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THE EXAMINATION OF THE PATTERNS OF BI-LATERAL GROUND
REACTION FORCES DURING QUIET STANDING

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ABSTRACT

The main purpose of this study was to determine the degree of lower limb independence and relatedness during quiet standing. Nine healthy young adults (5 females, 4 males; 22 ± 1 year; 1.69 ± 0.08 m; 73 ± 13 kg) stood with one foot on each of two separate force plates where they performed three eyes-open and three eyes-closed quiet standing trials. Various traditional center-of-pressure (COP)-based measurements were made in order to determine if the subject pool demonstrated postural dynamics typical of healthy young adults. The focal point of the study, however, were the approximate entropy (ApEn) measurements, which were used to ascertain the orderliness of the vertical ground reaction forces created under each foot. Data were typical when assessed using traditional measures of COP motion. Results of the ApEn measures suggested that the left leg and right leg are controlled independently of each other during quiet standing. The findings of this study will be useful to researchers and clinicians who are interested in learning more about the possible control mechanisms employed during quiet stance. Added knowledge to the field will optimistically lead to improved balance therapies and evaluation tools.

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Chapter 1

Introduction

1.1 Introduction

Although a seemingly trivial process, researchers and clinicians have studied the topic of upright human posture for several decades. This work has been motivated by the desire to deepen the understanding of how humans remain balanced and to use this understanding in the development of clinical assessment tools and therapeutic techniques that are intended to evaluate the balance of an individual so as to predict a patient's fall risk and rehabilitate any balance deficits that may be present.

Quiet standing, which may be thought of as when a person stands upright in a comfortable postural position and tries to remain still, is a common condition from which researchers analyze the balance of an individual. Interestingly, quiet standing is considered as a dynamic process as there is a certain degree of postural sway associated with it. To elaborate further, postural sway is created as a subject shifts his/her weight - generally, this is an involuntary action that takes place naturally and unconsciously - in the forward/backward and left/right directions. Researchers are able to quantify the mechanics of postural sway and, in some cases, they are able to characterize what typical postural dynamics are for a certain population of subjects (e.g., young adults vs. elderly adults, or healthy patients vs. cerebral palsy patients). Large deviations from these typical postural dynamics can then be used as a sign that a pathologic condition is present or that a decline in postural control is occurring (e.g., aging is associated with a decline in postural control – See *Section 2.4.1*).

1.2 Purpose

Traditional studies of quiet standing have had subjects stand with both feet on one force plate, and examined the net force produced by both limbs (Winter et al., 1995). This type of analysis, however, may pre-suppose that both limbs are controlled in a similar fashion. In this project the vertical ground reaction forces beneath both feet were measured independently thereby permitting the comparison of the two sets of forces and the degree of their independence and relatedness. Therefore, the purpose of this project was to ascertain the degree of lower limb independence and relatedness during quiet standing. A number of control mechanisms are feasible and the analysis will attempt to identify the one at play with regards to these forces during quiet standing.

1.3 Experimental Overview

Subjects stood in quiet stance with one foot on one force platform and the other foot on a separate force platform. Using custom designed MATLAB code, various traditional center of pressure metrics were calculated from the data gathered by the force plates. The programming also employed approximate entropy algorithms to analyze the vertical ground reaction forces under each of the subjects' feet and the inter-relatedness of these signals. The basic rationale for data analysis was twofold: The center of pressure metrics that were computed could be compared to similar measures taken in other studies; thus, helping establish whether the subject pool of this study displayed typical quiet stance dynamics. The approximate entropy measurements allowed for the quantifying of lower limb independence and a basis from which conclusions could be drawn about the possible postural control mechanisms at work during quiet standing. Chapter 3 contains a more extensive description of the methods used in this study.

Chapter 2

Review of Literature

2.1 Balance

Despite its trivial appearance, quiet human standing is a complex process of great research interest. The following sections serve to provide a brief overview of how researchers define human balance and postural stability. The mechanics required to maintain postural stability and the clinical measures used to assess it are also reviewed.

2.1.1 Defining Human Balance

In the field of Newtonian mechanics, an object is assumed to be in a state of balance, if its center of mass (COM) falls within its respective base of support (BoS) (Pollock et al., 2000). Ergo, if its COM falls outside the BoS, the object is unbalanced.

It is quite important to differentiate between a few terms that will be frequently used in this text. The COM is the point where the total body mass resides in a global reference frame and is the weighted average of the COM of all body segments (Winter et al., 1995). The center of gravity (COG), on the other hand, may be considered as the point where the weight of the body acts and is not a fixed point as it is sensitive to the position of all the body segments (Rodgers and Cavanagh, 1984). The center of pressure (COP) is a parameter often measured using a force platform and is the equivalent to the weighted average of all the pressures created by all surfaces that are in contact with the ground (Winter et al., 1995). Thus, during quiet bipedal stance the

overall COP lies somewhere in between both feet. COM, COG, and COP are sometimes used interchangeably, and it may now be appreciated from their definitions that it is incorrect to do so.

While balance remains a trivial concept for inanimate objects, the subject of human balance is one that remains undefined. Various definitions for human balance have been proposed a few of which include:

- “Balance is a generic term describing the dynamics of body posture to prevent falling.” (Winter et al., 1995, pp. 194)
- “A multidimensional concept referring to the ability of a person not to fall.” (Pollock et al., 2000, pp. 405)
- “Unsupported, standing humans are in unstable equilibrium, or balance...” (Horak et al., 1987, pp. 1881)

Although the term balance is used quite frequently in many clinical settings, the term itself - with respect to human balance - does not have a universally accepted definition (Berg, 1989). This presents problems for therapists and physicians, who are often required to assess the balancing ability of their patients. This will be discussed in future sections. The intimate relationship between human balance and postural control will be explored next.

2.1.2 Defining Postural Control

Revisiting Newtonian mechanics, the notion of stability is straightforward: the further the COM may be displaced before the object becomes unbalanced, the greater the stability of that object (Pollock et al., 2000). A human in quiet stance is subject to this physical definition, as he/she is constantly adjusting their COMs: During quiet standing, the human body sways slightly anteriorly/posteriorly and medially/laterally. This is often referred to as postural sway.

However, unlike an inanimate object (unless acted upon by an outside force), a human may regain its stability by acting upon itself. While this may be through a conscious task (i.e., stepping a foot forward to avoid a fall) or an unconscious task (i.e., remaining upright during quiet stance), postural control has been considered by some as the ability to maintain equilibrium in a gravitational field by keeping or returning the center of body mass over its base of support (Horak et al., 1987). It has been suggested through prior work that postural control is maintained by a complex sensorimotor system that integrates information from the visual, vestibular, and somatosensory systems (Prieto et al., 1996).

2.1.3 Maintaining Postural Control

Simple mechanics govern upright biped stance. As long as an object's COM remains within its BoS, it will not fall. If the COM falls outside the BoS, then a moment will be created thus causing a disturbance in the object's equilibrium. If this perturbation is not addressed, then the object will topple. Pollock et al. (2000) have suggested that human postural control is required for the following: preserving a specific posture (i.e., sitting or standing), voluntary movements (i.e., walking or bending about the waist), and compensating in reaction to external disturbances (i.e., slipping or tripping). The goal of all of these actions is to have one's COM remain within his/her BoS.

In a study by Murray et al. (1975), normal upright posture was deemed to possess two distinguishing qualities: a relatively large area of stability over which weight may be shifted and maintained along with a COP, which is the point of application of the ground reaction forces with the support surface and is usually measured with a force plate, that is fluctuating while still remaining close to the mean COP. They also observed a mean COP that fluctuated slightly anterior to the vertical projection of the COM, which was referenced from the work of

Hellebrandt et al. (1938) as being approximately 5 centimeters anterior to the lateral malleolus, and suggested that normal posture may be preserved by muscular contractions that keep the COM behind the COP.

It may be inferred from the findings of Murray et al. (1975) and a simple knowledge of mechanics that normal postural stance desires a COP that remains well within a certain stability boundary. Slobounov et al. (1997) used this perspective in analyzing postural control. A two-dimensional stability boundary for the COP was defined based on the lateral edges of subjects' feet along with functional sway parameters gathered from other studies. Along with this stability boundary, a parameter known as virtual time-to-collision (VTC) was analyzed. VTC was used to relay the spatiotemporal proximity of the COP to the stability boundary. It was assumed that postural equilibrium would be maintained as long as the subject was able to avoid COP collisions with the stability boundary. A review of literature suggests that at the time this was the first time VTC was applied to postural control studies. Slobounov et al. (1997) promoted the use of the VTC parameter in future studies as their results proposed there was a threshold value of VTC when the COP approached certain areas of the stability boundary. This in turn prompted compensatory movements in order to maintain postural stability.

2.1.4 Clinical Measures

Due to the need for therapists to both improve postural stability and develop a useful evaluative tool to predict the risk of falls for at risk populations (e.g., the elderly; subjects with pathologies that cause a decline or cause an alteration in the neuromuscular, somatosensory, visual, or musculoskeletal systems), much attention has been invested in the developing and testing of clinical measures of balance.

Horak (1987) reviewed what were considered typical measures of balance. These tests (i.e., standing with one foot in front of the other, standing on a tilt board or ball, standing on narrow rails, pushing the trunk of a subject) usually involved measuring the length of time a subject could maintain a particular equilibrium position without faltering while certain inputs to the postural control system - like visual and somatosensory data - were altered. Horak argued that these tests lacked the systematic approach necessary to truly gauge balance as usually only large perturbations were used and thus created exaggerated subject responses not typical of true ambulatory behavior.

Horak also suggested that ideal tests for balance should include measures that reflect both the functional abilities and quality of movements of the postural control system. Tinetti et al. (1988) attempted to present such a tool. A balance test that observed typical functional actions, such as getting up from a chair, sitting down, turning the trunk while walking, and raising the feet while walking was compared against common neuromuscular examinations (i.e., manual muscle testing like hip flexion, lower extremity adduction/abduction, and knee flexion/extension). It was found that the neuromuscular findings alone were not sufficient enough to identify mobility problems; however, a correlation between these findings and the ability to perform specific mobility maneuvers was observed. These findings along with those of others would lead to the development of various functional mobility tests: the Tinetti Assessment Tool (Tinetti, 1986), Mathias and associates' Get-up and Go test (Mathias et al., 1986), Gabel and Simons's Balance Coding (Gabel and Simons, 1982), and the Berg Balance Scale are just some examples (Berg et al., 1989).

Measuring postural control is not simple and is complicated by various factors. For example, the location of the COM is not easily ascertained and postural adjustments may occur in movement ranges that are difficult to detect with simple observation (Horak, 1987). Thus, the efficacy, validity, reliability, and the objectivity of balance measures are an area of interest.

Giorgeti et al. (1998) studied the reliability of a few quiet balance and ambulatory tests (i.e., One Leg Standing, Tandem Gait, and Functional Reach). Reliability levels of 0.73-0.85 for the One Leg Standing and Functional Reach tests were found while Tandem Gait was deemed unreliable with a reliability level of 0.31. The intrarater and interrater reliability scores of other pertinent testing procedures have been mentioned in other studies (Brauer et al., 2000): The Berg Balance Scale ($r = .91$), Tinetti Mobility Index ($r = .91$), the Get-up and Go Test ($r = .76$).

It may, however, be important to keep in the mind the characteristics of the members of the subject population being observed in each respective study. Giorgeti et al. (1998) studied a dichotomous group of non-disabled and disabled subjects (although criteria used to delineate between the two groups was not given explicitly) who were of both genders while Brauer et al. (2000) studied one hundred elderly women (mean age = 73 +/- 5 years) who were independent community-dwellers. The latter noted that their testing population might have affected their results, as the population was comprised of healthy adults that appeared to be functioning at higher levels of postural control than the populations of other studies. This may be true for other studies (King et al., 1994; Prieto et al., 1996), as well. Therefore, it is important to be cognizant to the subjects being tested and to the measure being used to do so. It is only through this careful observation that the valid comparison of results from related studies and the evolution of new postural control tools may be made.

While laboratory measures that utilize advanced equipment, such as force plates or motion analysis systems, may provide accurate measurements of various COP-based and kinematic parameters, these methods may not be practical for therapists to use on a daily basis. In fact, Brauer et al. (2000) suggest that COP measures alone are not enough to do things like predict one's risk of falling. These ideas - along with the notion that clinical balance tests are limited in their ability to detect subtle changes in postural stability (Horak, 1987; Brauer et al. 2000) - are just some of the reasons that the usefulness of clinical assessments must be improved.

Doing so would allow therapists to both better predict the risk of falling and enhance the postural stability and gait patterns of their patients.

2.2 Balance Mechanisms

The subsequent sections offer a brief description of the means by which the musculoskeletal system maintains and the central nervous system controls upright stance. Models of and common strategies used in postural control are also considered.

2.2.1 Muscles

As the maintenance of posture heavily depends upon the position of every segment of the body, it also inherently involves the use of several muscles from all segments of the body. However, that is beyond the scope of this section. Instead, a brief overview of the most prominent muscles involved with postural stability will be discussed.

A description of the muscles involved in the preservation of postural stability cannot be made without a brief mentioning of the two most traditionally accepted mechanisms for postural stability: the ankle strategy and the hip strategy. Under the ankle strategy postural stability is assumed to be controlled mainly by those muscles that act on the ankle joint while under the hip strategy postural stability is assumed to be controlled by the muscles that act on the hip joint.

According to Winter et al. (1995) – who instituted one of the first studies in which two force plates were used (one for each foot) – both an ankle and hip strategy are used during normal stance with the feet side-by-side. The ankle strategy was found to be responsible for anterior/posterior (A/P) sway while the hip strategy was found to be responsible for medial/lateral (M/L) sway. Under these results, it may be inferred that the main muscles involved in the ankle

strategy are the plantar flexors (i.e., gastrocnemius, soleus, plantaris, tibialis posterior, flexor hallucis longus, flexor digitorum longus, peroneus longus, and peroneus brevis) and dorsiflexors (i.e., tibialis anterior, extensor hallucis longus, and extensor digitorum longus). Conversely, the muscles governing the hip strategy are the hip abductors (i.e., the gluteus group mainly) and the hip adductors (i.e., adductor brevis, adductor longus, adductor magnus, pectineus, gracilis, gluteus medius, psoas major, and the iliacus). Rectus femoris and biceps femoris are also commonly studied muscles associated with the hip strategy. It is interesting to note that Winter et al. (1995) also found that the strategies controlling A/P and M/L sway were switched when subjects stood in the Romberg stance (i.e., with one foot in front of the other).

The results of Winter et al. (1995) were observed while using an inverted pendulum (IP) model of quiet stance. This model, which is one of the most traditionally utilized models of quiet stance, will later be described, but it is important to now note that the IP model is centered on the ankle strategy.

Reiterating that during normal upright stance the COM falls slightly anterior to the ankles, it should be no surprise that the plantar flexors - especially the calf muscles (i.e., gastrocnemius and soleus) – receive the most attention, as they are believed to create a backward torque to counteract forward toppling. This schema is reinforced by the work of Loram et al. (2004), which examined the muscle length of the calf muscles *in vivo* during small voluntary sways of their subjects. They observed what they called paradoxical movement in the calf muscles: During forward sway, the calf muscles shortened and vice versa during backward sway.

The importance of the plantar flexors was further supported in a study conducted by Ushiyama and Masani (2011). Here the plantar flexor muscle volume of each subject was estimated and then divided by their body mass in order to normalize the measure so that it could be compared between-subjects. Time-domain COP data was then gathered during quiet stance using force plates, and the relationship between this data and plantar flexor muscle volume was

examined. The researchers observed that more muscle volume of the plantar flexor group - relative to body mass - required the subject to produce less relative plantar flexion torque to maintain quiet stance. In other words, this suggests that if one were to consider two subjects who have the same body mass but differing plantar flexor muscle volume, the one with more plantar flexor muscle volume would need to exert less relative plantar flexion torque - as a percentage of maximum voluntary contraction torque – compared to the other subject. In turn, it was observed that the more plantar flexor muscle volume a particular subject in this study had, the less fluctuation of his/her COP (i.e., a negative correlation was seen).

Further research into the plantar intrinsic foot muscles, which are muscles with origins and insertions within the foot (i.e., abductor hallucis, flexor digitorum brevis, and quadratus plantae), by Kelly et al. (2011) has suggested a slight nuance into the previously accepted scheme. Instead of controlling A/P sway like most of the extrinsic plantar flexor muscles are believed to do, the activity of the plantar intrinsic foot muscles were found to be moderately to strongly correlated with the M/L component of postural sway. As postural demand of subjects was increased (i.e., going from double limb support to single limb support), the activity of the plantar intrinsic foot muscles was also found to increase thereby further validating that they have a role in postural control.

2.2.2 Neural Control

Postural stability may be considered a dynamic activity in which the body must constantly maintain balance while in upright stance. As such, the central nervous system (CNS) plays a major role in stabilizing the upright body and much study is given to the manner in which it accomplishes this feat. A simple negative feedback loop would perhaps seem to be the most intuitive and plausible explanation. In this scheme, the CNS would use sensory information (i.e.,

visual, proprioceptive, or vestibular information) to compensate for external disturbances. For example, if sensory information relays that the body is leaning forward, then the CNS would activate efferent nerves that would, in turn, create the appropriate motor function causing the body to sway backward.

Assuming a feedback model is indeed the means by which the CNS controls quiet standing, the question then becomes – What and how many parameters must the CNS monitor to sustain standing balance? One of the most popular models used to attempt to answer this question is one that assumes the CNS employs a proportional-integral-derivative (PID) controller. For instance, Johansson et al. (1988) used a PID controlling regime along with an inverted pendulum model (See *Section 2.2.4* for more detail on this model) to explain quiet standing. Based on the idea that the angular deviation from upright posture is recognized by the CNS as a type of angular error signal, Johansson et al. suggests that three corrective torques are necessary to stabilize the body: one that is proportional to the angular error, a second that is proportional to the time derivative of the angular error, and a third that is proportional to the integral of the angular error. A PID control systems deems that the proportional term accounts for error at the present time, the integral term accounts for the past errors, and the derivative term accounts for future errors. According to Johansson et al., the CNS would generate these three corrective torques based on sensory feedback information. Figure 2-1 illustrates this concept.

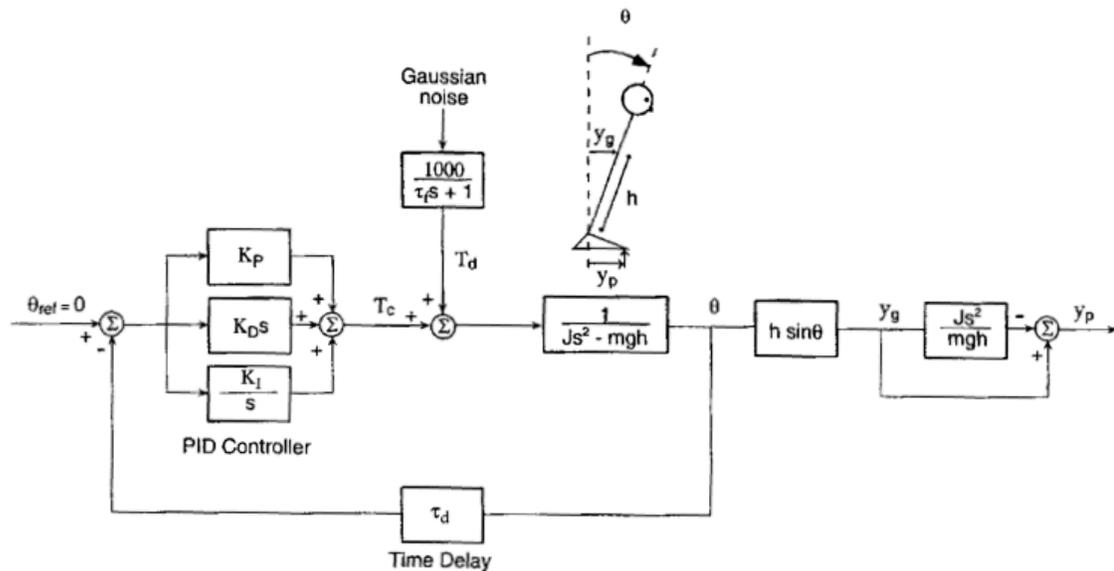


Figure 2-1. Inverted Pendulum Model of Quiet Stance Under PID Control. K_P , K_I , and K_D represent proportional, integral, and derivative terms of the PID (reproduced from Peterka, 2000).

While a negative feedback scheme seems feasible, many studies have found evidence that a feedforward strategy of postural control may be more likely. A feedforward strategy implies that an anticipatory nature is utilized regardless of the system output. In the case of postural stability, this would mean that potential postural disturbances are accounted for in a pre-defined manner before they even occur.

The results of Horak and Nashner (1986) support the notion of feedforward neurological control for upright stance. In brief, subjects were instructed to stand on two different types of support surfaces (i.e., a normal support surface and a support surface that was short in size in relation to the subjects' feet). Both types of surfaces were systematically moved using a hydraulic servomotor in order to generate A/P disturbances in the subjects' posture, and the corresponding muscle activation patterns used by the subjects were examined. Interestingly, on the normal support surface subjects exhibited muscle patterns associated with an ankle strategy while on the short support surface subjects used a hip strategy that activated muscles antagonistic for the ankle

strategy in an opposite proximal-to-distal fashion. These findings along with observations of muscle latency times led Horak and Nashner to conclude that the participants were able to create a continuum of different postural movements that may be employed in different magnitudes and in varying temporal relations. This evidence was consistent with their hypothesis that the CNS has a limited set of postural control programs that are selected in *advance* of movement.

Fitzpatrick et al. (1996) investigated the possibility that a feedback loop is responsible for preserving normal upright stance. The researchers were interested in calculating the loop gain of the postural control mechanism. This was calculated by multiplying the reflex gain and the muscle and load gain. The loop gain of the pathway would inherently describe the ability of the feedback control to resist disturbances and would have a value much greater than unity. On the other hand, an ideal feedforward pathway was described as having a loop gain of unity and would not introduce phase changes. The results of their study found a loop gain of approximately unity with only slight phase changes present in the data. As their results pointed towards a reflex response based on feedforward control, Fitzpatrick et al. suggested that perhaps afferent information may not only be used to correct past disturbances but also to anticipate future postural disturbances. The researchers were careful to note, however, that they do not completely reject the use of negative feedback in simple reflex pathways. Rather, they suggest that while postural stability cannot be achieved through a negative feedback loop alone, such a loop may operate in conjunction with a feedforward mechanism.

Various other studies (Gatev et al., 1999; Loram et al., 2004; Chvatal et al., 2011) presented data supporting some form of a feedforward mechanism. Gatev et al. (1999) inspected postural muscle activity and sway characteristics in different support (i.e., normal and narrow stance widths) and sensory (i.e., eyes open and eyes closed) conditions. The muscle activity results of their subjects' lateral gastrocnemius muscle suggested that a central program was active and working to both control ankle joint stiffness and anticipate the loading pattern. It may be

interesting to note that Gatev et al. found that vision affected this feedforward modulation as this is opposite to the conclusions made by Fitzpatrick et al. (1996).

While asking participants to voluntarily sway forward and backward slightly so as not to flex the hip or knee, Loram et al. (2004) investigated the muscular activity of postural muscles. They proposed that intrinsic ankle joint stiffness is not capable of sustaining human balance. With this as part of their basis and other findings involving muscle length, they concluded that muscle length rather than stiffness is the parameter that must be under predictive central control arguing this was the only way postural stability may be maintained. These results must be carefully considered as the subjects used by Loram et al. participated in voluntary A/P swaying, which may not be truly representative of normal quiet standing.

Chvatal et al. (2011) have further reaffirmed the notion that postural movements cannot be solely based on a direct somatosensory feedback loop. Studying both stepping and non-stepping postural responses (i.e., moving the base of support vs. normal quiet stance) of their subjects, they monitored the common muscle synergies used for each type of response. Their results showed that the same muscle synergies were recruited for both types of responses; however, this recruitment was found to be based on the desired direction of the COM movement and not by the perturbation direction. Arguing that a feedback-based system would display the same muscle synergies for the same perturbation directions, the researchers instead suggested a central control system based on flexible motor modules that are recruited to stabilize the COM. Chvatal et al. also proposed that the nervous system may operate on sensory input and create motor output separately.

The control of the CNS will undoubtedly continue to be a popular topic in the study of postural stability. Indeed, ascertaining the explicit means and degree to which various aspects of the CNS controls quiet standing would provide invaluable insight to researchers and clinicians

alike. From the evidence presented here previously, it seems the control scheme may perhaps involve the integrated action of both feedback and feedforward mechanisms.

2.2.3 Other Models

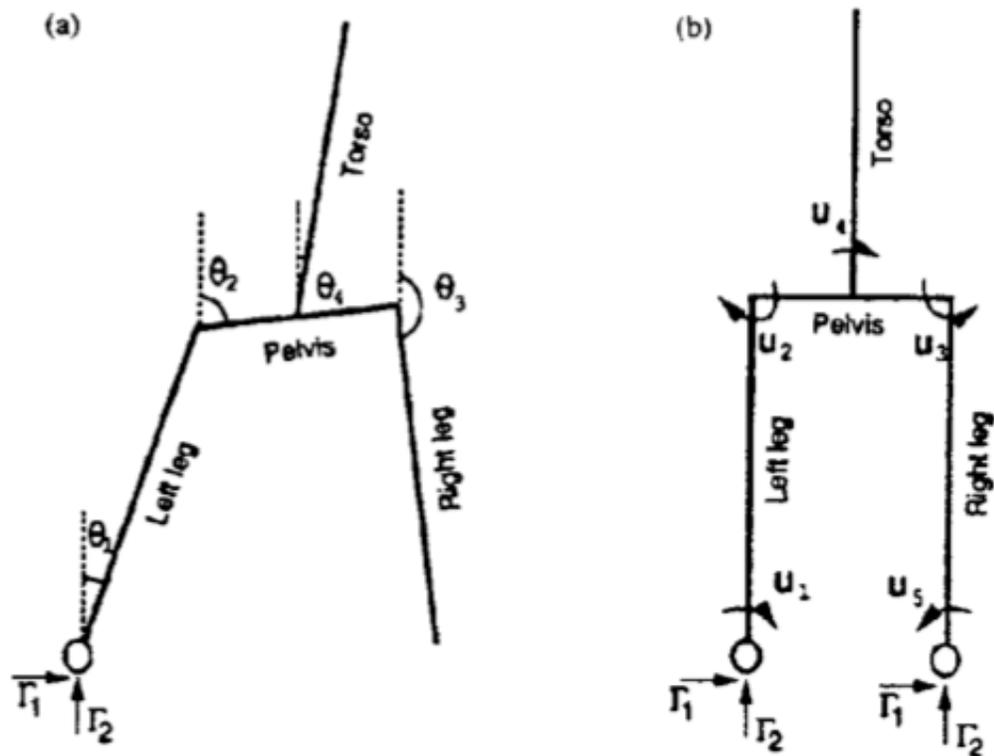


Figure 2-2. Four-link Biped Model in the Frontal Plane Consisting of Two legs, a Pelvis, and a Torso. (a) Model in unconstrained position. (b) Model constrained to equilibrium position. θ_{1-4} are position variables, u_{1-5} are the input of five moments of arm actuators, and $r_{1,2}$ are the output of tangential and normal ground reaction forces (reproduced from Iqbal et al., 1993).

Besides the IP model, the dynamic activity of quiet stance has been simulated using various other models. In an attempt to imitate motion in the frontal plane, Iqbal et al. (1993) proposed the use of a nonlinear, four-link biped system. Figure 2-2 illustrates this model.

Iqbal et al. used a computer simulation to predict the joint trajectories that their four-link biped model would produce as it operated under three different controlling mechanisms: intrinsic feedback, full decoupling feedback, and partial decoupling feedback. The notion of a full decoupling controller involves at least as many control inputs as degrees of freedom of the system of interest. Partial decoupling feedback, on the other hand, involves dividing the degrees of freedom of the system into two sets of inputs – one set is associated with the control of voluntary movements while the other set is associated with keeping system constraints in place. Both full and partial decoupling schemes rely on a central feedback mechanism. This differs from intrinsic feedback where the central controller (e.g., in biological systems the CNS is the central controller) is assumed to play a supervisory role and local control centers are responsible for regulating local sensory inputs. The simulation results were then compared to the measured joint trajectories of human subjects who participated in voluntary sway movements in the frontal plane. The results of their four-link biped model – under each controlling mechanism - were found to be quite analogous to the experimental results. Iqbal et al. concluded various points from these results: no matter which controlling mechanism was studied each consisted of a feedback and feedforward part; either an active or a supervisory role of the CNS is feasible to maintain the system; and with intrinsic feedback requiring the least amount of total system involvement while still maintaining a stable system even compared to more complex schemes, it may be the most plausible control mechanism for quiet standing. Their results also suggest that the four-link biped model that was used is a viable imitation of upright human posture.

Peterka (2000) presented another interesting way to model the dynamics of quiet stance. Using an inverted pendulum model and a PID control scheme for quiet stance, Peterka computer-simulated inverted pendulum motion that produced stabilogram diffusion functions (SDFs)—which are functions that describe the displacement of the mean COP and are dependent on the time interval between COP comparisons—closely resembled natural, physiological SDFs. This

was accomplished by setting the values of the proportional constants and a time delay term used in the PID scheme to critical values. Varying the proportional, integral, and derivative constants by 10% above and below critical values and the time delay term by 20% above and below its critical value still produced simulated-SDFs that resembled physiological SDFs.

2.2.4 Inverted Pendulum Model and The Ankle Strategy

Perhaps the widest used mechanical model of the human body in quiet standing is an inverted pendulum (IP). In its most common form, the IP model assumes the body is rigid and pivots about the ankle joint. Figure 2-3 depicts this classical form, which is shown for five instances in time.

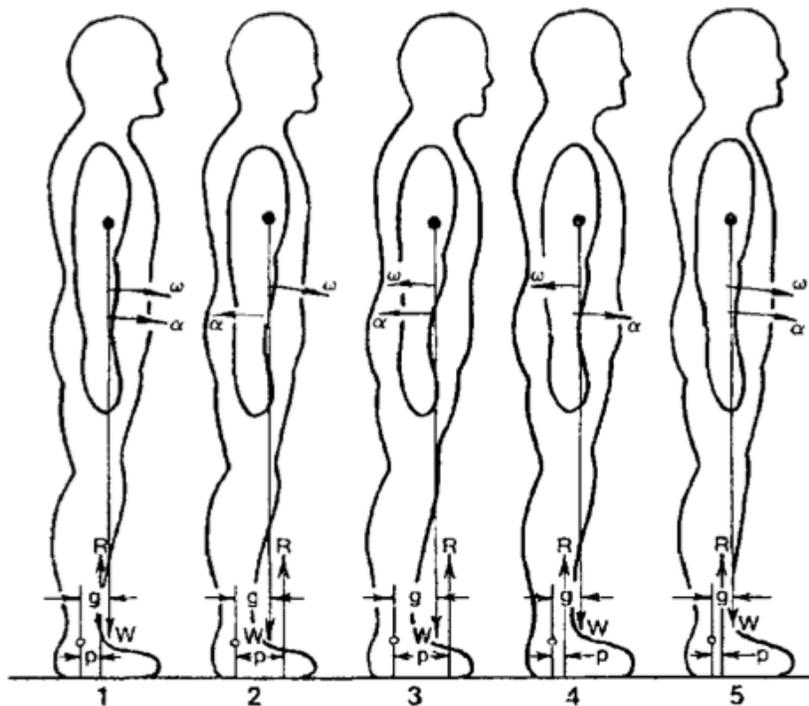


Figure 2-3. An Inverted Pendulum Model of Quiet Standing. Shown for five instances in time (reproduced from Winter et al., 1995).

The parameters in Figure 2-3 are as follows: vertical reaction force, R ; angular velocity, ω ; angular acceleration, α ; body weight, W ; COP (i.e., where R acts), p ; and COM (i.e., where W acts), g . This model is two-dimensional and is mainly concerned with the A/P displacement of the body. With W and R being equal and opposite during standing, Winter et al. (1995) presented the following basic equation for the IP:

$$Rp - Wg = I\alpha \quad [2.1]$$

where: I is the moment of inertia of the total body about the ankle joint ($\text{kg}\cdot\text{m}^2$), α has units of ($\text{rad}\cdot\text{s}^{-2}$), g and p have units of (m), and R and W have units of (N).

Assuming this IP model to be correct, it is easy to see from Equation 2.1 that when $Wg > Rp$ and if ω is initially clockwise, the body will begin to sway forward (Time 1). Automatically, the body will look to compensate for this movement by using the plantar flexors to move the COP anteriorly until it is ahead of the COM (Time 2). This will cause $Rp > Wg$ which will, in turn, lead to α and ω changing to a counterclockwise direction and backward sway for the body (Time 3). Thus, in another compensatory action, the body will decrease the action of the plantar flexors in order to move the COP posteriorly until it is more posterior than the COM. At this point, the body will desire to counteract this backward sway. Wg will eventually be greater than Rp , which means that α will now act clockwise again (Time 4). ω will follow suit and reverse its direction to a clockwise one, and the body will have returned to its original position (Time 5).

Further examining this IP model, Winter et al. (1995) approximated the horizontal linear acceleration of the COM (\ddot{x}) as:

$$\ddot{x} = \alpha d \quad [2.2]$$

where: d is the distance from the ankle joint to the total body COM.

Rearranging Equation 2.1 using information from Equation 2.2, the following is obtained:

$$p - g = \frac{I\ddot{x}}{Wd} = K\ddot{x} \quad [2.3]$$

Equation 2.3 clearly shows that the difference between the COP and COP positions is directly proportional to the horizontal linear acceleration of the COM. Winter et al. describe this difference as an “error” signal in the postural control system and deem that it is responsible for causing the COM’s horizontal acceleration.

It is apparent that in the sagittal plane this IP model is assumed to be under the control of the plantar flexors. In other words, an ankle strategy is used as the predominant control mechanism for this model. Indeed, Winter et al. (1995) suggest that for small perturbations the ankle strategy alone is responsible for stabilizing the IP in the A/P direction. However, while many researchers use the form of the IP model where the ankle strategy is dominant in the sagittal plane, it is important to note that other forms of the IP do exist. Figure 2-4 presents an additional IP model that describes movement in the frontal plane, which according to Winter et al. is often stabilized by the combination of an ankle and hip strategy. Here the body is assumed to pivot about both ankle joints and both hip joints.

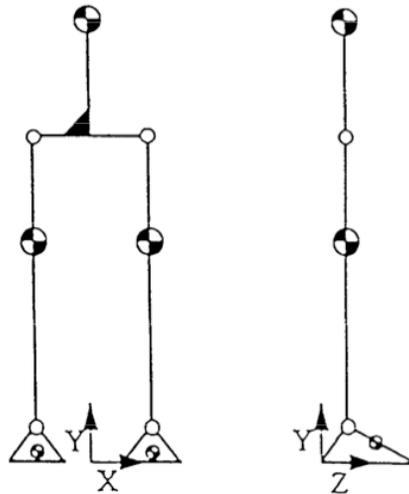


Figure 2-4. Two Biomechanical Models of the Inverted Pendulum. Left: IP model for the frontal plane where body pivots about both ankle joints and hip joints. Right: IP model for sagittal plane where body pivots about the ankle alone for small perturbations and about both the ankle and hip for large perturbations (reproduced from Winter et al., 1995).

In a study by Karlsson and Lanshammar (1997), the results of Winter et al. (1995) were further substantiated. Here, input data gathered from force plate recordings was inserted into an IP model where only movement in the sagittal plane was considered. The output of the model was the theoretical horizontal linear acceleration of the COM. Subjects of the study stood on a force plate where they were instructed to mimic the movement of an IP by keeping their body rigid and slowly swaying about their ankles in the sagittal plane. The true horizontal linear acceleration was discerned using data from the force plates and was referred to as reference data. The true and theoretical accelerations were compared. Results for when subjects used an ankle strategy may be seen in Figure2-5. The relative agreement of model data with reference data validates the use of an IP model where the body sways in the sagittal plane with the help of an ankle strategy.

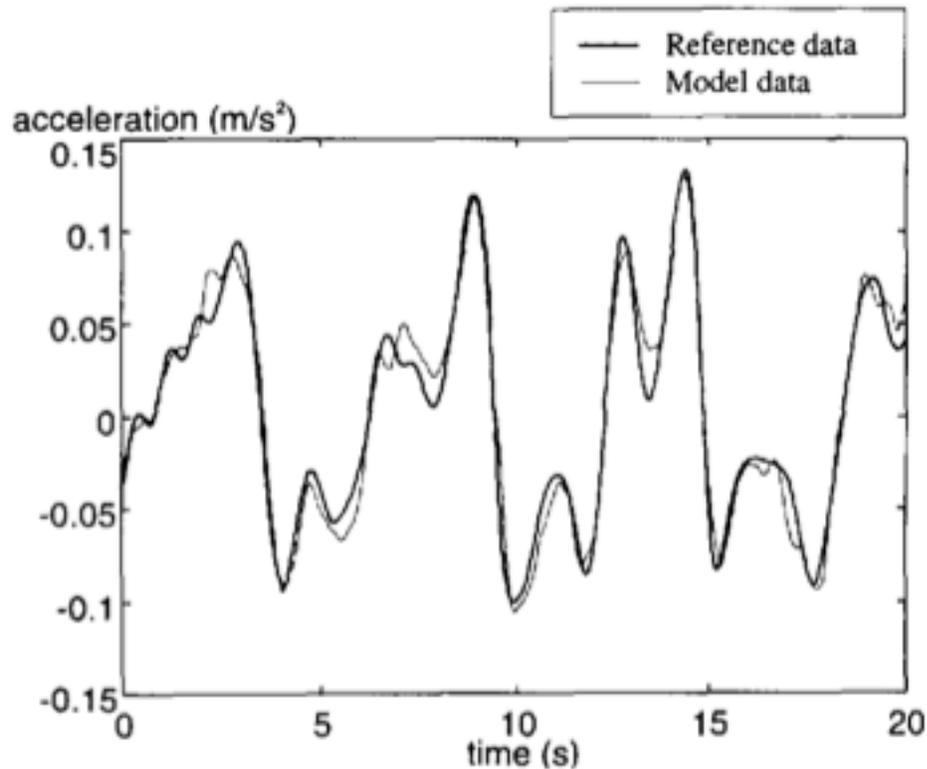


Figure 2-5. Quiet Standing Using an Ankle Strategy. The dark line represents reference data, and the lighter line represents the model data (reproduced from Karlsson and Lanshammer, 1997).

However, when subjects involved in the same study were instructed to simply stand as still as possible it was observed that the model and reference data did not coincide as well. It may perhaps be inferred then that an ankle strategy does not solely maintain quiet standing where movement is not limited to the sagittal plane and other muscles strategies (i.e., the hip strategy) may be active simultaneously. This inference is in agreement with the findings of Winter et al. (1995). Nonetheless, Karlsson and Lanshammer provided evidence supporting the validity of using an IP model to study upright posture.

The popular IP model, however, is not without its opponents. In reference to the findings of Winter et al. (1998), Morasso and Schieppati (1999) further examined the notion that muscle stiffness alone can stabilize upright posture in an IP model of quiet standing. Using an IP model

controlled by an open-loop system, they calculated that the theoretical minimum ankle stiffness necessary to maintain upright stance was approximately 1835 N-m/rad. Comparing this value to other literature values for the actual ankle stiffness displayed in the body a large discrepancy was seen. For example, Hof (1998) used a specialized ergometer to measure ankle to be within the range of 250-450 N-m/rad. Thus, the evidence presented by Morasso and Schieppatti does not support the idea that ankle stiffness alone is enough to stabilize upright stance in an IP model. However, the authors also noted that at the time methods to estimate ankle stiffness *during* the process of quiet standing were not available. Further research needs to be carried out so that the influence of ankle stiffness potential on upright stance may be addressed.

2.2.5 Hip Strategy

Aside from the ankle strategy, another postural control strategy that is commonly observed is that of the hip strategy. As its name implies, the hip strategy focuses on movement centered about the two hips; thus, it is employed with the use of the muscles that control the hip. During normal quiet stance of subjects standing with one foot on each of two separate force plates, Winter et al. (1995) found that both an ankle and hip strategy may be observed and that each strategy operates independently. An ankle strategy was found to dominate A/P movement while a hip strategy was the predominant mechanism used in M/L movement (See Figure 2-6).

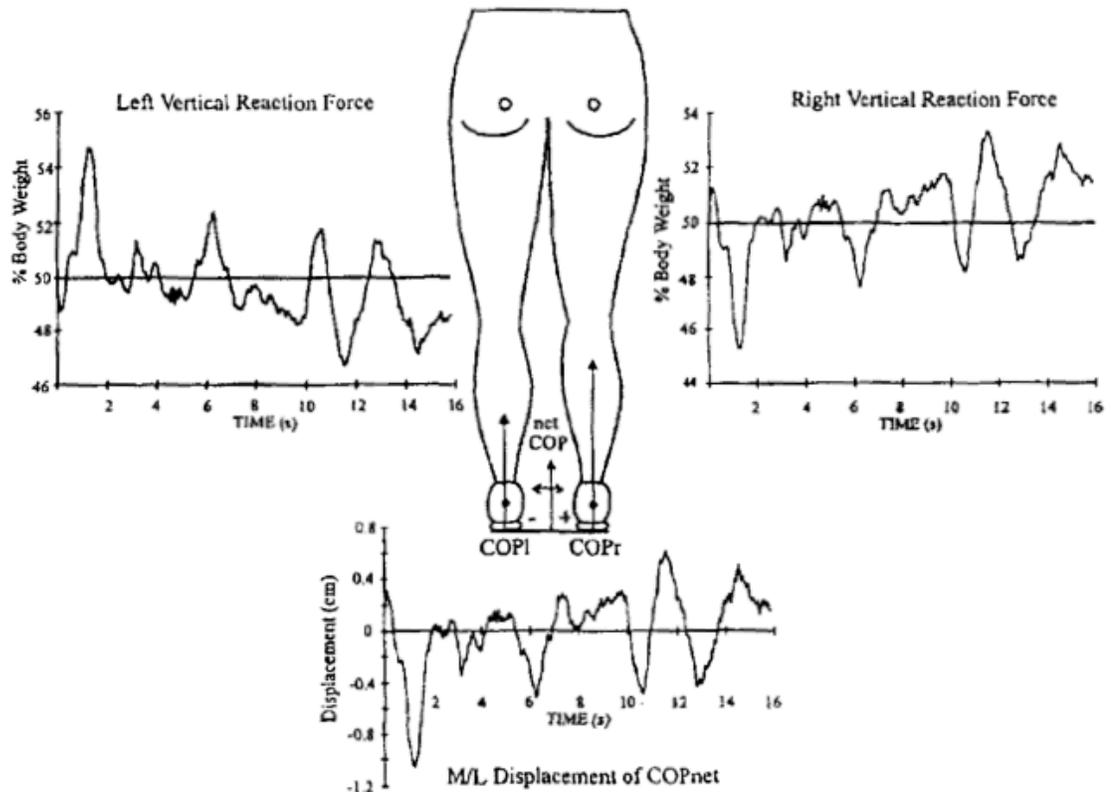


Figure 2-6. The Load/Unload mechanism. Shown are the results of quiet standing trials showing left/right vertical ground reaction forces and the M/L displacement of COP_{net} . It may be observed from these that the “load/unload” mechanism dominates control of the M/L COP_{net} (reproduced from Winter et al., 1995).

This hip strategy was referred to as a “load/unload” mechanism as it is associated with the loading of one limb and simultaneous unloading of the other. Under other types of stance (i.e., tandem stance, 45° stance), however, both mechanisms may still be observed but are no longer independent (Winter et al., 2003). Indeed, a combined ankle-hip strategy or sole hip strategy seems plausible especially to compensate for large postural perturbations. The ankle invertors/evertors may only generate a maximum moment of approximately 10 N-m while the hip adductors/abductors may create a maximum moment in excess of 100 N-m (Winter et al., 1995). Thus, for large perturbations that the ankles cannot handle, the hips would flex (e.g., move the

COM anteriorly) or extend (e.g., move the COM posteriorly) accordingly in order to maintain balance.

2.3 Investigative Measures of Balance

In a laboratory setting, the vast majority of efforts to quantify postural stability are based on COP measurements. As an aside, be sure to keep in mind the difference between the COP, COM, and COG as sometimes these terms are used interchangeably when, in fact, it is incorrect to do so (See *Section 2.1.1*). In order to gather information regarding the COP, it is most common for researchers to use force plate(s). Traditional studies have gathered COP_{net} (e.g., a combination of COP measures from both feet) information from subjects standing on a single force plate; however, two force plates must be used to study the individual COPs under each foot (Winter et al., 1995). The following sections outlines some of the time/frequency domain based measures that are commonly used to quantify data from quiet standing experiments.

2.3.1 Time Domain Based

Force plate technology typically enables researchers to observe the vertical ground reaction forces under each foot, determine the COP(s) from these forces, and track time intervals so that instantaneous points may be compared. Various time-domain based measures arise from these capabilities. Perhaps the most commonly used is that of the postural stabilogram, which illustrates the planar trajectory of the COP over the examined time interval. It may be seen as a summation of A/P and M/L COP displacement, and may be shown for individual feet (i.e., COP_{Left}, COP_{Right}) or both feet (e.g., COP_{net}). Figure 2-7 shows a typical stabilogram.

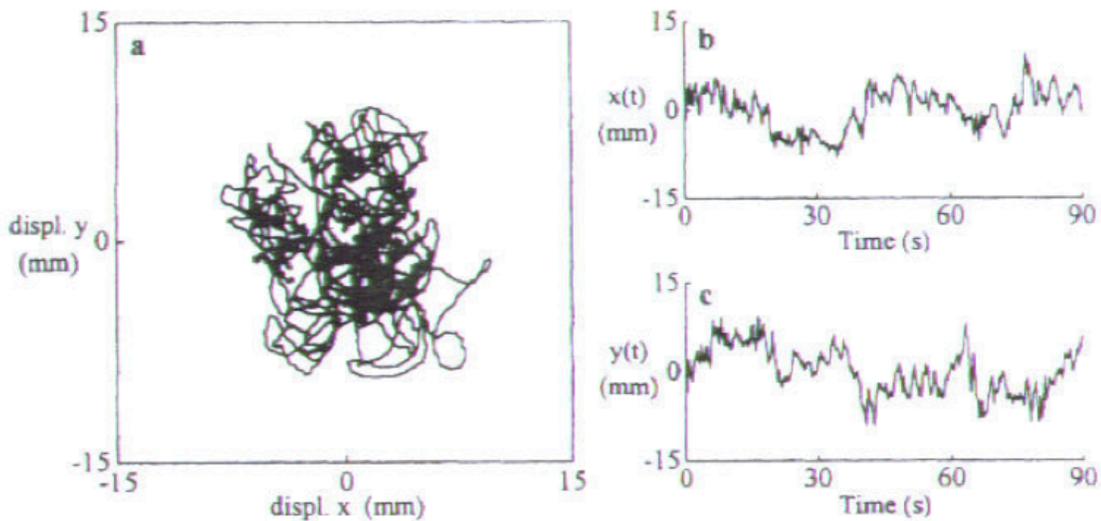


Figure 2-7. A Typical Postural Stabilogram. Left: Net COP displacement in the A/P and M/L directions. Right: The A/P and M/L time series used to create stabilogram (reproduced from Collins and De Luca, 1994).

Although the characteristic “spaghetti” pattern of stabilograms appears quite chaotic, Collins and De Luca (1994) have suggested that the COP displacement process should *not* be modeled as a chaotic process and is better represented as a stochastic process. The authors found that stabilograms may indeed be modeled as a system of bounded, correlated random walks. On a similar note, the SDFs discussed previously (See *Section 2.2.3*) attempt to provide a quantitative statistical measure of the apparent random variations of COP trajectories (Peterka 2000).

Other common time domain based measures used to characterize the stabilogram include: the mean COP position, the mean distance from the COP, the total length of the COP path, the mean velocity of the COP, sway area, and 95% confidence circles or ellipses (Prieto et al., 1996). From these commonly used measurements it may be noticed that a portion of them quantify the distance of the COP from a reference point and others (e.g., sway area and the 95% confidence circles/ellipses) quantify the area enclosed by the stabilogram. With special note to the 95%

confidence shapes, the ellipse can be computed using principal component analysis (Oliveira et al., 1996).

The vertical ground reaction forces (VGRFs) under each foot, which together add up to the weight of the subject, vary over time as weight is shifted between feet and are often of interest to researchers. For instance, Gutnik et al. (2008) studied quietly standing subjects and used VGRF data to calculate the angle of inclination from a vertical centroidal line exhibited by each of them. It is interesting to note that the results of this study found that all subjects, who were all right-handed males, showed a right-sided bias of the VGRFs. This supports the notion that asymmetry is a normal part of human quiet stance.

2.3.2 Frequency Domain Based

COP-based measurements are sometimes analyzed in the frequency domain. For instance, some studies (Prieto et al., 1996; Loughlin and Redfern, 2001) have computed the power spectral density of the COP signal. In essence, the power spectral density function quantifies the frequency content of a stochastic process and allows periodicities, if any, to be observed. Thus, assuming quiet stance to be a stochastic process as previously suggested, this parameter is well suited to describe it. Spectral moments, which essentially describe the overall shape of a group of points, are usually calculated first. The total power of the signal may then be calculated by integrating the area of the entire power spectrum (Prieto et al., 1996). The centroidal frequency, which is the mean rate of zero crossings, and the frequency dispersion, which is a measure of the variability in the frequency content of the power spectral density, are also sometimes computed (Prieto et al., 1996). However, since quiet stance may be described as a temporally dynamic process, its spectral characteristics may vary with time and time-varying spectral densities may be needed (Loughlin and Redfern, 2001).

Other rudimentary frequency based measures (i.e., mean frequency, RMS frequency, rotational frequency of the COP) are also commonly used to characterize quiet standing data.

2.4 Balance in Select Clinical Groups

Balance deficits are commonly observed in both older individuals and those afflicted with certain pathologies. With this being the case, the study and rehabilitation of balance in these clinical groups are of interest to researchers and therapists alike. Balance in the elderly, those with idiopathic scoliosis, and those with cerebral palsy are examined here.

2.4.1 Elderly

For older individuals, the risk of a fall is an important concern. According to recent statistics from the Center for Disease Control and Prevention (CDC), 1 in 3 people 65 and older experiences a fall each year. In this same age group, falling is the leading cause of injurious death. Injuries incurred by the elderly from falling can lead to a decrease in their quality of life and an increase in the chance for early death. Data also supports the notion that the risk of suffering serious injury from a fall increases with age (CDC, 2011). The consequences of falling are not only a significant problem for this group of individuals but for the healthcare community, who experiences a tremendous financial burden caring for fall victims (Alexander et al., 1992). In 2000, the care of fatal fall victims cost in excess of \$19 billion, which adjusted to 2010 dollars, equates to \$28.2 billion (CDC, 2011).

The declining mobility and overall stability of older age groups is a topic of popular study. It is suggested this decline may have a dual nature in the sense that some of the dysfunction may be due to age-related degeneration of the postural control system (i.e., visual,

vestibular, somatosensory systems) and some may – or may not – be due to specific pathologies unique to each individual (Prieto et al., 1996). The etiology of age-related deficits in postural control has not yet been determined, although certain postural parameters and their correlation with age have been investigated. In a study by King et al. (1994), the functional base of support (FBOS) was compared between 113 subjects with an age range of 20 to 91. The FBOS was defined as the difference between the mean COP positions during voluntary, maximum forward/backward leaning divided by foot length. In general, the FBOS was found to decrease with age while no correlation was found regarding postural sway. Prieto et al. (1996) reported that the mean velocity of the COP increased with age and others parameters like the mean distance traveled by the COP and area measures of stabilograms held no relation with age. In a more recent work, Tucker et al. (2010) compared the voluntary sway of elderly subjects who were divided into low- and high-risk falling groups. Results showed that reaction times to initiate/terminate sway were higher and COP-COM separation during M/L sway was smaller for the high-risk fall group, who also exhibited a higher median subject age, compared to the low-risk fall group.

It is apparent that falling is a serious issue for elderly people. The ability to predict falls and monitor signs of decline in the postural control system in a clinical setting would be most advantageous, but the methods currently available need improving (See *Section 2.1.4*).

2.4.2 Idiopathic Scoliosis

Idiopathic scoliosis (IS) – as its name implies – is a disease of unknown origin that presents as a three-dimensional deformation – especially in the lateral directions – of the spine and rib cage. It has been documented that the misalignment caused by IS may contribute to the decline of postural stability (Chen et al., 1998; Gruber et al., 2011). In particular, Chen et al.

(1998) observed an increase in the M/L sway of adolescent IS subjects versus normal subjects as they attempted to maintain various postures during quiet standing. The authors suggest that standing with the hips in full flexion is the best means of delineating differences between IS and normal subjects. On a similar note, Gruber et al. (2011) found that M/L COP range and sway area were higher in adolescent IS subjects compared with the control group. Interestingly enough, the authors also suggested that their results may support the idea that IS patients with more severe curvatures exhibit greater postural stability than those with less severe curvatures. In other words, an adapted control strategy may be available based on the severity of the disease.

Using postural stability tests to ascertain the severity of IS in patients would be a most useful and non-invasive method (Zabjek et al., 2005). Efforts to develop such tools are certainly warranted.

2.4.3 Cerebral Palsy

Those afflicted with a form of the neurological disease Cerebral Palsy (CP) may experience cognitive, visual, and/or motor function deficits. The symptoms of CP depend on the form with which one is diagnosed. For instance, motor skills on one side of the body may be affected (spastic hemiplegia) or both sides (spastic diplegia). Regardless, of the specific form of CP much attention is paid to the postural control of children with CP. These children often suffer from poor postural muscle modulation patterns following perturbations, such as improper activation of antagonists before agonists and delayed muscle activation times (Woollacott et al., 2005). It has been documented that common postural sway parameters may be used to discriminate between normal and CP subjects (Harris et al., 1993) and that rehabilitative efforts to improve quiet stance have had some success (Woollacott et al., 2005). In regard to the study done by Woollacott et al. (2005), it was found that repetitive balance training exercises in both spastic

hemi-/diplegia subjects might, in fact, facilitate neurological control patterns that result in improved postural stability – even over long periods of time. Such results should provide an optimistic feeling to the therapist wishing to improve the postural control of CP patients. Hopefully, future studies will provide similar results.

2.5 Summary

The study of quiet standing is no stranger to ambiguity. Although the capability of various models and quantifying measures of human stance to produce valid results has been seen, little explicit certainties have been shared between studies. The goal of understanding postural stability is driven in large part by the desire to create better clinical assessment and rehabilitative methods for those who suffer from balance deficits. Thus, the motivation of this work is to add novel information to the field in hopes of advancing towards the aforementioned goal.

Chapter 3

Methods

3.1 Overview

The content of this chapter describes the subjects who participated in this study, the equipment and software used, the quiet standing trials performed, and the methods of data analysis. Table 3.1 offers a brief outline of the protocol that was followed.

Table 3.1. A Summary of the Experimental Procedure.

Stage	Explanation
Participant Consent	Experiment was explained and written consent obtained from participants.
Age and Mass Information	Age and mass of subjects measured.
Quiet Standing Trials	Each subject performed a total of 6 quiet standing trials – 3 eyes open and 3 eyes closed trials – with one foot on each of two separate force plates. Each trial lasted 40 seconds.
Traditional COP measurements	Numerous parameters describing the motion of the COP computed.
Entropy measurements	Approximate entropy measures calculated to quantify the orderliness, or lack thereof, of vertical ground reaction forces. Done for each foot separately and then a cross-entropy measure determined for both signals.
Statistical Analysis	Statistical analysis conducted in order to determine the variance of results and if results were indeed significant.

3.2 Participants

Nine subjects (5 females, 4 males) were recruited for this study. They had a mean age of 22 ± 1 year, a mean height of 1.69 ± 0.08 m, and a mean mass of 73 ± 13 kg. The Institutional

Review Board approved all procedures, and all subjects provided written informed consent for participating in this study.

3.3 Force Plate Equipment

The main data collection hardware used in this study were two Bertec force plates (#N50201 Type 4060S). One (force plate A) had dimensions of approximately 80 cm by 40 cm while the other (force plate B) had dimensions of approximately 60 cm by 40 cm. The plates were separated by 1.27 cm. Figure 3-1 illustrates the orientation of force plates A and B with respect to one another and the coordinate system used. The two force plates were aligned such that the x-axes of their respective coordinate systems overlapped. Both plates had the same plate thickness and topmost surfaces that were aligned in height.

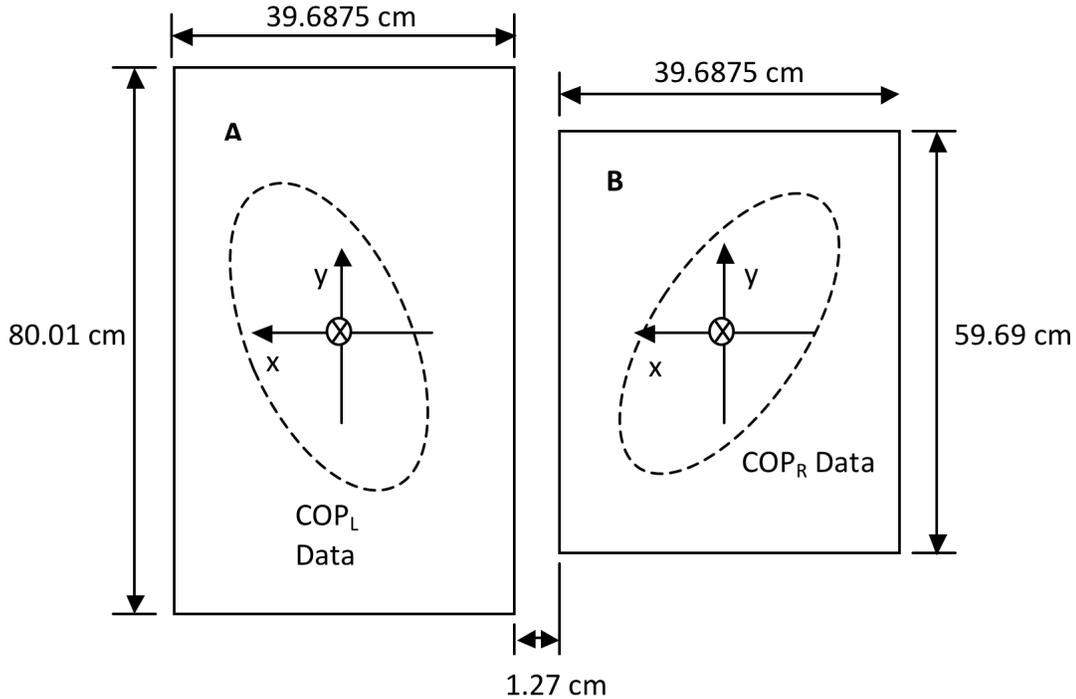


Figure 3-1. A Schematic of the Force Plates Used. Force plate A was used to collect COP data associated with the left foot (COP_L), and force plate B was used to collect COP data associated with the right foot (COP_R). The two dashed ellipses show the approximate orientation of the subject's feet on each platform. The positive directions of the Cartesian coordinate system used is also shown on each plate.

The primary functional components of the force plate are the load cells, which are located beneath the surface of the platform. These load cells contain strain gauges that respond to the loads applied to the force plate surface. The signal generated by the strain gauges is proportional to the load on the force plate (i.e., the greater the load the higher signal that is generated). These signals are sent to an amplifier and a calibration matrix, which is unique to each force plate, is used which results in the computation of three ground reaction forces (i.e., F_x , F_y , and F_z in units of N) and three ground reaction moments (i.e., M_x , M_y , and M_z in units of N-m). Equations 3.1 and 3.2 may then be used along with the force and moment data to ascertain the point of application of the force on the plate (See Figure 3-2), that is the center of pressure (COP), in the two horizontal directions (x and y).

$$x = \frac{-h \cdot F_x - M_y}{F_z} \quad [3.1]$$

$$y = \frac{-h \cdot F_y + M_x}{F_z} \quad [3.2]$$

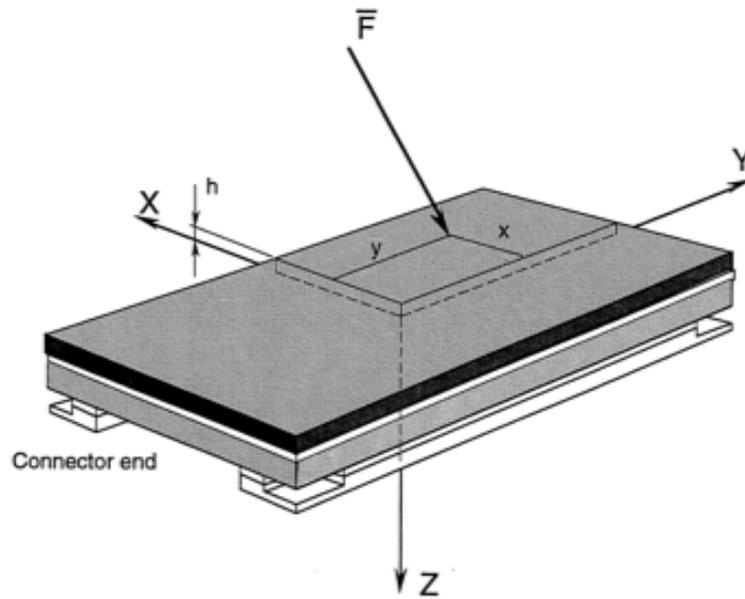


Figure 3-2. An Illustration of a Force Plate. The positive directions of the Cartesian coordinate system utilized are shown. The origin of the coordinate system is located at the center of the topmost surface of the plate. The thickness of the plate is represented by h . A force vector, \bar{F} , is and the coordinates, x and y at which it acts (i.e., the COP) are also shown (reproduced from Bertec Force Plate User's Manual).

3.4 Data Collection Software

National Instruments LabVIEW 6.1 software was used in conjunction with the force plates. This software acquired data at a rate of 1000 Hz and allowed for the calibration matrix discussed previously to be applied to the incoming signals. The force and moment data it collected was compiled into a text file and were later transferred to an Excel file.

3.5 Quiet Standing Trials

Before subjects were asked to stand on the force plates, he/she was asked to remove their shoes and then their height and mass were measured and recorded. Next he/she was asked to place each of his/her feet on a separate force plate. In order to standardize foot position between separate trials, tape was used to demarcate the location of the big toe and the heel of each of the subject's feet. Thus, for each separate trial the subject was required to step onto the force platforms so that his/her big toe and heel of each foot was in contact with the associated tape strips. After establishing the preferred foot position, the subject was given a one-minute break.

Two sets of trials were then conducted: eyes-open (EO) and eyes-closed (EC). For the EO trials, the participants were asked to step onto the force plates, stand in a comfortable posture with their arms at their sides, and look forward at a designated target on the wall. Next data collection was commenced using LabVIEW software for a time interval of forty seconds. After this time had elapsed, subjects were allowed to sit down and rest for one-minute. This same procedure was repeated for an additional two trials. For the EC trials, a similar protocol was followed with the exception that once the subjects were in position, he/she was instructed to close their eyes for the duration of the trial. Each subject performed three EC trials. In summary, each subject completed six trials – 3 EO and 3 EC.

3.6 Data Analysis

All of the collected data sets were then analyzed using custom written MATLAB software. A detailed description of the actual MATLAB code used may be found in the Appendix. The subsections below describe the two main categories of metrics employed to

quantify the test results. These measurements were compared on both an intra-subject and inter-subject basis (See *Section 3.3.5*).

3.6.1 COP Measurements

The data associated with the left foot and right foot of the subjects was first evaluated separately, that is, Equations 3.1 and 3.2 were used to compute the COP under each foot.

Stabilograms were then generated for each foot. Next the data from each force plate was shifted into a global reference frame (See Figure 3-3) so that COP_{NET} could be computed using Equation 3.3.

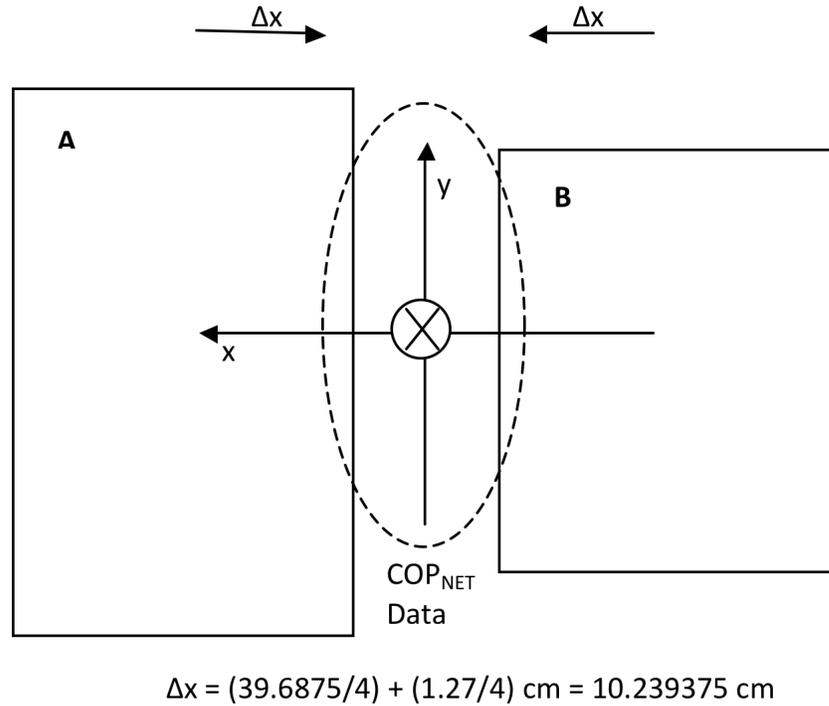


Figure 3-3. A Diagram of the Global Reference Frame Used. All the x-coordinates of the COP data from force plate A and force plate B was shifted horizontally by a value (Δx), which was calculated from the dimensions of the force plates (See Figure 3-1), into a global reference frame such that this frames origin resides in between both plates. The direction of this shift was negative for force plate A data and positive for force plate B data. The y- and z-coordinates of the COP data did not need to be shifted as these axes of both plates' individual reference frames were already aligned. The dashed ellipse shows the approximate location of the system's overall COP, or COP_{NET} .

$$COP_{NET} = \frac{COP_L \cdot F_{zL}}{F_{zL} + F_{zR}} + \frac{COP_R \cdot F_{zR}}{F_{zL} + F_{zR}} \quad [3.3]$$

where COP_L and COP_R are the center of pressure under the left and right foot, respectively; and

F_{zL} and F_{zR} are the vertical ground reaction forces under the left and right foot, respectively.

COPnet was then used to create a postural stabilogram and various other traditional COP parameters were determined for the global system. All COP parameters were based on the work

done by Prieto et al. (1996) except the confidence ellipse area, which is based on the work of Oliveira et al. (1996). Some of these COP parameters were applied separately to the COP motion in the anterior-posterior (AP) direction and motion in the medial-lateral (ML) direction. A summary and brief explanation of them may be found in Table 3.3.

Table 3.2. Parameters Used to Quantify Motion of COP_{NET}.

Parameter	Abbreviation	Explanation	Units
Resultant Distance	RD	Vector distance from the mean COP to each pair of points in AP- and ML-time series	mm
Mean Distance	MDIST	Average distance from the mean COP	mm
Mean Distance-AP*	MDIST _{AP}	Average AP distance from the mean COP	mm
RMS Distance	RDIST	Root Mean Square distance from the mean COP	mm
RMS Distance-AP	RDIST _{AP}	Root Mean Square AP distance from the mean COP	mm
Total Excursions	TOTEX	Total length traveled by the COP	mm
Total Excursions-AP	TOTEX _{AP}	Total length traveled by the COP in the AP direction	mm
Mean Velocity	MVEL	Mean velocity of the COP	mm/s
Mean Velocity-AP	MVEL _{AP}	Mean velocity of the COP in the AP direction	mm/s
Confidence Ellipse Area	Area _{net}	Area encompassed by confidence ellipse fit to COP data	mm ²

*Note: For each anterior-posterior (AP) parameter a medial-lateral (ML) one also exists and is simply not listed here.

3.6.2 Entropy Measurements

Four sets of entropy computations were made for each trial: approximate entropy (ApEn) of the left foot vertical ground reaction force (VGRF), ApEn of the right foot VGRF, ApEn of the total VGRF (i.e., the sum of the VGRF from each foot), and the approximate cross-entropy between the VGRF under each foot. For the force plate system used in this study, the VGRF corresponded to the force in the z-direction (i.e., F_z). The ApEn of the VGRFs was taken instead of the COP because these forces are not coupled unlike the COP, which depends upon other forces and moments. The entropy measures employed were based on the work presented by Pincus and Singer (1996). The goal of these measures was to reveal the orderliness of the signals and thereby help quantify the degree of lower limb independence and relatedness during quiet standing – the very purpose of this work.

The basis behind the ApEn measurements is such: The logarithmic frequency of a data block - of a specified length - is computed and compared to the logarithmic frequency of the data block that comes next in the series. Thus, low values of ApEn correlate with regularity while high values correlate with irregularity. In the case of the approximate cross-entropy, the same notion holds true. It should be kept in mind, however, that the approximate cross-entropy value should be compared with respect to the values of the individual ApEns of the signals being compared. In other words, one may only deem the approximate cross-entropy to be high (or low) only if the individual approximate entropy of the signals being compared is much lower (or higher).

To compute the ApEn on a signal two parameters must be specified: m and r . The ApEn algorithm takes a sequence of m data points and determines the logarithmic likelihood that this sequence is similar to other sequences of data points in the data set. For the comparisons made in this study m was set to 2 (Pincus, 1991). When computing ApEn a parameter r is used to determine the closeness of data sequences, which effectively filters out sequences that are not

close. For the comparisons made in this study r was set at 0.0717. This particular value of r was computed in an additional set of experiments where a known weight was placed on a force plate and the standard deviation of the signal created was determined. This procedure was carried out for both of the force plates used in this study. The highest standard deviation value that was computed was used as r , which effectively is the noise threshold – changes less than this threshold were ignored in the computation of ApEn.

If the analyzed data set is regular, the value of ApEn is small, while with increasing irregularity of the data set the value of ApEn will be greater. For example, a sine wave is regular and has an ApEn close to zero. In contrast, white noise consists of random independent samples and has an ApEn value close to 2.

3.7 Statistical Analysis

The statistical method applied to the COP metrics and the approximate entropy measurements was a two-factor (i.e., eyes-open vs. eyes-closed) repeated-measures analysis of variance. A significance level of $p < 0.05$ was used in the analysis. Minitab 16.1.0 software was utilized to carry out the statistical analysis of the results.

Chapter 4

Results

4.1 COP Metrics Results

COP data were collected independently from the right foot and left foot using two separate force plates. Figure 4-1 shows the vertical ground reaction forces (VGRFs) created under each foot during a quiet standing trial. The sum of the VGRFs under each foot directly reflects the weight of the subject plus the product of their mass and any acceleration of their center of mass. Observing Figure 4-1 reveals that maximums in the left foot data approximately correspond to minimums in the right foot data, and that the subject's weight was distributed more on one foot as opposed to the other (e.g., in Figure 4-1 this corresponds to more weight being distributed over the left foot than over the right foot). The latter observation, however, does not always hold true for every trial (or every subject), as the distribution of weight between the feet appears to fluctuate more evenly in some trials compared to others (See Figure 4-2). The two figures below represents two quiet standing trials performed by the same subject.

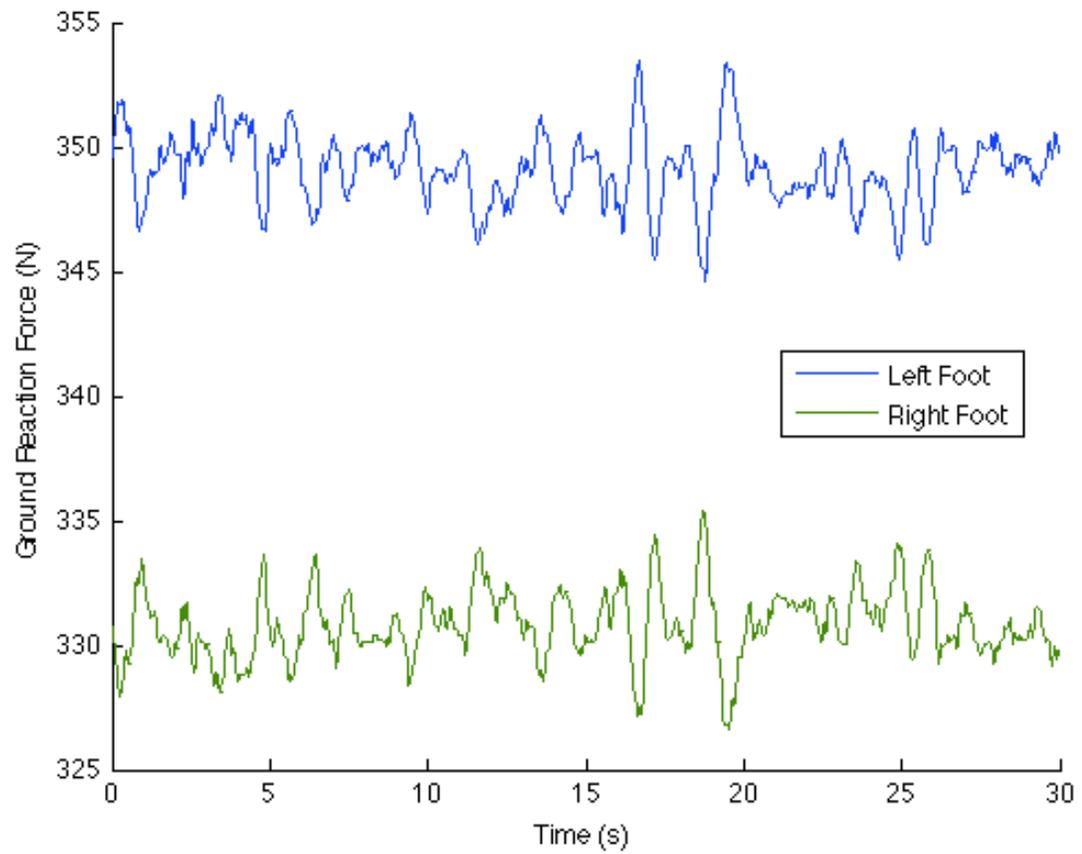


Figure 4-1. Temporal Plot of a Subject's Vertical Ground Reaction Force Under Each Foot.

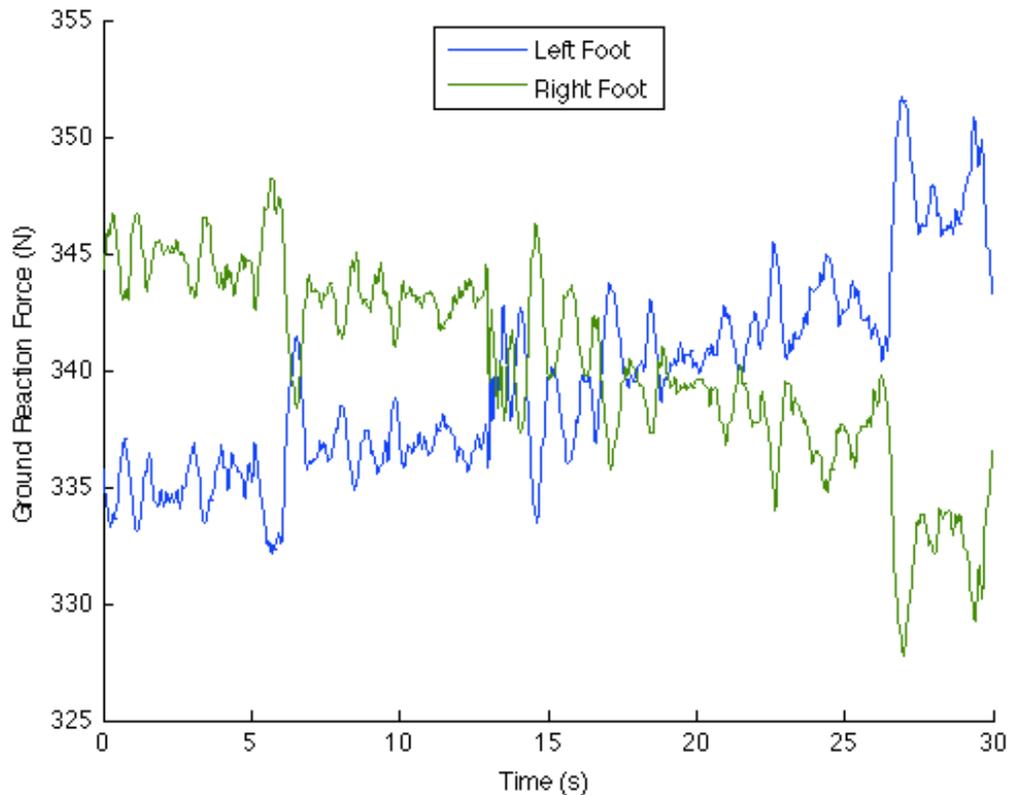


Figure 4-2. The Vertical Ground Reaction Forces Created During a Second Quiet Standing Trial. The data shown here is from the second eyes-open trial of the same subject whose data from the first eyes-open trial is shown in Figure 4-1. The subject was given a one-minute break between trials.

The COP data from the left and right foot was used to create two individual stabilograms, which is shown in Figure 4-3. COP_{NET} data was used to create its own separate stabilogram, which is shown in Figure 4-4. Confidence ellipses are drawn around the data in each of these figures, and the data shown by them is from one subject. The stabilograms created from the data of other subjects appear quite similar to the one's shown. For instance, the stabilograms of each individual foot lie along a titled axis that reflects the orientation of each foot, and the stabilograms of COP_{NET} clearly show more anterior-posterior displacement than medial-lateral displacement. Both observations reflect characteristics that are typical of stabilograms (e.g., Winter, 1995).

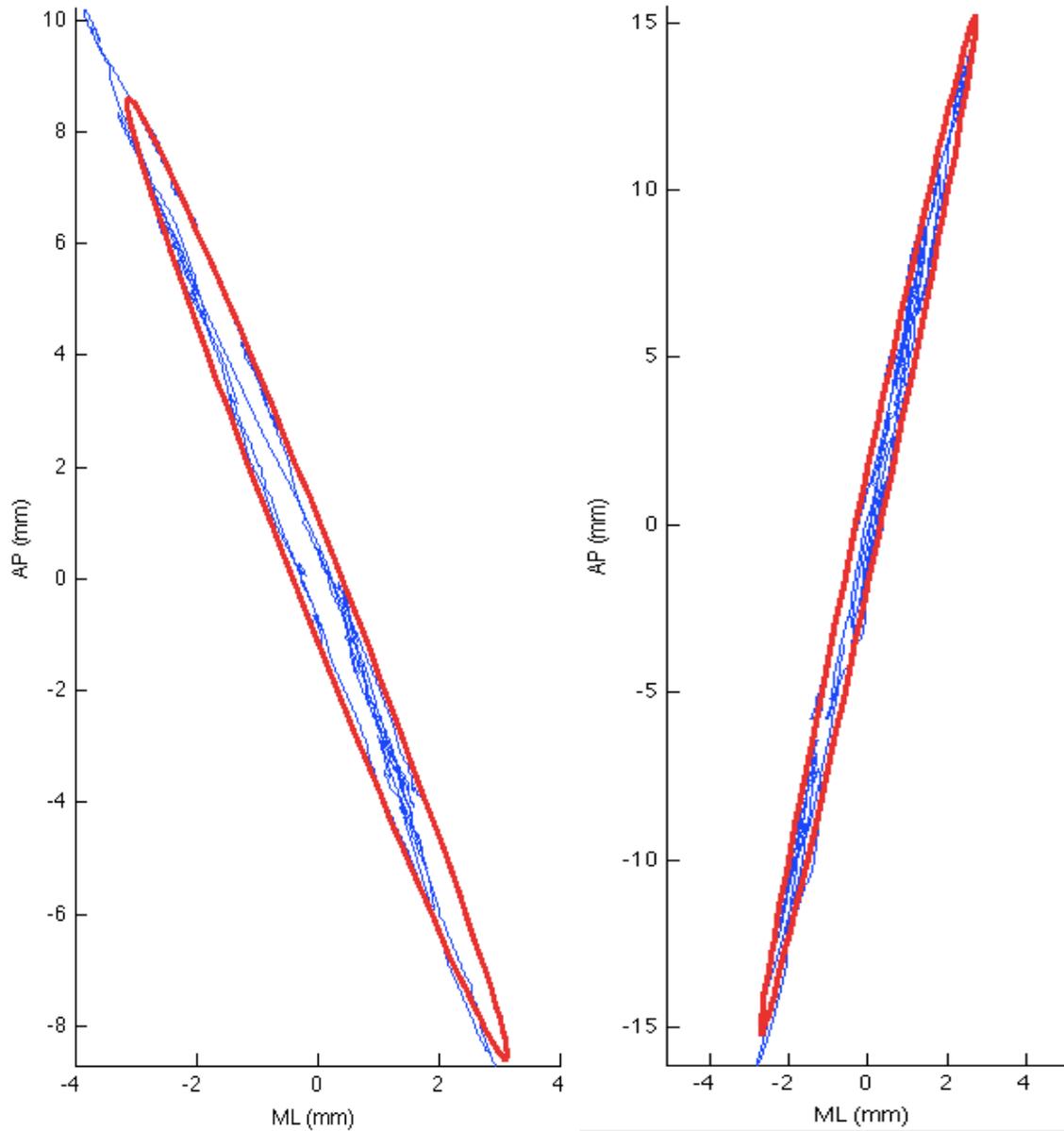


Figure 4-3. Stabilograms depicting the COP Trajectory Under the Left and Right Foot. The figure on the left corresponds to left foot COP data and vice versa for the figure on the right. A confidence ellipse is drawn around each data set. The data shown is from one eyes-open trial performed by one subject.

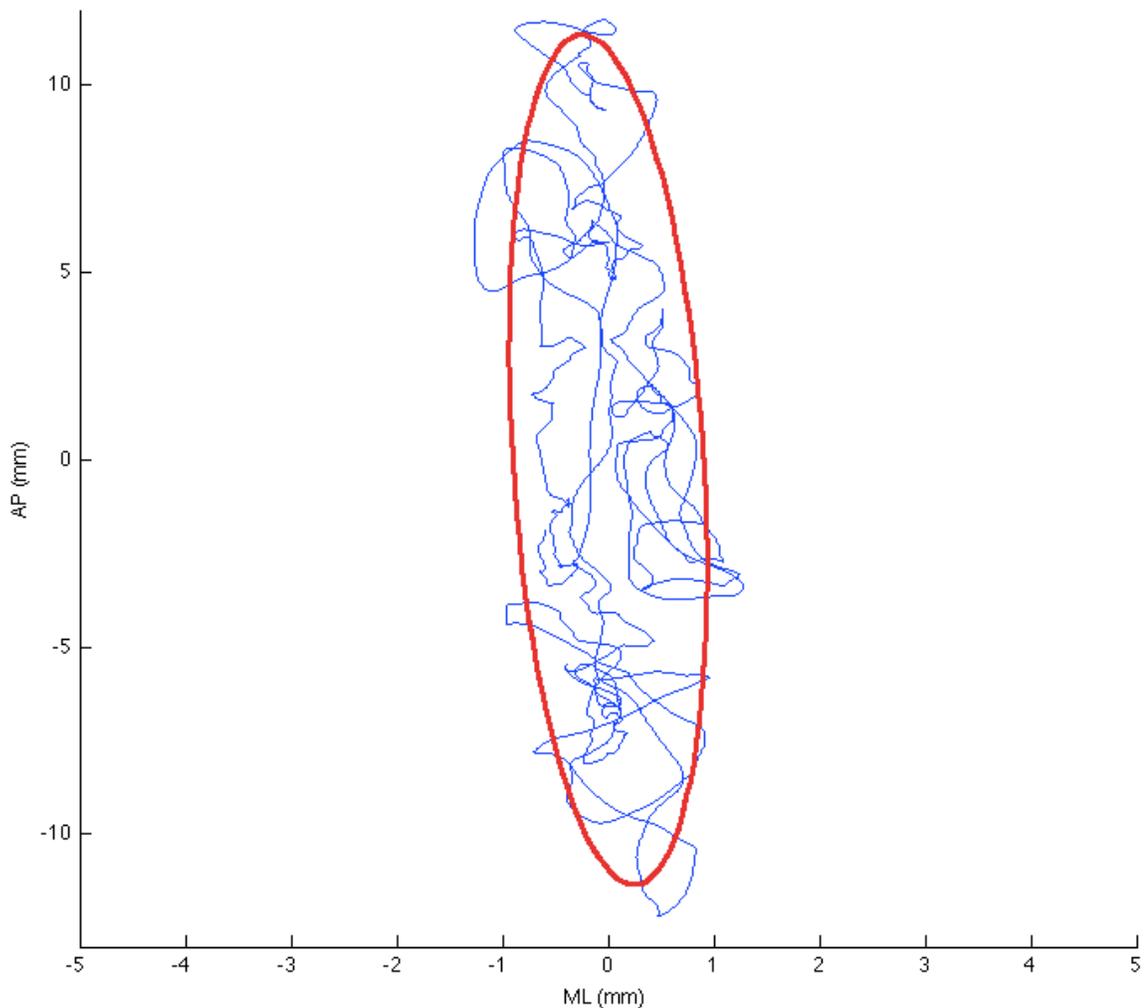


Figure 4-4. COP_{NET} Stabilogram. This plot is a summation of the left and right foot COP data. A confidence ellipse is drawn around the data set. The data shown is from one eyes-open trial performed by one subject.

The various metrics that were computed using COP_{NET} were averaged for all trials of each subject and separated according to eyes-open and eyes-closed conditions. The results are shown in Table 4.1. Explanations of each parameter listed in the table may be found in Table 3.3. Upon simple observation it may be seen that the mean of all parameters increased in the eyes-closed trials while the standard deviations show more of a mixture with some increasing and some decreasing. It may also be pertinent to note that all subjects voiced that he/she felt that they

experienced greater postural sway in their first eyes-closed trial compared to the previous three eyes-open trials, although there was no statistically significant difference ($p < 0.05$).

Table 4.1. A Summary of COP_{NET}-based Measures Computed During Quiet Standing. Results from both eyes-open and eyes-closed conditions are shown. The values shown in the Eyes-Open and Eyes-Closed columns are the mean \pm the standard deviations of all trials in each condition performed by all subjects. The last two columns show the change in the mean and the standard deviations between the two different visual conditions. In these columns a positive value indicates an increase in the eyes-closed condition while a negative value indicates a decrease in the eyes-closed condition.

Parameter	Eyes-Open	Eyes-Closed	Change in Mean	Change in Stand. Dev.
MDIST (mm)	3.38 \pm 1.85	3.53 \pm 1.31	0.15	-0.54
MDIST _{AP} (mm)	3.30 \pm 1.85	3.43 \pm 1.32	0.13	-0.53
MDIST _{ML} (mm)	0.41 \pm 0.21	0.47 \pm 0.20	0.06	-0.01
RDIST (mm)	4.04 \pm 2.21	4.27 \pm 1.53	0.23	-0.68
RDIST _{AP} (mm)	3.99 \pm 2.21	4.22 \pm 1.54	0.23	-0.67
RDIST _{ML} (mm)	0.52 \pm 0.27	0.60 \pm 0.26	0.08	-0.01
TOTEX (mm)	176 \pm 49.00	228 \pm 63.80	52	14.80
TOTEX _{AP} (mm)	156 \pm 43.10	207 \pm 55.20	51	12.10
TOTEX _{ML} (mm)	52.5 \pm 30.0	61.6 \pm 33.80	9.1	3.80
MVEL (mm/s)	5.87 \pm 1.63	7.60 \pm 2.13	1.73	0.50
MVEL _{AP} (mm/s)	5.20 \pm 1.44	6.89 \pm 1.84	1.69	0.40
MVEL _{ML} (mm/s)	1.75 \pm 1.00	2.05 \pm 1.13	0.3	0.13
Ellipse Area (mm ²)	25.0 \pm 20.20	27.5 \pm 14.90	2.5	-5.30

4.2 Approximate Entropy Results

The results of the approximate entropy (ApEn) measures were averaged amongst all subjects and separated between eyes-open and eyes-closed conditions. The results are shown below in Table 4.2. The ApEn of the individual VGRFs from the left and right foot are low in value, as they both tend toward zero therein reflecting a relatively ordered signal. When

compared to ApEn-Right and ApEn-Left, the approximate cross-entropy of the two VGRF signals is much higher. Switching between eyes-open and eyes-closed conditions does not seem to vary any of the ApEn values ($p < 0.05$).

Table 4.2. Summary of Approximate Entropy Measures Computed During Quiet Standing. Results from both eyes-open and eyes-closed conditions are shown. Approximate entropy (ApEn) of the left foot vertical ground reaction force (VGRF), the right foot VGRF, the combined VGRFs (VGRF_{NET}), and between the VGRF of each foot (Cross-ApEn) are shown.

Parameter	Eyes-Open	Eyes-Closed
ApEn_Right	0.137 ± 0.073	0.146 ± 0.071
ApEn_Left	0.142 ± 0.072	0.149 ± 0.070
ApEn_VGRF _{NET}	0.154 ± 0.062	0.159 ± 0.058
Cross_ApEn	2.45 ± 0.15	2.48 ± 0.13

4.3 Statistical Analysis Results Summary

A two-factor repeated-measures analysis of variance was applied to the results. According to the results of this analysis, no statistically significant results were observed for the relationships of interest: Changing between eyes-open and eyes-closed conditions did not significantly effect the COP measurements, nor did the order in which trials were performed by the subjects, nor did the relationship between the individual ApEn values of the left and right foot, nor did the ApEn VGRF_{NET} compared to individual ApEnS, and nor did Cross-ApEn in either visual condition. Statistical significance was determined using $p < 0.05$. A summary of these results may be found in Table 4.3.

Table 4.3. Results of Two-factor Repeated-measures Analysis of Variance. A significance level of $p < 0.05$ was used.

Relationship of Interest	Statistical Significance
COP Metrics in Eyes-Open Vs. Eyes-Closed	None
Test Order	None
ApEn _{RIGHT} Vs. ApEn _{LEFT}	None
ApEn VGRF _{NET} Vs. Individual ApEns	None
ApEn VGRF _{NET} between Eyes-Open and Eyes-Closed	None

4.4 Summary

The COP data gathered was used to create stabilograms of the left and right foot as well as stabilograms of the net center of pressure between both feet. The general shapes of these stabilograms reflected the expected orientations and deviations of the COP typically seen during quiet standing trials. The stabilograms were further characterized or “unfolded” by calculating several COP-based metrics. The COP-based metrics showed an increase going from eyes-open to eyes-closed conditions, even though no statistical significance was observed ($p < 0.05$).

Individual ApEn measurements of the left and right foot VGRFs were low in value indicating relatively ordered signals while Cross-ApEn measurements were high in value indicating that the left and right foot signals were relatively unrelated to each other. No statistical differences ($p < 0.05$) were observed for any of the ApEn results.

Chapter 5

Discussion

5.1 Overview

This chapter presents a discussion of the results of this study, limitations of the study, potential future studies, and conclusions. Various deductions are made despite the fact that no statistically significant results were present. Justifying this action is the following adage: a lack of statistical significance, which may be the consequence of an insufficient sample size, does not directly correspond to a lack of clinical importance. Thinking otherwise is a common misinterpretation made in clinical experiments (Altman and Bland, 1995). This study strived to avoid any statistical misinterpretations and gather any evidence that may indeed be clinically valuable with respect to a further understanding of postural control.

Another important concept to keep in mind in regard to analyzing statistical results is power, which is the probability that a null hypothesis is rejected when it is false. Experiments without an adequate degree of power may fail to find a significant effect even though one may be present. Power is proportional to the defined error rate, to the magnitude of the difference one wishes to detect, and to the number of observations being made. Power is inversely proportional to the amount of variability present in the sample population (Curran-Everett, 2010). Thus, these factors must be considered when assessing the statistical significance of results.

5.2 COP Metrics

The premise underlying the calculated COP metrics was that these results could be compared to the results found in other studies that performed similar experiments. This comparison allowed for the assessment of whether the subjects who participated in this experiment displayed “normal” quiet stance. In other words, it provided a basis via which to determine whether the subjects exhibited typical quiet standing dynamics. The following subsections discuss the COP results within the context of this study and then compare these results to those found by other research.

5.2.1 Within-experiment Observations

Observing the change in magnitude of the means of the COP parameters in Table 4.1 it was readily seen that all parameters increased in switching from the eyes-open to eyes-closed conditions. This may indicate that suppressing the visual facet of the proprioceptive system led to a greater degree of postural sway. However, the standard deviation magnitudes exhibited a different trend between the two visual conditions: The standard deviation decreased for all parameters except measures of COP length (i.e., TOTEX measures) and measures of COP velocity (i.e., MVEL measures). Thus, despite the lack of vision, the fact that the COP was moving at a higher velocity and traversed a longer path, and that the subjects stated that he/she experienced a noticeable increase in postural sway during their first eyes-closed trial, the subjects still demonstrated less variation - in general - in their postural sway. The decrease in variation could perhaps be explained by accepting the notion that the subjects employed a type of adaptation or a sort of learning mechanism as they completed two more eyes-closed trials. This adaptation may be seeded in the reweighting of sensory inputs by the proprioceptive system – a

hypothesis shared by Gatev et al. (1999). It should be mentioned here that the effect of test order (i.e., three consecutive eyes-open trials followed by three consecutive eyes-closed trials) did not show statistical significance.

5.2.2 Comparison to Past Studies

Contrary to the results of this study, others have documented that changing from eyes-open to eyes-closed conditions has a significant effect on the COP metrics displayed by young healthy adults, which is how the subject population of this study may be described (mean age of 22 ± 1 year). For example, in a similar experiment Prieto et al. (1996) examined twenty young adults (10 females, 10 males – recall, the current study involved 9 subjects: 5 females, 4 males) who had a mean age of 26.4 ± 4.9 years. These subjects performed quiet standing trials with both feet on one force plate for a duration of thirty seconds. The subjects performed one eyes-open and one eyes-closed trial. The same COP metrics that were computed in this study were also computed in the Prieto et al. study. While they reported an increase in all COP parameters in the eyes-closed condition – similar to the results of the current study - these researchers documented a significant effect between visual conditions – unlike the results of the current study. Prieto et al. reported that a significant effect with respect to visual status for the following: MDIST ($p < 0.01$), MDIST_{AP} ($p < 0.01$), RDIST ($p < 0.01$), RDIST_{AP} ($p < 0.01$), TOTEX ($p < 0.01$), TOTEX_{AP} ($p < 0.001$), MVEL ($p < 0.01$), MVEL_{AP} ($p < 0.001$), and confidence ellipse area ($p < 0.01$). Recall that significance was determined in the current study by using $p < 0.05$.

In agreement with the findings here, the results of Gatev et al. (1999) reported a lack a statistical significance between COP metrics for eyes-open and eyes-closed conditions. Gatev et al. observed a group of seven male subjects, who had a mean age of 42 ± 3 years, as they stood quietly on one force plate. The subjects carried out five eyes-open and five eyes-closed trials that

lasted for fifty seconds. The researchers monitored the subjects' COM using cameras and body markers as it deviated in the anterior-posterior and medial-lateral directions. The motion of the COM was cross-correlated to the COP motion that was tracked by the force plate. Using a confidence level of $p < 0.05$ - the same that was used in this study – Gatev et al. did not find a significant effect of vision (or lack of) on quiet stance measurements.

When one compares the results of this study to the ones described previously, deciding if the subject pool demonstrated typical quiet stance dynamics is not a straightforward task. Ideally, to deem the quiet standing of a subject to be typical would mean it has characteristics that are representative of the majority of the population. Due to the different methods and slight nuances used in the quantifying of quiet standing dynamics, however, makes this an impractical criterion. Perhaps the most practical means of comparison lies within COP-based measurement, as these are among the most popular means of studying quiet standing. While no standardized list of COP metrics exists to define what normal postural dynamics are, tracking the effect of vision on COP-based measures is common and could serve as a sensible criterion for distinguishing between typical and atypical postural dynamics.

With this in mind along with the previous works described, the Prieto et al. (1996) experiment appears to provide a suitable baseline for comparison as it investigated a greater number of subjects compared to both the current study and the one by Gatev et al. (1999); and, therefore may exhibit a higher statistical power (recall that power is proportional to the number of observations made) making it more likely to predict a significant effect when one is present. Then, just as in the Prieto et al. paper, one could expect that a significant effect of eyes-open versus eyes-closed conditions with respect to COP-based measurements is typical in the quiet standing of young healthy adults. Ergo, the lack of significance found between these two conditions indicates that the subject pool of the present study displayed atypical postural dynamics.

However, it is possible that the results of Prieto et al. were influenced by only conducting one trial in each visual condition for each subject (1 trial x 20 subjects = 20 observations). Since Gatev et al. conducted five trials (5 trials x 7 subjects = 35 observations) and the present study conducted three trials (3 trials x 9 subjects = 27 observations) in each visual condition, it may also be construed that these studies had a higher statistical power than that of the Prieto et al. study. On this basis then, a different conclusion would be drawn: The visual condition has no significant effect on the measures of COP motion of healthy young adults; therefore, the subject pool of the current study demonstrated typical postural dynamics.

Assuming that the visual condition of a subject does not affect his/her postural stability to a significant degree, it may be inferred that some clinical groups may not be as at risk for falling as previously thought. For example, subjects with visual deficits due to pathological disorders or from environmental factors (e.g., a dimly lit room) may not be at a significant risk to fall. This could suggest that the visual component of sensory information does not play a major role in postural control.

Despite the presence or lack of statistical significance, the results of the current study should not be discarded. In its defense, the subject pool was small and therefore may have lacked the statistical power necessary to reveal significant results even if they were indeed present. While this statement certainly points out a downside of the current study that could be improved upon in future studies, potential evidence regarding postural control mechanisms may indeed lie within the present results and may hold importance in a clinical setting. It was on this foundation that the conclusions in the subsequent sections were built.

5.3 Approximate Entropy and Control Mechanisms

The approximate entropy (ApEn) measurements were crucial in accomplishing the purpose of this present study: to ascertain the degree of lower limb independence and relatedness during quiet standing. In order to do this, the ApEn of the vertical ground reaction force (VGRF) underneath each foot was computed individually ($ApEn_{LEFT}$ and $ApEn_{RIGHT}$), the approximate entropy of the sum of the two VGRFs ($ApEn_{VGRF_{NET}}$), and the approximate cross-entropy (Cross-ApEn) comparing the two VGRFs. These ApEnS were examined over the entire test interval. Before moving into what conclusions concerning postural control can be elicited from the results, three imperative points should be clarified: Firstly, the ApEn of the VGRFs was taken instead of the COP because these forces are not coupled unlike the COP which depends upon other forces and moments. Secondly, the Cross-ApEn analysis monitored the development of disorder (or order) in one signal and provides information of whether a corresponding amount of disorder (or order) simultaneously develops in the other signal. Thirdly, bear in mind that a low value of ApEn corresponds to order whereas a high value of ApEn corresponds to disorder.

It was observed that $ApEn_{LEFT}$ and $ApEn_{RIGHT}$ displayed relatively low values that tended towards zero. This indicated that when the signals were analyzed separately a high degree of order existed. However, when the Cross-ApEn of the two VGRFs was examined it displayed a relatively high value, which was even higher than the previously reported value of ApEn for white noise (i.e., $Cross-ApEn > 2$), compared to $ApEn_{LEFT}$ and $ApEn_{RIGHT}$. If two signals developed disorder in a similar fashion (i.e., as chaos develops in one signal, a proportional amount of chaos develops in the other signal), then one would expect the Cross-ApEn of the signals to be low in value, which would be reflective of the notion that the two signals were similar to one another. Thus, the high value of Cross-ApEn observed in the current study implies that the VGRF under the left foot is very much unrelated to the VGRF under the right foot.

Summarizing the ApEn results: $ApEn_{LEFT}$ and $ApEn_{RIGHT}$ were low in value therein reflecting that the individual VGRF signals were orderly. Cross-ApEn of the two VGRF signals was high in value therein indicating the signals were very disorderly or unrelated compared to each other. Comparison of these ApEn results allowed for the purpose of this study to be accomplished: The calculated ApEn results suggest that the left leg and the right leg are controlled *independently* of one another during quiet standing. This claim was further substantiated by the ApEn VGRF_{NET} measures, which were slightly higher than $ApEn_{LEFT}$ and $ApEn_{RIGHT}$. This outcome intuitively makes sense if the previous conclusion holds true because one would expect that if two unrelated signals are combined, then they should exhibit a greater degree of disorder - ergo, a greater ApEn - than when examined separately.

As was the case with the COP metrics, no statistically significant effects were observed for any of the ApEn results (See Table 4.3). However, this should not overshadow the potential clinically valuable evidence concerning postural control detected in this study.

5.4 Limitations of the Present Study

There were several limitations of this study. Firstly, the small number of subjects, which was a product of time limitations, may not have provided the best representation of the general population. Secondly, the subject pool was not as ideally controlled as perhaps it could have been leading to a larger degree of variance than was acceptable. For instance, using a mixed group of males and females possibly could have skewed results. Due to innate anatomical differences (e.g., the wider pelvis generally seen in females or the larger anthropometrics seen in males), it is no surprise that differences in postural characteristics arise between males and females. As mentioned previously, it is possible that an insufficient sample size and/or a large degree of variance could have certainly led to insufficient statistical power. Calculating statistics that

indicate levels of significance was not the crux of this study, but computing significant results surely helps increase confidence that the obtained results are both valid and reproducible.

Additionally, as this was the first time ApEn measurements were applied to the dynamics of quiet standing, it was not possible to compare them to any references. In other words, it was not possible to deem the ApEn results typical or atypical as was done with the COP-based measures. This fact, undoubtedly, merits more study to further understand and/or characterize the ApEn results exhibited by healthy young adults. Another limitation of this study in regard to the ApEn measurements was that while they do suggest that the lower limbs are controlled independently of each other, it was beyond the scope of the present study to suggest that this conclusion can be generalized to all young, healthy subjects. Once again, more study is warranted to reveal if this is true; nonetheless, the ApEn results do start to build a promising case.

5.5 Suggestions for Future Studies

The work done in the present study has laid the groundwork from which several, similar studies may be formulated. For example, since this study examined healthy young adults, it would be interesting to analyze a group of elderly individuals to see if the degree of lower limb independence is different or even the same. On a similar note, it would also be interesting to track the degree of lower limb independence throughout different stages of a subject's development (i.e., early childhood to adolescence to early adulthood to late adulthood), although carrying out such an experiment may be impractical.

Besides varying the type of subjects tested, the testing procedure used in this study could be varied slightly. For example, repeating the same protocol that was used here except asking participants to stand on one leg in both eyes-open and eyes-closed conditions could help further substantiate the significance (or lack thereof) of vision during quiet standing. Another interesting

procedure variation would be to apply a slight unexpected perturbation (i.e., a light push) to the subjects during their quiet standing trials and observe how this disturbance influences the degree of lower limb independence as the subjects stabilize themselves.

Chapter 6

Conclusion

This study demonstrated that the lower limbs might be controlled independently of each other during quiet standing. This asymmetry was quantified in a group of young healthy adults, who displayed typical postural dynamics, by taking approximate entropy (ApEn) measurements of the vertical ground reaction forces (VGRFs) created under the left and right foot. As the VGRF signals are a direct consequence of the control required to maintain quiet stance, the ApEn measures allowed for the examination of the possible control mechanisms at work and to quantify the degree to which the control of the left foot was related to the control of the right foot. Individual ApEn measures tended toward zero while approximate cross-entropy measures were greater than two. These ApEn values indicated order in the individual signals and disorder when the signals were analyzed with respect to one another. Altogether these results implied the VGRF under the right foot were very much unrelated to the VGRF under the left foot. ApEn results of the two VGRFs summed together (ApEn VGRF_{NET}) also support this conclusion.

Although no statistically significant results were generated, the evidence observed and the implications associated with this evidence were not ignored as they may hold clinical importance. In particular, understanding how the left and right leg are controlled during quiet standing will ideally lead to the development of better clinical assessment tools and therapeutic techniques that are aimed at improving postural stability for those who suffer balance deficits. Perhaps ApEn measures could be used in conjunction with traditional COP-based measurements as an added layer of analysis to help characterize typical postural dynamics. Future studies designed to validate the interesting results obtained here are undoubtedly warranted.

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Appendix

MATLAB Pseudocode

The purpose of the MATLAB code used in this study was to take the raw force plate data collected by the Labview 6.1 software and use it to compute various, traditional COP parameters as well as entropy-based measurements of the vertical ground reaction forces created by the subjects during quiet standing. The COP parameters were computed using MATLAB and then compared to the results seen in other studies, which utilized experimental procedures that were similar to the ones used here. The relative and cross-entropy measures that were calculated allowed for the quantifying of lower limb independence and relatedness during quiet standing. Along with the aforementioned measurements, the code block was also used to create various plots including stabilograms and vertical ground reaction force data vs. time. The succeeding sections provide more detailed descriptions and the pseudocode associated with all the code blocks used in this study.

A.1 FP_Analysis.m

This is the main portion of code used to analyze the data gathered from the force plates. One of the purposes of this code is to calculate the respective COP under the right foot and under the left foot. A collective COP created by both feet – COP_{net} – is also computed. Plots of the left foot COP, right foot COP, COP_{net} , and the vertical ground reaction forces under each foot are output, as well. There are various call-statements to other m-files found within this code. Descriptions of these other files may be found throughout this appendix.

A.1.1 Pseudocode

Clear workspace, close open files, and clear screen

Set path to locate directories containing code

Display description of FP_Analysis.m to screen

Get and load Excel file containing raw force plate data:

```
for i = 1:6      % create loop such that all 6 trials for each subject are analyzed
```

```
    data = xlsread(filename, i, 'B1:M40020')
```

Assign filter parameters

```
    cutoff frequency = 6 Hz
```

```
    interval between samples: dt = 0.001
```

```
    filter order: forder = 2
```

```
    time base t = 0:dt:((length (data) - 1)*dt)'
```

Send data to be filtered forward and reverse using a 2nd order butterworth filter

```
    Call filtering file: [data] = Filtmat (dt, cutoff, forder, data) (See A.2)
```

Disregard first 5 seconds and last 5 seconds worth of data. Recall 1000 Hz sample rate:

```
    temp = data(5001:35000, :);
```

```
    clear data
```

```
    data = temp;
```

Extract data from filtered matrix:

```
    Small Force Plate data      % Right foot data
    Fx1 = data(:,1);            % ML Ground Reaction Force
    Fy1 = data(:,2);            % AP Ground Reaction Force
    Fz1 = data(:,3);            % Vertical Ground Reaction Force
    Mx1 = data(:,4);            % ML Moment
    My1 = data(:,5);            % AP Moment
    Mz1 = data(:,6);            % Vertical Moment
```

```
    Large Force Plate data      % Left foot data
    Fx2 = data(:,7);            % ML Ground Reaction Force
```

```

Fy2 = data(:,8);           % AP Ground Reaction Force
Fz2 = data(:,9);           % Vertical Ground Reaction Force
Mx2 = data(:,10);          % ML Moment
My2 = data(:,11);          % AP Moment
Mz2 = data(:,12);          % Vertical Moment

```

Calculate Large Plate (Left foot) COP:

factor in force plate covering thickness: $h = 0.005$ m

COP x-coordinate: $\text{copml_L} = ((-h * Fx2) - My2) ./ Fz2$

COP y-coordinate: $\text{copap_L} = ((-h * Fy2) + Mx2) ./ Fz2$

Convert COP coordinates from m to mm by multiplying by 1000

Calculate Small Plate (Right foot) COP:

factor in force plate covering thickness: $h = 0.005$ m

COP x-coordinate: $\text{copml_S} = ((-h * Fx1) - My1) ./ Fz1$

COP y-coordinate: $\text{copap_L} = ((-h * Fy1) + Mx1) ./ Fz1$

Convert COP coordinates from m to mm by multiplying by 1000

Shifting Force Plate COP data sets into a Global Reference Frame:

$xG(:,1) = \text{copml_L} - 102.3975$

$xG(:,2) = \text{copml_S} + 102.3975$

$yG(:,1) = \text{copml_L}$

$yG(:,2) = \text{copml_S}$

Calculating Net COP in Global Reference Frame:

Fraction of left foot weight: $A = Fz2 ./ (Fz2 + Fz1)$

Fraction of right foot weight: $B = Fz1 ./ (Fz2 + Fz1)$

COPnet x-coordinate: $\text{COPnet_ml} = (xG(:,1) .* A) + (xG(:,2) .* B)$

COPnet y-coordinate: $\text{COPnet_ap} = (yG(:,1) .* A) + (yG(:,2) .* B)$

$\text{COPnet} = [\text{COPnet_ml} \text{ COPnet_ap}]$

Reference COP data to mean COP:

$$AP = \text{COPnet}(:,2) - \text{mean}(\text{COPnet}(:,2))$$

$$ML = \text{COPnet}(:,1) - \text{mean}(\text{COPnet}(:,1))$$

Plot COP data:

One subplot with left foot COP data

One subplot with right foot COP data

One subplot with COPnet data

Confidence ellipse added to each plot:

$$[\text{area}, \text{axes}, \text{angles}, \text{ellip}] = \text{ellipse}(\text{M/L data}, \text{A/P data}, 1) \text{ (See A.4)}$$

COPnet Metrics calculated: [RD, MDIST, MDISTap, MDISTml, RDIST, RDISTap, RDISTml,

TOTEX, TOTEXap, TOTEXml, MVEL, MVELap, MVELml] = COP_Metrics(AP, ML, t)

COP metrics displayed to screen

Vertical ground reaction force data for both feet plotted: plot(t, Fz2, t, Fz1)

Approximate Entropy for Right and Left Foot:

Right leg VGRF normalized: $x1 = Fz1./(BW*0.5)$ %BW = mass measured on scale
 multiplied by gravitational
 acceleration

Run length: $m = 2$

Noise threshold: $r = 0.0717./(BW*0.5)$

Call Approximate Entropy m-file: [ApEn_Right] = Approxen(x1,m,r) (See A.6)

Left leg VGRF normalized: $x2 = Fz2./(BW*0.5)$

Call Approximate Entropy m-file: [ApEn_Left] = Approxen(x2,m,r) (See A.6)

Cross-Entropy:

Right leg VGRF normalized: $x = Fz1./(BW*0.5)$

Left leg VGRF normalized: $y = Fz2./(BW*0.5)$

Call Cross-Entropy m-file: [Cross_ApEn] = cross_approxen(x,y,m,r); (See A.7)

Collect COP metric data:

```
data_out = [MDIST MDISTap MDISTml RDIST RDISTap RDISTml TOTEX
TOTEXap TOTEXml MVEL MVELap MVELml area_net ApEn_Right ApEn_Left
```

Cross_ApEn] (See A.5)

Create labels for Excel columns:

```
labels = {'MDIST (mm)' 'MDISTap (mm)' 'MDISTml (mm)' 'RDIST (mm)' 'RDISTap
(mm)' 'RDISTml (mm)' 'TOTEX (mm)' 'TOTEXap (mm)' 'TOTEXml (mm)' 'MVEL
(mm/s)' 'MVELap (mm/s)' 'MVELml (mm/s)' 'Ellipse Area (mm^2)' 'ApEn_Right'
'ApEn_Left' 'Cross_ApEn'}
```

Write data to Excel file:

```
xlswrite( filename, labels, 7, 'B1:Q1')
if i == 1
    xlswrite( filename, data_out, 7, 'B2:Q2')
elseif i == 2
    xlswrite( filename, data_out, 7, 'B3:Q3')
elseif i == 3
    xlswrite( filename, data_out, 7, 'B4:Q4')
elseif i == 4
    xlswrite( filename, data_out, 7, 'B5:Q5')
elseif i == 5
    xlswrite( filename, data_out, 7, 'B6:Q6')
elseif i == 6
    xlswrite( filename, data_out, 7, 'B7:Q7')
end
```

end % ends loop designed to gather all 6 trials performed by one subject

Remove path

A.2 Filtmat.m

This m-file is responsible for filtering the raw force plate data forward and in reverse using a Butterworth filter. For this particular study, a 2nd order Butterworth filter was used with a cutoff frequency of 6 Hz. The main output of this code is a matrix containing the filtered data.

A.2.1 Pseudocode

Allow for double pass: $\text{cutoff} = \text{cutoff} / (\text{sqrt}(2) - 1) ^ (0.5/\text{forder})$

Use built-in MATLAB function to compute coefficients: $[\text{b},\text{a}] = \text{butter}(\text{forder}, 2*\text{cutoff}*\text{dt})$

Pad data to account for endpoint problems $[\text{data}, \text{nadd}] = \text{Paddon}(\text{data})$ (See A.3)

Determine size of data: $[\text{m},\text{n}] = \text{size}(\text{data})$

Use built-in MATLAB digital filter:

```
for i = 1:n
    fdata(:,i) = filtfilt(b,a, data(:,i))
end
```

Unpad data: $\text{fdata} = \text{Paddoff}(\text{fdata}, \text{nadd})$ (See A.3)

A.3 Paddon.m and Paddoff.m

The purpose of these code blocks is to pad a matrix with extra data in order to avoid endpoint problems that may arise with use of a Butterworth filter. Paddon.m adds extra data while Paddoff.m removes the extra data.

A.3.1 Pseudocode Paddon.m

Find size of input matrix: $n = \text{size}(\text{data}, 1)$

Set size of padding: $\text{nadd} = \text{round}(n * 0.3)$

Pad beginning, middle, and end of input matrix:

```

b(nadd,:) = a(1,:) - (a(2,:) - a(1,:));
for i=1:nadd-1
    diff = a(i+1,:) - a(i,:);
    b(nadd-i,:) = b(nadd-i+1,:) - diff;
end
b(nadd+1:nadd+n,:) = a(1:n,:);
b(n+nadd+1,:) = a(n,:) + (a(n,:) - a(n-1,:));
for i=1:nadd-1
    diff = a(n-i,:) - a(n-i-1,:);
    b(n+nadd+i+1,:) = b(n+nadd+i,:) + diff;
end

```

A.3.2 Pseudocode Paddoff.m

Find size of input matrix: $n = \text{size}(\text{fdata}, 1)$

Find start and finish: start = nadd+1, fini = n-nadd

Unpad: b = fdata(start:fini,:)

A.4 ellipse.m

The code contained in this section is based on an algorithm presented in: *Oliveira LF, Simpson DM, Nadal J. 1996. Calculation of area of stabilometric signals using principal component analysis. Physiological Measures 17: 305-312.* In brief, principal component analysis is used to fit a confidence ellipse to the stabilograms generated by FP_Analysis.m. The axes, angles, and area of this ellipse are also computed.

A.4.1 Pseudocode

number of inputs (x- and y-matrices) found

covariance matrix calculated: $V = \text{cov}(x,y)$

eigenvectors/values of V: $[\text{vec}, \text{val}] = \text{eig}(V)$

axes, angles, and area of ellipse computed:

```
axes = 1.96*sqrt(svd(val))
```

```
angles = atan2(vec(2,:), vec(1,:))
```

```
area = pi*prod(axes)
```

ellipse data:

```
t = linspace (0, 2*pi)
```

```
ellip = vec*1.96*sqrt(val)*[cos(t); sin(t)] + repmat( [mean(x); mean(y)],
```

```
1, 100)
```

ellipse data plotted

A.5 COP_Metrics.m

This code block calculates various COP-based parameters and is based off of the work presented in: *Prieto, T. E., Myklebust, J. B., Hoffmann, R. G., Lovett, E. G., & Myklebust, B. M. (1996). Measures of postural steadiness: Differences between healthy young and elderly adults. IEEE Transactions on Bio-Medical Engineering, 43(9), 956-966. doi:10.1109/10.532130.* It is important to note that all the COP data input to this file is from the previously computed COP_{net} data and that this data has been referenced to the mean COP.

A.5.1 Pseudocode

resultant distance: $RD = \sqrt{AP.^2 + ML.^2}$

average distance from mean COP: $MDIST = \text{mean}(RD)$

average AP and ML distance from mean COP:

$MDISTap = \text{mean}(\text{abs}(AP))$

$MDISTml = \text{mean}(\text{abs}(ML))$

RMS distances from mean COPnet:

$RDIST = \sqrt{\text{mean}((RD.^2))}$

$RDISTap = \sqrt{\text{mean}((AP.^2))}$

$RDISTml = \sqrt{\text{mean}((ML.^2))}$

total length of COPnet path:

$TOTEX = TOTEXap = TOTEXml = 0$

for ((length of data) - 1)

$TOTEX = \sqrt{(AP(i+1) - AP(i))^2 + (ML(i+1) - ML(i))^2} + TOTEX$

$TOTEXap = (\text{abs}(AP(i+1) - AP(i))) + TOTEXap$

$$\text{TOTEXml} = \text{abs}(\text{ML}(i+1) - \text{ML}(i)) + \text{TOTEXml}$$

end

mean velocity of COP:

T = time of test interval

MVEL = TOTEX / T

MVELap = TOTEXap / T

MVELml = TOTEXml / T

A.6 Approxem.m

This code block's purpose is to calculate the approximate entropy of a signal, and it is based on work presented in *Pincus, S. M., & Singer, B. H. (1996). Randomness and degrees of irregularity. Proceedings of the National Academy of Sciences, 93, 2083-2088.* There are a few important things to keep in mind: If the approximate entropy of a signal tends to be low, then the signal is highly regular. Conversely, if the approximate entropy of a signal is high, then the signal is highly irregular. The noise threshold, r , was computed in an additional set of experiments where a known weight was placed on the force plate and the standard deviation of the signal was determined. This procedure was carried out for both of the force plates used in this study. The highest standard deviation value that was computed was used as the noise threshold – any data value exceeding this value was discarded in the calculation of approximate entropy. It is also important to note that the VGRFs were first normalized by dividing them by half of the subject's measured body weight before their approximate entropy was computed. The noise threshold was also scaled by half of the subjects body weight.

A.6.1 Pseudocode

Number of vectors to compare:

```

n = length( x );
nsum = n - m;
for i=1:nsum,
    id(i) = 0;
    ic(i) = 0;
    for j=1:nsum,
        jj = 0;
        for k=1:m,
            dif = abs( x(i + k - 1) - x(j + k - 1) );
            if (dif > r)
                jj = 1;
                break
            end
        end
        end
    if jj == 0
        ic(i) = ic(i) + 1;
        dif = abs( x(i + m) - x(j + m) );
        if dif < r | dif == r
            id(i) = id(i) + 1;
        end
    end
end
end

```

end

Compute ApEn(m,r,n) from id(.) and ic(.). Take mean of logs of conditional probability of staying close, if already close:

```

ApEn = 0.0;
for i=1:nsum,
    ratio = id(i) / ic(i);
    ApEn = ApEn + ( log( ratio ) );
end
ApEn = -1.0 * (ApEn / nsum );

```

A.7 cross_approxen.m

This m-file is designed to calculate the cross-entropy between two signals and is based on the work presented in: *Pincus, S. M., & Singer, B. H. (1996). Randomness and degrees of irregularity. Proceedings of the National Academy of Sciences, 93, 2083-2088.* If the cross-entropy of the signals tends to be higher (or lower) than the individual approximate entropies of the two signals, then the two signals may be considered highly irregular (or regular) with respect to each other. Once again, the VGRFs were normalized prior to the entropy calculations.

A.7.1 Pseudocode

Normalize input data:

$$xx = (x - \text{mean}(x)) / \text{std}(x)$$

$$yy = (y - \text{mean}(y)) / \text{std}(y)$$

Number of vectors to compare:

```
n = length( x );
```

```
nsum = n - m;
```

Dimension arrays used for comparisons:

```
phi0 = zeros( nsum, 1);
```

```
phi1 = zeros( nsum, 1);
```

Compute how many points are close:

```
for k=1:m
```

```
    for i=1:nsum
```

```
        for j=1:nsum
```

```
            d=1;
```

```
            diff1 = abs( xx(i+d-1) - yy(j+d-1) );
```

```
            if diff1 <= r
```

```
                phi0(i) = phi0(i)+1;
```

```
                diff2 = abs( xx(i+m) - yy(j+m) );
```

```
                if diff2 <= r; phi1(i) = phi1(i) + 1; end
```

```
            end
```

```
        end
```

```
    end
```

```
end
```

Compute probability of staying close, if already close:

```
g = 1;
```

```
for f=1:nsum
```

```
    if phi0(f) > 0.5
```

```
        ic(g) = phi0(f);
```

```
        id(g) = phi1(f);  
        g = g + 1;  
    end  
end  
ratio = ( id ) ./ ( ic );  
for q=1:length(ic)  
    if ratio(q) < 0.00001  
        ratio(q) = 0.5 / ( ic(q) );  
    end  
end
```

Calculate cross approximate entropy: $\text{Cross_ApEn} = -1 * (\text{mean}(\log(\text{ratio})))$;

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