 2 trials under different weight distribution levels: 50/50, 30/70 and 70/30. For example, 70/30 required the participants to partition 70% of their body weight on the left/ front foot and to partition 30% of the body weight on their right/ rear foot. Our previous studies showed that the participants tend to load 65-75% of their body weight on the rear foot when they were instructed to comfortably stand in staggered and
tandem postures (Wang et al. 2012). As a result, the 30/70 distribution level is the most comfortable body weight partition in these two postural tasks whereas the 70/30 level is the hardest one for participants to maintain their balance. Participants reported no preference on a particular weight distribution level in the side-by-side standing.

Table 3-1. Fz ratio (mean± SE) of the left/front and right/rear foot during the quiet stances

<table>
<thead>
<tr>
<th></th>
<th>50/50 (1.00)</th>
<th>30/70 (0.43)</th>
<th>70/30 (2.33)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Side-by-side</td>
<td>0.988±0.026</td>
<td>0.451±0.007</td>
<td>2.283±0.035</td>
</tr>
<tr>
<td>Staggered-R</td>
<td>0.995±0.012</td>
<td>0.442±0.003</td>
<td>2.316±0.028</td>
</tr>
<tr>
<td>Tandem-R</td>
<td>1.012±0.012</td>
<td>0.450±0.006</td>
<td>2.294±0.036</td>
</tr>
</tbody>
</table>

* Values within the parentheses represent the weight distribution requirement for each stance

Each participant’s body weight was recorded before the data collection and loading of the left/ front foot was monitored throughout each trial. There was a 10% tolerance for each weight distribution level: in the 50/50 level, the loading of the left/front foot was limited within 45-55% and in the 30/70 level, the loading was restricted to 25-35% on that foot. An auditory beep was provided if the participants’ loading ratio was outside the assigned tolerance range. Table 3-1 shows the vertical ground reaction force ratio of the left/ front foot to the right/ rear foot for each condition. Given different body weight distribution requirements, participants maintained their balance on each stance successfully throughout the trial. There were 2 trials at each of the 3 (foot position) × 3 (weight distribution) conditions. Each trial was 45 s in duration. The conditions were randomly assigned to the participants and the 2 trials at each
condition were blocked. A 5 s break was given between each trial of a condition and there was a 1 min break between conditions.

A piece of paper was spread flat on each force platform (one piece for each platform) and an outline of the participant’s feet was traced for each stance allowing foot position to be accurately repeated for each trial of a stance. For each postural stance the participant was asked to stand with their feet at a comfortable distance apart in the stance required. The mean ± SD of foot distance in both ML and AP directions for each stance are reported in the caption of Figure 3-1. The participants were instructed to stand with their arms folded in front of the chest and keep their knees straight during the testing. Data recording was initiated 5 s after the participants positioned themselves comfortably in the respective posture on the force platforms.

Data Analysis

Center of pressure

The initial and final 5 s of the force and moment data were removed from data analysis in order to avoid any transient contributions to the COP signals generated from either initiating the postural stance or fatigue at the end of the data recording. The raw kinetic data were down-sampled to 50Hz and then low-pass filtered by a 4th order double pass Butterworth filter with 6 Hz cut-off frequency. The COPs in the ML and AP directions were calculated independently for each limb (COP_L and COP_R). The COP_NET was derived by the equation of Winter et al. (1993).
**COP mean velocity**

COP velocity has been related to the amount of regulatory activity associated with postural stability measured by COP mean distance or root mean square distance (Hufschmidt, Dichgans, Mauritz, & Hufschmidt, 1980; Maki, Holliday, & Fernie, 1990). It is also frequency sensitive and more reliable in the clinical sense of the quantification of postural control than COP amplitude or variation (Geurts, Nienhuis, & Mulder, 1993; Goldie et al., 1989). As a result, COP mean velocity was used to quantify the linear properties of the COP trajectories in this study. COP mean velocity is the average velocity of the COP excursion in the ML and AP directions, respectively (Prieto, Myklebust, Hoffmann, Lovett, & Myklebust, 1996),

\[
MVELO_{AP} = TOTEX_{AP} / T
\]

\[
TOTEX_{AP} = \frac{\sum_{n=1}^{N-1} [AP[n+1] - AP[n]]}{N-1}
\]

where \(MVELO_{AP}\) is the COP mean velocity in the AP direction, \(TOTEX_{AP}\) is the total excursion of the COP path in the AP direction and \(T\) is the testing duration. Calculation of the COP mean velocity in the ML direction follows the same equations. Data analysis focused on the mean velocity of the COP \(L\), COP \(R\) and COP \(NET\) trajectories as a function of foot position and weight distribution in each individual direction.

**Coupling between COP\(L\) and COP\(R\)**

*Relative phase.* The coupling effects between COP\(L\) and COP\(R\) in both directions were quantified by the circular mean and standard deviation (SD) of the Hilbert
transformed relative phase (Fuchs, Jirsa, Haken, & Kelso, 1996; Rosenblum & Kurths, 1998; Mardia, 1975). The directedness of the COP_L—COP_R relative phase time series was tested by Rao’s spacing test and the non-unimodally distributed data (12 trials out of the total 432 trials, 2.8%) were removed from this analysis (Rao, 1976). It was interpreted that the circular mean of a relative phase $\Delta \varphi = \varphi_L(t) - \varphi_R(t)$ distribution around 0º represents an in phase coupling between COP_L and COP_R whereas a distribution around 180º represents an anti-phase coordination (Haken, Kelso, & Bunz, 1985; von Holst, 1973; Bardy, 2004). A low circular SD indicates a strengthened coupling whereas a large value reveals a more flexible coupling dynamics (Kelso, 1995). Oriana 2.02e (Kovach Computing Services, Anglesey, Wales) was used for circular statistic analyses.

**Phase synchronization detection.** Given the relatively large COP_L—COP_R circular SD shown from our previous study, further coupling information was investigated by analyzing the time evolutionary properties of the COP_L—COP_R relative phase $\Delta \varphi$ time series. These properties are characterized as the total number of phase synchronization epochs and total phase synchronization duration by our customized phase synchronization detection method and analyzed as function of foot position and weight distribution.
Figure 3-2. COP_{L} and COP_{R} of side-by-side (50/50) postural position in the AP direction under different analysis manipulations. The 1\textsuperscript{st} panel displays COP_{L} and COP_{R} time series; the 2\textsuperscript{nd} panel displays the Hilbert transformed COP_{L}—COP_{R} relative phase $\Delta\phi$; the 3\textsuperscript{rd} panel shows the unwrapped $\Delta\phi$ time series and the 4\textsuperscript{th} panel shows the phase synchronization epochs detected by the slope analysis (solid line: the start of a phase synchronization epoch; dotted line: the end of the phase synchronization epoch).

Figure 3-2 shows the time series of the COP_{L} and COP_{R} of the side-by-side (50/50) postural position under different analysis manipulations. The COP_{L}—COP_{R} coupling $\Delta\phi$ is wrapped within the range from $-\pi$ to $\pi$ revealing limited information of the phase synchronization (2\textsuperscript{nd} panel). As a result, a Matlab R2010b built-in unwrap function was applied to unfold $\Delta\phi$ time series. This manipulation corrects the phase angles $\Delta\phi$ (in radian) by adding multiples of $\pm 2\pi$ when absolute jumps between consecutive elements.
of $\Delta \phi$ are greater than or equal to the default jump tolerance of $\pi$. In this way, the $\Delta \phi$ time series can be “smoothed” and the “phase synchronization” epochs could be visually observed (3rd panel).

The “phase synchronization” epochs of each unwrapped $\Delta \phi$ time series were identified by a moving window slope analysis (4th panel). A window size of 120 samples (1750 samples in total for each trial) was pre-selected to define the minimum amount of time a phase synchronization epoch can be expected. The program then walks through the $\Delta \phi$ time series, calculates the slope angle for samples within the window. An angular threshold and circular SDs of different foot position $\times$ weight distribution conditions were introduced as the threshold on the Y-axis to detect the “flat” phase synchronization epoch for both ML and AP directions. The angular threshold was validated as the following: 18 trials (6 from SS-50/50, 6 from SR-30/70 and 6 from TR-70/30) were randomly picked from the original data pool and split into either the testing or the validation groups. Different angular thresholds were tested on the first group and validated on the second group in two ways: 1) KPSS stationary test was conducted for each detected phase synchronization epoch from the validation group (Hamilton, 1994); 2) paired t test was conducted for comparing the circular SD difference between the phase synchronization and phase transition epochs (other epochs except the locking ones). If the synchronization epochs are validated as stationary and the circular SDs of the synchronization epochs are significantly different from that of the transition epochs, the angular threshold can be accepted. In our data, we set $20^\circ (\pi/9)$ as the angular threshold.
**Statistical analysis**

The independent factors were foot position and body weight distribution. The dependent variables were the mean velocity of COP\textsubscript{L}, COP\textsubscript{R} and COP\textsubscript{NET}, circular SD of the COP\textsubscript{L}—COP\textsubscript{R} relative phase, the total number and duration of the COP\textsubscript{L}—COP\textsubscript{R} phase locking epochs in both ML and AP directions. Due to the discrete and non-normally distributed property of the number of phase synchronization epochs, this variable was transformed to global ranks to perform the standard parametric 2-way ANOVA for foot position× weight distribution interaction effect for both directions (Akritas, 1990)\textsuperscript{1}. For COP mean velocity, the average across the two trials was calculated. A 3 (foot position) × 3 (weight distribution) × 3(COP trajectory) fixed effect repeated measures ANOVA was performed in the ML and AP directions, separately. A 3 (foot position) × 3 (weight distribution) fixed effect repeated measures ANOVA was performed for other variables (circular SD and total phase synchronization duration) in both directions. The Bonferroni post-hoc test was used to determine the differences on all pairs of levels of independent variables \( \alpha \) was set at the level of 0.05 and any effects were statistically lower than that are reported. Where Mauchly’s test indicated violation of sphericity, the Greenhouse-Geisser estimate was used to provide a conservative test of ANOVA main and interaction effects.
Figure 3-3. Mean velocity ± SE for COP (COP\textsubscript{L}, COP\textsubscript{R} and COP\textsubscript{net}) in both ML and AP directions as a function of foot position and weight distribution.
Results

COP mean velocity

In the ML direction (Figure 3-3), there were significant main effects for foot position ($F_{1.245, 12.447} = 137.959, p<.001$), COP ($F_{1.112, 11.118} = 43.299, p<.001$) and interaction effects for foot position $\times$ COP ($F_{1.288, 12.883} = 9.646, p=.006$) and weight distribution $\times$ COP ($F_{1.621, 16.213} = 4.905, p=.027$). The tandem stance displayed the largest COP mean velocity ($M = 2.092$ cm/s) followed by the staggered and side-by-side stances (staggered $M = .685$ cm/s; side-by-side $M = .216$ cm/s); whereas the side-by-side stance displayed the lowest COP mean velocity. The COP_{NET} mean velocity was significantly larger than that of the COP_{L} and COP_{R} ($COP_{L} M = .948$ cm/s; $COP_{R} M = .779$ cm/s and $COP_{NET} M = 1.265$ cm/s). Post-hoc analysis of the COP trajectory revealed that the mean velocity of COP_{NET} was significantly larger in comparison to the COP under each individual foot in side-by-side and staggered postures. In tandem stance, the COP mean velocity of the rear foot showed the lowest value which was significantly different than the COP velocities of the front foot and COP_{NET}. In both 50/50 and 30/70 loading levels, COP_{NET} mean velocity displayed the highest values over the velocities of each individual foot. When partitioned 70% of the body weight on the front foot, the COP_{NET} velocity still displayed the highest value; whereas the rear foot COP mean velocity was the lowest in comparison with that of the front foot and COP_{NET}.

In the AP direction (Figure 3-3), the main effect of all three factors was significant ($foot position F_{2, 20} = 149.866, p<.001$; weight distribution $F_{2, 20} = 10.573$, 33
p=.001; COP trajectory $F_{2, 20}=31.262$, $p<.001$). There was a significant interaction for foot position $\times$ COP ($F_{1.845, 18.454}=16.496$, $p<.001$), weight distribution $\times$ COP ($F_{2.031, 20.310}=11.977$, $p<.001$) and foot position $\times$ weight distribution $\times$ COP ($F_{2.742, 27.242}=4.091$, $p=.018$). The results were similar to those in the ML direction in that the tandem stance displayed the largest COP mean velocity ($M=2.545$ cm/s) followed by staggered and side-by-side stances (staggered $M=1.661$ cm/s; side-by-side $M=.577$ cm/s). The 70/30 loading level induced the highest COP mean velocity ($M=1.719$ cm/s) in comparison with the other levels ($50/50 M=1.569$ cm/s; 30/70 $M=1.495$ cm/s). In the AP direction, the left/ front foot COP velocity ($M=2.057$ cm/s) was significantly larger than that of the right/rear foot and the COP$_{\text{NET}}$ (right/ rear foot COP $M=1.337$ cm/s; COP$_{\text{NET}}$ $M=1.389$ cm/s). Post hoc analysis showed that the front foot COP velocity was significantly larger as compared to that of the rear foot and the COP$_{\text{NET}}$ in both staggered and tandem stances. The left/ front foot COP velocity was the highest when the participants loaded evenly on both feet and distributed 70% on the left/ front foot; whereas the COP$_{\text{NET}}$ velocity was the lowest when they partitioned 30% on the left/ front foot. In both side-by-side and staggered stances, COP mean velocity of the more loaded foot displayed the largest value in comparison with the velocities of the less loaded foot COP and COP$_{\text{NET}}$. However, in tandem stance, the front foot COP velocity was significantly larger than that of the rear foot COP and COP$_{\text{NET}}$ in the three different loading levels. In the 50/50 weight distribution level, the front foot COP velocity of the staggered stance was also significantly larger than the velocity of the COP$_{\text{NET}}$. 

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Table 3-2. Circular mean angle (mean±SD) of COP<sub>L</sub>—COP<sub>R</sub> relative phase

<table>
<thead>
<tr>
<th></th>
<th>AP</th>
<th></th>
<th>ML</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Side-by-side</td>
<td>Staggered</td>
<td>Tandem</td>
<td>Side-by-side</td>
</tr>
<tr>
<td>50/50</td>
<td>0.007 ° ± 10.540 °</td>
<td>359.978 ° ± 8.544 °</td>
<td>179.562 ° ± 81.899 °</td>
<td>193.400 ° ± 59.586 °</td>
</tr>
<tr>
<td>30/70</td>
<td>17.396 ° ± 35.134 °</td>
<td>356.514 ° ± 24.737 °</td>
<td>313.787 ° ± 77.975 °</td>
<td>178.386 ° ± 56.040 °</td>
</tr>
<tr>
<td>70/30</td>
<td>344.786 ° ± 18.81 °</td>
<td>17.646 ° ± 60.671 °</td>
<td>167.22 ° ± 64.779 °</td>
<td>173.331 ° ± 59.185 °</td>
</tr>
</tbody>
</table>

- Each unit circle represents each individual weight distribution level
- Arrows indicate the circular mean angles for each foot position × weight distribution condition
**Coupling between COP\textsubscript{L} and COP\textsubscript{R}\**

*Relative phase.* Table 3-2 illustrates the circular mean of COP\textsubscript{L}—COP\textsubscript{R} relative phase as a function of foot position and weight distribution in both ML and AP directions. In the ML direction, the coupling of COP\textsubscript{L}—COP\textsubscript{R} is mediated by foot position such that COP\textsubscript{L} and COP\textsubscript{R} were in phase coupled in staggered and tandem postures and anti-phase coupled in side-by-side stance across all loading levels. In the AP direction, the coordination dynamics for the tandem stance is driven by body weight distribution as compared with the other postures. COP\textsubscript{L} and COP\textsubscript{R} were in-phase coupled in all stances when the participants loaded 30% of their body weight on the left/ front foot. However, in 50/50 and 70/30 loading levels, side-by-side and staggered stances displayed an in-phase coupling of the COP\textsubscript{L}—COP\textsubscript{R} whereas the tandem stances exhibited an anti-phase coupling of the two individual foot COPs.

Figure 3-4 shows the circular SD of the COP\textsubscript{L}—COP\textsubscript{R} relative phase as a function of foot position and weight distribution. In the ML direction, the main effect of foot position ($F_{1.182, 21.280} = 10.006, p= .003$) and weight distribution ($F_{2, 36}= 6.986, p= .003$) was significant. Circular SD of the tandem stance ($M= 41.977^\circ$) was significantly lower than that of the side-by-side ($M= 61.397^\circ$) and staggered ($M= 58.601^\circ$) stances. In addition, circular SD of the 70/30 level ($M= 62.797^\circ$) was the largest as compared to the other distribution levels (50/50 $M= 48.112^\circ$; 30/70 $M= 51.066^\circ$). No foot position × weight distribution interaction was found in this direction.
Figure 3-4. Circular SD ± SE of the COP$_L$—COP$_R$ relative phase time series in both ML and AP directions as a function of foot position and weight distribution.

In the AP direction, there were significant main effects for foot position ($F_{2,34} = 74.208$, $p<.001$) and weight distribution ($F_{2,34} = 11.414$, $p<.001$), and an interaction effect for foot position $\times$ weight distribution ($F_{4,68} = 11.209$, $p<.001$). The tandem stance displayed the largest circular SD of COP$_L$—COP$_R$ (M= 79.883°) followed by staggered and side-by-side stances (staggered M= 55.976°; side-by-side M= 45.679°). 70/30 distribution level (M= 70.622°) displayed the largest circular SD as compared to the other loading levels (50/50 M= 53.583°; 30/70 M= 57.332°). Post hoc analysis showed that circular SD of the 70/30 level was significantly larger than that of the 50/50 and 30/70 loading levels in the staggered stance.
Figure 3-5. Averaged (22 trials) total number of phase synchronizations (A) and total phase synchronization duration (B) in both ML and AP directions as a function of foot position and weight distribution. Data shown are in the mean and standard error format.
Phase synchronization. Figure 3-5 (A) illustrates the number of phase synchronization epochs as a function of foot position and weight distribution. In the ML direction, the main effect of foot position yielded an F ratio of $F_{2, 189} = 8.970, p < .001$, indicating that the number of phase synchronization epochs was significantly greater for tandem stance ($M = 121.908$ in rank) than for the other postures (side-by-side $M = 84.879$; staggered $M = 92.523$ in rank). The main effect of weight distribution yielded an F ratio of $F_{2, 189} = 8.186, p < .001$, indicating that the number of phase synchronization epochs was the lowest for 70/30 weight distribution level ($50/50 M = 107.841$; $30/70 M = 112.076$; 70/30 $M = 78.583$ in rank). The foot position $\times$ weight distribution interaction effect was non-significant ($F_{4, 189} = 2.066, p = .087$).

In the AP direction, there were significant main effects for foot position ($F_{2, 189} = 36.809, p < .001$) and weight distribution ($F_{2, 189} = 11.129, p < .001$), and an interaction effect for foot position $\times$ weight distribution ($F_{4, 189} = 3.471, p = .009$). The number of phase lock epochs of tandem posture ($M = 59.970$ in rank) was significantly lower than that of side-by-side ($M = 120.591$ in rank) and staggered stances ($M = 117.939$ in rank). And the number of phase synchronization epochs was the lowest for 70/30 loading level ($M = 77.765$ in rank) in comparison to the other two ($50/50 M = 111.129$; $30/70 M = 109.606$ in rank). Post hoc analysis showed that the number of phase synchronizations of 70/30 distribution level was significantly lower than that of 50/50 and 30/70 levels in staggered stance and significantly lower than that of 30/70 level in tandem position.

Figure 3-5 (B) displays the total phase synchronization duration as a function of foot position and body weight distribution in both directions. In the ML direction, the
main effect of weight distribution yielded an F ratio of $F_{1,281,12.810} = 8.657$, $p = .008$, indicating that the total of phase synchronization duration for the 30/70 level ($M = 25.498$) was significantly greater than for the 70/30 loading level ($M = 17.556$). The foot position $\times$ weight distribution interaction effect was also significant ($F_{4,40} = 2.877$, $p = .035$). Post hoc analysis showed that the total of phase synchronization duration of 70/30 distribution level was significantly lower than that of 30/70 level in staggered stance and significantly lower than that the other two levels in tandem position.

In the AP direction, the main effect of foot position yielded an F ratio of $F_{2,20} = 54.412$, $p < .001$, indicating that the total phase synchronization duration was significantly lower for tandem stance ($M = 7.825$) than for the other postures (side-by-side $M = 26.407$; staggered $M = 21.475$). The main effect of body weight distribution yielded an F ratio of $F_{2,20} = 20.896$, $p < .001$, indicating that the total phase synchronization duration was the lowest for 70/30 loading level ($50/50 M = 21.641$; $30/70 M = 20.797$; $70/30 M = 13.269$). The foot position $\times$ weight distribution interaction effect was significant ($F_{4,40} = 5.197$, $p = .002$). Post hoc analysis indicated that the total duration for the 70/30 distribution level was the lowest in comparison to the other loading levels in staggered position.

**Discussion**

The purpose of this study was to examine how the asymmetries of foot position and weight distribution influence postural control and the inter-leg coordination dynamics during quiet upright stance. Participants were instructed, while standing in either a side-by-side, staggered or tandem right foot forward position, to intentionally produce three
different levels of body weight distribution (50/50, 30/70 and 70/30). The interaction
effects of the two mechanical constraints were revealed in both linear and nonlinear
analyses. The COP mean velocity was predominantly influenced by body weight
distribution in the side-by-side stance, whereas foot position was more influential in the
tandem stance. The coupling of COP$_L$—COP$_R$ displayed a relative coordination pattern
indicating “flexible” and “stable” characteristics of postural control. The staggered
stance represents a “hybrid” blend of the dynamic properties of the side-by-side and
tandem positions in both analyses.

In side-by-side stance, our findings at 50/50 body weight level are comparable
with the COP displacement results of Winter et al. (1995), in which COP$_{NET}$ variation
was approximately the average of COP$_L$ and COP$_R$ in the AP direction and was the
largest in the ML direction due to the “canceling out” effect between COP$_L$ and COP$_R$.
With the asymmetric weight distribution (30/70 and 70/30 levels), the overall control of
postural sway velocity COP$_{NET}$ was significantly larger than that of each individual foot
in the ML direction. This finding is consistent with the observation of Winter and
colleagues that the ML control of COP$_{NET}$ is predominantly under the control of the
loading/ unloading of each limb. However, we did not find a COP velocity difference
between the more loaded and less loaded foot in the ML direction. One reason for this
could be that asymmetrical weight distribution induced postural sway in the ML was
compensated by the COP velocity in the AP, in which the asymmetry of control of
postural sway velocity increased with increasing weight distribution asymmetry in favor
of the more loaded foot followed by velocity of COP$_{NET}$ and COP of the less loaded foot.
This observation is comparable with the findings by Anker et al. (2008), who reported increased COP mean velocity in the AP direction with the increase of asymmetrical weight-bearing. They also proposed the reduced efficiency of hip load/unload mechanisms with an increase of the external load on one side. Even though the components of sway of these two orthogonal directions have been reported to be relatively independent (Winter et al., 1996; Winter, Patla, Ishac, & Gage, 2003), it has been observed that when postural instability is introduced to a standing task (e.g., reducing base of support at one dimension, depriving certain sensory feedback), sway constraints can be changed in the way to enhance the cross talk between the two orthogonal directions. For example, changes in the dimension of the support surface in one direction affect stability when the other dimension of the support surface was much smaller or kept constant (Day, Steiger, Thompson, & Marsden, 1993; Gatev, Thomas, Thomas, & Hallett, 1999; Mochizuki et al., 1999; King et al., 2012). Postural sway velocity is sensitive to the mean frequency of postural regulation, especially the higher frequencies within the COP excursion path that reflect the stabilizing torque generated from the ankle joints (Prieto et al., 1996; Geurts et al., 1993). Given that the hip loading/unloading mechanism in the frontal plane is constrained by the required body weight distributions, it is possible that the ankle joint of the more loaded foot generated higher torques in the sagittal plane to stabilize the side-by-side stance.

There are a few differences in the experimental designs of Anker et al. (2008), Genthon & Rougier (2005) and that of the current study. In Anker et al. (2008), the bandwidth of each loading level was 5% (e.g., 47.5%-52.5% for the 50/50 reference level) so that the participants had to rely on visual feedback to achieve the task goal, which
might induce larger COP amplitude variations in the ML direction. In contrast, in our experiment, the loading threshold provided to the participants was 10% so that they can more easily access the required loading level with minimal audio warning, especially in the side-by-side position (Table 3-1). In addition, we only applied two weight distribution levels in our study (i.e., 50/50 level requires the participants partitioned evenly on both leg whereas the 30/70 and 70/30 levels only varies the more loaded leg from the right-side to the left-side of the body) in comparison with 5 asymmetrical loading levels introduced by Anker et al. (2008) and 3 levels by Genthon & Rougier (2005). In other words, we did not channel the weight distribution to the extreme to reflect the characteristics of one-leg standing reported by Genthon & Rougier (2005). Genthon & Rougier (2005) proposed that there is a direct linkage between the activation of the plantar cutaneous mechanoreceptors under the foot and the COP trajectory variation. The somatosensory inputs from the sole of the unloaded foot are decreased when the participant gradually redistributes body weight to the other side and this introduces more variation of COP motion on the unloaded leg. Thus, the weight distribution levels in our study may not strongly have driven the activation of the mechanoreceptors under the less loaded support to the threshold of reducing activity to consequently provoke qualitative changes of the motor output.

In tandem stance, the COP mean velocity changes are opposite to the side-by-side stance in that the rear foot showed the lowest value that was significantly different than the sway velocities of the front foot and COP$_{\text{NET}}$ in the ML direction; and COP velocity of the front foot displayed the highest value in the AP direction. The COP mean velocity patterns are predominantly influenced by weight distribution for the side-by-side stance.
and influenced by foot position for the tandem position (Figure 3-3). With the change of foot position from the side-by-side to tandem stance, the challenge of postural instability was reflected by the gradually increased spontaneous body sway in addition to gradually decreased dimension of the support surface in the ML direction. Tandem stance is the least experienced foot position and balance in the sagittal plane is dominated by the hip/ankle flexion and extension whereas balance in the frontal plane is maintained through the ankle invertors/evertors and hip abductor/adductor (Day et al., 1993; Gatev et al., 1999). Shifting body weight forward in this position requires participants to stabilize their postural sway by gradually increasing the front foot’s ankle/hip torques in both ML and AP directions (King et al., 2012).

An interesting finding from our study was that the changes of postural sway velocity of the staggered stance, which revealed a “hybrid” blend of the patterns of the side-by-side and tandem stances (Figure 3-3). In our task, 30/70 is the most comfortable weight distribution level, similar to the side-by-side standing, postural sway velocity of the COP_{NET} in ML and COP of the more loaded foot in AP displayed the largest values. In contrast, 70/30 is the least experienced level for all participants, which illustrates a postural sway pattern similar to the tandem stance—the rear foot COP has the lowest mean velocity in ML whereas the front foot COP displays the largest value in the AP direction. When both limbs were evenly loaded, sway velocity of the COP_{NET} was still significantly larger than that of the COP_{L} and COP_{R} in ML whereas COP of the front foot showed the largest value in AP. In comparison with the side-by-side and tandem stances (i.e., postural sway is predominantly constrained at either the sagittal or frontal plane), mechanical constraint of the staggered position has more flexibility on both planes.
(Winter et al., 1993, 1996). When lower limbs are loaded close to the pattern of a natural stance (e.g., in a loading of the rear foot in expectation to perturbation or step initiation), the participants tend to rely on the control strategy applied for the side-by-side stance to achieve the task goal possibly because the side-by-side strategy is the well practiced and easily adapted one, and moreover still accessible. With the gradually increased loading on the front foot, demands on the front foot’s ankle/hip control gradually increase in both directions.

In the present study, we assumed that COP$_L$ and COP$_R$ are two interacting control variables driven by the torso and limb motion of the degree of freedoms of the body in order to maintain the vertical projection of the COM within the base of support. In other words, the body sway motion of COM is driven by the coordination dynamics of COP$_L$ and COP$_R$. Our results showed that coordination mode between the left and right leg in both directions is predominantly determined by foot position and the anatomical characteristics of the lower limb. In side-by-side stance, participants tend to load evenly on both limbs. Postural sway is primarily controlled by the instantaneous torque generated from the ankle plantar- and dorsi-flexors in the sagittal plane and the hip loading/unloading mechanism in the frontal plane (Winter et al., 1995, 1996).

Regardless of the body weight distribution, the inter-leg coordination is in-phase in the AP direction and anti-phase in the ML direction (Table 3-2). However, in staggered and tandem positions, when participants were required to stand in their natural and comfortable manner, they tend to load more on the rear foot (Jonsson et al., 2005; Wang et al., 2012). In other words, participants prefer to stand in the way to keep the intrinsic dynamics of the two feet asymmetric. In tandem stances, the inter-leg coordination
pattern is opposite to the side-by-side stance due to the aforementioned dimension changes of the support surface and the anatomical constraints of the lower limb. The coordination mode of the staggered position represents a transition from the side-by-side to tandem stance.

The strength of inter-leg coupling was indexed by circular SD of COP_L—COP_R relative phase (Figure 3-4). Our results showed relatively large circular SD for all foot position × weight distribution conditions indicating a relative coordination pattern of COP_L and COP_R. Relative coordination is different from absolute coordination in bimanual oscillation tasks (Haken et al., 1985; Kelso, 1995). Indeed, according to von Holst (1973), coordination reflects the complementary tendency of a self-organized system including maintenance tendency (the tendency for each individual oscillator to continue the rhythmic motion at its natural frequency—a competitive process) and magnet effect (the tendency that different oscillators tends to vibrate in the same tempo—a cooperative process). He interpreted relative coordination as the organization attainable when the competitive tendency was equal to or greater than the cooperative process. In this situation, the system was not fixed at a certain attractor or phase relation (even though the most attended phase relations were still proximate to those of abisolution coordination) instead phase wandering and drifting occurs (Kelso, DelColle, & Schöner, 1990; Kelso, 1995).

Comparatively, when the postural difficulty was gradually increased across the side-by-side to tandem stances, the coordination dynamics of the two lower limbs displayed a strengthened coupling in the ML direction whereas a weakening relation in the AP direction. In other words, the two orthogonal directions were compensated in the
way that the maintenance tendency from one direction, which challenges the stability of the self-organized system, can be compensated by a fairly strong magnetic trend from the other direction. Interestingly, the coupling strength of the staggered stance, again, displayed a “hybrid” blend of the patterns of the two extreme postures, in which circular SD patterns at 30/70 and 50/50 levels was similar to the side-by-side stance and at 70/30 level was close to the tandem foot position. The foot position × weight distribution interaction effects are more prominent at certain standing posture and influenced by the dimension of the support area.

The phase wandering was quantified by the time evolutionary properties of the COP_L—COP_R coupling as the number and total duration of the phase synchronization epochs. Wang & Newell (2012b) proposed that the phase synchronization epochs reflect the capability of the self-organized system to retain its intrinsic attractive states (i.e., the COP_L—COP_R coordination dynamics) at certain phase relation during postural control (Table 3-2). However, the control system does not hold the steady state for long periods of continuous time due to the relative coordination properties of COP_L—COP_R coupling. Instead, one foot frequently leads off to break up the current state, opening the system to access information and/or explore the boundaries of the intrinsic attractor dynamics. The alternation of the phase synchronization and transition continuous throughout the trial indicates “stable” and “flexible” postural control dynamics. As a result, the strengthened coupling of COP_L—COP_R, or the magnetic trend of the system, can be indexed by longer total phase synchronization duration.

Indeed, our findings indicate that tandem stance and staggered position with 70/30 loading level had significantly weakened coordination dynamics (indexed as larger
circular SD), less number phase synchronization epochs (the $\text{COP}_L—\text{COP}_R$ relative phase time series tends to drift away from the intrinsic attractor and never “stabilized” at certain phase level) and shorter total synchronization duration as compared with the other conditions in the AP direction (Figure 3-5). The maintenance tendency or the “flexible” coordination dynamics in this direction, however, are compensated by a relatively more “stable” pattern in the ML direction. Our findings on inter-leg coordination dynamics indicated again the existence of a “tight” linkage between ML and AP postural control mechanisms which can be enhanced as a function of the change of foot position (Mochizuki et al., 1999; King et al., 2012).

In summary, the asymmetry of foot position plays a primary role and induces qualitative changes in postural control and inter-leg coordination dynamics. The foot position $\times$ body weight distribution interaction is predominantly represented in the staggered stance displaying a “hybrid” blend of both linear and nonlinear properties of the side-by-side and tandem stances. Our results reveal that the ML and AP motions in the two orthogonal directions are not absolutely independent; instead, their cross talk can be enhanced and channeled as a function of the asymmetry of foot position. The coupling of $\text{COP}_L—\text{COP}_R$ represents a relative coordination dynamic pattern characterized as phase wondering and drifting. The organization of the control system is a blend of maintenance and magnetic tendency of $\text{COP}_L$ and $\text{COP}_R$, which indicates ”flexible” and “stable” properties of the system.
CHAPTER 4: THE EFFECTS OF FOOT POSITION AND ORIENTATION ON INTER- AND INTRA-FOOT COORDINATION IN STANDING POSTURES: A FREQUENCY DOMAIN PCA ANALYSIS

Abstract

We investigated the effect of foot position and foot orientation on asymmetrical body weight loading and the inter- and intra-foot coordination dynamics of standing postures. The participants were instructed to stand with the feet side-by-side and in staggered and tandem positions with the right foot oriented at different angles (30°, 60° and 90°). The results showed that the participants naturally loaded more on their left foot when positioned with their right foot forward. As the right foot was gradually oriented from 90° to 30°, they loaded a significantly large proportion of their body weight on their left foot. Foot position played a more important role than foot orientation in channeling the inter- and intra-foot coordination dynamics due to the fact that it constrains both the area of the base of support and the loading of the feet. In particular, when postural stance was challenged by the limitation of the base of support, the COPs in the unstable plane (inter-foot coordination) had larger factor weightings. In contrast, when standing posture was not challenged by the base of support boundary, the COPs of the more loaded foot (intra-foot coordination) dominated foot coordination in postural control. These findings show that the mechanical constraints of foot position and orientation interact to channel the inter- and intra-foot coordination dynamics of standing postures.
Introduction

In postural control, a destabilizing torque due to gravity must be counter-balanced by a corrective torque exerted by the feet against the base of support (Winter et al., 1998; Massion, 1992). As such, postural sway is typically described as the motion of the center of mass (COM) while the center of pressure (COP\textsubscript{NET}) that represents the position of the weighted average of the collective pressure over the support surface indicates the location of the ground reaction force (Winter et al., 1998). The collective COP\textsubscript{NET} incorporates the contribution of the ground reaction force and the COP of each foot (Winter, 1995). However, information on the body weight distribution and the left and right foot COP (COP\textsubscript{L} and COP\textsubscript{R}) coordination cannot be separated apart through the investigation of the COP\textsubscript{NET} pattern. In particular, loading over the feet and the COP\textsubscript{L} and COP\textsubscript{R} patterns might not be similar when humans adapt to different foot positions and foot orientation while standing (Mizrahi & Susak, 1989; Lacquaniti & Maioli, 1994; Wang et al., 2012). Consequently, understanding postural control requires an examination of the inter- and intra-foot coordination dynamics along with the body weight loading in different quiet stances.

Previous studies have shown the independence of the sagittal and frontal plane COP\textsubscript{NET} in the side-by-side stance (Kapteyn, 1973). Indeed, most postural studies have examined the COP\textsubscript{NET} motion only in the sagittal plane probably because it has been viewed as less stable or due to the assumption that postural control strategies in these two planes are identical (e.g., inverted pendulum model—Winter et al., 1998). Evidence exists, however, that there is inter-dependence of the spontaneous COP\textsubscript{NET} sway in both planes, particularly when postural stability is challenged by reducing the area of the base.
of support (Aruin, Forrest, & Latash, 1998; Mochizuki et al., 1999, 2006; Freitas et al., 2006). At the muscular-articular level, it has been shown that there is a direct linkage between the COP\textsubscript{NET} patterns and the lower limb muscle activations (Winter et al., 1998; Casadio et al., 2005; Loram & Lakie, 2002). However, recent investigations suggest that the high firing rate of the lower limb muscle occurs when the upper body leaning or being transported towards the directions other than the anterior-posterior (AP) and medial-lateral (ML) of the transvers plane (Henry, Fung, & Horak, 1998; Imagawa, Hagio, & Kouzaki, 2012). Nevertheless, how the coordination of the COPs under each individual foot (i.e., COP\textsubscript{L}-AP, COP\textsubscript{R}-AP, COP\textsubscript{L}-ML and COP\textsubscript{R}-ML) assists balance has not yet been fully explored.

The asymmetrical body weight loading serves not only as an independent mechanical factor in postural control (i.e., by requiring the participants intentionally load unevenly on the feet) but also can be a dependent variable that covaries with other mechanical factors or task constraints (Jonsson et al., 2005; Genthon & Rougier, 2005; Anker et al., 2008; Wang & Newell, 2012a, b). Anker et al. (2008) reported an overall COP velocity increase when participants are required to load most of their body weight on one side of the body when standing in a side-by-side manner. Evidence exists that people prefer to load evenly in the side-by-side stance whereas for a staggered or tandem foot position load more on the rear foot (Jonsson et al., 2005; King et al., 2012). The asymmetrical body weight loading channels the sensitivity of the cutaneous receptors under the sole of the more loaded foot thus could increases muscle stiffness of the limb (Genthon & Rougier, 2005; Reynolds, 2010; Sozz, Honeine, Do, & Schieppati, 2013).
Varying foot position has been hypothesized to selectively activate specific lower limb muscles so that different foot positions can be applied to strengthen the knee joint after injury (Signorile, Kacsik, Perry, Robertson, & Williams, 1995). Clinically, physicians have required patients adapt to the Romberg tandem stance to provoke instability (Hu & Woollacott, 1994). However, when required to stand in a self-preferred comfortable manner, a high degree of between-subject variability is present in both standing width and foot angle for healthy individuals (Mcllroy & Maki, 1997). By gradually increasing the distance between the feet in the frontal plane, it has been observed that COP\text{\textsubscript{NET}} variation decreases in the ML direction along with the compensated increase in its orthogonal AP direction (Kollegger, Wober, Baumgartner, & Deecke, 1989; Day et al., 1993). Other studies, however, have shown that the most COP\text{\textsubscript{NET}} variation is in the extreme stances, such as the feet together or 45 cm apart in the lateral direction in comparison with a position of the feet being 15 cm apart (Kirby et al., 1987; Holbein & Chaffin, 1997). Watanabe, Takeya & Baratto (1979) observed that the mean value of COP path in both AP and ML directions is identical when the feet are positioned in 30° fan-shape.

The focus of the current study was to investigate the role of foot position and orientation on the inter- and intra-foot coordination dynamics and body weight loading of the feet in quiet standing. We investigated three standard foot positions (i.e., side-by-side, staggered and tandem stances) with the manipulation of the participants’ right foot orientation (i.e., 30°, 60° and 90°). Therefore, the body weight distribution in this study is a dependent variable that could be influenced by the interactive effects of foot position and orientation.
Previous studies have shown the pairwise coordination of two-COP time series evolves across several time scales and involves sequential time lagged covariations (Kinsella-Shaw et al., 2011; Wang & Newell, 2012b; King et al., 2012). However, it is the instantaneous dynamics of the four COPs together channel the COP\textsubscript{NET} patterns in quiet stance to accommodate different mechanical constraints. The sequential time scales exist not only within each COP time series but also in-between these COPs. Given the aforementioned properties of the data set (including four time invariant COPs), we used a multivariate approach (a frequency domain PCA analysis—$f$PCA) to reveal the inter- and intra-foot coordination as a function of foot position and orientation (for a brief introduction on the $f$PCA, see appendix; for details, see Molenaar, 1985, Molenaar, Wang, & Newell, 2013). As a result, the analyses were not isolated as they typically have been to a single dimension (AP or ML) or the AP and ML being considered independently.

The $f$PCA compresses the dimension of the multivariate data set in determining the number of principal components while accommodating the problem of sequential lead/lags in the time series (Anderson, 1963; Molenaar, 1985; Molenaar et al., 2013). This contrasts with the time domain PCA analysis (Daffertshofer et al. 2004), where determination of the principal components does not take into account the time lead/lag relations and thus induce a biased estimate. The $f$PCA technique also affords, at each individual principal component, an estimate of the collective degree of coupling of the four COP time series (that represented by the factor weighting of each COP time series) across certain frequency range. For example, if the weightings of the four COPs are equivalent at the 1\textsuperscript{st} principal component, then the conclusion can be made that the four COPs displayed an equal contribution to the total variance interpreted by PC1. If, on the
other hand, the factor weighting of the COP$_L$-ML is the smallest whereas that of the COP$_R$-AP is the largest, the interpretation would be that these two COP time series had the least and the most contribution to the 1$^{st}$ PC, respectively. Nevertheless, the factor weighting of each COP time series is not a fixed or an averaged parameter, instead they change as a function of frequency.

**Methods**

**Subjects and Apparatus**

Twelve right-footed healthy subjects (foot preference determined by the Waterloo Footedness Questionnaire, see Elias et al 1997), four female and eight male [mean age 29 yr (range 25-35 yr); mean height 169.3 m (range 155.5-183.0 m); mean weight 65.6 kg (range 45-85 kg)], participated in this study. All subjects provided informed consent for the present study that was approved by the Institutional Review Board (IRB) of Pennsylvania State University. The foot kinetic data were collected by two adjacent AMTI force platforms (Advanced Mechanical Technology Inc., OR6-5-1000) at a sample rate of 100 Hz. The platforms were calibrated and synchronized for data collection.

**Tasks and Procedures**

There were 3 foot positions tested in this experiment: feet side-by-side standing about hip width apart (side-by-side), staggered (staggered-R) and tandem (tandem-R) stances each with the right foot forward. The participants were allowed to choose their
own comfortable distance for the feet to be apart. In the tandem stance, the participants were instructed to position their right foot directly in front of the left foot. In addition, the orientation of the right foot was restricted to 90°, 60° and 30°, whereas the left foot was instructed to face forward throughout all conditions (Figure 4-1). During the testing, the participants were required to hang their arms beside the trunk and keep their knees straight. The task requirement was to stand as still as possible. The participants were free to adjust their body weight distribution on the feet in a preferable manner before the data collection. Once the data collection started, the participants were not allowed to shift their body weight distribution any more.

There were 2 trials at each of the postural stances (foot position × foot orientation). Each trial was 60 s in duration. Data recording was initiated 5 s after the participants were positioned comfortably in the respective posture on the force platforms. The postural conditions were randomly assigned to the participants and the 2 trials at each postural condition were blocked. A 10 s break was given between each trial of a condition.
Figure 4-1. Schematic of right foot orientation in: A) side-by-side; B) staggered and C) tandem foot positions.
Data Analysis

The raw kinetic data were filtered with a 4\textsuperscript{th}-order 6-Hz low-pass zero-lag Butterworth filter. The COP signals in both AP and ML directions were derived and centered independently\textsuperscript{3}.

Ground reaction force ratio

The ground reaction force ratio (Fz ratio) of the left to the right foot was calculated:

$$Fz\% = \frac{F_{z\text{Left}} - F_{z\text{Right}}}{F_{z\text{Left}} + F_{z\text{Right}}} \times 100\% \quad (1)$$

where $F_{z\text{Left}}$ and $F_{z\text{Right}}$ represent the ground reaction force of the left and right foot, respectively. The ratio will have a value ranges from -1 to 1. Note that when the body weight is evenly distributed on the feet, the ratio will be 0; when the body weight is asymmetrically distributed on the left foot, the ratio will be larger than 0; otherwise, the ratio will be less than 0.
Figure 4-2. Example of group averaged COP power spectrum of: A) the side-by-side; B) staggered and C) tandem stance all with the right foot oriented at 60°.
**Frequency domain principal component analysis (fPCA)**

The fPCA provides the factor weightings spectra for the four COPs (at each principal component). According to the length of our multivariate data set, the frequency can be ranged from 0.0167 Hz to 50Hz (3000 frequency bins). However, most of the COP power concentrates at the very low frequencies (< 1Hz, for a review see Winter 2005). Therefore, an observation of the COP power spectrum to localize the frequency range capturing most of the signal power is the first step conducting the fPCA. Figure 4-2 showed the Fast Fourier Transformed COP power spectra (0.050-1.833 Hz) of three side-by-side postures with the frequency resolution of 0.0167Hz. The first two frequency bins past the DC component (0.0167 and 0.0334Hz) were removed from the power spectra and also the further fPCA analysis due to the fact that they represent events that occur every 60 and 30 s, and probably do not provide significant information about the postural control system (Prieto et al. 1996).

On the COPs’ power spectra, two frequency ranges (Low and High) was determined by the frequencies corresponding to 50% and 95% of the total power (f50, f95). Table 4-1 showed the group-averaged f50, f95 power frequency of the four COP time series at different postural conditions. Therefore, the averaged frequency ranges for different foot positions were: side-by-side: Low 0.050-0.150Hz, High 0.167-0.98Hz; staggered-R: Low 0.050-0.200Hz, High 0.217-1.200Hz; tandem-R: Low 0.050-0.250Hz, High 0.267-1.550Hz.

Accordingly, the percentage of total variance interpreted by the first two PCs and the factor weightings of the COP time series from the fPCA was averaged and reported across these two frequency ranges that each covered roughly 50% of the total power.
spectrum of a particular foot position. Table 2 shows the mean percentage of the total variance explained by PC1 and PC2 as a function of postural stance at both low and high frequency ranges. PC1 represented the largest proportion of the total variance of the multivariate data set (i.e., 72.45—87.43%). As a result, statistical analyses on the four COP factor weightings were conducted only for PC1.

**Statistical analyses**

A 3 (foot position) × 3 (foot orientation) fixed effect repeated measures ANOVA was conducted for Fz ratio. A series of multivariate ANOVA (MANOVA) were conducted with the COP factor weightings of PC1 as the dependent variables, and with postural stance and foot orientation as the independent variables at each frequency range, respectively. A two-way univariate test was performed afterwards to determine the foot position × COP and foot orientation × COP interaction effects, individually, at both low and high frequencies. The LSD post-hoc analysis was used to determine the differences on all pairs of levels of independent variables. In all analyses, only effects that were statistically significant at p < 0.05 are reported (a Bonferoni adjustment was conducted on the significant level for pairwise comparisons). In the case that homogeneity of the covariance matrices was violated, Pillai’s trace estimate was used to provide a conservative test of MANOVA main and interaction effects.
Table 4-1. Group averaged f50, f95 power frequency (mean ±se) as a function of posture, foot position and frequency epoch

<table>
<thead>
<tr>
<th></th>
<th>COP&lt;sub&gt;L&lt;/sub&gt;-AP</th>
<th>COP&lt;sub&gt;R&lt;/sub&gt;-AP</th>
<th>COP&lt;sub&gt;L&lt;/sub&gt;-ML</th>
<th>COP&lt;sub&gt;R&lt;/sub&gt;-ML</th>
<th>COP&lt;sub&gt;L&lt;/sub&gt;-AP</th>
<th>COP&lt;sub&gt;R&lt;/sub&gt;-AP</th>
<th>COP&lt;sub&gt;L&lt;/sub&gt;-ML</th>
<th>COP&lt;sub&gt;R&lt;/sub&gt;-ML</th>
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</thead>
<tbody>
<tr>
<td><strong>Side-by-side</strong></td>
<td></td>
<td></td>
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<td></td>
<td></td>
</tr>
<tr>
<td>90º</td>
<td>0.123±0.012</td>
<td>0.139±0.015</td>
<td>0.145±0.017</td>
<td>0.159±0.017</td>
<td>0.840±0.052</td>
<td>0.848±0.056</td>
<td>1.270±0.164</td>
<td>1.200±0.124</td>
</tr>
<tr>
<td>60º</td>
<td>0.153±0.013</td>
<td>0.132±0.012</td>
<td>0.160±0.018</td>
<td>0.140±0.013</td>
<td>0.919±0.051</td>
<td>0.812±0.051</td>
<td>1.153±0.083</td>
<td>0.890±0.057</td>
</tr>
<tr>
<td>30º</td>
<td>0.165±0.013</td>
<td>0.158±0.014</td>
<td>0.178±0.019</td>
<td>0.160±0.015</td>
<td>0.853±0.054</td>
<td>0.833±0.052</td>
<td>1.133±0.065</td>
<td>0.915±0.051</td>
</tr>
<tr>
<td><strong>Staggered</strong></td>
<td></td>
<td></td>
<td></td>
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<td></td>
<td></td>
</tr>
<tr>
<td>90º</td>
<td>0.200±0.013</td>
<td>0.206±0.014</td>
<td>0.197±0.012</td>
<td>0.187±0.018</td>
<td>1.076±0.061</td>
<td>1.291±0.065</td>
<td>1.076±0.054</td>
<td>1.343±0.086</td>
</tr>
<tr>
<td>60º</td>
<td>0.201±0.022</td>
<td>0.157±0.017</td>
<td>0.213±0.021</td>
<td>0.154±0.013</td>
<td>1.138±0.049</td>
<td>1.283±0.080</td>
<td>1.162±0.054</td>
<td>1.103±0.063</td>
</tr>
<tr>
<td>30º</td>
<td>0.212±0.018</td>
<td>0.194±0.017</td>
<td>0.187±0.016</td>
<td>0.185±0.016</td>
<td>1.219±0.052</td>
<td>1.395±0.073</td>
<td>1.140±0.064</td>
<td>1.203±0.065</td>
</tr>
<tr>
<td><strong>Tandem</strong></td>
<td></td>
<td></td>
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<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>90º</td>
<td>0.172±0.013</td>
<td>0.331±0.034</td>
<td>0.271±0.021</td>
<td>0.319±0.029</td>
<td>1.531±0.101</td>
<td>1.856±0.127</td>
<td>1.662±0.095</td>
<td>1.838±0.128</td>
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<tr>
<td>60º</td>
<td>0.197±0.021</td>
<td>0.263±0.028</td>
<td>0.258±0.024</td>
<td>0.257±0.026</td>
<td>1.415±0.059</td>
<td>1.604±0.083</td>
<td>1.665±0.078</td>
<td>1.547±0.070</td>
</tr>
<tr>
<td>30º</td>
<td>0.155±0.019</td>
<td>0.238±0.026</td>
<td>0.221±0.025</td>
<td>0.226±0.025</td>
<td>1.231±0.064</td>
<td>1.438±0.074</td>
<td>1.422±0.068</td>
<td>1.372±0.071</td>
</tr>
</tbody>
</table>
Table 4-2. Mean percentage of variance (mean ± SE %) interpreted by PC1 and PC2 as a function of postural stance and foot orientation

<table>
<thead>
<tr>
<th></th>
<th>Low frequencies</th>
<th>High frequencies</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Side-by-side</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>90° PC1</td>
<td>83.05±2.33</td>
<td>81.28±2.10</td>
</tr>
<tr>
<td>90° PC2</td>
<td>14.21±1.91</td>
<td>15.19±1.83</td>
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<tr>
<td>60° PC1</td>
<td>84.70±1.62</td>
<td>82.37±2.61</td>
</tr>
<tr>
<td>60° PC2</td>
<td>12.74±1.59</td>
<td>13.70±2.00</td>
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<tr>
<td>30° PC1</td>
<td>87.21±2.93</td>
<td>82.39±2.80</td>
</tr>
<tr>
<td>30° PC2</td>
<td>10.17±2.27</td>
<td>13.95±2.04</td>
</tr>
<tr>
<td><strong>Staggered</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>90° PC1</td>
<td>74.91±3.84</td>
<td>79.25±3.17</td>
</tr>
<tr>
<td>90° PC2</td>
<td>17.68±2.68</td>
<td>14.27±2.38</td>
</tr>
<tr>
<td>60° PC1</td>
<td>78.63±2.35</td>
<td>78.51±2.23</td>
</tr>
<tr>
<td>60° PC2</td>
<td>18.19±1.71</td>
<td>18.55±2.15</td>
</tr>
<tr>
<td>30° PC1</td>
<td>82.81±2.32</td>
<td>84.44±2.22</td>
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<tr>
<td>30° PC2</td>
<td>14.98±2.07</td>
<td>13.13±1.76</td>
</tr>
<tr>
<td><strong>Tandem</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>90° PC1</td>
<td>72.45±2.72</td>
<td>77.62±2.06</td>
</tr>
<tr>
<td>90° PC2</td>
<td>17.69±1.83</td>
<td>15.41±1.59</td>
</tr>
<tr>
<td>60° PC1</td>
<td>78.87±2.94</td>
<td>83.55±1.86</td>
</tr>
<tr>
<td>60° PC2</td>
<td>16.84±2.21</td>
<td>11.94±1.28</td>
</tr>
<tr>
<td>30° PC1</td>
<td>86.78±1.48</td>
<td>87.43±1.16</td>
</tr>
<tr>
<td>30° PC2</td>
<td>10.14±1.20</td>
<td>8.89±0.64</td>
</tr>
</tbody>
</table>

**Results**

Table 4-1 shows that $f_{50}$ and $f_{95}$ power are dominated at the slow time scale in the side-by-side stance. Changing the position of the feet from the side-by-side to tandem stance, $f_{50}$ and $f_{95}$ progressively shifted as additional fast time scale components were introduced into the COP time series. Figure 4-3 shows the COP_L and COP_R patterns of a representative subject standing in the side-by-side foot position with the right foot oriented at different angles. The COP_L trajectory in the AP direction gradually increased as the position of the right foot changed from 90° to 30°.
Figure 4-3. Example of COP_L and COP_R patterns in side-by-side stance. A) 90°; B) 60° and C) 30° (staggered and tandem positions not shown).
Fz ratio

Figure 4-4 exhibits the Fz ratio as a function of foot position and orientation.

There were significant main effects of foot position and right foot orientation, and their interaction on body weight loading [foot position: $F_{(2,22)}= 80.736$, $p<0.001$; foot orientation: $F_{(2,22)} = 14.552$, $p<0.001$; foot position × foot orientation: $F_{(2.524, 27.767)}=3.443$, $p=0.037$]. The participants loaded more on their left foot in the staggered and tandem stances [side-by-side $M=0.046$; staggered $M=0.395$; tandem $M=0.364$]. Moreover, they loaded less on their right foot when it was positioned at 30° [30° $M=0.305$; 60° $M=0.269$; 90° $M=0.231$]. Post-hoc comparisons indicated that the mean score of the Fz ratio at 30° foot orientation level [30° $M= 0.456$] was the largest followed by 60° and then 90° in the staggered stance [60° $M=0.408$; 90° $M=0.320$].

![Figure 4-4. Fz ratio between the left and right foot as a function of postural stance and foot orientation.](image)
MANOVA

Figure 4-5 shows the factor weightings of each COP time series to the 1st principal component as a function of foot position and orientation at each frequency epoch. The MANOVA analyses showed that there were significant main effects of foot position and orientation together with an interaction between these two factors on the four COP weightings at both frequency epochs [Low: foot position $F_{(8, 194)} = 16.382, p<0.01$, orientation $F_{(8,194)} = 0.847, p<0.05$, foot position× orientation $F_{(16, 396)} = 3.072, p<0.01$; High: foot position $F_{(8, 194)} = 17.946, p<0.01$, orientation $F_{(8, 194)} = 2.894, p<0.01$, foot position× orientation $F_{(16, 396)} = 2.649, p<0.01$]. All COP factor weightings interacted with foot position. Although the foot orientation was only manipulated for the right foot, it also influenced the COP factor weightings of the left foot. The interaction effect was significant only for the right foot COPs with the COPL-ML an exception at high frequencies.
Figure 4-5. Factor weighting of each individual COP (arbitrary unit) as a function of postural stance (A: side-by-side; B: staggered; C: tandem), foot orientation and frequency epoch.
When the right foot was positioned at 90°, the COP main effect and foot position \times COP interaction were significant at both frequency epochs [Low: COP F(3,132)=3.039 p<0.05, foot position \times COP F(6,132)=13.923 p<0.01; High: COP F(3,132)=5.152 p<0.05, foot position \times COP F(6,132)=11.251 p<0.01]. Post hoc analyses showed that the COP_{L-ML} had a larger factor weighting than the COP_{R-AP} at low frequencies [COP_{L-AP} M=0.506, COP_{R-AP} M=0.470, COP_{L-ML} M=0.511 and COP_{R-ML} M=0.479], whereas the COP_{L-AP} had the least weighting at high frequencies [COP_{L-AP} M=0.427, COP_{R-AP} M=0.507, COP_{L-ML} M=0.492 and COP_{R-ML} M=0.503]. In the side-by-side stance, the factor weighting of the COP_{L-ML} was smaller than that of the COP-AP at low frequencies and was smaller than that of the right foot COPs at high frequencies. In the staggered position, the COP_{R-ML} was smaller than that of the left foot COPs at low frequencies and was smaller than the COP_{R-AP} at the high frequency epoch. In the tandem foot position, the COPs in the AP direction showed the lowest weightings for both frequency epochs.

When the right foot was oriented at 60°, the COP main effect and foot position \times COP interaction effects were significant only at the high frequency epoch [High: COP F(3,132)=3.689 p<0.05, foot position \times COP F(6,132)=15.821 p<0.01]. Post hoc analyses showed that the COP_{L-ML} had the least factor weighting in the side-by-side stance. The right foot COPs had smaller weightings as compared with the left foot in the staggered foot position. In the tandem stance, the weighting of the COP_{L-AP} was smaller than that of the COPs in the ML direction.
In the 30° foot orientation condition there was a significant COP main effect and foot position × COP interaction for both frequency epochs [Low: COP F\,(3, 132)=5.041 p<0.01, foot position × COP F\,(6, 132)=6.4841 p<0.01; High: COP F\,(3, 132)=5.552 p<0.01, foot position × COP F\,(6, 132)=5.824 p<0.01]. Post hoc analyses showed that the weighting of the COP\(_L\)-ML was the smallest as compared with other COP time series in the side-by-side stance at both frequency epochs. At high frequencies, the COP\(_L\)-AP had the smallest weighting in the tandem foot position.

Foot orientation × COP univariate test (Figure 4-5 read by row)

For the side-by-side stance, there was a significant COP main effect and foot orientation× COP interaction at low frequencies [COP: F\,(3,132)=48.956, p<0.01; foot orientation× COP: F\,(6,132)=2.935, p<0.05]. However, only the COP main effect was significant at high frequencies [COP: F\,(3, 132)=20.979, p<0.01]. Post hoc analyses revealed that factor weighting of the COP\(_L\)-ML was significantly lower than other COP signals in all foot orientation levels at both frequency epochs [Low: COP\(_L\)-AP M=0.494, COP\(_R\)-AP M=0.553, COP\(_L\)-ML M=0.396, COP\(_R\)-ML M=0.528; High: COP\(_L\)-AP M=0.490, COP\(_R\)-AP M=0.521, COP\(_L\)-ML M=0.408 and COP\(_R\)-ML M=0.539]. When the right foot was positioned at 60°, weighting of the COP\(_L\)-AP was significantly smaller than the right foot COP at low frequencies.

For the staggered foot position, the effects of COP and foot orientation × COP interaction were significant at both frequency epochs [Low: COP F\,(3, 132)=8.392 p<0.01, foot orientation × COP F\,(6, 132)=4.187 p<0.01; High: COP F\,(3, 132)=8.279 p<0.01, foot orientation × COP F\,(6, 132)=7.161 p<0.01]. Post hoc analyses revealed that the
weightings of the left foot COPs were significantly larger than that of the right foot COPs in both 60° and 90° conditions at both frequency epochs.

For the tandem stance, there was a significant main effect of COP and an interaction between foot orientation and COP at both frequencies [Low: COP F (3, 132) = 13.513 p<0.01, foot orientation × COP F (6, 132) = 3.539 p<0.01; High: COP F (3, 132) = 35.293 p<0.01, foot orientation × COP F (6, 132) = 2.255 p<0.05]. Post hoc analyses showed that the weighting of the COP_L-AP was the smallest in all foot orientation levels at both frequency epochs. When the right foot was positioned at 90°, the COP_R-AP had smaller weighting as compared with that of the COP signals in the ML direction.

**Discussion**

The present experiment was set up to investigate the effect of different foot positions and orientations on the body weight loading and foot coordination dynamics in quiet standing postures. To address this question, the participants stood quietly in three foot positions while the right foot was oriented at different angles. The ground reaction forces and COP time series were recorded and derived respectively from two synchronized force platforms. These findings provide supportive position that the mechanical constraints of foot position and orientation interact to channel the inter- and intra-foot coordination dynamics of standing postures.

The Fz ratio showed that the participants loaded more on their left foot when they positioned their right foot forward (Figure 4-4). Consistent with previous studies, the participants loaded 65%-75% of their body weight on the rear foot when they positioned
one foot in front of the other (Jonsson et al., 2005; King et al., 2012). In the current study, we did not request our participants voluntarily load more on one side of their body so that the asymmetrical body weight loading reflects the interactive effect of the mechanical constraints. A related finding was that the participants gradually shifted their body weight to the left side when their right foot was oriented from 90º to 30º. The body weight shifting could due to the fact that the participants utilize their left foot for stabilization and the right foot for flexibility of adaptation (for reviews, see Peters, 1988; Sadeghi, Allard, Prince, & Labelle, 2000).

By progressively increasing the difficulty of standing posture, the power of the slow time scale components, in particular, of the four COP time series increased (Figure 4-2). The increased frequency power can be a consequence of increased lower limb muscle activation. In EMG studies of the side-by-side stance, it has been shown that the tonic co-activation of the ankle flexors is in-phase with forward body sway and the rectus femoris activity is in-phase with backward body lean at 0.05Hz (Winter et al., 1996; Saffer, Kiemel, & Jeka, 2007). In the tandem foot position, the more loaded left limb shows enhanced tonic soleus muscle activation than that of the right leg. High correlations between peroneus longus and tibialis bursts and the COP_{NET} motion in the frontal plane have also been observed. (Sozzi et al., 2013).

Table 4-2 reveals that more than 95% total variance of the multivariate COP data set was accommodated by PC1 (72.45—87.43%) and PC2 (8.89—18.55%) whereby postural control has been simplified to roughly two “dynamical” dimensions (Daffertshofer et al., 2004; Hong & Newell, 2006). This finding is consistent with our previous nonlinear analysis in which the correlation dimension of the healthy participants’
COP_{NET} trajectory is about 2 indicating a two dimensional attractor dynamic (Newell, van Emmerik, Lee, & Sprague, 1993). Therefore, the four COP time series are not independent as previous research has implied (Kapteyn, 1973). Rather, they are coordinated substantially to influence the patterns of the COP_{NET}. Furthermore, foot position played a significant role in the foot coordination dynamics for the side-by-side and tandem stances whereas body weight loading exhibited a significant effect for the staggered stance.

The findings showed that different inter- and intra-foot coordination dynamics (indexed as the COP factor weighting in the f/PCA) in quiet stance are realized by the manipulation of the mechanical factors arising from foot position and orientation (Figure 4-5). The MANOVA findings revealed that, as compared with foot orientation, foot position had a stronger mechanical effect on foot coordination dynamics. Although foot orientation plays an important role channeling the body weight distribution, foot position directly induces postural instability and weight asymmetry by significantly changing the area of the base of support (Jonsson et al., 2005; King et al., 2012).

In the side-by-side stance, the factor weightings of the COPs in the AP direction were significantly larger than that of the COPs in the ML direction at both frequency ranges. The asymmetries can be attributed to the limited base of support dimension in the sagittal plane and the geometry of the lower limb. The anatomical structure of the ankle is that of a hinge joint introducing more substantial instability in the sagittal plane (Winter et al., 1996). Therefore, consistent to previous study (Kinsella-Shaw et al., 2011), the inter-foot coordination in the sagittal plane dominated postural stability in the side-by-side foot position.
In the tandem stance, however, the foot coordination dynamics is in contrast to that of the side-by-side standing in that the weightings of the COPs in the ML direction were significantly larger than the COPs in the AP direction. Due to the narrowed base of support in the frontal plane, more internal instability is generated laterally (Day et al., 1993; Genthon & Rougier, 2005; Reynolds, 2010; Sozzi et al., 2013). Previous investigations have shown that narrow standing width of foot position decreased the role of ankle mechanisms in the sagittal plane and increased the role of ankle and hip mechanisms in the frontal plane (Gatev et al., 1999). The homogeneous muscles of the lower limbs are active in a mutual push-pull manner in order to generate the alternative impulses for postural stability in the frontal plane (Sozzi et al., 2013).

In the staggered stance, COP of the left foot revealed larger factor weightings at both frequency ranges than the right foot COP time series. The foot coordination difference between the staggered and the other foot positions arises when the postural stability is not challenged by the reduced area of the base of support so that the body weight loading tends to be an important factor influence the foot coordination dynamics. Further, the asymmetrical body weight distribution is a by-product of the interactive effect of foot position and orientation (Figure 4-4). The “load receptors” (e.g., Gogi tendon organs) located within the extensor muscles activate postural reflex directly (Berger, Discher, Trippel, Ibrahim & Dietz, 1992). The decrease of afferent feedback of the cutaneous mecanoreceptors under the sole of the less loaded foot introduces sensory information re-weighting of the postural control system (Genthon & Rougier, 2004).

A striking finding was that the four COP weightings of the staggered stance were essentially identical when the right foot was oriented at 30° indicating an equal
contribution to PC1. Previous studies have shown that indices capturing the COP_{NET} variability changed according to foot distance and orientation (Watanabe et al., 1979; McLlroy & Maki, 1997; Chiari, Rocchi, & Cappello, 2002). In an early observation, Watanabe et al. (1979) reported that the COM motions in both sagittal and frontal planes are identical in a 30º fan-shaped foot position (i.e., heels close together with the toes open at 30º) implying a “regular squared” body sway area. Our findings revealed that this change is not only related to the characteristics of COP_{NET} but also to the inter- and intra-foot coordination dynamics. Given the fact that the self-chosen preferred foot angle between the long axes of the feet is 14º, whether a self-chosen foot position or a “regular squared” base of support facilitates postural stability is an open question. Future investigations can examine the impact of the required and self-chosen foot position and orientation on the inter- and intra-foot coordination dynamics.

Previously, it has been shown the existence of the interdependence of COP_{NET} in the two orthogonal planes when the area of the base of support is reduced (Aruin et al., 1998; Mochizuki et al., 1999, 2006; Day et al., 1993). Our findings directly revealed the inter- and intra-foot coordination dynamics in different standing postures. The only postural stance showing an equal contribution of the four COP time series was the staggered foot position with the right foot oriented at 30º.

In summary, the present experiment has shown that the asymmetrical body weight loading and foot coordination dynamics are differentially modulated by mechanical constraints derived from the interactive effects of foot position and orientation. The participants loaded more on the left foot when positioned their right foot forward or at a non-90º foot angle. Collectively, when postural stance is challenged by the limitation of
the base of support, the COPs of the unstable plane (inter-foot coordination) displayed larger factor weightings so that dominated the postural control system. In contrast, when standing posture is not restricted by the support area, the COPs of the more loaded foot (intra-foot coordination) played an significant role in postural control. These findings show that foot position and orientation interact to channel the inter- and intra-foot coordination dynamics of standing postures.
CHAPTER 5: INTER- AND INTRA-FOOT COORDINATION STANDING ON
AN UNSTABLE SURFACE

Abstract

The experiment was set up to investigate the influence of a reduced base of support (beam width: 2.5, 4.0 and 8.5cm) together with its orientation (beam orientation: longitudinal and horizontal) on the inter- and intra-foot coordination dynamics. A frequency domain PCA analysis was applied on four COP time series (COP\(_L\)-ML, COP\(_R\)-ML, COP\(_L\)-AP and COP\(_R\)-AP), collected from two synchronized force platforms, to reveal their contributions to postural stability as a function of base of support. The orientation of the base of support played a more significant role channeling the foot coordination dynamics as compared with the width of the support surface. When the shortened beam was oriented along the horizontal axis, especially in the Horizontal (2.5cm) condition, the four COP time series revealed a parallel contribution to the 1\(^{st}\) principal component indicating an inter-dependence of the inter- and intra-foot coordination. When the beam was positioned along the longitudinal axis, the COPs in the AP direction showed larger weightings to PC1 implying an inter-foot coordination in the sagittal plane. In addition, the COP range of motion in the AP direction exceeded the beam width boundaries in the Horizontal (2.5cm) condition revealing the “searching” or “detecting” properties of the postural control system for the base of support limitations. These findings provide further evidence that the inter- and intra-foot coordination operate in adaptive cooperative ways to sustain postural stability in light of changing support surface constraints to standing.
Introduction

Even though human bipedal stance is inherently unstable, given that approximately two-thirds of our body mass is located two-thirds of the body height above the ground, postural control for healthy adults is typically subconscious and takes little effort or attention. It has been proposed that standing on an unstable support surface (e.g., support surface shorter than the length of the feet) can stress the postural control system to a greater degree than standing on a stable surface (Massion, 1992; Riccio, 1993). Evidence exists that a reduced support surface limits the anti-gravity torque generated from the ankle joint by increasing the shear force provoked by the hip (Horak & Nashner, 1986). Standing on an unstable support surface channels both feedforward and feedback postural control mechanisms, including attenuating the early components of the cerebral potential (N32-P39), reduced or disappeared anticipatory postural adjustments (APAs), multi-sensory afferent re-weighting and even pronounced EEG modulation before falling down (Gavrilenko, Gatev, Gantchev, & Popivanov, 1991; Aruin et al., 1998; Krizkova, Hlavacka, & Gatev, 1993; Ivanenko, Levik, Talis, & Gurfinkel, 1997; Trimble & Koceja, 2001; Streepey, Kenyon, & Keshner, 2007; Slobounov et al., 2009).

Previous studies with standing on a shortened base of support (BOS) have predominantly focused on the control mechanisms in the sagittal plane (Ivanenko et al., 1997; Aruin et al., 1998; Krishnamoorthy, Latash, Scholz, & Zatsiorsky, 2004). In an early study of the anticipatory postural adjustments, Aruin et al. (1998) proposed that the central nervous system (CNS) “protects” the stability of upright standing by suppressing the APAs when the participant stood on a shortened support surface. This “protection” is more pronounced and compensatory when the direction of the APAs is in the plane of the
postural instability. During standing on an unstable base of support, increased sway in one direction may by itself lead to loss of balance. Therefore, it is possible that the postural control mechanism operates by the inter-dependence of the inter- (left and right) and intra- (AP and ML) foot coordination to maintain postural stability. Indeed, it has been observed that changes in the area of the base of support in one direction affect stability when the other direction of the support surface was much smaller or kept constant (Day et al., 1993; Mochizuki et al., 1999, 2006).

The focus of the present experiment is the influence of the reduced base of support on participants’ inter- and intra-foot coordination dynamics (Wang & Newell, 2013). Previously we have shown that information and foot position manipulations modulate the inter-foot coordination of foot dynamics in quiet standing (Wang & Newell, 2012b; Wang et al., 2012). Here, we extended this work by testing the hypothesis that the inter-dependence of the foot coordination is enhanced in postural control when standing on a reduced support surface because postural instability in one direction can be compensated by its orthogonal direction. We investigated three shortened standing widths (i.e., 2.5, 4.0 and 8.5 cm) to the base of support with also the manipulation of the support surface orientation (i.e., along its longitudinal and horizontal axes). We tested the hypothesis that the inter- and intra-foot coordination will increase as the postural instability increases. In particular, increased foot coordination dynamics is expected when the shortened support surface is oriented along the horizontal axis due to the fact that posture is less stable in the sagittal plane in the side-by-side stance (Bottaro et al., 2008; Suzuki et al., 2013).
A frequency domain PCA analysis (fPCA) was conducted to reveal the dynamic coordination of the foot center of pressure (COP) time series (COP\(_L\)-ML, COP\(_R\)-ML, COP\(_L\)-AP and COP\(_R\)-AP). The reasons for applying fPCA were due to: 1) the multivariate properties of the data set (rather than treating each COP separately as is usually the situation in postural studies); 2) the time lead-lag relation within and between the multivariate time series (that has been shown to influence the determination of PCA in the time domain); and 3) the capacity of the analysis to compress the dimension of the data set (Anderson, 1963; Molenaar, 1985; Daffertshofer et al., 2004; Molenaar et al., 2013). The contribution of each COP signal to the principal component is represented by its respective factor weighting as a function of frequency.

**Methods**

**Subjects and Apparatus**

Twelve right-footed healthy subjects, five female and seven male [mean age 29 yr (range 25-34 yr); mean height 167.25 cm (range 155.0-184.0 cm)], participated in this study. All subjects provided informed consent for the present study that was approved by the Institutional Review Board (IRB) of Pennsylvania State University. The force and moment data of each foot were collected by two adjacent AMTI force platforms (Advanced Mechanical Technology Inc., OR6-5-1000) at a sample rate of 100 Hz. The platforms were calibrated and synchronized for data collection.
Tasks and Procedures

The experimental testing started with two baseline trials of the side-by-side standing on the platforms followed by 12 trials of the pseudo-randomized unstable side-by-side stances with the shortened base of support. For the baseline testing, the distance of the first metatarsophalangeal articulations between the left and right foot was 25cm.

For the unstable conditions, a rectangular beam that was in direct contact with the force platform was fixed under the center of a board (Figure 5-1A). The beam was oriented either along the longitudinal or the horizontal axis of the board. Therefore, the length of the beam was the same with the length of the board in the Longitudinal condition and was the same with the width of the board in the Horizontal condition.

The participants stood on the boards either with the beam oriented along the longitudinal or the horizontal axis. Note that postural instability is significant in the frontal plane when the beam is positioned along the longitudinal axis whereas postural instability is enhanced in the sagittal plane when the beam is oriented along the horizontal axis of the unstable board. The width of the beam varied at 2.5cm (Narrow), 4.0cm (Medium) and 8.5cm (Wide). Therefore, there were 2 (beam orientation) × 3 (beam width) unstable postural stances.
Figure 5-1. Experiment setup. A demonstrates the dimension of the beam underneath the board. The left shows the beam oriented along the longitudinal axis of the board and the right shows the beam oriented along the horizontal axis of the board. The width of the beam can be 2.5cm, 4.0cm or 8.5cm (pointed by the arrows). B illustrates a participant standing on the unstable boards (two sets of a board, one on each force platform) positioned in parallel. The beams are oriented along the longitudinal axis of the boards introducing instability in the frontal plane. The distance between the beams was 25 cm. The boards are 4cm above the force platforms, which positioned together.
In addition, there were two sets of a board for each standing condition, one for each foot and placed at the top of each force plate (Figure 5-1 B). The boards were positioned in parallel throughout the testing. The distance between the beams under the boards was also 25cm. During the testing, the participants were required to hang their arms beside the trunk and keep the edges of the boards in no contact with the force platforms throughout the testing. The task requirement was to stand as still as possible. In the case the postural stability was challenged, there was no restriction on introducing the knee and other joints for maintaining balance.

There were 2 trials at each of the postural stances. Each trial was 60 s in duration. Data recording was initiated 5 s after the participants were positioned comfortably in the respective posture on the force platforms. The 2 trials at each unstable condition were blocked. A 10 s break was given between each trial of a condition and 30 s breaks were given between each condition.

Data Analysis

The raw kinetic data were filtered with a 4th-order 6-Hz low-pass zero-lag Butterworth filter. The COP signals in both directions were derived from each force platform by $COP_{AP} = \frac{(-h \cdot F_x - M_y)}{F_z}$ and $COP_{ML} = \frac{(-h \cdot F_y + M_x)}{F_z}$, where $h$ ($h=4.0\text{cm}$) is the height of the board over the force platform in the unstable condition. In the baseline trials, the participants stood on the platforms directly ($h=0\text{cm}$). The four COP time series were centered independently.
Table 5-1. Group averaged f50, f95 power frequency (mean ±se) as a function of COP type, beam orientation and width

<table>
<thead>
<tr>
<th></th>
<th>COP_L-AP</th>
<th>COP_R-AP</th>
<th>COP_L-ML</th>
<th>COP_R-ML</th>
<th>COP_L-AP</th>
<th>COP_R-AP</th>
<th>COP_L-ML</th>
<th>COP_R-ML</th>
</tr>
</thead>
<tbody>
<tr>
<td>SS</td>
<td>0.122±0.013</td>
<td>0.138±0.015</td>
<td>0.137±0.018</td>
<td>0.163±0.016</td>
<td>0.813±0.053</td>
<td>0.870±0.071</td>
<td>1.190±0.117</td>
<td>1.523±0.206</td>
</tr>
<tr>
<td>Narrow</td>
<td>0.141±0.016</td>
<td>0.149±0.014</td>
<td>0.137±0.016</td>
<td>0.164±0.015</td>
<td>0.903±0.045</td>
<td>0.906±0.043</td>
<td>1.512±0.183</td>
<td>2.034±0.341</td>
</tr>
<tr>
<td>L</td>
<td>0.192±0.030</td>
<td>0.160±0.021</td>
<td>0.190±0.036</td>
<td>0.184±0.031</td>
<td>1.007±0.052</td>
<td>1.047±0.055</td>
<td>1.447±0.124</td>
<td>1.912±0.320</td>
</tr>
<tr>
<td>Medium</td>
<td>0.147±0.017</td>
<td>0.168±0.023</td>
<td>0.134±0.016</td>
<td>0.171±0.019</td>
<td>0.949±0.057</td>
<td>0.990±0.071</td>
<td>1.258±0.097</td>
<td>1.599±0.236</td>
</tr>
<tr>
<td>Wide</td>
<td>0.348±0.049</td>
<td>0.347±0.047</td>
<td>0.262±0.052</td>
<td>0.244±0.028</td>
<td>2.033±0.183</td>
<td>2.096±0.174</td>
<td>1.905±0.146</td>
<td>1.966±0.150</td>
</tr>
<tr>
<td>H</td>
<td>0.271±0.045</td>
<td>0.263±0.051</td>
<td>0.232±0.036</td>
<td>0.187±0.032</td>
<td>1.631±0.186</td>
<td>1.659±0.182</td>
<td>1.585±0.114</td>
<td>1.658±0.131</td>
</tr>
<tr>
<td>Medium</td>
<td>0.235±0.035</td>
<td>0.250±0.039</td>
<td>0.217±0.040</td>
<td>0.192±0.028</td>
<td>1.333±0.149</td>
<td>1.319±0.140</td>
<td>1.492±0.134</td>
<td>1.358±0.092</td>
</tr>
</tbody>
</table>
**COP range of motion**

The COP range was calculated by taking the difference between the maximum and minimum values of the COP time series indicating the range of the COP migration of the trial.

**Frequency domain principal component analysis (fPCA)**

The power spectrum of the COP time series was derived first by the Fast Fourier Transform (FFT) with the frequency resolution of 0.0167Hz. The first two frequency points past the DC component (0.0167 and 0.0334Hz) were not included in the fPCA analysis due to the fact that they represent events that occur every 60 and 30 s, and probably do not provide significant information about the postural control system (Prieto et al., 1996). The frequencies \( f_{50}, f_{95} \) corresponding to 50% and 95% of the total power of the each COP signal and were defined for each postural condition (Table 5-1). Mean frequencies were derived for each standing posture. Therefore, the fPCA analysis was conducted within two frequency epochs that each covered roughly 50% of the total power on the power spectrum (baseline: Low 0.050-0.150Hz, High 0.167-1.10Hz; Longitudinal condition: Low 0.050-0.150Hz, High 0.167-1.300Hz; Horizontal condition: Low 0.050-0.250Hz, High 0.267-1.667Hz).
The eigenvalues of the first two PCs were reported in Table 5-2, according to which PC1 represented the greatest proportion of the total variance of the multivariate data set (i.e., 71.98-83.67%). As a result, the weightings of the COP time series for PC1 were reported and further analyzed on each individual frequency epoch.

Table 5-2. Mean percentage of variance (mean ± SE %) interpreted by PC1 and PC2 as a function of beam orientation and width

<table>
<thead>
<tr>
<th></th>
<th>Low frequency epoch</th>
<th>High frequency epoch</th>
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<tbody>
<tr>
<td></td>
<td>PC1</td>
<td>PC2</td>
</tr>
<tr>
<td>SS</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Narrow</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>78.57±2.75</td>
<td>16.37±2.00</td>
</tr>
<tr>
<td></td>
<td>82.07±2.56</td>
<td>13.16±1.68</td>
</tr>
<tr>
<td>Narrow</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>72.62±2.77</td>
<td>19.61±2.29</td>
</tr>
<tr>
<td>Medium</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>71.98±2.40</td>
<td>20.43±1.32</td>
</tr>
<tr>
<td></td>
<td>75.61±2.67</td>
<td>17.10±1.83</td>
</tr>
<tr>
<td>Wide</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>73.652.28</td>
<td>20.08±1.77</td>
</tr>
<tr>
<td></td>
<td>79.14±1.94</td>
<td>15.03±1.35</td>
</tr>
<tr>
<td>Narrow</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>73.803.14</td>
<td>18.39±2.62</td>
</tr>
<tr>
<td></td>
<td>80.66±2.65</td>
<td>12.20±1.77</td>
</tr>
<tr>
<td>Medium</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>72.49±3.66</td>
<td>17.59±2.43</td>
</tr>
<tr>
<td></td>
<td>75.50±3.69</td>
<td>16.39±2.51</td>
</tr>
<tr>
<td>Wide</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>73.89±4.03</td>
<td>18.16±2.62</td>
</tr>
<tr>
<td></td>
<td>83.67±2.68</td>
<td>11.54±1.71</td>
</tr>
</tbody>
</table>
**Statistical analyses**

A series of 2 (beam orientation) × 3 (beam width) fixed effect MANOVA were conducted on the COP weightings of PC1 at each frequency epoch, respectively. A two-way univariate test was conducted afterwards to determine the beam orientation × COP type and beam width × COP type interaction effects at both low and high frequencies. The Bonferroni post-hoc analysis was used to determine the differences on all pairs of levels of independent variables. In all analyses, only effects that were statistically significant at p <0.05 are reported (a Bonferroni adjustment was conducted on the significant level for pairwise comparisons). In the case that homogeneity of the covariance matrices was violated, Pillai’s trace estimate was used to provide a conservative test of MANOVA main and interaction effects.
Figure 5-2. Exemplar of the $\text{COP}_L$, $\text{COP}_{\text{Net}}$ and $\text{COP}_R$ patterns in Longitudinal-Narrow (A-top), Longitudinal-Wide (A-bottom), Horizontal-Narrow (B-top) and Horizontal-Wide (B-bottom) standing condition. The width of the beam was 2.5cm in the Narrow and 8.5cm in the Wide conditions. Note that the range of the COP motion of each foot and the $\text{COP}_{\text{Net}}$ in the AP direction is beyond the width of the beam in Horizontal-Narrow condition.
Results

COP$_L$, COP$_{Net}$ and COP$_R$ trajectories for a representative participant standing on the unstable boards are shown in Figure 5-2. The COP migration in both AP and ML directions were significantly increased in the Horizontal-Narrow condition. More importantly, in the least stable condition (i.e., Horizontal-Narrow), the COP migration was not restricted within the base of support area and tended to move beyond the boundaries.

Figure 5-3 exhibits the COP range of motion as a function of the COP type, beam width and beam orientation. The COP range patterns in the baseline trials were similar to the unstable board stances with the beam oriented along the longitudinal axis so that they were not influenced by the instability induced from the frontal plane. In contrast, however, the COP range was substantially larger in the horizontal stances in both planes as compared with the baseline condition. As a result, the instability in the sagittal plane (Horizontal conditions) induced larger COP motion than that in the frontal plane (Longitudinal conditions).

For the COP range of motion, there were significant main effects of beam orientation, beam width and COP type together with the interactions of beam orientation with beam width and COP type, respectively [beam orientation: $F_{(1, 264)} = 148.261, p<0.01$; beam width: $F_{(2, 264)} = 14.373, p<0.01$; COP type: $F_{(3, 264)} = 384.910, p<0.01$;
beam orientation × beam width: \( F_{(2, 264)} = 18.682, p < 0.01 \); beam orientation × COP type: \( F_{(3, 264)} = 7.565, p < 0.01 \). The post hoc analyses showed that COP had larger range of motion when the beam was oriented along the horizontal axis of the board [Longitudinal: M=1.110cm; Horizontal: M=1.835cm] and when the width of the beam was smaller [8.5cm: M=1.255cm; 4.0cm: M=1.526cm; 2.5cm: M=1.636cm]. In addition, the COP time series in the AP direction displayed a larger range of motion than that in the ML direction [COP\(_L\)-ML: M=0.436cm; COP\(_R\)-ML: M=0.488cm; COP\(_L\)-AP: M=2.384cm; COP\(_R\)-AP: M=2.581cm]. In the Horizontal condition, the four COP time series had the largest motion when the width of the beam was the smallest, followed by 4.0cm and then 8.5cm beam width [2.5cm: M=2.229cm; 4.0cm: M=1.876cm; 8.5cm M=1.400cm]. In both Horizontal and Longitudinal conditions, the COPs in the AP direction showed larger range of motion as compared with the COPs in the frontal plane.
Figure 5-3. The range of the COP\textsubscript{L} and COP\textsubscript{R} motion as a function of beam orientation and width in both AP and ML directions.

Figure 5-4 exhibits the 50% and 95% power frequency of the COP time series as a function of beam orientation and width. The ANOVAs showed that there were significant beam orientation main and beam orientation \times beam width interaction effects for the $f_{50}$ [beam orientation: $F_{(1, 264)} = 46.532, p < 0.01$; beam orientation \times beam width: $F_{(2, 264)} = 5.074, p < 0.01$]. The post hoc analyses revealed that the $f_{50}$ was smaller when the beam was positioned along the longitudinal axis as compared with that in the Horizontal condition [Longitudinal: $M=0.162\text{Hz}$ and Horizontal: $M=0.254\text{Hz}$]. The 2.5cm beam
width induced the largest $f_{50}$ when the beam was oriented along the horizontal axis
[2.5cm: $M=0.300\text{Hz}$; 4.0cm: $M=0.238\text{Hz}$; 8.5cm: $M=0.224\text{Hz}$].

For the $f_{95}$, there were significant main effects of beam orientation, beam width
and COP type together with the interactions of beam orientation with beam width and
COP type, respectively [beam orientation: $F_{(1, 264)}=31.986$, $p<0.01$; beam width: $F_{(2, 264)}$
=11.214, $p<0.01$; COP type: $F_{(3, 264)}=9.826$, $p<0.01$; beam orientation $\times$ beam width: $F_{(2, 264)}$
=5.008, $p<0.01$; beam orientation $\times$ COP type: $F_{(3, 264)}=10.708$, $p<0.01$]. The post
hoc analyses showed that the $f_{95}$ was smaller when the beam was oriented along the
longitudinal axis of the board [Longitudinal: $M=1.297\text{Hz}$; Horizontal: $M=1.670\text{Hz}$] and
was the smallest when the width of the beam was 8.5cm [8.5cm: $M=1.288\text{Hz}$; 4.0cm:
$M=1.493\text{Hz}$; 2.5cm: $M=1.669\text{Hz}$].
Figure 5-4. Power frequency of the four COPs as a function of beam orientation and width ($f_{50}$: top; $f_{95}$: bottom).
The COP$_R$-ML displayed a larger $f_{95}$ mean value as compared with the COPs in the AP direction [$\text{COP}_L$-ML: $M=1.309\text{Hz}$; COP$_R$-ML: $M=1.336\text{Hz}$; COP$_L$-AP: $M=1.533\text{Hz}$; COP$_R$-AP: $M=1.755\text{Hz}$]. Moreover, in the Horizontal condition, the $f_{95}$ was the largest when the width of the beam was 2.5cm [2.5cm: $M=2.000\text{Hz}$; 4.0cm: $M=1.633\text{Hz}$; 8.5cm: $M=1.376\text{Hz}$]. When the beam was positioned longitudinally, the COPs in the ML direction revealed larger $f_{95}$ than that of the COPs in the AP direction [$\text{COP}_L$-ML: $M=1.406\text{Hz}$; COP$_R$-ML: $M=1.848\text{Hz}$; COP$_L$-AP: $M=0.953\text{Hz}$; COP$_R$-AP: $M=0.981\text{Hz}$].

Figure 5-5 shows the weightings of each COP time series to the 1$^{st}$ principal component as a function of beam orientation and width at both frequency epochs. The MANOVAs showed that there was a significant main effect of beam orientation at both frequency epochs [Low: $F_{(4, 63)}=3.779, p<0.01$; High: $F_{(4, 63)}=3.952, p<0.01$]. The mean weighting of the COP$_L$-ML was smaller when the beam was positioned longitudinally as compared with horizontally for both frequencies [Low: Longitudinal $M=0.368$, Horizontal $M=0.463$; High: Longitudinal $M=0.381$, Horizontal $M=0.454$]. In contrast, the COP$_R$-AP showed a larger weighting score when the beam was positioned along the longitudinal axis of the unstable board at the low frequency epoch [Low: Longitudinal $M=0.572$, Horizontal $M=0.510$].
Figure 5-5. Weighting of each COP time series (arbitrary unit) as a function of beam orientation (Longitudinal conditions: upper; Horizontal conditions: lower), width and frequency epoch.
In the Longitudinal conditions, the COP type main effect was significant for both frequency epochs [Low: $F_{(3,132)} = 18.298$, $p<0.01$; High: $F_{(3,132)} = 10.401$, $p<0.01$]. The width of the base of support did not have any significant influence on the four COP weightings. At the low frequency epoch, the COPs in the AP direction showed larger weightings than that in the ML direction [$\text{COP}_{L}-\text{ML}: M=0.368$; $\text{COP}_{R}-\text{ML}: M=0.450$; $\text{COP}_{L}-\text{AP}: M=0.523$; $\text{COP}_{R}-\text{AP}: M=0.572$]. In contrast, at the high frequencies, the $\text{COP}_{L}-\text{ML}$ displayed the least weighting as compared with other COPs [$\text{COP}_{L}-\text{ML}: M=0.381$; $\text{COP}_{R}-\text{ML}: M=0.475$; $\text{COP}_{L}-\text{AP}: M=0.489$; $\text{COP}_{R}-\text{AP}: M=0.547$].

In the Horizontal conditions, the COP type main effect was significant only for the high frequency epoch [$F_{(3,132)} = 12.021$, $p<0.01$]. The post hoc analysis showed that the COPs in the ML direction had smaller weighting as compared with the COPs in the AP direction [$\text{COP}_{L}-\text{ML}: M=0.454$; $\text{COP}_{R}-\text{ML}: M=0.414$; $\text{COP}_{L}-\text{AP}: M=0.527$; $\text{COP}_{R}-\text{AP}: M=0.538$].

When the width of the beam was 2.5cm, there was significant COP type main effect for both frequency epochs and beam orientation \(\times\) COP type interaction effect at the low frequencies [Low: COP type $F_{(3,88)} = 14.593$, $p<0.01$, beam orientation $\times$ COP type $F_{(3,88)} = 3.167$, $p<0.05$; High COP type $F_{(3,88)} = 11.122$, $p<0.01$]. The post hoc analyses showed that, in the Longitudinal conditions, the COPs in the ML direction had smaller weightings than that of the COPs in the AP direction at the low frequencies [$\text{COP}_{L}-\text{ML}: M=0.353$; $\text{COP}_{R}-\text{ML}: M=0.414$; $\text{COP}_{L}-\text{AP}: M=0.550$; $\text{COP}_{R}-\text{AP}: M=0.596$].

When the width of the beam was 2.5cm, only the COP type main effect at the low frequencies was significant [$F_{(3,88)} = 3.267$, $p<0.05$]. The mean weighting score of the $\text{COP}_{R}-\text{AP}$ was significantly larger than the $\text{COP}_{L}-\text{ML}$ [$\text{COP}_{L}-\text{ML}: M=0.430$; $\text{COP}_{R}-\text{AP}$...
M=0.522]. When the width of the beam was 8.5cm, both COP type main and beam orientation × COP type interaction effects were significant at both frequency epochs [Low: COP type $F_{(3, 88)}$=3.825, $p<0.05$, bean orientation × COP type $F_{(3, 88)}$=3.507, $p<0.05$; High COP type $F_{(3, 88)}$=7.013, $p<0.01$, bean orientation × COP type $F_{(3, 88)}$=4.216, $p<0.01$]. The post hoc analyses showed that, when the beam was positioned along the longitudinal axis of the board, the COP_{L}-ML had the least weighting as compared with the COPs in the AP direction for both frequencies [Low: COP_{L}-ML $M=0.339$, COP_{L}-AP $M=0.518$, COP_{R}-AP $M=0.576$; High: COP_{L}-ML $M=0.349$, COP_{L}-AP $M=0.462$, COP_{R}-AP $M=0.574$].

Discussion

The present experiment investigated the influence of a reduced base of support surface and its orientation on inter- and intra-foot coordination dynamics. We investigated the hypothesis that with an increase of postural instability the pattern of inter- and intra-foot coordination would be modified adaptively to the surface of support constraints. To address this question, we manipulated the standing width (2.5, 4.0 and 8.5 cm) to the base of support that also varied in orientation relative to the unstable board (along the longitudinal or the horizontal axis). A frequency domain PCA analysis was applied to reveal the contribution of each individual COP time series to the 1st principal component of the multivariate data set. The central finding was the inter- and intra-foot coordination dynamics of the COP time series were systematically modified by the interaction of beam width and orientation constraints to standing posture.
The COP range

Consistent with previous studies, the COP range of motion increased when the participants stood on the unstable boards with shortened support area (Mochizuki et al., 2006). In particular, the range of COP motion exceeded the sagittal plane base of the support boundary in the Horizontal-Narrow condition (Figure 5-2 B-top). This effect has been interpreted as reflecting the consequence of a “searching mechanism” where the CNS searches to detect the limits of stability for the upright standing posture (Mochizuki et al., 1999; 2006). More generally, postural stability is realized throughout the time evolutionary changes of the foot coordination dynamics by a composition of a “stable” and “flexible” coordination pattern (Wang & Newell, 2012b; Suzuki et al., 2013). The “stable” coordination represents the magnet effect of the system in which the feet cooperate to maintain postural stability. The “flexible” pattern reflects the exploration tendencies of the foot dynamics to both internal and external environments. With the decrease of the base of support area, foot coordination behaved more “flexibly” in the shortened support dimension exploring the stability boundary and performed more “stably” in its orthogonal dimension to compensate postural instability induced by the exploration activities (Wang & Newell, 2012b).

In contrast to Mochizuki et al. (2006) we only observed an increased COP migration when the beam was oriented along the horizontal axis of the board (Figure 5-3). Even though the range of COP-AP migration was significantly larger than that of the COP-ML for all standing conditions, the width of the beam challenged the postural control system only in the Horizontal conditions. When the beam was positioned along the longitudinal axis of the board, the COP migration of the feet was comparable to the
baseline stance even in the condition when the width of the beam was 2.5cm. Therefore, the dimension of the base of support displayed a pronounced effect on the COP range as compared to the width of the beam (Wang et al., 2013).

This observation can be attributed to the anatomical structure of the lower limb. In the side-by-side stance, postural sway in the sagittal plane is significant due to the fact that it is controlled by an in-phase motion of the ankle joints. In contrast, postural sway in the frontal plane is predominantly controlled by an anti-phase hip loading-unloading mechanism with a relatively wide support area (Winter et al., 1996; Day et al., 1993; Wang & Newell, 2012b). The interesting finding is that when postural sway in the sagittal plane was challenged by the shortened base of support, the “exploratory” properties of the COP time series were evident in both directions.

\textit{f}_{50} \text{ and } \textit{f}_{95} \text{ of the COP power spectrum}

Our findings showed that in the Longitudinal standing conditions, the COPs in the frontal plane displayed larger \textit{f}_{95} as compared with the COP time series in the sagittal plane (Table 5-1). This observation is probably due to the fact that the gradually shortened beam width in the frontal plane induces more EMG activation of the lower limb muscles such as the peroneus longus and tibialis (Sozzi et al., 2013). Ivanenko et al. (1997) also found a directionally specific ankle torque-COM motion relation organized as a function of the height of the unstable seesaw surface indicating an ankle strategy for maintaining balance.
The increased contribution of fast time scale components can also attribute to the enhanced feedback for postural adjustment. For example, Streepey et al. (2007) found that when standing on a shortened support surface, the participants relied more on visual information to maintain balance even when the information itself was unstable (the visual screen oscillating sinusoidally at 0.1Hz). Due to the mechanical constraint on the base of support, limited tactile information from the sole of the feet and proprioceptive feedback can be expected as a result (Krizkova et al., 1993). In this view, the CNS re-evaluates and re-adjusts the appropriate amount of sensory information utilized for postural control.

In contrast, when the beam was oriented along the horizontal axis, the COP time series of the feet showed the equivalent $f_{50}$ and $f_{95}$ power frequency whereas the 2.5cm beam width condition showed significantly increased slow and fast time scale components. The horizontal standing condition challenged the postural stability in both planes. Due to the shortened base of support dimension in the sagittal plane, the ankle joint failed to effectively generate the anti-gravity torque to maintain postural stability. Instead, postural instability in the AP direction can be compensated by the gradually increased hip flexion and extension and also the increased compensatory muscle activation and inter-foot coordination dynamics in the frontal plane (Horak & Nashner, 1986; Horak & Moore, 1993; Wang & Newell, 2012b; Sozzi et al., 2013).
Figure 5-6. Box plot of the Fz ratio as a function of standing condition. The $F_{z\text{Left}}$ and $F_{z\text{Right}}$ represent the ground reaction force of the left and right foot, respectively. Circles show the Fz ratio of the outliers identified for each standing condition.

$$F_z\% = \frac{F_{z\text{Left}} - F_{z\text{Right}}}{F_{z\text{Left}} + F_{z\text{Right}}} \times 100\%$$

**Couplings of the COPs to PC1**

The COP$_{\text{Net}}$ derived from a single force platform incorporates the contribution of the body weighting loading and the COP of each foot (Winter et al., 1996). Even though the participants loaded evenly on their feet in all standing conditions, the inter- and intra-
foot coordination dynamics play an important role to maintain postural stability (Figure 5-6). Table 5-2 showed that more than 95% total variance of the multivariate COP data set was accommodated by the first two PCs whereby postural control has been simplified by roughly two “dynamical” degrees of freedom (Daffertshofer et al., 2004; Hong & Newell, 2006). In particular, PC1 explained 71.98-83.67% of total variance indicating significant covariation and correlation within the four COP time series. Furthermore, the MANOVAs showed that this inter-dependence changed as a function of task constraint—beam orientation had a more pronounced influence on the COP weightings as compared to the beam width (Newell, 1986; Davids et al., 2007).

The contribution of each individual COP time series on the postural control system is predominantly influenced by the dimension of the base of support (Wang, Molenaar, & Newell, 2013). When the beam was positioned along the horizontal axis of the unstable board, the four COPs revealed equivalent contributions to PC1 given the range of the COP-AP was larger than that of the COPs in the frontal plane. However, when the beam was oriented along the longitudinal axis, COPs in the sagittal plane displayed larger weighting to PC1, which is consistent with our previous observations of the side-by-side stance on the floor. Previously, we found that, in both frequency epochs, the COP_{L-ML} displayed the least weighting score in the side-by-side stance with varied right foot orientations (Wang et al., 2013). In the current study, a consistent finding is that the least COP_{L-ML} weighting was in all the Longitudinal orientation conditions. This pattern of findings could be driven by the different activation patterns of the plantar cutaneous mechanoreceptors under the sole of the feet (Genthon & Rougier, 2005).
In summary, the current experiment has been shown that the orientation of the base of support orientation plays a more significant role than beam width on the COP weightings in the inter- and intra-foot coordination dynamics in the side-by-side postural stance. In particular, the four COP time series revealed a parallel contribution to the 1st principal component when the beam was positioned along the horizontal axis of the board indicating an inter-dependence of the inter- and intra-foot coordination. When the beam was positioned along the longitudinal axis, the COPs in the AP direction showed larger weightings to PC1 implying an inter-foot coordination in the sagittal plane. These findings provide further evidence that the inter- and intra-foot coordination operate in adaptive cooperative ways to sustain postural stability in light of changing support surface constraints to standing.
CHAPTER 6: GENERAL DISCUSSION

This dissertation examined the influence of mechanical and task constraints on the body weight distribution and foot coordination dynamics of quiet postural stances. The $\text{COP}_{\text{Net}}$, as a controlled variable, is the composite of the proportion of body weight and the COP of each foot (Winter, 1995). By using two force platforms, the effects of the body weight loading and COP pattern of the feet can be separated apart. In particular, these effects can be significant with the manipulation of mechanical and task constraints such as foot position, foot orientation and the size of the base of support. The body weight distribution serves not only as an independent variable by requiring the participant intentionally load more weight on one side of the foot but also as a dependent variable due to the covariation of the weight loading to the other task constraints.

Three questions were primarily addressed in this dissertation. Firstly, we investigated the role of foot position and intentionally induced asymmetrical body weight loading on the inter-foot coordination dynamics in the upright stances. The inter-foot coordination was quantified by both linear and nonlinear analyses. Secondly, we examined how the interactive effects of foot position and orientation influence the inter- and intra-foot COP coordination dynamics. Lastly, we investigated how the width and orientation of the reduced base of support channels the inter-and intra-foot coupling in a side-by-side foot position. A frequency domain PCA was applied in the 2$^{rd}$ and 3$^{rd}$ experiments due to the multivariate and time invariant properties of the data set.
Influence of Mechanical Constraints on Foot Coordination Dynamics

**Foot Position and Orientation.** The side-by-side and tandem foot positions reflect the two extremes of the continuum of the base of support dimension. The side-by-side stance limits the width of the support area in the sagittal plane whereas the tandem stance minimizes the base of support dimension in the frontal plane. The staggered foot position, on the other hand, is located in the middle of the continuum and, therefore, displayed a “hybrid” blend of the dynamic properties of both the side-by-side and tandem stances. The results of Experiment 1 showed that the COP mean velocity was predominantly influenced by the intentionally induced body weight distribution in the side-by-side stance, however, it was more influenced by the foot position in the tandem stance. In the staggered stance, when the participant’s body weight distribution satisfied their natural preference (30/70), the participants tend to use a control strategy similar to the side-by-side foot position, otherwise to the tandem stance.

Previously, it has been reported that postural sway in the AP and ML directions is relatively independent (Kapteyn, 1973; Winter et al., 1996). Mochizuki et al. (1999, 2006), Kinsella-Shaw, Harrison & Turvey (2011) and King et al. (2012) showed a pattern of the intra-foot coordination in quiet stance using both linear and nonlinear analyses. In Experiments 2 and 3, we found that the inter- and intra-foot coordination co-existed even in the side-by-side stance. It appears, therefore, that even though the inter-foot coordination in the AP direction displayed larger weighting as compared with that in the ML direction, the COP$_{Net}$ pattern reflects the dynamic coupling of these four COP time series. The aforementioned “hybrid” blend dynamics was also observed in Experiment 2,
where the 4 COPs shared the contribution to the COP$_{\text{Net}}$ in the staggered foot position with the right foot oriented 30º.

The results of Experiment 2 revealed the interactive foot position and orientation effect on body weight distribution. Consistent with the previous studies (Jonsson et al., 2005; King et al., 2012), the participant loaded more on their rear foot when positioned one foot in front of the other. In addition, they shifted a greater proportion of their body weight to the left when their right foot was oriented from 90º to 30º. For the inter- and intra-foot coordination dynamics, however, foot position played a predominantly stronger role than foot orientation because it directly constrains the base of support dimension. In particular, when postural stance was challenged by a limited support area, the COPs of the unstable plane (inter-foot coordination) displayed a significant contribution to the foot coordination dynamics. In contrast, when standing posture was not challenged by the base of support boundaries, the COPs of the more loaded foot (intra-foot coordination) dominated the foot coordination in quiet stance.

**Body Weight Distribution.** In Experiment 1, the body weight distribution was one of the independent mechanical factors that channeled the inter- and intra-foot coordination dynamics of the postural control system. However, the body weight distribution served as a dependent variable in the 2nd and 3rd experiments. In fact, the body weight distribution covaries with foot position and foot orientation. In the side-by-side stance, the participants tend to load evenly on both feet, but load asymmetrically when one foot is positioned in front of the other (King et al., 2012; Wang et al., 2012; Johsson et al., 2005). Even though the body weight distribution served as an independent variable in Experiment 1, a significant foot position effect was still realized, especially
when the standing difficulty of the task increased. Further, we observed that the interactive effects of foot position and orientation on the body weight loading in Experiment 2. Experiment 3 showed that this evenly distributed body weight loading preference in the side-by-side stance was not influenced by the width and orientation of the reduced base of support.

Overall, mechanical factors, especially foot position or the dimension of the support area, played an important role channeling the loading preference in quiet stance. We proposed that the asymmetrical body weight distribution could present by differentiating the shortened base of support orientation in the side-by-side foot position (e.g., beam oriented along the longitudinal axis for the left and horizontal axis for the right foot, respectively).

**Base of Support.** The dimension of the base of support can be manipulated in two different ways: 1) by changing the foot position (Experiment 1 and 2); and 2) by reducing the size of the support area directly (Experiment 3). In Experiment 1, we found that the staggered stance revealed a “hybrid” postural control strategy of the side-by-side and tandem foot positions in both linear and nonlinear analyses. In Experiment 2, we found that there was an essentially equal contribution of the four COPs on postural stability in the staggered 30º right foot orientation condition. Consistent with previous studies, the base of support dimension is one of the significant mechanical factors channeling the body weight distribution and foot coordination dynamics (Day et al., 1993; Winter et al., 1993, 1995; Gatev et al., 1999; Aruin et al., 1998; Mochizuki et al., 2006).

The results of Experiment 3 revealed that when the shortened base of support was oriented along the horizontal axis, especially in the Horizontal (2.5cm) condition, the four
COP time series had a similar level of contribution to postural stability. In the past literature, evidence exists that the interdependence of the spontaneous COP_{Net} sway in both planes increases by reducing the area of the base of support (Aruin et al., 1998; Mochizuki et al., 1999, 2006). In our study, we directly showed that the increased postural stability induces an enhanced inter- and intra-foot coordination dynamics.

To sum up, our investigations showed that the equal contribution of the four COP time series on postural stability occurs, at least, in two circumstances: 1) that the dimensions of the support area in the AP and ML directions are similar (e.g., Experiment 2, staggered 30° condition); and 2) when postural stability is challenged by the increased task difficulty (e.g., Experiment 3, Horizontal 2.5cm condition).

**Foot coordination dynamics.** The upright equilibrium in the sagittal plane is a saddle-type dynamic, involving both stable and unstable manifolds, i.e., regions where the postural state transits close to an equilibrium point (i.e., the points where the sagittal plane shear force equals to zero) and regions where it falls away from the upright position, respectively (Zatsiorsky & Duarte, 1999, 2000; Morasso & Sanguineti, 2002; Hsu et al. 2007). Postural control at the unstable manifolds involves feedback processes with time delays (Peterka, 2002). According to Experiment 1, the postural control process can also be interpreted as a set of “running solutions” combined with multiple “stable” (or mode-locked states) and “flexible” inter-foot coordination epochs. The “stable” characteristic reflects the magnet property of these two COP time series whereby they cooperate with each other to move at the same cadence. The “flexible” strategy, on the other hand, represents the competitive process of the self-organized system where one signal tries to
lead the other and finally escapes from the previous steady state (von Holst, 1973; Haken et al., 1985).

The cooperative process is present in the dimension where the support area is relatively small accompanied by an increased demand of consistent inter-foot coordination. The competitive process, on the other hand, reflects the flexibility of the system exploring the internal and external perturbations (Kugler & Turvey, 1987; Kelso, 1993). Since the inter-foot coordination dynamics can be stabilized at different phase relations, the postural control system essentially displays a feature of “multistability” with multiple stable states or attractors; the stability of a state depends on how quickly the system returns to a state or another state following a perturbation (phase transition) [Kelso, 2012]. Future study can be focused on whether there is a direct mapping between the nonlinear “multistability” and the above mentioned stable manifold close to the equilibrium.

In Experiment 2 and 3, we further investigated the foot coordination dynamics in a way to collaborate COP time series in both AP and ML directions. As most of the previous studies showed, postural control mechanisms are primarily investigated in the sagittal plane that discount the control process in the frontal plane and further the cooperative process between the frontal and sagittal planes (Riccio, 1993; Massioin, 1992). We showed that the COP time series of the two orthogonal directions are not independent as the previous studies implied instead they are coordinated in a certain way to satisfy the task constraints (e.g., foot position, orientation and so on). Again, future study can be conducted by examine the inter- and intra-foot coordination dynamics from
different levels of analyses (e.g., kinematic joint motion) and also the linkages between these levels.

**Limitations and Future Directions**

There are a few limitations in these studies. Firstly, the kinematic joint motions and muscle activities of the lower limbs were excluded from the investigations. The experiments were conducted based on the framework of the multi-linkage model (Scholz & Schöner, 1999; Park & Horak, 2004; Hsu et al., 2007) and assumed that the anti-gravity torque is generated from the motions of the joints and the muscle-activations at the lower limbs of the postural control system. However, there is no direct examination on how the COP\textsubscript{Net} pattern and foot coordination dynamics relate to the redundant joint kinetic and kinematic motions. This is important because it has been observed in our testing that, in the tandem foot position, the participants utilized their upper limb to compensate for perturbations in the frontal plane. In addition, when standing on the horizontally oriented narrowed support, the participants introduced the hip and knee motions to assist balance. Further studies are anticipated to investigate: 1) how synergies at the muscular-articular level contribute to the inter- and intra-foot coordination dynamics and, therefore, the COP\textsubscript{Net} pattern; and 2) how the synergies adapt to satisfy different task, environmental and organismic constrains.

Secondly, given the multivariate property of the COP time series, appropriate data analysis techniques should be selected and interpretation of the results should be cautiously addressed. For example, according to the HKB model (Kelso, 1984; Haken et
al., 1985), a direct mapping has been observed between the index fingertips’ relative motion and their muscle activities. The in-phase coupling represents the activation of the homologous muscles whereas the anti-phase coordination indicates the activation of the non-homologous muscles. However, due to the properties of the redundant tasks (James, 2009; James & Newell, 2011), discrepancies exist between the HKB model and our findings, in that the lower limb muscle activation patterns cannot be interpreted from the COP_L-COP_R relative phase dynamics (Sozzi et al., 2013). More specifically, the COP_L-COP_R relative phase of the frontal plane showed in-phase coordination in tandem stance. However, previous studies reported that the peroneus longus (PER) muscles of both legs displayed a reciprocal activation pattern (Sozzi et al., 2013).

Thirdly, it is well known that postural control is mediated by various sources of sensory feedback: visual, vestibular, tactile and proprioceptive information. It has been shown that decreasing the body weight applied on one side of the limbs decreases the activation of the plantar cutaneous mechanoreceptors in the sole of the foot and consequently decreases the quality of the motor response (Okubo, Watanabe, & Baron, 1980; Genthon & Rougier, 2004; Anker et al., 2008). When standing in a tandem foot position, Golgi tendon organs (“load receptors”) located within the lower limb extensors provide additional proprioceptive information (Diener, Dichgans, Guschlauer, & Mau, 1984; Dietz, Gollhofer, Kleiber, & Trippel, 1992). With the manipulation of different mechanical factors, the amount of sensory information provided to our postural control system also changed. Future studies can be conducted with the sensory information integration and re-weighting perspective to investigate their effects on the inter-and intra-
foot coordination dynamics (Oie, Kiemel, & Jeka, 2002; Peterka, 2002; Carver, Kiemel, & Jeka, 2006; Polastri, Barela, Kiemel, & Jeka, 2012).

**General Conclusions**

In summary, this dissertation examined the interactive effects of the mechanical and task constraints on the body weight distribution and the inter- and intra-foot coordination dynamics of postural stance. Firstly, the body weight distribution and foot position were not controlled in parallel as a series of animal studies have shown (Lacquaniti & Maioli, 1992, 1994). In fact, their interactive effects have been observed in both linear and nonlinear analyses whereby the inter-foot coordination was significantly influenced by the intended body weight distribution in the side-by-side foot position whereas it was dominated by foot position in the tandem stance. The staggered stance, however, displayed the “hybrid” characteristics of the side-by-side and tandem foot positions.

Secondly, compared with foot orientation, foot position played a predominant role channeling the inter- and intra-foot coordination. In particular, when postural stance was challenged by the limitation of the base of support, the COPs in the unstable plane (inter-foot coordination) had larger weightings. In contrast, when standing posture was not challenged by the base of support boundary, the COPs of the more loaded foot (intra-foot coordination) dominated foot coordination in postural control.

Thirdly, the orientation of the base of support was more important to the foot coordination dynamics as compared with the width of the support surface. When the
shortened beam was oriented along the horizontal axis, the inter-dependence of the inter- and intra-foot coordination increased. When the beam was positioned along the longitudinal axis, the inter-foot coordination in the AP direction was observed.
REFERENCES


FOOTNOTES

1. In applied statistics, when the assumption of normality has been violated, nonparametric statistics are mostly recommended. However, often it is assumed that nonparametric methods lack statistical power and that there is a paucity of techniques in more complicated research designs, such as in testing for interaction effects. In these situations, two distinct approaches can be applied: 1) transform the data to a form more closely resembling a normal distribution framework, such as log transformation, square root transformation, etc. or 2) use a distribution free procedure such as rank transformation (RT). Rank transformation procedures are ones in which the usual parametric procedure (e.g., the standard analysis of variance—ANOVA) is applied to the ranks of the data instead of to the data themselves. One form of the rank transformations is to rank the entire set of observations from its smallest to largest, with the smallest observation having rank 1, the second smallest rank 2, and so on. Average ranks are assigned in case of ties.

2. Non-redundant coordination tasks are those in which a specific pattern of coordination between two effectors is the task goal. Studies of bimanual coordination have typically used non-redundant coordination tasks (e.g., Kelso, 1984). Redundant coordination tasks are those in which multiple, or even an infinite number of, potential coordination solutions could potentially be used to satisfy the task goal. These tasks include actions such as throwing, locomotion and maintaining posture in the face of internal and external perturbations.
3. The COP\textsubscript{L}-AP, COP\textsubscript{R}-AP, COP\textsubscript{L}-ML and COP\textsubscript{R}-ML are calculated according to the global coordinate reference of the participant.
APPENDIX : Compressing movement information via principal component analysis (PCA): comparison between the time and frequency domain PCA

PCA is a multivariate data transformation technique that is a form of factor analysis (Anderson, 1963; Morrison, 2004) and has its origins with Pearson (1901). The data are considered to be vector-valued random variables, where the entries of the vectors are mutually correlated. PCA transforms p-variate data vectors with correlated entries into p-variate data vectors with uncorrelated entries.

Let \( \mathbf{y}_i = [y_{i1}, y_{i2}, \ldots, y_{ip}]^T \) be a p-variate data vector associated with the i-th subject (replication) in a homogeneous population of subjects (the superscript \( ^T \) denotes transposition). To ease the presentation it is supposed that \( \mathbf{y} \) is centered, i.e., the probability distribution across subjects of each \( y_{ik} \), \( k=1,2,\ldots,p \), has mean zero. Let \( \mathbf{V} \) denote the (p,p)-dimensional covariance matrix associated with \( \mathbf{y}_i \), \( i=1,2,\ldots \). Then the Spectral Theorem for finite-dimensional symmetric matrices (Halmos, 1974) implies that \( \mathbf{V} \) can be uniquely decomposed as follows:

\[
\mathbf{V} = \mathbf{EDE}^T
\]  

(1)

where \( \mathbf{E} \) represents a (p, p)-dimensional ortho-normal matrix. That is, \( \mathbf{EE}^T = \mathbf{E}^T \mathbf{E} = \mathbf{I}_p \) [ \( \mathbf{I}_p \) is the (p,p)-dimensional identity matrix]. In addition, \( \mathbf{D} \) is a (p,p)-dimensional diagonal matrix. The Spectral Theorem implies the important result that the decomposition given by (1) is unique. Each particular covariance matrix \( \mathbf{V} \) yields unique matrices \( \mathbf{E} \) and \( \mathbf{D} \). The p-variate columns of \( \mathbf{E} \) are called the eigenvectors. The entries along the diagonal of \( \mathbf{D} \) are called the eigenvalues. If \( \mathbf{V} \) is a positive-definite covariance matrix then all eigenvalues are strictly larger than zero.
The PCA transformation of vectors with correlated entries into vectors with uncorrelated entries is defined as follows:

\[ z_i = E^T y_i, \quad i=1,2,\ldots \]  

(2)

In (2) \( z_i = [z_{i1}, z_{i2}, \ldots, z_{ip}]^T \) denotes the transformed p-variate vector of subject i. It is called the vector of principal component scores, where the principal component scores are \( z_{ik} \), \( k=1,2,\ldots,p \). It can be shown, using simple matrix algebra, that the covariance matrix of \( z_i \) is \( D \), hence the entries \( z_{ik} \) of \( z_i \) are mutually uncorrelated.

PCA can be used for data reduction with minimal loss of information. Let the eigenvalues \( d_{kk}, \quad k=1,2,\ldots,p \), along the diagonal of \( D \) in (1) be ordered in decreasing amplitude. That is, \( d_{11} \geq d_{22} \geq \ldots \geq d_{pp} \). This can always be achieved by means of suitable permutation. Let S be the sum of the p eigenvalues \( d_{kk} \), \( k=1,2,\ldots,p \), and suppose that the sum of the first \( q \) eigenvalues \( d_{11} + \ldots + d_{qq} \) equals 0.8S. Then it is said that the first \( q \) eigenvectors “explain 80% of the variance of \( V \)”. This implies that the remaining \( (p-q) \) eigenvectors together explain 20% of the variance. One, therefore, can consider a dimension reduction by defining the \( (p, q) \)-dimensional matrix \( E^{(q)} \) consisting of the first \( q \) eigenvectors of \( E \) (the ones explaining 80% of the variance of \( V \)). In addition, the remaining \( (p-q) \) eigenvectors are collected in the \( (p, p-q) \)-dimensional matrix \( E^{(p-q)} \).

Hence \( E \) is decomposed as \( E = E^{(q)} + E^{(p-q)} \). Also, analogously, decompose \( z_i \) as \( z_i = [z_i^{(q)}, z_i^{(p-q)}]^T \). Pre-multiply (2) by \( E \):

\[ z_i = E^T y_i, \quad E z_i = E E^T y_i \rightarrow E z_i = y_i \]

Hence, pre-multiplication of \( z_i \) by \( E \) yields back the original data vectors \( y_i \). Next, substitute the decomposition \( E = E^{(q)} + E^{(p-q)} \) into \( E z_i = y_i \):

\[ y_i = E z_i = E^{(q)} z_i^{(q)} + E^{(p-q)} z_i^{(p-q)} \]  

(3)
In (3) denote $E^{(q)}$ as $\mathbf{A}$, denote $z^{(q)}_i$ as $\eta_i$, and $E^{(p-q)}z^{(p-q)}_i$ as $\varepsilon_i$. Accordingly (3) can be rewritten as:

$$y_i = \Lambda\eta_i + \varepsilon_i \quad (3')$$

that resembles a q-factor model (Anderson, 2003; Morrison, 2004).

Denote the (p,p)-dimensional covariance matrix of $\Lambda\eta_i = E^{(q)}z^{(q)}_i$ by $V^{(q)}$ and the (p,p)-dimensional covariance matrix of $\varepsilon_i = E^{(p-q)}z^{(p-q)}_i$ by $V^{(p-q)}$. Then it holds that:

$$V = V^{(q)} + V^{(p-q)} \quad (4)$$

and the choice of $\Lambda\eta_i = E^{(q)}z^{(q)}_i$ minimizes $V^{(p-q)} = V - V^{(q)}$. In this sense dimension reduction by means of PCA is optimal (Ahmed & Rao, 1975).

**PCA in the time domain.** PCA in the time domain of a p-variate time series $y(t)$, $t=0,1,\ldots,$ proceeds in the same way as described above. Again, it is assumed, without affecting generality, that $y(t)$ is centered in that the mean of $y(t)$ is zero. In addition, it is assumed that $y(t)$ is weakly stationary (Brillinger, 1975), which implies that the sequential covariance of $y(t_1)$ and $y(t_2)$ for all times $t_1$ and $t_2$ only depends upon the lag $u = t_1 - t_2$. Consequently, the sequential covariance structure of $y(t)$ is given by:

$$\text{cov} [y(t), y(t-u)^T] = V(u), u=0,\pm 1, \ldots \quad (5)$$

Accordingly, $V(0) = \text{cov}[y(t), y(t)^T]$ is the (p,p)-dimensional covariance matrix of $y(t)$ at lag $u=0$. Then PCA in the time domain (PCA$_t$) proceeds as described above by substituting $V(0)$ for $V$, yielding the analogue of (3'):

$$y(t) = \Lambda\eta(t) + \varepsilon(t) \quad (6)$$

In psychometrics PCA$_t$ is known as $P$-technique (Cattell 1952, 1963). Anderson (1963) and Holzman (1963) presented a critique of $P$-technique because it neglects the sequential covariance structure at lags $u \neq 0$. Molenaar and Nesselroade (2009) showed
that neglecting the sequential covariance structure at lags \( u \neq 0 \) in PCA yields invalid results if the relationship between the \( q \)-variate component series \( \eta(t) \) and the \( p \)-variate observed series \( y(t) \) is not given by (6) but by

\[
y(t) = \Lambda(0)\eta(t) + \Lambda(1)\eta(t-1) + \ldots + \epsilon(t) = \sum_{u=0}^{\infty} \Lambda(u)\eta(t-u) + \epsilon(t), u=0,1,\ldots
\]  

(7)

According to (7) the regression of the observed series on the component series is characterized by differential lead-lag relationships. If the observed series \( y(t) \) obeys (7) then determination of the dimension \( q \) of the component series \( \eta(t) \) is biased upwards. That is, the estimated value of the dimension \( q \) of component series \( \eta(t) \) will tend to be too high.

**PCA in the frequency domain.** A version of PCA that is valid for any weakly stationary time series is PCA in the frequency domain (Brillinger, 1975). Denote by \( y(\lambda_k) \) the discrete Fourier transform (DFT) of \( y(t) \), \( t=0,1,\ldots,T-1 \):

\[
y(\lambda_k) = T^{-1/2} \sum_{t=0}^{T-1} y(t)e^{-i2\pi\lambda_k t}
\]

(8)

\[\lambda_k = \frac{k}{T}, k = 0,1,\ldots,T-1\]

\( i = \sqrt{-1} \) is the imaginary unit and \( e^{-i2\pi\lambda_k t} = \cos(2\pi\lambda_k t) - i \sin(2\pi\lambda_k t) \). The DFT given by (8) is a one-to-one invertible transformation from the time domain (indexed by \( t \)) to the frequency domain (indexed by \( \lambda_k \)). The inverse discrete Fourier transformation (IDFT) is given by

\[
y(t) = T^{-1/2} \sum_{k=0}^{T-1} y(\lambda_k) e^{i2\pi\lambda_k t}, t = 0,1,\ldots,T-1
\]

(9)
It is apparent that the DFT does not affect the information content of a time series \( y(t) \), but only represents it in a different (frequency) domain. A very useful characteristic of the DFT is that application to (7) yields (Ahmed & Rao, 1975):

\[
y(\lambda_k) = A(\lambda_k)\eta(\lambda_k) + \epsilon(\lambda_k)
\]

Hence, the DFT transforms (7) in which the regression of the observed series on the component series is characterized by arbitrary differential lead-lag relationships into a complex-valued analogue of (6) which is valid for application of PCA.

PCA proceeds by carrying out the complex-valued analogue of standard PCA to the \((p,p)\)-dimensional complex-valued covariance matrices of \( y(\lambda_k) \), \( k=0,1,...,T-1 \). These covariance matrices are called the spectral density matrices and will be denoted by \( V(\lambda_k) \), \( k=0,1,...,T-1 \). Analogous to (6) each \( V(\lambda_k) \) is decomposed as:

\[
V(\lambda_k) = V(\lambda_k)^{(q)} + V(\lambda_k)^{(p-q)}
\]

based on the analogue of (3):

\[
y(\lambda_k) = E(\lambda_k)z(\lambda_k) = E(\lambda_k)^{(q)}z(\lambda_k)^{(q)} + E(\lambda_k)^{(p-q)}z(\lambda_k)^{(p-q)}
\]

The \( n \)-th entry of the \( j \)-th column of \( E(\lambda_k) \) is \( e_{nj}(\lambda_k) = \text{Re} e_{nj}(\lambda_k) - i\text{Im} e_{nj}(\lambda_k) \). This so-called Euclidean representation of the elements of the complex-valued eigenvectors in \( E(\lambda_k) \) can be changed into polar representation as follows:

\[
e_{nj}(\lambda_k) = |e_{nj}(\lambda_k)|e^{-iq_{nj}(\lambda_k)}; n, j = 1,2,...,p
\]

\[
|e_{nj}(\lambda_k)| = \sqrt{\text{Re}(e_{nj}(\lambda_k))^2 + \text{Im}(e_{nj}(\lambda_k))^2} \quad \text{and}
\]

\[
q_{nj}(\lambda_k) = \tan^{-1}\left[-\frac{\text{Im}(e_{nj}(\lambda_k))}{\text{Re}(e_{nj}(\lambda_k))}\right]
\]

The absolute value \( |e_{nj}(\lambda_k)| \) is called the amplitude of the \((n,j)\)-th entry of \( E(\lambda_k) \) and \( q_{nj}(\lambda_k) \) is called its phase. If the amplitude of \( e_{nj}(\lambda_k) \) is high then the \( j \)-th component
explains a considerable amount of the variance of the \(n_{th}\) observed series at frequency \(\lambda_k\).

The information about complex-valued eigenvectors will be presented in the form of amplitude and phase difference spectra.

To sum up, the main steps in PCA of a given \(p\)-variate time series \(y(t)\), \(t=0,1,\ldots,T-1\) are: a) estimation of the \(V(\lambda_k)\), \(k=0,1,\ldots,T-1\); b) determining the eigenvalue decomposition of each \(V(\lambda_k) = E(\lambda_k)^*D(\lambda_k)E(\lambda_k)\), where the superscript * denotes the conjugate transpose; c) carrying out the unitary rotation to minimum phase described in the Appendix and determining the number of important components \(z(\lambda_k)^{(q)}\); and d) taking the IDFT of \(z(\lambda_k)^{(q)}\), \(k=0,1,\ldots,T-1\), to obtain the time-domain \(q\)-variate component series \(z(t)^{(q)}\), \(t=0,1,\ldots,T-1\). Because each \(V(\lambda_k)\) is a \((p,p)\)-dimensional Hermitian covariance matrix, it follows that the eigenvalues along the diagonal of \(D(\lambda_k)\) are real-valued. Estimation of the spectral density matrices \(V(\lambda_k)\), \(k=0,1,\ldots,T-1\) can be accomplished in several distinct ways (Brillinger, 1975). Here, we use the simplest approach possible by taking the DFT of \(y(t)\) defined by (8), constructing the \((p,p)\)-dimensional so-called periodogram matrices \(H(\lambda_k) = y(\lambda_k)y(\lambda_k)^*\), and estimating each spectral density matrix by

\[
V^{est}(\lambda_k) = (2M + 1)^{-1} \sum_{m=M-k}^{M+k} H(\lambda_m), k = 0,1,...,T-1 \quad (14)
\]
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