GENERALIZATION OF MEDIOLATERAL AND ANTEROPOSTERIOR POSTURAL CONTROL RESPONSES FOR DIFFERENT STANCE POSITIONS IN QUIET STANCE

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ABSTRACT

GENERALIZATION OF MEDIOLATERAL AND ANTEROPOSTERIOR POSTURAL CONTROL RESPONSES FOR DIFFERENT STANCE POSITIONS IN QUIET STANCE

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Human balance in quiet stance is the result of continuous center of pressure (COP) adjustments in anteroposterior (AP) and mediolateral (ML) directions and controlled by ankle and hip mechanisms. These mechanisms joins the balance control in different combinations, depending on the stance position. The purpose of this study is to determine the participation of these mechanisms on different stance positions. 22 healthy subjects (20-28 years) were required to maintain their balances on a dual force plate for 60 seconds in 9 different positions (feet side-by-side, intermediate stance: 30°, 45° and 60° right/left foot forward, tandem left, tandem right). Three postural parameters, COP_{net}, net COP change, COP_c, ankle activity and COP_v, hip activity and dependency or independency of COP_c and COP_{net} or COP_v and COP_{net} were calculated in addition to the amplitude of contributions. In ML direction, tandem stance is maintained mainly by ankle mechanism ($p<0.001$). For all conditions except tandem stance, ML balance is controlled by hip mechanism ($p<0.001$). In AP direction, tandem stance is controlled primarily by the hip mechanism ($p<0.001$), but all the other conditions are controlled dominantly by the ankle mechanism ($p<0.001$). Yet, secondary mechanisms also joins to this control. An effect of stance position was identified ($p<0.001$). Moreover, for intermediate stance conditions, this effect was evident only for the secondary mechanisms. Dominant mechanisms are not affected from the adopted position as much as secondary mechanisms. Secondary mechanisms on the other hand, are adjusted to adapt to new conditions.

Keywords: Postural Control, Mediolateral Balance, Anteroposterior Balance, Center of Pressure, Quiet Stance

ÖZ

SAKİN DURUŞTA MEDİOLATERAL VE ANTEROPOSTERİOR POSTÜRAL KONTROL YANITLARININ FARKLI DURUŞ POZİSYONLARI İÇİN GENELLENMESİ

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Sakin duruş esnasında insan dengesi basınç merkezinin anteroposterior (AP) ve mediolateral (ML) yönlerde sürekli olarak kontrolünün bir sonucudur ve bu kontrol ayak bileği mekanizması ve kalça mekanizmaları ile sağlanır. Bu iki mekanizmanın postüral kontrole katılım oranları kişilerin duruş pozisyonlarına göre değişir. Bu çalışmanın amacı farklı duruş pozisyonlarında ayak bileği ve kalça mekanizmalarının rolünü tespit etmektir. 22 sağlıklı katılımcıdan (20-28 yaş) dual kuvvet ölçer yüzey üzerinde 9 farklı pozisyonda (ayaklar yan yana, ara duruş: 30°, 45°, 60° sağ/sol ayak önde, tandem sağ, tandem sol), 60 saniye boyunca dengelerini korumaları istenmiştir. Katılımcıların COP_{net}, net basınç merkezi değişimi, COP_c, ayak bileği aktivitesi ve COP_v, kalça aktivitesi olmak üzere postüral parametreleri; COP_c ve COP_v ile COP_{net} arasındaki korelasyon ve her iki mekanizmanın katılım oranı hesaplanmıştır. ML yönde tandem duruşu sağlayan esas mekanizmanın ayak bileği olduğu görülmüştür ($p<0.001$). Tandem duruş haricindeki tüm duruşlar için ML yönde postüral kontrolü sağlayan ana mekanizma kalça mekanizmasıdır ($p<0.001$). AP yönde tandem denge kalça mekanizması ile kontrol edilmektedir ($p<0.001$). Tüm diğer duruşlar ise ayak bileği mekanizması tarafından kontrol edilmektedir. İncelenen bütün pozisyonlarda ikincil mekanizmanın da postüral kontrole belirgin ölçüde katılımı gözlenmiştir. Bununla birlikte duruş pozisyonunun mekanizmaların aktiflik ve katılımları üzerinde etkisi olduğu tespit edilmiştir. ($p<0.001$). Dahası bu etki ara duruşlar için yalnızca ikincil mekanizmalar arasında gözlenmiştir. Dominant mekanizmanın aktifliği ve katılım miktarı kişilerin adapte ettikleri duruş pozisyonundan ikincil mekanizmalar kadar etkilenmemektedir. Öte yandan ikincil mekanizmalar yeni durumlara adaptasyon amacıyla kullanılmaktadır.

Anahtar Kelimeler: Postüral Kontrol, Mediolateral Denge, Anteroposterior Denge, Basınç Merkezi, Sakin Duruş

To my family

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LIST OF ABBREVIATIONS

AP: Anteroposterior

ML: Mediolateral

COP: Center of Pressure

COG: Center of Gravity

COM: Center of Mass

IR: Intermediate Right

IL: Intermediate Left

SbyS: Side-by-side

TR: Tandem Right

TL: Tandem Left

SD: Standard Deviation

CHAPTER 1

INTRODUCTION

1.1. Background: Posture Control in Quiet Stance

Humans learn to stand on two legs at a very early age, and they are used to it so much that they do not notice how important this ability is and how it is maintained. Although bipedal standing seems quite straightforward, maintaining it is a very complex task for the body. Since two-thirds of the body mass is carried above a distance of two-thirds of body height from the ground, the system has to be controlled continuously to maintain its stability (Winter, 1995).

The importance of postural stability as a field of research is based on the impacts in different areas and different aspects. The upright stance is the first skill gained on two legs during motor development. Since motor development is a cumulative progress, all of the activities developed after upright stance such as walking depends on the success of upright stance. Although postural stability is the same ability that we gain in childhood, we see different aspects of it in wirewalking or wave surfing. As we age, it takes many forms in different postures and, maintenance of erect posture creates a basis for many activities. Hence, the deficiencies in postural stability result in problems like loss of balance control and falls. The research on postural stability, postural control, and factors affecting it helps us to understand the mechanism and prevent undesirable consequences due to altered posture. Thus, the topic attracted many interests over the years from various fields such as rehabilitation science, sports sciences, or robotics.

To maintain and improve the stability of human are the main interests in order to succeed in various physical activities, yet the human body is an inherently unstable system. Being in a static equilibrium means the resultant forces and moments acting on the body is zero. However, many factors prevent it, such as external forces like gravity or nonlinear behavior of muscles. When humans are standing erect, they do not stand still; instead, continuous muscle contractions occur, and they sway. This is called postural sway, and although it seems like an artifact, it is the difference between a human and an inanimate object. The postural sway mechanism is the reason behind we walk as bipedal creatures. Walking is the exaggerated version of this mechanism, and it is a continuous process of falling.

To understand how postural sway mechanisms work and how upright stance is maintained, it is necessary to be familiar with the terminology.

As mentioned above, postural sway is the continuous oscillatory action of the center of mass and center of pressure as a result of gravitational force.

Posture corresponds to the orientation of the body with respect to the gravitational vector (Winter, 1995).

Balance is a concept describing the dynamics of body posture, which is related to inertial forces acting on the body and inertial characteristics of the body, in order to prevent falling. It is the ability of not to fall from a clinical point of view. Inanimate objects fall or move due to gravitational force when their gravity line comes out of the base of support while the human body can counteract the gravitational force to prevent falling with continuous muscular activities. Thus, human beings have control over their balance, which we call it as postural control (Winter, 1995; Pollock, Durward, Rowe & Paul, 2000). In other words, it is the action or ability to preserve a state of balance with minimal postural sway (Pollock et al., 2000; Cho, K. Lee, B. Lee, H. Lee & W. Lee, 2014).

Center of mass (COM) is the point representing the weighted average of locations of each body segment in 3D space with respect to mass. It is the point at which the total body mass is concentrated and represented in the global reference system. COM is a passive or controlled variable, and it is controlled by the postural control activities to maintain a state of balance (Winter, 1995).

Center of gravity (COG) is the vertical projection of COM. Both COM and COG are positions, and they are expressed in meters as units (Winter, 1995).

Center of pressure (COP) is the point of application of the vertical ground reaction force vector. It represents the weighted average of the distributed pressure between the support surface and the contact area. It is independent of the COM. Changes in COP can be used as an indirect indicator of postural sway, and consequently, it represents the state of balance. When a person stands on one foot, then the COP lies somewhere within that foot. When both feet are on the ground, then the COP is somewhere in between two feet depending on the weight exerted by each foot. Since COP is a measure of balance, analysis of the COP is used in studies on balance and postural control, and it is commonly collected using force plates or pressure mats. One force plate can only yield the total COP of the body. However, when both feet are in contact with the ground, each has its own COP. In order to measure each COP individually, two force plates are needed (Winter, 1995).

As mentioned earlier, people sway when they are standing still, and this sway occurs in two cardinal planes of body. Sways in anterior-posterior direction occur in the sagittal plane and sways in medial-lateral direction occurs in the frontal (coronal) plane. The location of the COP depends only on the activity of ankle muscles. While the activity of plantar and dorsal flexor muscles moves the COP in the anteroposterior (AP) direction, evertor and invertor muscle activity move the COP in the mediolateral (ML) direction. Like COM and COG, COP is also a representation of a location, and its unit is meters (Winter, 1995).

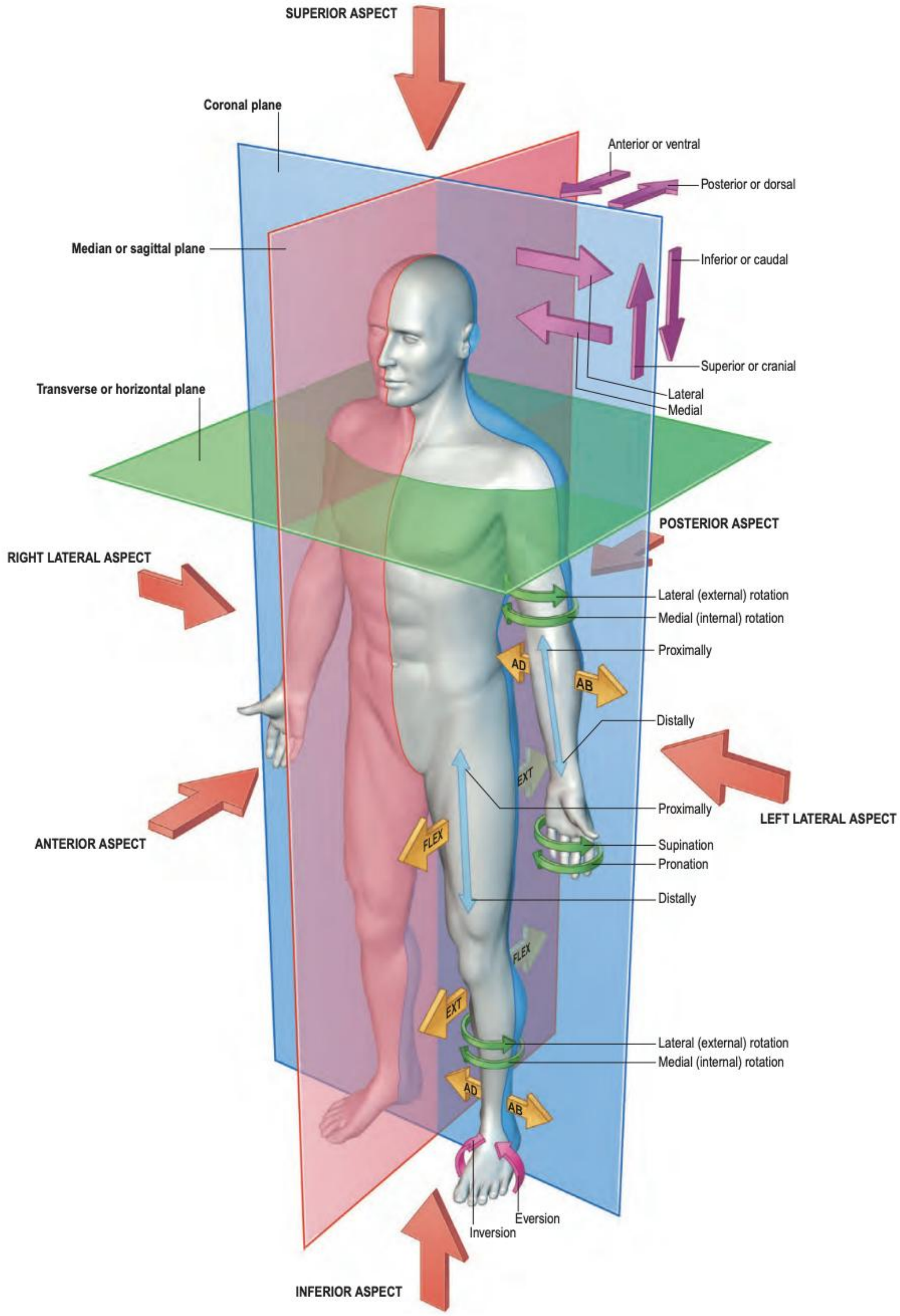

Figure 1.1. Cardinal planes and axes of movement for human body. Abbreviations shown on arrows: AD, adduction; AB, abduction; FLEX, flexion; EXT, extension. (Standring, 2016)

The posturl sway and ıts control is achieved by the integration of the information collected from proprioceptive, somatosensory, vestibular, and visual systems in the central nervous system.

As stated previously, even in quiet stance, postural sway occurs in both AP and ML directions. Only in particular circumstances, the COG is located directly over the ankle joint axis, and there is no need for any muscular activity for ankle joint to maintain balance (Nashner and McCollum, 1985; McCollum and Leen, 1989). Other times, one may say that there are always adjustments of COP and consequently, a postural sway to maintain balance.

Stability in both AP direction or sagittal plane and ML direction or frontal plane is critical to talk about fully achieving a state of balance. Lack of either one may result in poor balance or falls. A full understanding of postural control is necessary to prevent these outcomes and it is essential to analyze it in both planes.

In the sagittal plane, researchers treat the body as a 2D inverted pendulum moving around ankles since the rotation axes of ankle joints coincide. Having a look at the COP adjustments in this plane shows continuous COP regulation with the activity of ankle plantar and dorsal flexors. If the body COG is ahead of the COP with a clockwise angular velocity, in other words, when the body starts to lean in the forward direction, a clockwise angular acceleration is present. In order to correct this forward sway, the body will increase plantar flexor activity to move the COP ahead of the COG. With the activity of plantar flexor muscles, the COP being ahead of the COG will result in a counterclockwise angular acceleration, and at some point, it will reverse the angular velocity in the counterclockwise direction. Now the body is swaying backward; the plantar flexor activity will be decreased to correct this sway. This action moves the

COP behind the COG, and the body will experience a counterclockwise angular acceleration. This angular acceleration will decrease the angular velocity and reverse it. At this point, all the process has reversed into the beginning position as the body is forward swaying. These series of events occur almost in an infinite loop during quiet stance (Winter, 1995; Day, Steiger, Thompson & Marsden, 1993).

The described event can be seen in the frontal plane as well. Movements in the frontal plane can also be represented as an inverted pendulum, as long as the feet are not in a stance position wider than normal stance. Then, all of the processes mentioned above occur in the frontal plane about the ankle joint, as mentioned above. If feet are further apart, this time the movements are not only controlled by the ankle muscles but controlled by both ankle and hip muscles to move the COP towards the right or left of COG (Winter, 1995; Day et al., 1993).

1.2. Motivation and Purpose of the Thesis

Successful postural control constitutes a basis for many daily activities. Even standing still is a hardship for human beings, although it does not seem to. The body continuously and inherently puts an effort to control the quiet stance. Loss of this fundamental ability may lead to troubles while performing many activities and even may result in falling. In order to increase the life quality of the patients who experience a balance problem and to decrease health care costs, the best solution is to prevent them in the first place. Prevention of the problems related to the control of posture depends on understanding the control of posture. To be specific, it is impossible to avoid something which is not completely understood.

Naturally, many studies have been carried out to understand the postural control, and in light of these studies, it is known that there are two postural control mechanisms of quiet stance: an ankle or COP location control mechanism and a hip loading-unloading or bodyweight distribution mechanism. The activity of postural control mechanisms are specific to the distance and angle between feet or to be precise, specific to the stance position. Due to the importance of postural control in daily life activities and clinical practices in addition to the scientific interests, many scientists tried to determine the relative role of these two mechanisms in some fundamental stance positions such as normal stance, wide stance, narrow stance etc. However, the question of determination of a relationship between the results obtained from these various stance positions has remained unanswered. The insufficient number of studies in this particular topic confronts the researchers in the field with only a few studies focusing on the distribution of roles of these two mechanisms on postural control for different distances between feet in both mediolateral and anteroposterior directions. Though a frame has been drawn to elucidate the postural control in a typical quiet stance, the findings presented by studies yield controversial results in some particular positions. Most important of all, the main assets of the body to control the posture and the behavior of the body under various stance conditions adapted are known, which may help to understand the postural control of particular activities; but, it is not known if there is a relationship between these behavioral changes under different conditions. There is no data that can enable the professionals to have estimations on the postural control of a stance position that has never been studied.

This study is based on the anticipation that there should be a relationship between the relative positions of the feet and the proportional contribution of postural control mechanisms to sustain this position. This relation between stance position and postural control may be systematized according to the patterns observed, if any, in each position.

As mentioned earlier, based on the importance of the postural control mechanisms and the lack of sufficient studies about the effects of various common stance positions on those mechanisms, the purpose of this thesis is to determine the contribution of the ankle and loading-unloading mechanisms to the postural control during different stance positions adopted by individuals. With this motivation, three main stance positions are identified for investigation (Fig. 3.3) , which are side-by-side stance, tandem stance, and intermediate stance and the hypotheses are established accordingly.

1.3. Research Hypotheses

There are two postural control mechanisms of upright quiet stance: ankle mechanism and loading-unloading mechanism. Postural control can be interpreted in two planes in which body oscillations occur during upright stance: frontal plane and sagittal plane. These mechanisms contribute to the control of upright posture in each cardinal plane in different ratios relative to each other. The hypotheses of this thesis which are listed below, are established based on this information. They are also visualized in Figure 1.2.

H1: Postural control mechanisms are affected by the emplacement of the feet, and the observed effect is not arbitrary, instead there is a pattern between related conditions.

H2: The intermediate stances are controlled in such a way that the dominant control mechanism will be similar to the most lookalike border condition (i.e., side-by-side stance or tandem stance).

Five sub-hypotheses were also tested under the second hypothesis:

1. In a side-by-side stance, ankle mechanism controls the AP balance, and the loading-unloading mechanism controls ML balance.

2. In tandem stance, the AP balance is controlled by the loading-unloading mechanism, and ankle mechanism dominates the ML balance control.

3. In the 45° intermediate stance position (the angle is defined as the angle between the x-axis and the line passing through medial malleoli of the feet), both AP and ML plane postural controls are achieved with the almost equal contribution of both mechanisms.

4. For the 30° intermediate position, in the AP and ML directions, the postural control is dominated by the ankle mechanism and loading-unloading mechanism, respectively. This control pattern is following the most similar border condition, side-by-side stance.

5. In the 60° intermediate position, the control pattern is closer to the most similar border condition, tandem stance. The dominance of the loading-unloading mechanism in AP direction and the dominance of the ankle mechanism in ML direction are expected.

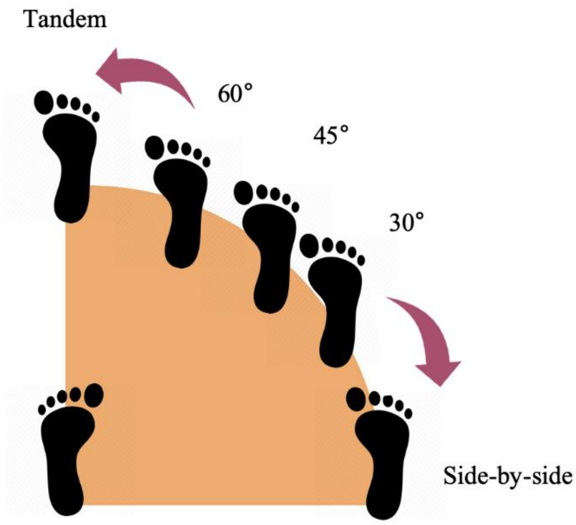

Figure 1.2. Visual representation of the hypotheses

1.4. Significance and Contribution of the Thesis

There are many studies in the posture and balance literature, illustrating that different stance positions affect the relative contribution of postural control mechanisms. However, to the best of author's knowledge, no study has been conducted to explore if there is a relationship between the relative contributions of these mechanisms under different stance conditions. To be precise, it is believed that there should be a certain behavior in the changes of the relative contributions of the postural control mechanisms when one foot is moved from, for instance, side-by-side stance to an intermediate stance position (where there is an angle between frontal axis and the line passing through medial malleoli of the feet). If there is a foreseeable relationship between stance positions, this can provide both clinicians and researchers an insight to estimate the dominant controller of different stance postures adopted in many different activities. With the identification of control mechanisms of the aforementioned stance positions, it will be possible to see the similarities and differences between all these positions. In addition, many disorders causing balance problems result in various posture and stance patterns specific to those disorders. For instance, neural disorders in the cerebellum can cause severe postural ataxia (Dichgans and Fetter, 1993; Diener et al. 1994). Identifying the control mechanisms of each condition and the relationship between these conditions can help health professionals to have an idea about the control process of a novel posture and balance adapted by the patients and improve the postural control mechanisms according to the needs of the patients.

This topic is worth studying in many aspects. First, this study is a comprehensive validation attempt that presents new data in side-by-side stance, tandem stance, and 45° intermediate stance positions. In addition, data from those three situations are collected from the same sampling group for the first time. By doing so, it helps to prevent the variability of the results between different study groups. This study will be the first study that points out the postural control mechanisms used in intermediate positions other than the 45° position. The collection of these data is a significant contribution on its own since, while walking, many different intermediate stance positions are evident in both healthy subjects and subjects with various kinds of disorders. By understanding the primary control mechanism during those particular positions, clinicians can develop compensatory mechanisms to improve the walking pattern of patients.

1.5. Thesis Outline

This thesis consists of five chapters and it is organized as follows:

Chapter 1 is the introduction to the thesis. In this chapter, an overview of the scope of this thesis is presented. Key terms and concepts related to postural control and balance are explained.

Chapter 2 presents an extensive survey of the literature in the postural control field. Earlier studies on postural control, studies investigated postural control mechanisms and the literature related to the effects of foot placement on postural control mechanisms are reviewed.

Chapter 3 is a detailed explanation of the methodology followed. It includes participant characteristics, experimental setup, and protocol, the method used for calculation of the contribution of the postural control mechanisms and data analysis procedure.

Chapter 4 presents results, discusses the postural control mechanisms in different stance positions concerning two cardinal planes, and demonstrates the relationship between the two postural control mechanisms in different stance positions.

Chapter 5 presents the summary and conclusions of the study. Main findings and results are discussed, and recommendations for further research in this field are given, and also the implications of this study are presented.

CHAPTER 2

LITERATURE REVIEW

2.1. Postural Control Mechanisms in Sagittal and Frontal Planes

The upright stance is one of the essential abilities of humans, yet it is difficult to maintain. Maintenance of upright stance and balance are the basis of most daily activities, but because of the body mass distribution, the body is unstable by nature. Human beings continuously work on keeping the body center of mass within the base of support even in quiet stance, although it seems there is no motion. Due to this inherent instability, humans are prone to fall when there is an internal or external intervention to the balance control system.

When standing still, the human body is not stationary like inanimate objects, but oscillates. The oscillations are the results of linear and angular movements of the body which are created by the postural muscles and are the responses of a postural control system to maintain a state of balance by controlling the displacement of the center of pressure to keep the center of mass in the base of support. The center of pressure displacement control is a result of different postural control mechanisms (Winter, Prince, Stergiou, and Powell, 1993). These mechanisms control the displacement of the center of pressure in two cardinal planes; frontal plane, and sagittal plane while standing still. These mechanisms are loading-unloading mechanism and ankle mechanism. The loading-unloading mechanism controls the bodyweight carried by each limb. It is also called a bodyweight distribution mechanism. When one limb is loaded, the other limb is unloaded proportionally. The ankle mechanism or COP location mechanism, on the other hand, determines the location of the center of pressure under each foot. The center of pressure displacements are a result of both mechanisms working together.

2.2. Previous Work on Postural Control Mechanisms

Before it is identified that two postural control mechanisms exist to control posture in sagittal and frontal planes, the majority of the researches on postural control were carried out only in the sagittal plane. Earlier studies on postural control mostly focused on the electromyographic activity of the muscles and provided valuable information on muscle activities during postural control. Okada (1972) studied relative muscular load in different human postures by collecting the surface EMG data from various muscles in the trunk, hip, and lower leg. It is concluded that in an upright stance, muscles are quite inactive, but the relative activity of soleus (ankle plantar flexor) is noticeable, which constitutes 10-20% of total muscular activity. Also, an activity of biceps femoris (knee extensor) can be seen in an amount of 5% compared to the total. As participants lean forward, the activity of soleus and gastrocnemius (ankle plantar flexor) dramatically increases.

Furthermore, a noticeable activity of tibialis anterior (ankle dorsiflexor) and quadriceps (knee extensor) and a small contribution of abdominal muscles are evident. Considering the results, it is evident that plantar flexor and dorsiflexor activity is present in an upright stance and forward lean condition. However, the results do not provide information on how the activity of these muscles control posture. Furthermore, the results are limited in the sagittal plane only, since the muscles in which activities are recorded are ankle muscles mainly and only hip muscles located in the anterior and posterior faces of the thigh which are responsible for the extension and flexion of the knee. The activity of muscles responsible for the abduction and adduction of the hip was not measured. Besides, the leaning posture is only confined in the forward-backward direction. Consequently, the information provided by this early EMG study is valuable yet limited.

Dietz and Berger (1982) studied the integration between left and right limbs in sagittal plane postural control. For this purpose, they created inequality between limbs by either stimulating the tibial nerve of one leg or by tilting the surface under one foot in the forward direction, causing a forward body lean. They measured the EMG activity of the tibialis anterior and gastrocnemius. As a result of the stimulation of the tibial nerve, triceps surae activation, i.e., plantar flexor activity increases in both limbs. The rotation of the platform in the forward-down direction, on the other hand, causes the activity of the tibialis anterior in both sides and results in dorsiflexion. It can be seen that, even if the feet are not in the same condition, they tend to respond symmetrically in the anterior-posterior plane. These results yield better insights into the control of anterior-posterior control of balance in terms of leg coordination and muscle activity; still, it lacks a comprehensive approach since it is confined only with ankle muscles.

The research on postural control mechanisms is essential since it is not a voluntary behavior; instead, it is a fast, continuous, inherent act that prevents humans from falling. Considering the activities of the body to maintain the balance in quiet stance are not obvious, earlier studies of postural control confined to determine which muscles are keeping humans in upright posture rather than for which aim do these muscle activities serve. The goal of the muscle activities is keeping the body posture in an upright position. However, the aim is to keep the body center of mass within the base of support with minimum oscillations, and this aim must be obtained by the organization of some mechanisms. The muscle activities then serve to these mechanisms to change the body COP and keep the projection of the COM in the support surface. These series of events are the reason which human beings can maintain an upright stance. Hence, although the muscle activity studies are essential to understand the mechanisms, they are just a starting point. As researchers were able to see which muscles are more active during the upright stance, they started to ask questions such as in which organization scheme do these muscles get activated? To this end, Nashner and McCollum (1985) studied the organization of postural control in the sagittal plane, but under perturbation conditions. They concluded that people mainly use a strategy they ended up calling an ankle strategy which consists of the plantar and dorsiflexor activity of the foot when the support surface moves with force in a rotational or translational manner. When the threat to the state of balance is more significant, for example, the support surface is smaller; they tend to use a pure hip movement or hip strategy to maintain balance. Then again, some people use a mixed strategy which includes the activation of both ankle and hip muscles, but it is denoted that a mixed strategy is only a transition strategy for people who cannot adopt a pure hip strategy and can be seen in people who are more prone to fall. The outcomes of this study create a new point of view in the postural control field since this research introduces an organization in which muscle activities occur. The two different postural control strategies introduced are the answer to the question of how the posture is controlled when there exists a threat to the steady-state, of course, if upright stance can be called as it a steady-state. However, as it is shown later, these strategies defined by Nashner and McCollum (1985) apply only to dynamic conditions, and they are some reflexive responses to recover the body posture after perturbations. They are the mechanisms used when there is external intervention to the normal postural sway. That said, these mechanisms do not give information on how balance control is sustained when there is not an intervention, i.e., an upright stance condition. In addition, these mechanisms do not give us any insight about the control of stance in the frontal plane, not necessarily in standard situations as quiet stance, even in compromised conditions. However, it is a valuable study showing which muscular groups are more involved in postural control.

Following this study, McCollum and Leen (1989) also studied the fast movements of postural control, i.e., the postural control mechanisms of disturbed stance in the sagittal plane. They have concluded that all of the movements generated to keep a state of the balance must be within a 360° stability cone including both sagittal and frontal plane although the study is conducted in the sagittal plane only. They used the results of Nashner and McCollum (1985) as a basis to define postural control strategies, but they used these strategies in a broader means to define all of the movements occurring to maintain balance which may include quiet stance as well. At the same time, they underlined that most subjects use an ankle strategy to preserve posture.

Macpherson (1988) carried out a study to investigate postural control mechanisms in quadrupedal stance and carried out the analysis in both sagittal and frontal plane by collecting EMG data from cats' limbs. IThe study focuses on the data obtained from hind limbs of cats since these limbs work together in postural control to displace the center of mass in the necessary direction. This pattern can also be seen in humans, i.e., interlimb coordination such as a cooperation in the sagittal plane and a reciprocal activity in the frontal plane. The study also points to the existence of balance control mechanisms for both planes. According to this research, postural control in the sagittal plane is achieved by flexors and extensors of hip and ankle in a cooperative action in both limbs, while in the frontal plane hip, knee and ankle flexors and extensors are working to control the posture in a reciprocal pattern. In the end, the study suggests an ankle and hip control of posture in the sagittal plane and a hip, knee, and ankle control over posture in the frontal plane. Although the study was carried out in cats, this is one of the earliest research on postural control mechanisms in both cardinal planes.

Up until this point, studies were carried out mostly by using mainly EMG data, rather than making COP analysis. COP analysis, on the other hand, gives a chance to analyze the controlling parameter of the COM, which is the main variable that is desired to be kept in the base of support. On attempts to collect COP data from a force plate, studies mainly focus on different characteristics of the COP data such as dimensions of sway path or average radius. Some attempts even take into account frontal plane COP changes during the analysis of COP signals (Black, Wall, Rockette & Kitch, 1982). Unfortunately, from a postural control point of view, those studies did not result with any comment on the mechanisms of control neither in static nor in perturbed conditions, even though they collect the methodologically required information on the COP time series.

Other researchers who collected COP data to analyze postural control tend to carry out investigations in postural control under various external perturbations (Moore, Rushmer, Windus & Nashner, 1988; Woollacott, von Hosten & Rösblad, 1988; Diener, Horak & Nashner, 1988; Diener, Horak, Stelmach, Guschlbauer & Dichgans, 1991; Horak, Nashner & Diener, 1990; Dietz, Trippel, Discher & Horstmann, 1991). These studies may give insights about the feedback system of human postural control but do not provide information on quiet stance control when there is no disturbance to the system from the outside environment. Despite the significant results they brought into the literature, they do not provide a basis to understand how unhealthy individuals sway more or lose their balance and fall even during quiet stance when there is no external disturbance. This focus on the perturbed conditions may be because of the difficulties of doing such dynamical balance experiments on unhealthy or disabled individuals. Nonetheless, those individuals do not necessarily lose their balance only after perturbations. Thus, static conditions must be analyzed in terms of postural control.

Collins and De Luca (1993) collected COP signals from a force platform and attempted to analyze postural control in static conditions. They interpreted the signals in sagittal and frontal planes to understand the COP behavior in both directions. From the analysis of the data, they concluded that collecting COP data from a force platform could give the information on the stochastic activity of COP along both mediolateral and anteroposterior axis. These data can be used to evaluate the postural instability of subjects. Still, these data by itself, unfortunately, cannot provide information on how postural control is achieved and by which mechanisms. Another key point De Luca and Collins discussed is about the control of these variations in COP signals. Though the data cannot provide information on the mechanism used, the researchers inferred some thoughts. The results were implying that mediolateral stability is more excellent than anteroposterior stability. They interpreted these results as anteroposterior stability is harder to achieve because of the coinciding ankle joints in the sagittal plane. In the sagittal plane, ankle joints act as a single hinge joint which allows rotation in this plane, and this is the reason for the excessive sway in that direction. While in the mediolateral plane, ankle joints are not coincident during actions of inversion or eversion. According to the author of this thesis, the misinterpretation in this conclusion is that lower extremity balance cannot be reduced only at the ankle joint level and relatively stable behavior in mediolateral direction cannot be explained only by joint alignments. That is to say, the problem in this conclusion is, it is based on the assumption of an ankle controlled balance in both planes without having the supporting data.

As can be seen and mentioned above, not many investigations were carried out to analyze postural control in the frontal plane neither by recording the EMG activity nor by interpreting the COP signals. Even the behavior under perturbation is focused only on the sagittal plane by researchers. However, Brunt et al. (1992) focused on the postural responses generated under lateral perturbations. They believed that lateral perturbations are critical from the aspects of lateral ankle complex injuries. Therefore, they recorded the EMG activity of ankle muscles (peroneus longus, posterior tibialis, and tibialis anterior) to define the muscle response to lateral perturbations better. EMG activity reports showed that posterior tibialis activity increased in the loaded limb while in the unloaded limb, only minimal peroneus longus activity was apparent.

On the other hand, the tibialis anterior activity was bilaterally similar. By looking at the results, it can be understood that invertor muscle activity controls the balance on the loaded limb, and everted, unloaded limb is only needed to be controlled minimally. The researchers implied that in the sagittal plane there is a coordinated activity of both legs, while in the frontal plane, the activity is reciprocal. The reason behind these results, as evaluated by the authors of the study, is probably the existence of a loading-unloading pattern to maintain postural control in the frontal plane. The author of this thesis founds this interpretation significant since the evidence can lead one to conclude that a loading-unloading mechanism exists and control frontal plane balance. The study was conducted under perturbed conditions, but the outcomes are critical since it is one of the earliest studies in which not only perturbations applied in the mediolateral direction but also the reactions in that plane, i.e., in the frontal plane, are also analyzed.

One of the most significant studies exploring the postural control mechanisms in the frontal plane investigated the effects of stance width and vision on the control of lateral sway (Day et al., 1993). The data were collected from a force platform and markers placed on the trunk, hips, knees, and ankles of subjects. Results show that, for the frontal plane, the ankle joint is active in the postural control only for the narrow stance width condition. In a narrow stance, the body behaves like an inverted pendulum in the frontal plane. For all other conditions, hip activity is dominant, but also the activity of ankle exists. However, in the sagittal plane, in contrast to the studies presented before, Day et al. (1993) suggested a hip control on anteroposterior movements. According to the data collected, in all conditions, including different stance width and vision conditions, most angular displacements occur in the hip region. It should be pointed out that, in this study, the hip region includes the movements of the pelvis together with vertebrae. The hip control becoming dominant as stance gets wider is related to joint alignment according to the authors, since none of the joints, neither ankle nor hip joints coincide as the distance between feet increases. As stance gets wider, lateral movements are increasingly controlled by muscles that abduct and adduct the hips instead of the evertors and invertors of the ankle. This theoretical framework created in this study may have an implication for clinical practices in patients with different lesions of the nervous system since different lesions affect balance in different directions such as increased lateral sway, anteroposterior sway or both. This work helps to understand which mechanical factors can be rehabilitated in order to make the balance system work properly. Even though this study does not take into account the behavior of COP signals, the researchers tracked the hip joint activity during quiet stance. It can be said that these outcomes help to reach a more general idea about the postural control mechanisms since the results cover all lower extremity and both cardinal planes.

Researchers investigated postural control by analyzing EMG signals, COP signals, and kinematic data obtained from joints to explore postural control in humans. Most of the studies were informative, yet somehow, they are not comprehensive. Earlier studies focused on postural control in the sagittal plane, and most of them tried to identify the active muscles during postural control by taking EMG data. Some later studies took the frontal plane into account; still, they were also limited to the EMG activity of the muscles. Some studies interpreted COP data, but their analysis lack of information, especially on the control of mediolateral plane balance. As mentioned earlier, the behavior of each limb is reciprocal when observed from the frontal plane. By using only one force platform, it is impossible to see the effect of the behavior of each limb separately. Instead, this methodology gives only the information about the COP displacement on each plane, which is the imbalance of the subject in those directions. Understanding the contribution of each leg in a quiet stance is only possible by using two separate force platforms for each foot.

Winter et al. (1993) realized that and introduced a groundbreaking concept. In their study, they first revalidated the concept presented by many researchers that anteroposterior balance is controlled somehow by cooperative activity of both limbs while a reciprocal activity between limbs controls the mediolateral balance by collecting COP data using two separate force plates for each limb. Many researchers denote that in quiet stance both anteroposterior and mediolateral balance in narrow stance conditions can be investigated by treating the body as an inverted pendulum where the control is at the ankle muscles. However, Winter et al. (1993) imply that this assumption may not be true unless a full analysis, including each limb, is carried out. With this in mind, it should be stated that the unilateral stance is controlled directly by the activity of ankle muscles. In a situation like this, the COP displacements in the anteroposterior direction are controlled by the ankle plantar flexors and dorsiflexors, and COP displacements in the mediolateral trajectory are controlled by the activity of ankle evertors and invertors. Since a normal, bilateral quiet stance is controlled by the net response of both limbs, individual responses from each limb must be collected. They proposed that the resultant COP in both cardinal planes is the weighted sum of the COP changes under each foot relative to the load carried by each limb. By taking a single limb stance as a base and carrying out a biomechanical analysis (Winter, 2009), they suggested that the ankle muscles control COP locations, but the contribution of COP changes under each foot is added to the resultant COP proportional to the load carried by each limb. From that point, two postural control mechanisms for quiet stance are presented. An ankle control mechanism and a loading-unloading mechanism. Both mechanisms control the resultant COP. Ankle muscles are responsible for the COP location, and this control mechanism is called the ankle mechanism while the loading-unloading control mechanism is described as a weighted average determinant. However, this study does not present any controlling muscle or joint activity responsible for the loading-unloading mechanism. In order to analyze the contribution of these two mechanisms to the control of quiet stance in both sagittal and frontal planes, the researchers extracted both mechanisms from the weighted sum of the resultant COP. The mechanism called the loading-unloading mechanism is a response created to keep the body in the base of support. Assuming that each foot carries a 50% proportion of the body weight on average, the loading-unloading mechanism represents the deviations from the average load carried by each limb. The loading-unloading mechanism shows the intentional effort of the body to change the load under each foot in order to preserve the state of balance. Hence, they assumed that a healthy person carries half of the body weight with right limb and the other half with the left limb and they calculated the fluctuations of the weight distribution by taking the 50% as a base and extracted them as the effort spent to keep the body in balance by the loading-unloading mechanism. After this extraction, they subtracted the contribution of the loading-unloading from total COP changes to obtain the contribution of the ankle. This method can be applied to both changes in the anteroposterior direction and the mediolateral direction to see the contribution of the ankle mechanism and the loading-unloading mechanism to the total postural control in each plane. The results show that, in a normal quiet stance position, postural control is obtained by pure ankle control in the sagittal plane. Each limb moves in cooperation, and the effect of the loading-unloading mechanism is negligible. The mediolateral control of posture is, on the other hand, controlled mainly by the loading-unloading mechanism with the small contribution of the ankle control mechanism. In their analysis, they found that in mediolateral plane COP changes with respect to time for each foot are in antiphase and with a high correlation. For example, ankle eversion in the right foot is somehow responded with the eversion of the left foot. If the load carried under each foot were the same for all time series, the resultant COP would be zero since the COP locations under each foot would result in the center as an average. However, they got a resultant COP far away from being zero, rather it fluctuated. This implies that the changes in resultant COP are due to the loading-unloading mechanism rather than the ankle mechanism. Provided that ankle muscles are not the main reason behind the COP displacements in ML direction, it can be concluded that the ankle evertors and invertors are not the muscles behind the loading-unloading mechanism. This leads to a thought that knee or hip muscles might be responsible for the loading-unloading mechanism. At this point, it should be underlined that the knee does not have a degree of freedom in the mediolateral direction and knee muscles are only responsible for the movements in the anterior-posterior direction, i.e., knee flexion and extension. Thus, this study claims that hip muscles are the reason behind the loading-unloading mechanism, and most probably the abductors and adductors of the hip which have control over movements in the mediolateral direction. In practice, this might be verified since a full abduction of one hip will lead the other limb to carry all the body weight and will unload the abducted knee. As a summary, this study by Winter et al. (1993) showed that the left and right limb cooperates in the anteroposterior direction and acts reciprocally in the mediolateral direction. The cooperation in the sagittal plane is the result of the ankle plantar flexor and dorsiflexor activity only. The reciprocal behavior seen in ML direction, on the other hand, is resulting from the activity of both ankle muscles and hip muscles; however the mechanism in charge of the changes in resultant COP is the loading-unloading mechanism, and this mechanism is the result of the activity of hip abductors/adductors. These results were of vital importance in many aspects. First, the study was carried out in undisturbed, quiet stance conditions, which is the basis for many activities and must be analyzed to understand the static balance of individuals fully. Second, this study is the first, which tries to define strategies that create postural control. Rather than the active muscles, this study focuses on why these muscles are activated in the first place. From the results of this study, we know that the purpose of the activation of ankle muscles is to change the location of COP to control COM while the purpose of the activity of hip muscles or muscles responsible for the control of body weight distribution is to keep the COP in the base of support by effecting COP location proportional to the load carried by each foot.

Another study is carried out by Winter, Prince, Frank, Powell, and Zabjek (1996) to expand the results obtained into different standing positions and to investigate the muscles responsible from the loading-unloading mechanism. In order to do that they took multiple records and carried out computational analyzes. First, they recorded EMG activity of hip abductors and adductors, but the activity of these muscles was subthreshold compared to the reaction forces in quiet stance. Thus no conclusion was carried out from that point. Then, they carried out a biomechanical analysis and calculated the moments on the hips in the frontal plane and compared the fluctuations in those moments for the reaction force changes. According to those analyzes and the implications of their prior study (Winter et al., 1993), they conclude that hip abductors and adductors are responsible for the loading/unloading mechanism, the mechanism responsible from the COP control in ML direction. The main purpose of the study was to apply the same approach used in the prior study in different stance positions. Thus, with the method introduced by Winter et al. (1993) contribution of the loading-unloading mechanism and contribution of the ankle mechanism to the postural control were calculated for tandem stance and an intermediate position of approximate 45º. Postural control mechanisms were interpreted for frontal and sagittal planes separately. The results showed that in tandem stance, anteroposterior balance is under the control of the loading-unloading mechanism and the contribution of the ankle mechanism was negligible. The mediolateral balance, on the other hand, was controlled by the ankle mechanism, and the role of the loading-unloading mechanism was small. Since the muscles responsible from the ankle movements in mediolateral direction are the ankle evertors and invertors, this result is interpreted as the mediolateral balance in tandem stance position is controlled by the ankle mechanism with the activity of ankle evertors and invertors. In the intermediate stance position, the mediolateral balance was controlled by the collaboration of both mechanisms where the contribution of the loading-unloading mechanism was dominant with a proportion of 60%. Both mechanisms were contributing to the change of net COP in an additive manner. However, in the mediolateral direction, the results were different. Both mechanisms were found to be responsible for the changes of the resultant COP, but in a different manner, compared to the mediolateral direction. The results showed that this time, there is a subtractive relationship between these two mechanisms. The ankle mechanism was dominant, but this was to decrease the effect of the loading-unloading mechanism and to create a much smaller change in total COP. In an intermediate stance position, both mechanisms were found responsible for the postural control in anteroposterior and mediolateral directions with a canceling and reinforcing behavior, respectively. The loading-unloading mechanism was thought to be activated by the hip flexors-abductors and extensors-abductors in each limb while the ankle mechanism was thought to be controlled by the canceling effort of the plantar flexors and dorsiflexors for anteroposterior direction and reinforcement of invertors and evertors of the ankle for the mediolateral direction. In summary, this study shows that the stance position influences the choice of appropriate postural control mechanisms. In tandem stance, balance in frontal plane and sagittal plane are controlled by ankle mechanism and loading-unloading mechanism respectively. In this stance position, these mechanisms control the posture independently. In an intermediate stance position where neither the ankle joints nor the hip joints coincide, these mechanisms act together. However, depending on the plane of motion, those mechanisms reinforce or cancel each other. As can be seen, these results are important, especially in terms of the effects of stance position in the control of posture. This study showed not only that postural control mechanisms exist, but also, these mechanisms depend on the stance position taken by individuals.

Understanding the postural control in different stance positions is important in many aspects. The human quiet stance is a fundamental skill of human beings for many activities, not only for static activities but also for dynamic activities such as walking especially double stance phase where both feet are in contact with the ground. The results in the presented study indicate the importance of analyzing the postural control mechanisms in intermediate stance positions which are used while walking. Also, investigations in different positions other than side-by-side stance are important for understanding different imbalances and falls which are seen in different kinds of disorders. The patients with cerebellar degeneration in anterior lobe present increased sways in the anteroposterior direction (Mauritz, Dichgans & Hufschmidt, 1979). In Friedreich's ataxia, body oscillations in mediolateral direction are more evident (Diener, Dichgans, Bacher & Gompf, 1984). Wallenberg's syndrome patients experience a diagonal pattern of sway after lateral medullary infarcts (Dieterich & Brandt, 1992). Patients having unilateral thalamic lesions (Masdeu & Gorelick, 1988) and unilateral basal ganglia lesions (Labadie, Awerbuch, Hamilton & Rapcsak, 1989) have an imbalance which leads lateral falls in the counter direction of the lesion. Analyzing the postural control in both frontal and sagittal planes and in different stance positions helps us to understand the affected mechanisms in those patients better. Besides, these analyses can be used to generate compensatory treatment programs for many patients who lost one of the postural control mechanisms like lower extremity amputees, diabetic patients and peripheral neurologic disorders, or hemiplegic patients who lost both postural control mechanisms in the affected side of the body. The analysis in different positions can be used to rehabilitate the patients by knowing the dominant strategy that must be improved. Also, the analyses conducted in different stance positions are not only important for walking, but they can also be used to understand the postural control in different sports.

The postural control mechanisms of quiet stance defined and investigated by Winter et al. (1993, 1996), provided a basis for studies in the detailed analysis of quiet stance. Many authors acknowledged these mechanisms and even observed in different patient populations in many publications. Every complementary study can also be thought of as a revalidation of the methodology and results presented by Winter et al. (1993, 1996). Studies validated that ankle mechanism and loading-unloading mechanism are the main controlling mechanisms of static balance in different disorders and disabilities such as diabetic sensory neuropathy, Parkinson's disease, transtibial and transfemoral amputations and total hip arthroplasties (Lafond, Corriveau & Prince, 2004; Termoz et al., 2008; Rougier & Bergeau, 2009; Rougier et al., 2008). Even though the contribution of each mechanism changed between disorders, the dominant strategy in the concerned plane did not change. Indeed the ankle mechanism and the loading-unloading mechanism are the dominant control mechanisms of feet side-by-side stance posture in anteroposterior and mediolateral directions respectively in both healthy subjects and individuals who have a disorder which might affect postural control. These studies are conducted in subjects who experience an ankle or hip disorder shows that the affected joint joins less in the postural control, and the unaffected joint compensates it. For example, in diabetic sensory neuropathy patients, sensation in peripheries of the body is affected. Thus, as can be expected the contribution of ankle mechanism decreased in both frontal and sagittal planes. Studies conducted in different age populations also show similar results. Young, middle-aged, and elderly people use the same postural control mechanisms (Termoz et al., 2008; Bonnet, Mercier & Szaffarczyk, 2013; Bonnet, Delval, & Defebvre, 2014). Still, as in unhealthy populations, aging also affects the contribution of each mechanism to the postural control, in fact, only ankle control is affected with aging, still not the dominant strategy in the concerned plane. The dominant strategies were found to be the same, even in different static postures like stooping and crouching (Weaver, Glinka, & Laing, 2017). All of those researches are not only explored the postural control mechanisms in different populations they also showed that these mechanisms exist by showing that for example a problem affecting the hip leads to a decrease in the contribution of the loading-unloading mechanism to the total postural control in a practical, clinical perspective.

Although studies conducted in different situations, it is obvious that one of the most confounding factors which may affect the relative contribution of postural control mechanisms is not disabilities or pathologies, rather it is the stance position of the participants as shown by Winter et al. (1996).

2.3. Effects of Stance Position on Postural Control

It is known for a long while that the position of feet affects the posture and balance (Fearing, 1924; Kirby, Price, & MacLeod, 1987; McIlroy, & Maki, 1997). Some researchers even tried to standardize the placement of feet during balance and posture tests to prevent between-subject variability in results. Not surprisingly, this influence of stance on balance took the interest of researchers who are trying to explore control mechanisms of posture.

To the best knowledge of the author, the first study investigating the effects of stance position on postural control is the study by Day et al. (1993). They investigated the effects of stance width on the postural control by using kinematic data. Although they did not identify specific mechanisms for posture control, they implied the control of ankle and trunk-leg complex on posture and tried to interpret the changes in the activities of ankle and trunk-leg complex according to the stance position. According to the results presented, the activity of the ankle controls the posture in frontal plane in a narrow stance. All other conditions in frontal and sagittal plane showed the control of trunk-leg activity.

After researchers discovered and acknowledged the postural control mechanisms in side-by-side stance as ankle and hip mechanisms, the question of whether these mechanisms are applied in the same way in different stance positions was raised.

Winter et al. (1996) conducted the first study exploring the effects of stance position on the postural control mechanisms. They investigated whether the ankle and loading-unloading mechanisms are present in tandem stance and an intermediate stance position and if present, how their contribution to postural control changes. The results of this study showed that both ankle and hip mechanisms control posture in tandem and intermediate stance; however, their contribution to balance is not the same as in the side-by-side stance position. In tandem stance, they found that the dominant controller of the frontal plane balance was ankle mechanism with the small contribution of the loading-unloading mechanism while the sagittal plane balance was controlled by the loading-unloading mechanism with the negligible participation of the ankle mechanism. In the intermediate stance position, postural control was the result of the contribution of both mechanisms almost equally, but the behavior of these mechanisms was totally different in sagittal and frontal planes. They reported that in the frontal plane both mechanisms reinforce to control mediolateral balance while in the sagittal plane, although both mechanisms collaborate, ankle mechanism has to dominate in order to cancel the effects of the loading-unloading mechanism. These conclusions were also validated through a study that expanded the investigation to young, elderly, and patients who have Parkinson's disease (Termoz et al., 2008).

The study carried out by Gatev, Thomas, Kepple, and Hallett (1999) explain further the results presented by Winter et al. (1996) and Day et al. (1993). Their effort to assess control mechanisms of balance during quiet stance and evaluate the influence of stance position on these mechanisms reveals the reason behind the controversial results of the aforementioned studies. First, the study was comprehensive with its method. They investigated the postural control mechanisms by using COP, EMG, and kinematic data for further analysis. Similar to the results presented by Day et al. (1993), hip angular motion is higher than ankle angular motion in side-by-side stance condition in the sagittal plane. However, they did not observe any corresponding linear hip motion, which would indicate a hip mechanism. Only the ankle angular motion is correlated with the COP excursions. In addition, the EMG data also supported these results.

Consequently, they conclude that in a side-by-side stance, the control of balance was under the activity of the ankle mechanism. In a narrow stance, on the other hand, the role of hip mechanism increased, and EMG activity of the lower leg muscles decreased. According to these data, they suggested that further narrowing the stance, as in tandem stance position, will lead the activity of hip increase so much that, the dominant control mechanism of the anteroposterior balance in the tandem position would be hip mechanism as suggested by Winter et al. (1996). When looked at the results in the frontal plane, they found the activity of both hip and ankle where the activity of the hip is higher as suggested by Day et al. (1993) and Winter et al. (1996). In the frontal plane, narrowing the support surface increases the role of the ankle mechanism and also increases the interaction between them. Authors suggested that this behavior reminds the intermediate stance position in the study of Winter et al. (1996), although one foot was not ahead of the other. This can be interpreted as the changes in the controlling mechanisms in the frontal plane is more sensitive to the changes in distance between feet rather than the alignment of the ankle joints. This conclusion is probably driven also by the researchers in the field since the studies on the effects of stance width carried out after Gatev focused only on the frontal plane. Bonnet, Cherraf, Szaffarczyk, and Rougier (2014) investigated only the mediolateral control of upright stance to analyze the effects of narrow and wide stances on the control mechanisms. They analyzed the effects of foot positioning on the control mechanisms in terms of the strength and the changes in the degree of active contribution of these mechanisms. According to the results presented, the strength of these mechanisms was higher in wide stance and lower in narrow stance compared with the standard stance. As the distance between feet increases, the mechanical contribution of the reaction forces under the feet increases as well. In terms of degrees of activity, both mechanisms were more active in narrow stance and less active in wide stance when compared to the standard stance. Since the area of the support surface increased during wide stance, the body needs less control over the balance. Looking at the results of the study for the dominant control mechanism in the mediolateral direction, shows that in accordance with the previous studies, the dominant control mechanism of the mediolateral plane is the body weight distribution mechanism (or the loading-unloading mechanism) in all stance conditions. The dominance was present in terms of both the degrees of active contribution of the mechanism and the strength of the mechanism. The contribution of the ankle mechanism decreased with the increasing stance width. However, as stated by other authors (Winter et al., 1993; Termoz et al., 2008), the contribution of the ankle mechanism was high enough to take into account. Although the contribution of the ankle mechanism is small compared to the loading-unloading mechanism, still mediolateral balance is controlled by both the ankle and the loading-unloading mechanisms. Therefore, this study showed not only the effects of stance positions on the postural control mechanisms it also revealed that the ankle mechanism has an important role in adjusting posture control in the challenging conditions in the frontal plane.

Complementary to this study, Bonnet (2014) conducted another study about frontal plane postural control to compare the differences in the degree of activity and strength of the postural control mechanisms in tandem, narrow and wide stances. The results were compliant with the previous studies on the tandem stance (Winter et al., 1996; Termoz et al., 2008). The dominant control mechanism of the tandem stance in the mediolateral direction was found to be the ankle mechanism. The narrow and wide stances were also compliant with the previous studies (Winter et al., 1993; Lafond et al., 2004; Termoz et al., 2008; Bonnet et al. 2014), as the controller of mediolateral balance was the loading-unloading mechanism. As in previous studies, the contribution of the ankle mechanism to the mediolateral control of posture increased with the decreasing distance between feet (from wide to tandem). However, surprisingly, the degree of active contribution of the loading-unloading mechanism did not decrease with the decrease in the support surface area in the mediolateral plane. Instead, the contribution of the hip mechanism decreased from wide stance to narrow stance but was higher than expected in tandem stance. Besides, the contribution of the ankle mechanism was also high during the tandem stance. This result contradicts with the previous studies which suggested a dominant ankle control in the mediolateral direction for tandem stance (Winter et al., 1996; Termoz et al., 2008). Nonetheless, given the controversial results as well as the small number of studies, it can be said that the tandem stance position should be further investigated to conclude on an exact control mechanism.

Another aspect of the positioning of the feet is the angle between the feet. Since different activities require the internal or external rotation of the feet, analyzing the effects of the angle between the inner borders of the feet to the contribution of the postural control mechanisms during quiet stance is important to reveal necessary adjustments in the mechanisms to improve balance in different activities or disorders. With this aim, Rougier (2008) hypothesized that how much forefeet are apart is an influencing factor on the contribution of postural control mechanisms. The study showed that contribution of the ankle mechanism in the frontal plane increases as the forefeet are spread apart from -30º to 120º. However, this effect was distinctive only in extreme conditions when all angles were compared. It has to be noted that although the contribution of the ankle mechanism increased with the increasing angle between feet, it was always under the 0.5 threshold in terms of contribution.

On the other hand, changes in the angle between feet did not result in any significant changes in the contributions of these mechanisms in the sagittal plane. In terms of dominant control mechanisms, this study showed again that the dominant postural control mechanism in the anteroposterior direction is the ankle mechanism and the dominant control mechanism in the mediolateral balance is the hip mechanism. Although the contribution of ankle increases when feet are spread apart, it was never higher than or equal to the contribution of the hip. On the other hand, the angle between feet can be interpreted as not crucial during balance tests unless an extreme external or internal rotation is present.

As presented so far, the studies which tried to investigate the effects of foot position focused on many aspects of the stance such as stance width, the angle between feet or distance between feet in both frontal and horizontal axes as in an intermediate stance position. These studies provide a very comprehensive basis to acknowledge the postural control mechanisms in those different stance conditions. However, the methodology used by the researchers in these studies is in lack of standardization. When some aspect of the stance position was investigated, the other aspects were not controlled in most of the studies. For example, Winter et al. (1996) did not specify any particular distance between feet during their intermediate stance position. Bonnet et al. (2014), on the other hand, let participants choose the distance between feet when investigating the effects of stance width. Termoz et al. (2008) did control the position the participant preferred for the sake of the data collected from the same participant, yet they did not specify any particular positions for all participants. On the other hand, Rougier (2008) defined specific positions for the feet placement and prevented the participants from adopting a more comfortable position for themselves, which may affect the data collected. Although the studies presented too much information on the effects of foot positioning on the postural control mechanisms, it is hard to claim that the results are standard, due to the different approaches used during data collection.

Two approaches can be used during data collection. One of them is to let the participants choose the most comfortable and natural stance position to prevent any effect resulting from the artificial condition created. However, it is known that stance position affects the contribution of the postural control mechanisms as shown by the presented studies so far. Letting the participant adapt a spontaneous position can result in the contribution of the ankle or loading-unloading mechanism in an undesired way. For instance, the results presented by Winter et al. (1996) have wide between-subject variability, especially in the intermediate position in which the authors let the participant decide the exact position taken.

The second approach is to make the participant adapt a predefined position. This time, there is a concern that the subject might be in an uncomfortable situation and may give an unnatural postural response or try to intervene in the automatic processes while trying to adapt to that particular position. However, increasing the number or duration of trials may give researchers a chance to let the participant adapt and acknowledge that particular predefined position. Altogether, a predefined position might be advantageous while working on balance and posture.

Finally, in light of the examined and presented literature, it can be said that the number of studies investigated the effects of different stance positions on postural control mechanisms are not much, and there exist controversial results in some particular positions such as tandem stance or intermediate stance positions. Also, some studies investigated the effects of foot positioning by defining a specific position before trial and some by letting the participants decide. The controversial results may be due to the methodological approach. All things considered, it can be said that more studies must be carried out to investigate the effects of stance position on the postural control mechanisms.

CHAPTER 3

METHOD

The purpose of the study is to determine the contribution of the ankle and loading-unloading mechanisms during quiet stance in different foot positions. In this chapter, research design, experimental setup, data collection process and finally analysis of the data are presented.

3.1. Design of the Study

In this study, human static balance was measured. In order to do that, data was collected from healthy adult subjects. For this purpose, approval of Human Researches Ethics Committee of Middle East Technical University was obtained. All participants were informed about the experimental procedure.

To fulfill the purpose of the study, healthy individuals with a normal balance system were required. Participants were selected according to their age and medical history. Since the purpose is to analyze the postural control mechanisms of quiet stance as in daily life, no training was needed before trials. Data were collected only once, and no consecutive recordings were needed since no additional, therapeutic applications were carried out during trials, which can cause short-term or long-term changes in the postural control mechanisms of subjects. The study was not blinded, but for the sake of data, people were only informed about the purpose of the study, but the details which may create an intentional intervention to the postural control were not given. All subjects were given the same instructions, and data were collected at the same conditions by the same researcher to eliminate the inter-experimenter validity.

All experiments were performed at the Biomechanics Laboratory of Middle East Technical University (METU), located in the Department of Mechanical Engineering, in one week.

3.2. Participants of the Study

The experiments were conducted on 22 healthy adults. Age range of the participants varies between 22-28 years. All subjects voluntarily participated, and written consents were obtained after informing about the purpose of the study before the experiment.

Subjects were included to study according to the following criteria: Being between ages 20 and 49, not having any neuromuscular condition or history, not having any type of physical disability, not having any ear infection/disorder history, having a healthy vestibular system, not having any pain that may be an obstacle for the experiment, having enough cognitive function to understand and obey commands. Subjects were not included under the conditions of alcohol or sedative use prior to the test within 48 hours. Anamnesis, demographic information and dominant upper and lower extremities were recorded.

3.3. Experimental Setup

The experiments were performed at the Biomechanics Laboratory of METU. Laboratory equipment was used to create the experimental setup. METU Biomechanics Laboratory holds a motion and gait analysis system named Kinematic Support System (KISS). The experiment was planned to proceed under quiet stance conditions. Two force plates and two amplifiers of KISS were used within the experimental setup. For data acquisition, a 16 channel multifunction I/O Device (USB-6212, National Instruments, Austin, TX, USA) was used to acquire data from force plates. The two strain gauge based force plates of KISS (40x60cm, type 4060, Bertec Corporation, Columbus, OH, USA) are positioned and embedded to the ground in a way that corresponds a regular step length of a gait cycle. Force plates were able to make measurements of force and moments from three cardinal axes. Two 6 channel amplifiers were used to amplify the voltage output obtained from the two force plates. Data acquisition device was used to convert analog signals to digital data. NI DAQmx (version 9.7.5, National Instruments, Austin, TX, USA) and LabVIEW (2013, National Instruments, Austin, TX, USA) software packages were used for data acquisition. Acquired digital signals were processed by LabVIEW. Calculations of contributions of postural control mechanisms in addition to the calculations of center of pressure signals and reaction forces were also carried out in LabVIEW software.

Figure 3.1. Placement of the embedded force plates (Çelik, 2008)

3.4. Experiment Procedure

Experiments are conducted in one session in Biomechanics Laboratory. One set of trial session was performed for each participant. A session consisted of 9 different conditions. Each condition was applied once. A total of 9 trials were conducted for each subject. Trials were performed for 60 seconds for each condition. The rest period was 30 seconds between the conditions. Participants were informed before the experiment to wear comfortable clothes. Participants were not allowed to wear any footwear during the experiments, to avoid any effect of footwear on postural stability.

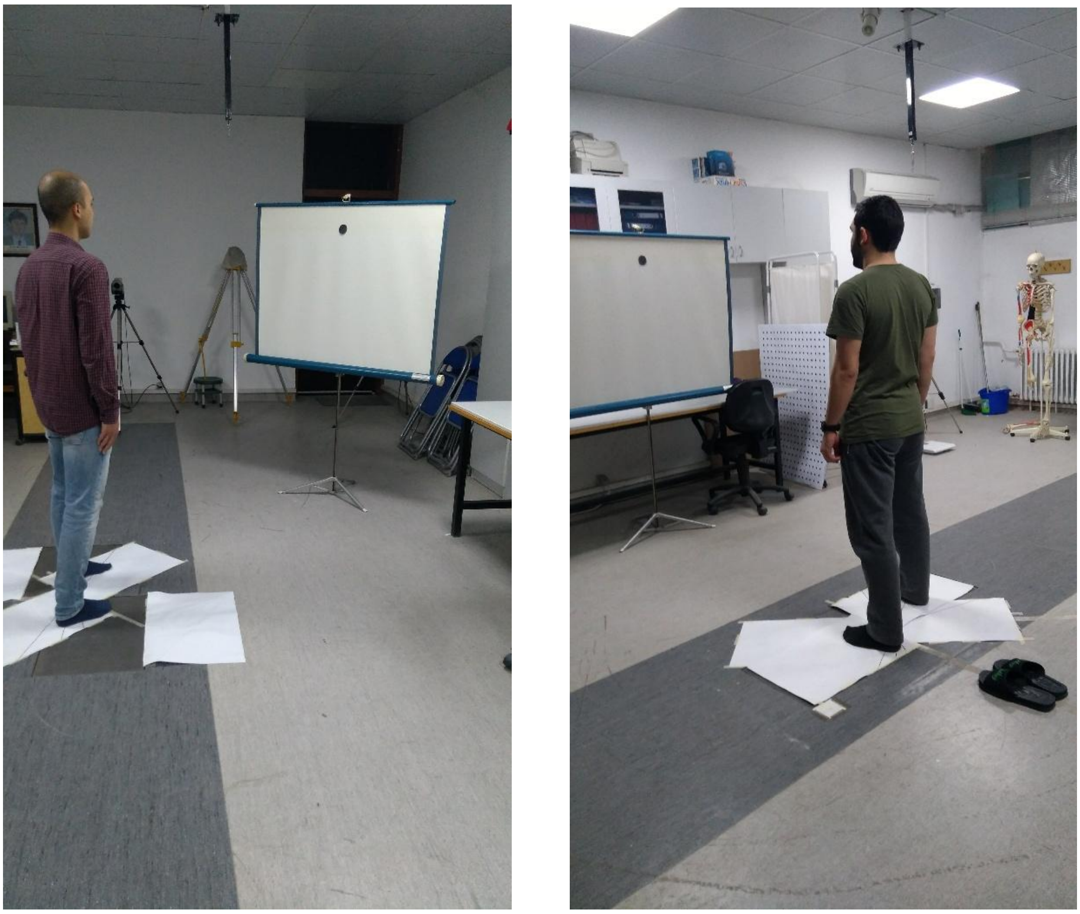

Figure 3.2. Subjects on force plate for IL and IR stance conditions, respectively

A session consisted of the following conditions in the following order:

1. Side-by-side stance : 0º foot position
2. 30R : 30º foot position, right foot ahead
3. 45R : 45º foot position, right foot ahead
4. 60R : 60º foot position, right foot ahead
5. TR : Tandem stance or 90º foot position, right foot ahead
6. 30L : 30º foot position, left foot ahead
7. 45L : 45º foot position, left foot ahead
8. 60L : 60º foot position, left foot ahead
9. TL : Tandem stance or 90º foot position, left foot ahead

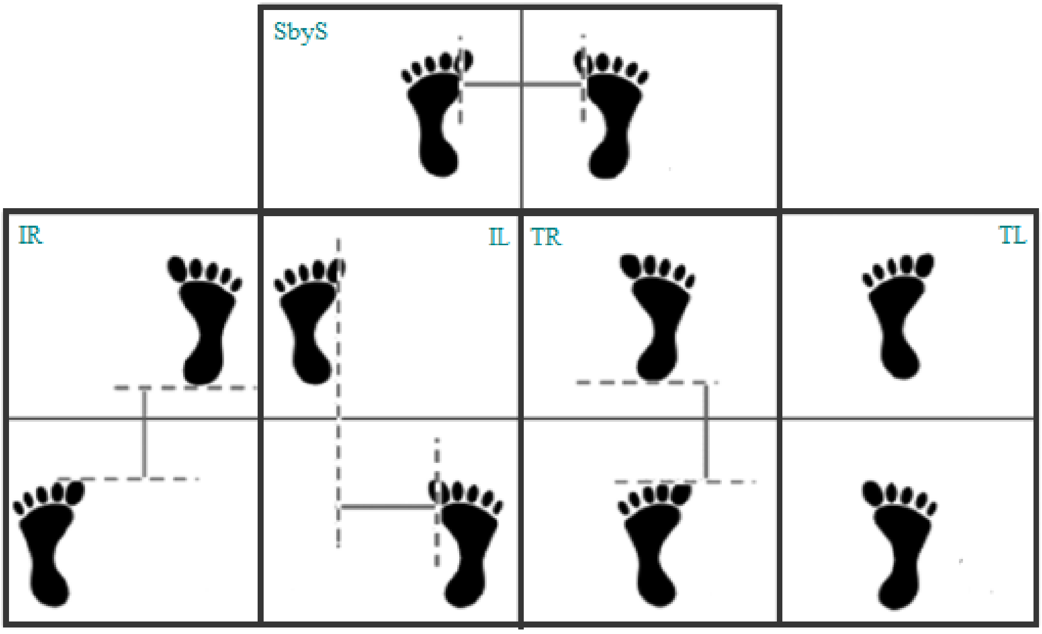

Figure 3.3. Placement of the feet. Abbreviations: SbyS, side-by-side stance; IR, intermediate stance (30°, 45°, 60° stances)left foot ahead; IL, intermediate stance right foot ahead; TR, tandem stance right foot ahead; TL, tandem stance left foot ahead

The listed nine stance conditions were adopted by the participants according to the predetermined reference points and lines. While determining foot positions for data collection, there were two things to be considered. The first is that using predetermined, constrained positions for the feet may be outside of the subject's preferred, natural posture. Due to this reason, letting the participant adopt a comfortable foot position is encouraged in many studies (Winter et al., 1993; Winter et al., 1996; Bonnet, Cherraf, Szaffarczyk & Rougier, 2014). The second issue is that this method may increase inter-subject variability and may even cause biased results since subjects tend to adopt the posture in which they perform the best (McIlroy & Maki, 1997). Thus in this study, predetermined foot positions were used to prevent both inter-subject variability and the voluntary intervention of individuals to the control of upright stance. The applied placement methods for all conditions such as distance between feet, internal or external rotation angle were all selected according to the methods used in the literature and adapted to this study. Also, the positioning methods which most individuals prefer (McIlroy & Maki, 1997) were chosen.

According to the literature, the changing distance between feet does not affect postural control in the AP direction and effects the ML postural control under only narrow stance conditions (Kirby et al., 1987). For stance widths larger than 15 cm there is not any significant difference between COP changes in the ML direction. In addition, an average of 17 cm distance between feet is preferred by most individuals during side-by-side stance (McIlroy & Maki, 1997). Thus, the stance width for side-by-side stance was predetermined as 17 cm distance between medial malleoli for all participants.

As stated by Kirby et al. (1987) in stance positions when one foot is farther forward than the other, significant changes are observed when the stance width is the largest, and these differences are not only observed in the ML direction which is already affected by the stance width, also observed in the AP direction. The postural control is most affected by 30 cm distance between feet, which was the widest stance condition in the mentioned study. Thus for intermediate stance conditions, taking into account that the force plates are embedded, the distance between feet was decided to be as close as possible to 30 cm. In this regard, for 30º stance the distance between feet was 35 cm, for 45º stance it was 25 cm and for 60º stance, since the aim was to imitate a similar condition to tandem stance, it was selected as 15 cm. In intermediate stance positions, the forward foot was required to be entirely ahead of the backward feet (Bonnet, Cherraf & Do, 2014).

For tandem stance, the distance between the calcaneus and second toe was set to 2.5 cm. The only condition needed to be satisfied here was to set feet ahead of each other for tandem right, and tandem left conditions. However, a 2.5 cm offset was set for the measurement concerns since two side-by-side force platforms were used.

As stated by both Kirby et al. (1987) and Rougier (2008), the angle between feet, i.e., internal or external rotation of feet, does not affect the postural control except for extreme conditions of toeing-in or toeing-out. Thus, all of the measurements were recorded at a talotarsal neutral position for all subjects where the line passing through calcaneus and second toe makes a 0° angle with the frontal axis, and bisection line of the lower leg coincides with the vertical axis.

Since force plates were embedded to the ground in such a position suitable for gait analysis, only the coinciding portion (24 x 40 cm) of the plates were appropriate to use in side-by-side and tandem stance conditions. However this was limiting the foot lengt of the participants. In order to prevent this and have a bigger coinciding area, coordinate axes of the force plates were rotated 45° to use them diagonally. This approach was also used for 30° and 45° stance conditions but this time to have more distance between feet in the AP direction. For tandem and 60° stance positions, that kind of an adjustment was not necessary. For tandem stance and all intermediate conditions, force plates were used as in the gait analysis experiments, i.e., AP direction was set to the direction of the length of the force plates. All of the stance positions were drawn and printed on papers which then covered the surface of the two force plates to determine the correct positions easily. The stance positions were also assessed with an eye examination.

There was a poster on the wall which participants were looking at with a black dot in the approximate eye height of the subject, displayed in Figure x. Participants were told to look at the dot during the experiment. The aim was to isolate the subject from the environment during the experiment to prevent distraction due to several objects in the laboratory. Another aim was to create an external focus of attention. As stated in many studies (Wulf, Shea, Park, 2001; Rotem-Lehrer, Laufer, 2007; Chiviacowsky, Wulf, Wally, 2010) external focus of attention prevents people from intervening with the automatic postural control mechanisms and improves performance during balancing and stability activities.

Participants stood steady on the force plates with eyes open and arms by the side. Each foot was placed in separate force plates. Participants were positioned on the force plate for each condition according to the session plan. Instructions were given to stand as still as possible while focusing on the black dot in front of them. For all trials, the same instructions were given, and no additional information or help was provided.

3.5. Data Acquisition and Analysis

Data were acquired in 50 Hz frequency. All data acquisition and calculation were made via Virtual Instruments (VI) created on LabVIEW. Virtual Instrument is the name of the code written in the G programming language of LabVIEW. Acquired data were separated into ankle and loading-unloading mechanism signals with the method provided in section 3.7. The strength of signals and cross-correlations between signals were also calculated in LabVIEW. Statistical analysis was carried out with SPSS (International Business Machines (IBM) Corporation, Armonk, NY, USA).

3.5.1. Data Acquisition in LABVIEW

Four VIs were created in LabVIEW to acquire digital signals and process the voltage signals to obtain force and center of pressure signals. Four different VIs were necessary since for some conditions coordinate axis were rotated as describedin section 3.4. According to this, three different VIs were created for data collection and one for data reading and calculations. In data collection VIs, 12 channels were created. Signals from these channels were amplified by multiplying with the appropriate gains. Amplified signals from each force plate were separately inserted into two separate matrices. Calibration matrix, C, of each force plate then multiplied with the new created signal matrix, S, of each force plate to obtain the matrix of forces and moments (6x1 matrix with the following elements, F_x, F_y, F_z, M_x, M_y, M_z). Matrix of force and moments, then separated to calculate the position of the center in x and y coordinate axes or in other words position of the center of pressure according to the undermentioned formulas where l and r in the formulas represents left and right feet respectively :

$$COP_{xl}(t) = \frac{-M_{yl}(t)}{F_{zl}(t)} \tag{3.1}$$

$$COP_{xr}(t) = \frac{-M_{yr}(t)}{F_{zr}(t)} \tag{3.2}$$

$$COP_{yl}(t) = \frac{M_{xl}(t)}{F_{zl}(t)} \tag{3.3}$$

$$COP_{yr}(t) = \frac{M_{xr}(t)}{F_{zr}(t)} \tag{3.4}$$

Coordinate axes of force plates were changed according to the trial condition during this step for each condition. Thus, three separate VIs with different coordinate axes were created ; one for side-by-side stance and 30L and 45L sub-conditions, one for tandem together with 60L sub-condition and the last one for 30R, 45R and 60R sub-conditions. A shift register loop was created for each force plate to display the change of center of pressure location in x and y axes for each foot, where x axis corresponds to frontal axis and y corresponds to the sagıttal axis of body. To achieve the continuous sampling, all code was implemented into a while loop. The loop was conditioned to run for 60 seconds unless it is stopped.

3.6. Methods for Calculation of Ankle and Loading-Unloading Contributions

As stated previously, Winter et al. (1993) identified the two postural control mechanisms, which are an ankle COP location mechanism and a bodyweight distribution mechanism, which is also known as the loading-unloading mechanism. These mechanisms are introduced according to some mathematical models. Before getting into the details of this model, it is important to understand the calculation of the COP from force platforms. By using only one force platform, the resultant center of pressure during bilateral stance can be calculated as follows :

$$COP_x(t) = \frac{-M_y(t)+F_x(t)*Z_0}{F_z(t)} + X_0 \tag{3.5}$$

$$COP_y(t) = \frac{M_x(t)+F_y(t)*Z_0}{F_z(t)} + Y_0 \tag{3.6}$$

In these equations M is the moment, F is the reaction force, x is the ML direction, y is the AP direction and z is the vertical direction. X_0, Y_0 and Z_0 are the offsets from the geometric center of the force platform.

As we know, the net COP measured by a force platform during quiet stance is actually a weighted sum of the net COPs under each foot. When we measure the COPs under each foot separately by using two force platforms we can clearly see the contribution of each limb to the total COP. In that situation COP_{net} can be calculated by using the following formula for both AP and ML directions :

$$COP_{net}(t) = COP_l(t) * \frac{F_{zl}(t)}{F_{zl}(t)+F_{zr}(t)} + COP_r(t) * \frac{F_{zr}(t)}{F_{zr}(t)+F_{zl}(t)} \tag{3.7}$$

$COP_l(t)$ and $COP_r(t)$ are the time-varying positions of COPs under the left and right foot, respectively. $F_{zl}(t)$ and $F_{zr}(t)$ are the vertical reaction forces under the left and right foot, respectively. Not to mention, $COP_l(t)$ and $COP_r(t)$ from each force plate must be calculated by equations (3.1) and (3.2) for ML direction and (3.3) and (3.4) for AP direction.

Since $F_{zl}(t)$ and $F_{zr}(t)$ are the loadings under each foot expressed as time-varying functions, the sum of $F_{zl}(t)$ and $F_{zr}(t)$ corresponds to the total body weight. Thus, $\frac{F_{zl}(t)}{F_{zl}(t)+F_{zr}(t)}$ and $\frac{F_{zr}(t)}{F_{zr}(t)+F_{zl}(t)}$ are the time-varying relative loads under each foot and their sum is equal to 1. When weight is equally distributed between legs, these ratios should be equal to 0.5.

$COP_{net}(t)$ depends on four time-varying variables, which can be seen from equation (3.7). As stated previously, during single limb stance $COP_l(t)$ and $COP_r(t)$ are totally under ankle control. In double limb stance, changes in the COP_{net} as a result of $COP_l(t)$ and $COP_r(t)$ are due to the ankle muscle activities. COP_l and COP_r changes in the ML direction are the results of the muscle activities of ankle evertors and invertors while changes in the AP direction are the result of the plantar flexor and dorsiflexor muscle activities. The muscle groups responsible for the loading-unloading of the limbs are hip muscles, which was also mentioned in the previous chapter. Thus, relative changes in the load distributions $\frac{F_z l(t)}{F_{zl}(t)+F_{zr}(t)}$ and $\frac{F_{zr}(t)}{F_{zr}(t)+F_{zl}(t)}$ are the result of the loading-unloading of each limb.

After this point, researchers suggest different methods to determine the contribution of each mechanism to the net COP. To understand what affects this weighted sum in equation (3.7) more ; i.e., the position of COPs or relative loads carried by each foot, Winter et al. (1993) made an assumption. If we assume that a healthy person carries the bodyweight equally in both limbs, then the ratios $\frac{F_z l(t)}{F_{zl}(t)+F_{zr}(t)}$ and $\frac{F_{zr}(t)}{F_{zr}(t)+F_{zl}(t)}$ both must be equal.

$$\frac{F_z l(t)}{F_{zl}(t)+F_{zr}(t)} = \frac{F_{zr}(t)}{F_{zr}(t)+F_{zl}(t)} = 0.5 \tag{3.8}$$

This will modify equation (3.7) as follows :

$$COP_{net}(t) = \ 0.5 * COP_l(t) + 0.5 * COP_r(t) \tag{3.9}$$

This means there is not a loading-unloading response, and the total change in the net COP is completely depends on the changes in COPl and COPr, which are controlled by the ankle muscles. This COP displacement explained by the COP location mechanism is then denoted as COP_c where c stands for changes :

$$COP_c(t) = 0.5 * COP_l(t) + 0.5 * COP_r(t) \quad (3.10)$$

Now, after subtracting the contribution of ankle mechanism, i.e., COP location effect denoted as COP_c from the net COP, we have left with the loading-unloading contribution due to the vertical reaction forces. The contribution of loading-unloading mechanism is denoted as COP_V where v stands for vertical.

$$COP_v(t) = COP_{net}(t) - COP_c(t) \quad (3.11)$$

After obtaining the signals for the contribution of ankle mechanism and loading-unloading mechanism, they computed the root mean squares of these signals to determine the average contribution of these mechanisms over the time series. They also carried a cross-correlation analysis between COP_c and COP_V to measure the dependence or independence of these two mechanisms.

The presented method here applies only to healthy people who carry an equal load under both limbs. However, even healthy individuals do not carry the bodyweight equally with each limb due to imbalances resulting from, for instance, dominant extremity use. Thus, this method is by design not applicable in experiments with unhealthy subjects. It is thought that this method is also not adequate to use in healthy subjects due to the assumption of equal load distribution. Because in different stance conditions other than side-by-side stance, load dıstribution between limbs are actually not equal. Also, Winter et al. (1996) must have realized this ; they presented a more reasonable method for the extraction of COP_c and COP_V from equation (3.7).

This time they do not assume the weight carried by each limb is equal. Instead, they modify the equation (3.10) according to the average loads carried by each foot among the time-series :

$$COP_c(t) = \bar{V}_l * COP_l(t) + \bar{V}_r * COP_r(t) \tag{3.12}$$

According to equation (3.12), the contribution of the COP location mechanism still depends on the time-varying COP coordinates under the left and right feet, but this time, there is not an assumption such as a healthy person carries an equal load with each limb. With this modification, a more precise calculation of both mechanisms is possible since even a weight distribution difference between limbs in an amount of 0.02 can be taken into account.

Still, this method applies the same logic with the previous one. It eliminates one of the mechanisms and then calculates it by subtracting the other one from the resultant COP. In this method, the load carried by each limb during the time series is still kept constant. By doing that, the contribution of the COP location mechanism when there is no change in the load distribution among time-series can be calculated. No change in the load distribution means there is no effort spent to control net COP by loading-unloading mechanism, i.e., the complete change in the net COP reflects the contribution of COP_l and COP_r or ankle COP location mechanism. In this situation, we count the fact that load may not be carried equally, still the total of average load carried by each foot gives us the %100 total body weight:

$$\bar{V}_l + \bar{V}_r = 1 \tag{3.13}$$

V_l and V_r are the average vertical loads on the tracked time-series, carried by left and right foot respectively.

After calculating the pure ankle mechanism contribution, to find the contribution of the loading-unloading mechanism, the same method applied as Winter et al. (1993). COP_c is subtracted from net COP to obtain and COP_v as in equation (3.11). Following the separation of ankle and loading-unloading signals, to determine how well they are contributing to net COP, Winter et al. (1996) made a cross-correlation analysis between $COP_c(t)$ and $COP_{net}(t)$ and $COP_v(t)$ and $COP_{net}(t)$ in addition to the root mean square calculation. Also, they carried out a cross-correlation analysis between $COP_c(t)$and $COP_v(t)$ to measure if these two mechanisms are working dependent or not.

A decade later, Rougier suggests a different method to calculate the contribution of the loading-unloading mechanism. In the suggested method, same principle is applied to calculate COP_c, as Winter et al. (1996) did. Then the effects of the loading-unloading mechanism is neutralised to calculate the contribution of the ankle mechanism, while keeping the load carried by each foot constant and equal to the average load. However, the very same approach is applied also to calculate the contribution of the loading-unloading mechanism. The difference between the method used by Winter et al. (1993,1996) and Rougier (2007) is the latter did not calculate the contribution of COP_v by subtracting the contribution of COP_c from the net COP. Instead, the second method neutralizes the effect of the ankle mechanism as well and calculates the pure contribution of the loading-unloading mechanism. In other words, this method calculates the average load carried by each foot among the time-series, but also calculates the mean position of the COP among the time series for both feet and uses them as coefficients in equation (3.7) to calculate COP_v. By doing that, COP_c can be calculated with equation (3.12) and COPv can be calculated as follows :

$$COP_v(t) = \overline{COP_l(t)} * \frac{F_{zl}(t)}{F_{zl}(t)+F_{zr}(t)} + \overline{COP_r(t)} * \frac{F_{zr}(t)}{F_{zr}(t)+F_{zl}(t)} \tag{3.14}$$

Where COP_l and COP_r are the mean positions of COPs under each foot in all of the time-series. With this method, Rougier avoided calculating the COP_{net} for each time interval in a trial. By calculating COP_c and COP_v separately, one might see that the sum of $COP_c(t)$ and $COP_v(t)$ are not simply equal to COP_{net} as in equation (3.11), since both mechanisms are calculated according to the averages of one another. This time, the question of how one can quantify the contribution of these mechanisms to total potural control arise. Instead of calculating root mean squares, they proposed that standard deviations (SD) of the computed $COP_c(t)$ and $COP_v(t)$ signals can be used to assess the respective contribution of these mechanisms according to the following formulas :

$$Contr_c = \frac{SD(COP_c)}{SD(COP_c)+SD(COP_v)} \tag{3.15}$$

$$Contr_v = \frac{SD(COP_v)}{SD(COP_c)+SD(COP_v)} \tag{3.16}$$

In these formulas, $Contr_c$ and $Contr_v$ correspond to the contribution of ankle and the contribution of the loading-unloading mechanism, respectively. With this method, $Contr_c$+$Contr_v$ is equal to 1.

$$Contr_c + Contr_c = 1 \tag{3.17}$$

Consequently, the dominant strategy can be determined by looking only at the results of one of these variables. In other words, if one of these contribution coefficients say $Contr_c$ is under 0.5 threshold, this implies $Contr_v$ is above this threshold from equation (3.17) and the loading-unloading mechanism is contributing more.

With this method, not only individual differences in load distribution has taken into account, but also the mean COP locations of each person is also considered. This is important because some people may tend to lean forward due to abnormal postures adopted. For example, an increased anterior pelvic tilt or a stooped posture can lead to a change in the mean position of COP and an anteriorly located COP can be normal for those subjects. Thus, instead of eliminating only the mean load shared by each limb, it is essential to eliminate the regular adapted COP for the sake of data. This also prevents another assumption such that COP is always located at the center for all healthy subjects. Hence, this method is more credible than its predecessors.

After Rougier (2007) presents an updated version of the method suggested by Winter et al. (1996) ; Bonnet, Cherraf, Szaffarczyk, Rougier (2014) suggested further improvements in terms of analysis of the contributions of these mechanisms. They used the same method introduced by Rougier (2007) for the separation of the contributions of each mechanism. However, they interpreted these contributions in two ways. They suggested that the correlation of $COP_c(t)$ or $COP_v(t)$ with COP_{net} indicate the degree of active contribution of the mechanism in question. If one curve was explaining the other ; for instance if the correlation of COP_c and COP_{net} is significantly higher, this is interpreted as ankle mechanism is more involved in the control of net COP. On the other hand, the contribution coefficients $Contr_c$ and $Contr_v$ is interpreted as the amplitude contribution of these mechanisms. In the case of $Contr_c$ being higher than $Contr_v$, ankle mechanism is interpreted as having more amplitude contribution, in addition to being more involved. This is also important in terms of analysis. If both of the mechanisms, highly correlates with COP_{net}, and if there is no significant difference between correlations, the contribution coefficients, i.e., the strength of the mechanisms helps one to differentiate which one is dominant.

In light of the proposed methods in the literature, this study used the method proposed by Rougier (2007) to calculate the contributions of ankle mechanism and loading-unloading mechanism since this method reckons postural variations between subjects.

Another alternative here is to use factor analysis. However, due to the fact that this study is collecting data for different intermediate stance positions for the first time in the literature to the best of author's knowledge, it is decided to follow the literature on selected method in order to compare the obtained results of this study with the literature. Because even healthy subjects may have some postural anomalies, it is more realistic taking both load distribution and COP location differences between individuals into account and calculating the contributions of each mechanism after eliminating the effects of postural variances for more correct results. In the comparison and analysis of the contributions obtained after the calculations, the perspective of Bonnet, Cherraf, Szaffarczyk, Rougier (2014) was found applicable and used in this study.

3.7. Procedure for Calculation of Ankle and Loading-Unloading Contributions

In order to obtain the contribution of ankle mechanism and loading-unloading mechanism separately, the following procedures were applied according to the method introduced by Rougier (2007), to the collected COP signals. The analysis was carried out for frontal and sagittal planes separately. The mediolateral direction is defined as the x-axis, and anteroposterior direction is defined as the y-axis. The following procedures were carried out in the same manner for all of the proposed stance conditions.

$COP_x(t)$ and $COP_y(t)$ were calculated for both force platforms under the left and right feet, in LABVIEW according to the following formulas, which are modified versions of equations (3.5) and (3.6) :

$$COP_{xl}(t) = \frac{-M_{yl}(t) - F_{xl}(t) * h}{F_{zl}(t)} \tag{3.18}$$

$$COP_{xr}(t) = \frac{-M_{yr}(t) - F_{xr}(t) * h}{F_{zr}(t)} \tag{3.19}$$

$$COP_{yl}(t) = \frac{M_{xl}(t) - F_{yl}(t) * h}{F_{zl}(t)} \tag{3.20}$$

$$COP_{yr}(t) = \frac{M_{xr}(t) - F_{yr}(t) * h}{F_{zr}(t)} \tag{3.21}$$

Since there is no covering on the force platforms in the vertical direction, h is taken as zero and the given equations were modified as follows :

$$COP_{xl}(t) = \frac{-M_{yl}(t)}{F_{zl}(t)} \tag{3.1}$$

$$COP_{xr}(t) = \frac{-M_{yr}(t)}{F_{zr}(t)} \tag{3.2}$$

$$COP_{yl}(t) = \frac{M_{xl}(t)}{F_{zl}(t)} \tag{3.3}$$

$$COP_{yr}(t) = \frac{M_{xr}(t)}{F_{zr}(t)} \tag{3.4}$$

COP_{xl} and COP_{xr} are the COP changes in the mediolateral direction under the left and right feet, respectively. Similarly, COP_{yl} and COP_{yr} are the COP changes in the anteroposterior direction under each foot. F_{xl}, F_{yl}, F_{zl}, M_{xl}, and M_{yl} are the acquired force and moment signals in mediolateral, anteroposterior, and vertical directions measured from the left foot. Equivalently F_{xr}, F_{yr}, F_{zr}, M_{xr}, and M_{yr} are the acquired force and moment signals of the force plate under the right foot.

After COP_x and COP_y for right and left feet were obtained, COP_{xl} and COP_{xr} signals were used to calculate the COP_{net} in the mediolateral direction, and COP_{yl} and COP_{yr} signals were used to calculate the COP_{net} in the anteroposterior direction by using equation (3.7).

$$COP_{xnet}(t) = COP_{xl}(t) * \frac{F_{zl}(t)}{F_{zl}(t) + F_{zr}(t)} + COP_{xr}(t) * \frac{F_{zr}(t)}{F_{zr}(t) + F_{zl}(t)} \tag{3.22}$$

$$COP_{ynet}(t) = COP_{yl}(t) * \frac{F_{zl}(t)}{F_{zl}(t) + F_{zr}(t)} + COP_{yr}(t) * \frac{F_{zr}(t)}{F_{zr}(t) + F_{zl}(t)} \tag{3.23}$$

In this equation, COP_{xnet} and COP_{ynet} resemble the resultant COPs in the mediolateral and anteroposterior directions, respectively.

According to the method used, neither contribution of the ankle mechanism nor the contribution of the loading-unloading mechanism is computed directly by extraction from the COP_{net} in x and y directions. Instead, two resultant theoretical COPs were calculated for each mechanism by eliminating the contribution of the other one. In this approach, COP_c or contribution of the ankle mechanism is thought to be the COP changes left after the elimination of the changes result from the loading-unloading mechanism. To do so, the mean load distribution of participants was calculated and implemented in equations (3.22) and (3.23) for both ML and AP directions to obtain the COP_c signal as follows, where $COP_{xc}(t)$ and $COP_{yc}(t)$ corresponds to the signals resulting from the ankle mechanism in ML and AP directions, respectively.

$$COP_{xc}(t) = \overline{V}_l * COP_{xl}(t) + \overline{V}_r * COP_{xr}(t) \tag{3.24}$$

$$COP_{yc}(t) = \overline{V}_l * COP_{yl}(t) + \overline{V}_r * COP_{yr}(t) \tag{3.25}$$

V_l and V_r are the average loads carried through the time series by left and right feet, respectively.

At the same time, COP_v was also calculated with the same approach, i.e., by eliminating this time, the changes in the COP_{net} which resulted from the ankle mechanism. In order to do that, the average COP location adopted by the subjects were calculated throughout the time series. Then, calculated averages were implemented in equations (3.22) and (3.23) to calculate COP_v in both ML and AP directions. The effects of the adopted forward/backward lean posture of participants were not taken into account by using the average COP locations of each participant as fixed coefficients. The changes in COP_{net} resulting from the ankle mechanism were eliminated to see the pure loading-unloading mechanism. According to this approach, COP_v for both ML and AP directions is calculated with the following formulas, where COP_l and COP_r are the average locations of the COPs under the left and right feet, respectively.

$$COP_{xv}(t) = \overline{COP_{xl}(t)} * \frac{F_{zl}(t)}{F_{zl}(t)+F_{zr}(t)} + \overline{COP_{xr}(t)} * \frac{F_{zr}(t)}{F_{zr}(t)+F_{zl}(t)} \tag{3.26}$$

$$COP_{yv}(t) = \overline{COP_{yl}(t)} * \frac{F_{zl}(t)}{F_{zl}(t)+F_{zr}(t)} + \overline{COP_{yr}(t)} * \frac{F_{zr}(t)}{F_{zr}(t)+F_{zl}(t)} \tag{3.27}$$

After COP_{net}, COP_c and COP_v signals were all calculated for both ML and AP axes, the amplitude contribution of ankle and loading-unloading mechanisms must be calculated. In order to do that, contribution coefficients were calculated. First, standard deviations of the obtained COP_c and COP_v signals for both ML and AP axis were calculated in LabVIEW. Then the contribution coefficients of these mechanisms were calculated according to the following formulas where the sum of contributions of the COP location mechanism and loading-unloading mechanism equals 1. Since the calculation of COP_c and COP_v signals were based on the elimination of the other factors, i.e., the substitution of one of the terms with its average, the resultant COP is not the sum of both signals as presented by Winter et al. (1993, 1996). With the amplitude contribution calculation, the total contribution of these mechanisms was still yielding the total change or effect seen in the COP_{net}.

$$Contr_c = \frac{SD(COP_c)}{SD(COP_c)+SD(COP_v)} \tag{3.15}$$

$$Contr_v = \frac{SD(COP_v)}{SD(COP_c)+SD(COP_v)} \tag{3.16}$$

$$Contr_c + Contr_c = 1 \tag{3.17}$$

In these equations, $Contr_C$ and $Contr_V$ correspond to the amplitude contributions of ankle mechanism and loading-unloading mechanism, respectively.

After the calculation of both signals and contribution coefficients, some statistical analysis was carried out to analyze the data collected.

3.8. Statistical Analysis

To compare the cross-correlations and contribution coefficients among all stance conditions, Friedman Test is used. Friedman is a non-parametric test used for data collected from one group for more than three different occasions. Wilcoxon Signed-Rank Test was used for posthoc analysis to reveal which stance conditions differed from each other. To clarify if there is a difference between left foot forward and right foot forward of the same condition, again Wilcoxon Signed-Rank Test is used. Wilcoxon Signed-Rank Test is a non-parametric test to compare two related, repeated or matched data collected from one group. In order to identify the effect of gender and dominant extremity, Mann Whitney U test is used. Mann Whitney U test is the equivalent of the Wilcoxon Signed-Rank Test for between-subject comparisons. It is used to compare one measurement for two populations. A significance level of 0.05 was used for all tests. All analysis was performed with IBM SPSS Statistics 24.

CHAPTER 4

RESULTS

In this chapter, first, all the stance conditions are investigated separately to determine the contribution of each mechanism (ankle and hip mechanisms) to the control of upright stance in that particular condition. In order to do so, both activity and amplitude contributions are investigated and presented by investigating correlation coefficients (COP_c, COP_v and COP_{net}) and contribution coefficients ($Contr_c$ and $Contr_V$). For each stance condition, primary control mechanism of quiet stance are presented.

For conditions except side-by-side stance, since there is right foot forward/left foot forward sub-conditions for each stance condition results for both sub-condition are presented at the same time, e.g. Tandem stance has two sub conditions such as tandem right (TR) and tandem left (TL). The difference between sub-conditions, if there is any, in other words the effect of forward foot preference are presented at the end of this chapter.

After the investigation of each stance condition separately and, identification of the dominant control mechanism of each stance condition, effect of stance position on the selected balance control mechanisms are presented by investigating the difference between the contributions of mechanisms through stance conditions.

Finally, effect of gender and handedness are investigated to identify if there is an effect of these factors to the results presented.

Through this chapter, results are separately presented for mediolateral (ML) and anteroposterior (AP) cardinal planes.

4.1. Contribution of Mechanisms in Different Stance Positions

4.1.1. Side-by-side Stance

In the ML direction, there is a very strong correlation relationship between COP_v and COP_{net} during side-by-side stance (.913 ± .257). That means COP controlled by loading-unloading mechanism is almost fully correlated with the COP_{net} signal. This pattern exists almost in all subjects. Table 4.1 presents the correlation coefficients for all subjects.

The correlation of COP_c and COP_{net} among subjects on the other hand is more variable but, overall correlation between COP_c and COP_{net} is weak (.337 ± .430) which means contribution of ankle mechanism is small according to correlation coefficient analysis. However, some subjects showed moderate to strong correlations as well as negative or positive correlations. It can be said that, for some subjects, both ankle and loading-unloading mechanisms are very active, and for some subjects these mechanisms cancel each other to create the net COP and for some add on each other. At the end, this is only valid for some subjects and not for the whole group. One representative trial graph showing COP_c, COP_v, and COP_{net} signals in ML direction for Subject 19 is presented in Figure 4.1.

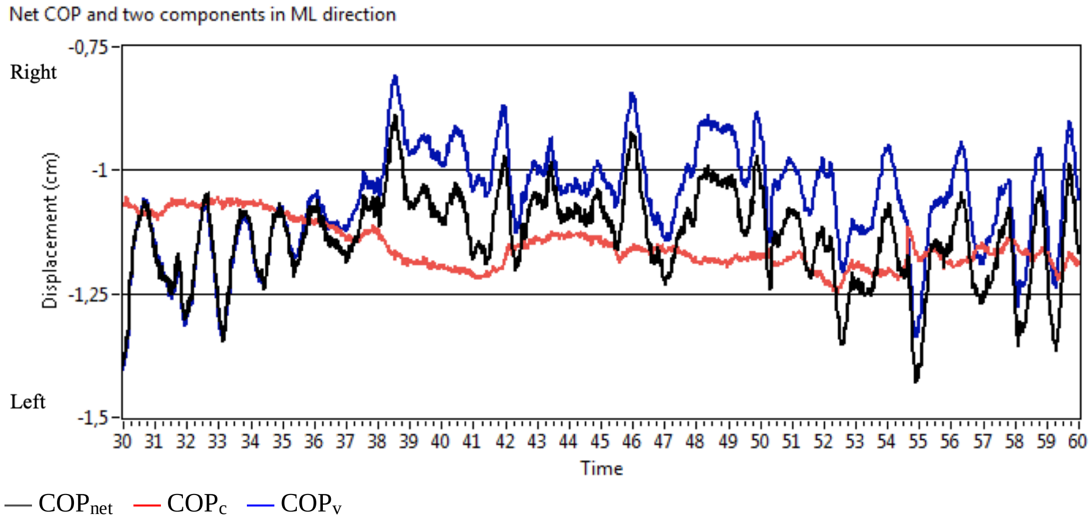

Figure 4.1. Net COP and two components in ML direction for side-by-side stance (S19)

Comparing the results of the correlation between COP_v and COP_{net}, and COP_c and COP_{net} both mean data and individual subject data shows that the correlation between COP_v and COP_{net} is almost perfect and higher than the correlation between COP_c and COP_{net} which indicates even in subjects with high ankle activity, loading-unloading mechanism matched the changes in COP_{net} better. Such high correlations of COP_v and COP_{net} indicates the dominant activity of the loading-unloading mechanism during side-by-side stance for the frontal plane, since the difference between correlations for COP_v vs. COP_{net} and COP_c vs. COP_{net} is statistically significant ($Z = -3.479$, $p < .001$). In addition to that, mean COP_c and COP_v showed positive correlation with COP_{net}, which means activity of both mechanisms sums up to total COP.

In order to say that there is a significant contribution of loading-unloading mechanism in side-by-side stance correlation analysis is not enough. Amplitude of these mechanisms should also be in accordance with the activity -the correlation relationship presented so far-. According to contribution coefficient ($Contr_C$ and $Contr_V$) calculations, amplitude or strength of loading-unloading mechanism is also higher than the amplitude of the ankle mechanism (.796 ±.118). Figure 4.2 presents the mean contribution coefficiets and standard deviations. In terms of both activity and amplitude contribution, loading-unloading mechanism is the dominant control mechanism in the ML direction.

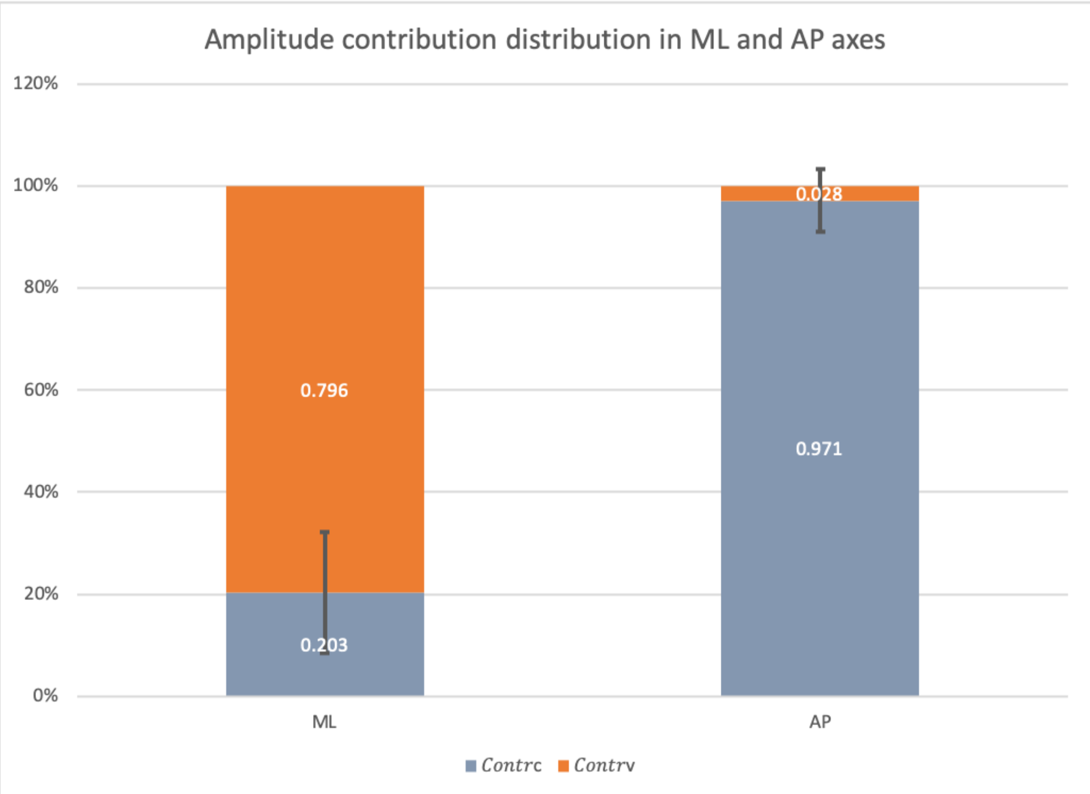

Figure 4.2. Mean amplitude contribution distribution and SD values of ankle and loading-unloading mechanisms in ML and AP direction for side-by-side stance.

In the AP direction there is a very strong relationship between COP_c and COP_{net} (.999 ± .003). For all subjects, correlation coefficients between COP_c and COP_{net} was almost 1. This means a very strong correlation and indicates a high activity of the ankle mechanism in the control of quiet stance. On the other hand, the correlation between COP_v and COP_{net} changes among participants. For some subjects, the loading-unloading mechanism adds on the activity of ankle mechanism when for some it subtracts from the activity of ankle mechanism. Alternatively, while for some subjects, the correlation coefficient between COP_v and COP_{net} is very strong, for some it is weak or even negligible. However, when the mean values are taken into account, overall activity of COP_v is negligible (0.060 ± 0.480) and there is a significant difference between contributions of COP_c and COP_v ($Z = 11.328$, $p < .001$). The activity of COP_{net} and two components (COP_c and COP_v) can be seen from a representative trial presented for Subject 21 in Figure 4.3.

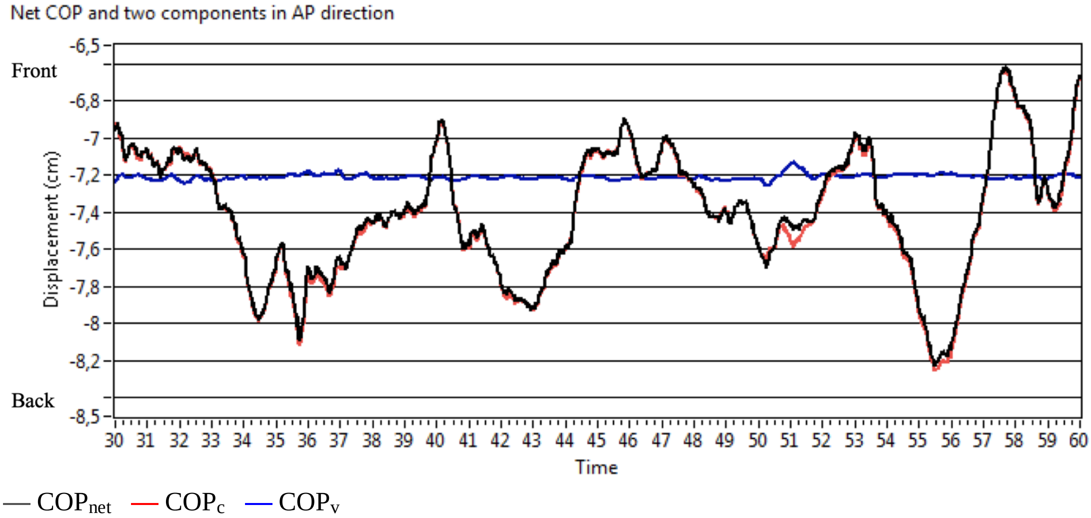

Figure 4.3. Net COP and two components in AP direction for side-by-side stance (S21)

The dominant activity of ankle mechanism is also supported by the contribution coefficients analysis. In terms of magnitude, ankle mechanism is by far the dominant control mechanism in the anteroposterior direction (.971 ± .062) where the contribution of the loading-unloading mechanism is negligible. Figure 4.2 presents the mean contribution coefficiets and standard deviations. This result implies the dominance of the ankle mechanism over loading-unloading mechanism for AP direction.

Table 4.1. *Correlation values for all subjects in side-by-side stance*

	ML		AP	
	COP_{net} vs. COP_c	**COP_{net} vs. COP_v**	**COP_{net} vs. COP_c**	**COP_{net} vs. COP_v**
S1	-0,173	0,990	0,99992	0,247
S2	-0,818	0,997	0,99991	0,883
S3	0,413	0,979	0,99998	-0,721
S4	-0,047	0,961	0,99999	-0,266
S5	0,116	0,981	0,99999	0,191
S6	0,672	-0,189	0,98495	0,912
S7	0,807	0,995	0,99998	-0,233
S9	0,628	0,989	0,99998	0,340
S10	-0,035	0,980	0,99997	-0,424
S11	0,635	0,980	0,99997	-0,140
S12	0,490	0,799	0,99997	0,002
S13	-0,388	0,991	0,99944	-0,476
S14	0,893	0,995	0,99995	0,596
S15	0,825	0,996	0,99964	0,791
S16	0,609	0,941	0,99965	-0,160
S17	0,406	0,993	0,99969	-0,067
S18	0,313	0,940	0,99991	0,222
S19	0,536	0,878	0,99998	0,450
S20	0,352	0,996	0,99992	-0,683
S21	0,527	0,983	0,99966	0,035
S22	0,305	0,989	0,99956	-0,241
Mean	0,3368	0,9130	0,999146	0,0599
SD	0,4301	0,2571	0,003256	0,4802

Table 4.2. *Contribution coefficient values for all subjects in side-by-side stance*

	ML		**AP**	
	$Contr_{\mathrm{c}}$	$Contr_{\mathrm{v}}$	$Contr_{\mathrm{c}}$	$Contr_{\mathrm{v}}$
S1	0.12	0.88	0.99	0.01
S2	0.11	0.89	0.98	0.02
S3	0.18	0.82	0.99	0.01
S4	0.22	0.78	1.00	0.00
S5	0.16	0.84	1.00	0.00
S6	0.57	0.43	0.70	0.30
S7	0.14	0.86	1.00	0.00
S9	0.16	0.84	0.99	0.01
S10	0.17	0.83	1.00	0.00
S11	0.20	0.80	0.99	0.01
S12	0.41	0.59	0.99	0.01
S13	0.12	0.88	0.96	0.04
S14	0.17	0.83	0.99	0.01
S15	0.13	0.87	0.96	0.04
S16	0.30	0.70	0.97	0.03
S17	0.11	0.89	0.98	0.02
S18	0.26	0.74	0.98	0.02
S19	0.36	0.64	1.00	0.00
S20	0.09	0.91	0.98	0.02
S21	0.18	0.82	0.97	0.03
S22	0.13	0.87	0.97	0.03
Mean	0.203	0.796	0.971	0.028
SD	0.118	0.118	0.062	0.062

4.1.2. Tandem Stance

In the ML direction, there is a very strong correlation between COP_c and COP_{net} (*TL : .994 ± .006*, *TR : .995 ± .005*). The relationship between COP_v and COP_{net} is variable among subjects, yet in overall there is a moderate correlation (*TL : .471 ± .404*, *TR : .413 ± .334*). Comparing the correlation results, correlation of COP_c and COP_{net} is significantly higher than correlation of COP_v and COP_{net} (*TL* : $Z = 7.909$, $p < .001$, *TR* : $Z = 8.444$, $p < .001$). According to these results, COP controlled by ankle mechanism is almost fully correlated with the COP_{net}. This means that the ankle mechanism was more active. The activity of COP_{net} and two components (COP_c and COP_v) can be seen from a representative trial in Figure 4.4 and Figure 4.5.

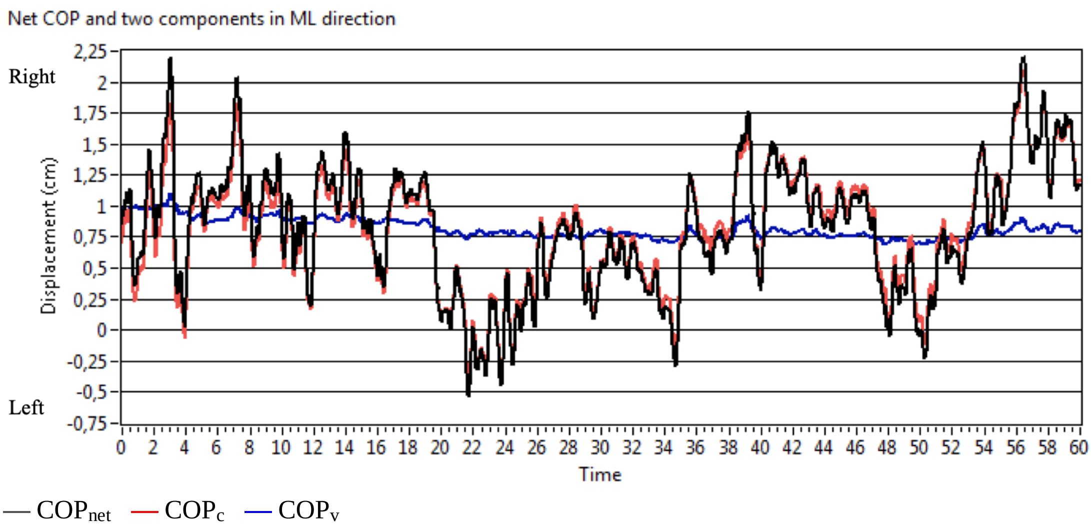

Figure 4.4. Net COP and two components in ML direction for TL (S11)

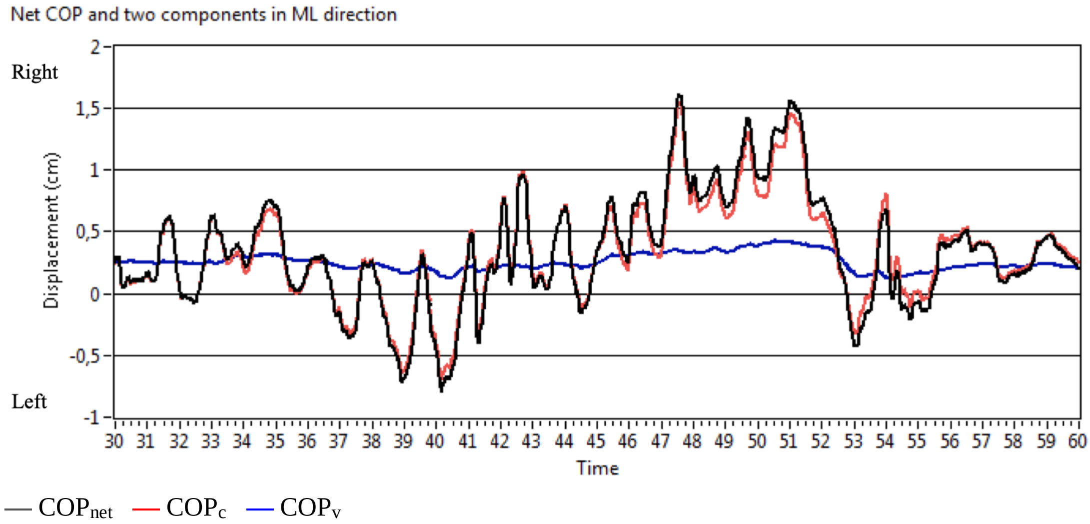

Figure 4.5. Net COP and two components in ML direction for TR (S21)

When the contribution coefficients are investigated, again in the ML direction, it can be seen that amplitude contribution of ankle mechanism is significantly higher (*TL :* .885 ± .068, *TR :* .910 ± .051) and the contribution of the loading- unloading mechanism is very weak or even negligible. Figure 4.6 presents the mean contribution coefficiets and standard deviations.

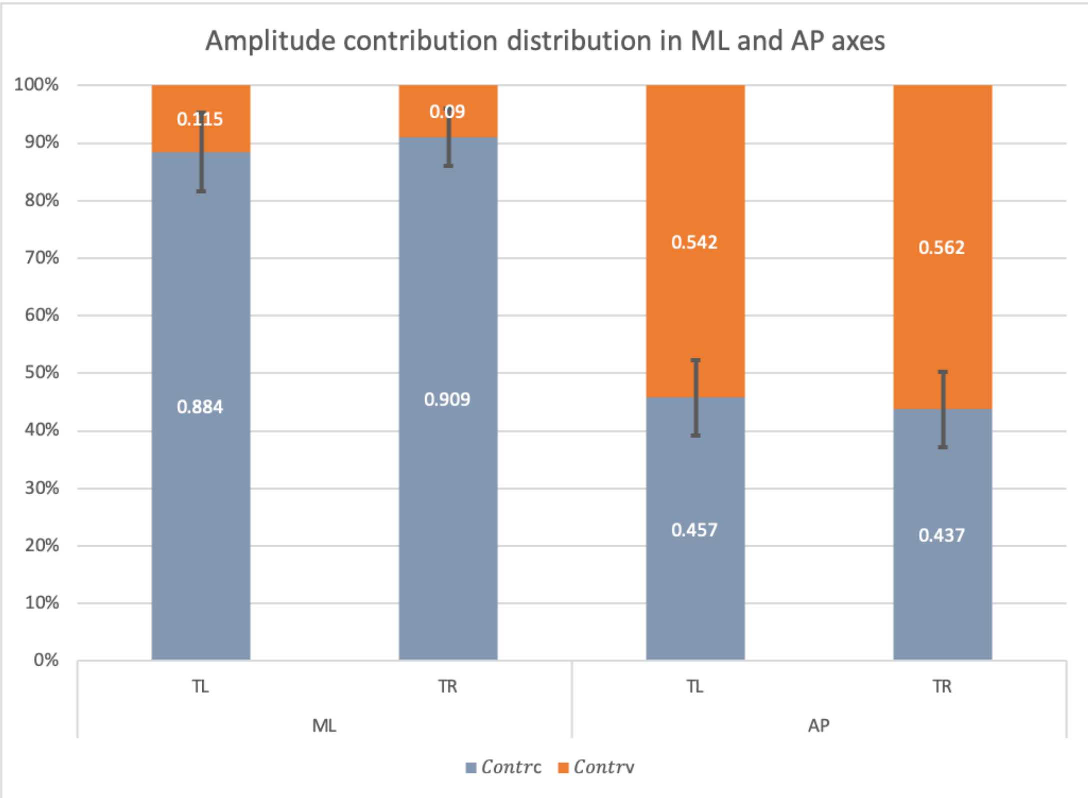

Figure 4.6. Mean amplitude contribution distribution and SD values of ankle and loading-unloading mechanisms in MLand AP direction for tandem stance.

These results reveal that ankle mechanism is dominant not only in terms of activity but also in terms of strength. It can be said that ankle mechanism is the dominant mechanism in the control of tandem stance in the ML direction.

In the AP direction, according to cross-correlation analysis, correlation relation between COP_c and COP_{net} is very weak (*TL* : .153 ± .265, *TR* : .060 ± .286) and between COP_v and COP_{net} is moderate/strong (*TL* : .473 ± .291, *TR* : .612 ± .207). In addition, the values of correlation amongst subjects vary both for COP_c vs. COP_{net} and COP_v vs. COP_{net}. For COP_c vs. COP_{net}, correlation values vary from weak to moderate and they are mostly additive yet sometimes subtractive. For COP_v vs. COP_{net}, correlation strength varies from weak to very strong among individuals and they are positive with only one exceptional subject (S21). By looking only to the cross-correlation data, it is hard to say that neither ankle, nor loading- unloading mechanism is dominantly active because, the cross-correlations between COP_c - COP_{net} and COP_v - COP_{net} are not significant in both TL and TR conditions (*TL :* Z = -.839, p = .201, *TR :* Z = -1.533, p = .063). However, for both TL and TR conditions the correlation between COP_v and COP_{net} is higher. The activity of COP_{net} and two components (COP_c and COP_v) can be seen from a representative trial in Figure 4.7 and Figure 4.8.

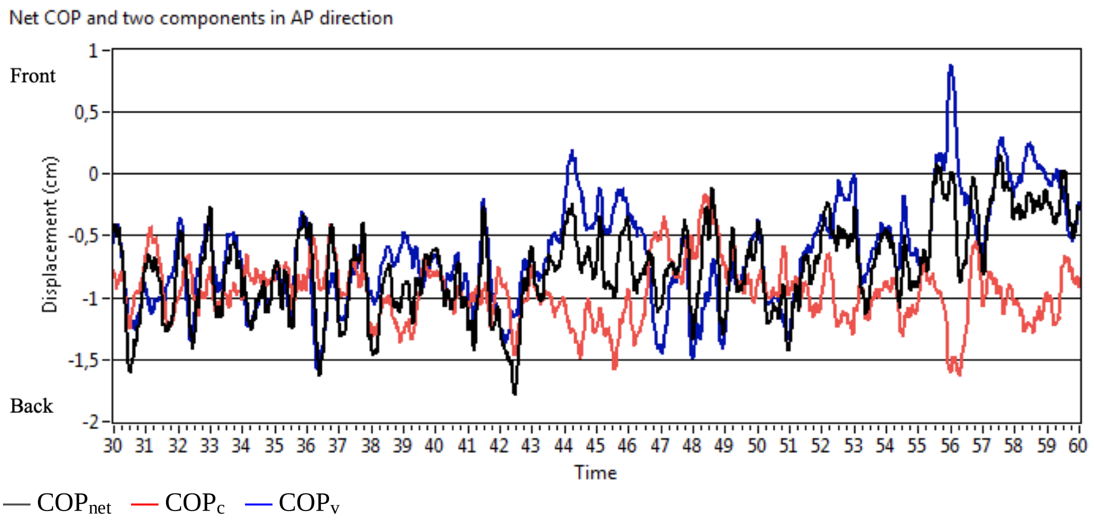

Figure 4.7. Net COP and two components in AP direction for TL (S4)

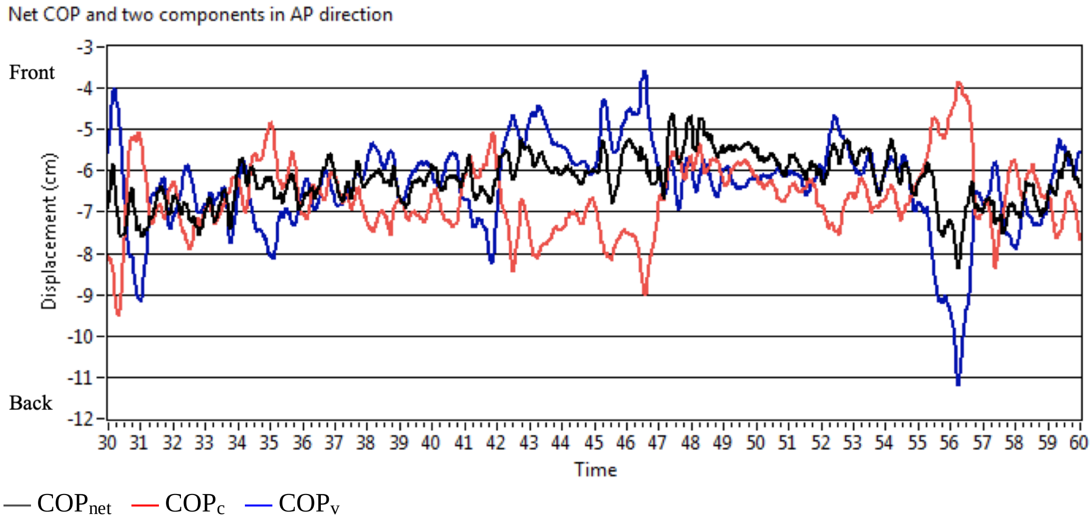

Figure 4.8. Net COP and two components in AP direction for TR (S10)

Similar results are obtained from the amplitude contribution analysis of these mechanisms. Figure 4.6 presents the mean contribution coefficiets and standard deviations. For both TL and TR conditions, contribution coefficients were almost equal (Contr$_V$ for *TL :* .542 ± .065, *TR :* .563 ± .065), yet the contribution of the loading-unloading mechanism is above the .50 level. Both mechanisms are contributing moderately where the contribution of the loading-unloading mechanism is slightly higher. Likewise it is seen in the correlation coefficient analysis, none of the mechanisms showed a higher contribution in terms of both activity and magnitude. Thus, even though the activity and strength of loading-unloading mechanism is higher than the ankle mechanism, this difference is too small to point out to a dominant mechanism for AP direction postural control in tandem stance.

Table 4.3. *Correlation values for all subjects in tandem stance*

	ML				AP			
	TL		TR		TL		TR	
	COP_{net} vs. COP_c	**COP_{net} vs. COP_v**	**COP_{net} vs. COP_c**	**COP_{net} vs. COP_v**	**COP_{net} vs. COP_c**	**COP_{net} vs. COP_v**	**COP_{net} vs. COP_c**	**COP_{net} vs. COP_v**
S1	0,986	0,735	0,996	0,649	0,237	0,550	0,767	0,340
S2	0,994	0,564	0,986	0,708	-0,037	0,460	0,011	0,587
S3	0,990	0,701	0,991	0,754	0,175	0,268	0,227	0,532
S4	0,997	-0,327	0,998	-0,201	0,117	0,593	-0,449	0,914
S5	0,992	0,862	0,999	0,363	-0,236	0,724	-0,054	0,709
S6	0,996	0,354	0,998	0,815	-0,095	0,896	0,358	0,153
S7	0,996	0,671	0,996	0,375	0,149	0,808	0,514	0,758
S8	0,999	0,724	0,996	0,316	-0,099	0,532	0,139	0,723
S9	0,999	-0,637	0,991	-0,032	0,114	0,332	-0,256	0,820
S10	0,987	0,422	0,990	0,715	0,278	0,832	0,059	0,664
S11	0,990	0,519	0,998	0,846	-0,083	0,763	-0,266	0,533
S12	0,980	0,691	0,999	0,622	0,149	0,512	-0,041	0,540
S13	0,984	0,346	0,999	0,051	-0,110	0,670	0,107	0,747
S14	0,993	0,837	0,991	0,212	0,053	0,373	-0,006	0,651
S15	0,999	0,267	0,990	0,675	0,446	0,481	-0,021	0,588
S16	0,999	0,302	0,982	0,403	-0,407	0,870	-0,148	0,896
S17	0,993	0,849	0,999	0,467	0,240	0,068	0,128	0,509
S18	0,994	0,931	0,998	-0,242	0,448	-0,160	0,092	0,661
S19	0,999	0,765	0,998	-0,082	0,517	0,353	0,067	0,780
S20	0,987	-0,074	0,997	0,525	0,432	0,067	0,016	0,446
S21	0,997	0,155	0,993	0,439	0,566	0,084	-0,375	0,756
S22	0,998	0,688	0,997	0,697	0,497	0,330	0,438	0,140
Mean	0,9936	0,4705	0,9951	0,4128	0,1525	0,4734	0,0595	0,6117
SD	0,0056	0,4040	0,0046	0,3342	0,2646	0,2910	0,2862	0,2066

Table 4.4. *Contribution coefficient values for all subjects in tandem stance*

	ML				AP			
	TL		TR		TL		TR	
	$Contr_{\mathrm{c}}$	$Contr_{\mathrm{v}}$	$Contr_{\mathrm{c}}$	$Contr_{\mathrm{v}}$	$Contr_{\mathrm{c}}$	$Contr_{\mathrm{v}}$	$Contr_{\mathrm{c}}$	$Contr_{\mathrm{v}}$
S1	0.81	0.19	0.91	0.09	0.46	0.54	0.60	0.40
S2	0.90	0.10	0.83	0.17	0.47	0.53	0.45	0.55
S3	0.84	0.16	0.83	0.17	0.50	0.50	0.46	0.54
S4	0.94	0.06	0.96	0.04	0.44	0.56	0.31	0.69
S5	0.81	0.19	0.96	0.04	0.42	0.58	0.42	0.58
S6	0.92	0.08	0.93	0.07	0.31	0.69	0.52	0.48
S7	0.90	0.10	0.92	0.08	0.37	0.63	0.43	0.57
S8	0.97	0.03	0.92	0.08	0.45	0.55	0.41	0.59
S9	0.99	0.01	0.89	0.11	0.49	0.51	0.36	0.64
S10	0.85	0.15	0.84	0.16	0.37	0.63	0.43	0.57
S11	0.86	0.14	0.94	0.06	0.39	0.61	0.46	0.54
S12	0.79	0.21	1.00	0.00	0.46	0.54	0.46	0.54
S13	0.84	0.16	0.96	0.04	0.43	0.57	0.41	0.59
S14	0.83	0.17	0.89	0.11	0.48	0.52	0.43	0.57
S15	0.97	0.03	0.84	0.16	0.50	0.50	0.45	0.55
S16	0.99	0.01	0.83	0.17	0.35	0.65	0.31	0.69
S17	0.82	0.18	0.96	0.04	0.51	0.49	0.47	0.53
S18	0.77	0.23	0.94	0.06	0.53	0.47	0.43	0.57
S19	0.97	0.03	0.95	0.05	0.53	0.47	0.38	0.62
S20	0.86	0.14	0.93	0.07	0.53	0.47	0.47	0.53
S21	0.92	0.08	0.88	0.12	0.55	0.45	0.42	0.58
S22	0.93	0.07	0.92	0.08	0.52	0.48	0.53	0.47
Mean	0.884	0.115	0.909	0.09	0.457	0.542	0.437	0.562
SD	0.068	0.068	0.05	0.05	0.065	0.065	0.065	0.065

4.1.3. Intermediate Stance

4.1.3.1. 30 ° Stance

In the ML direction, there is a moderate correlation between COP_c and COP_{net} in the 30L condition (.582 ± .352) and a strong correlation in the 30R condition (.631 ± .246). The correlation coefficient values varied among subjects in both 30L and 30R conditions. For most of the subjects, the correlation between COP_c and COP_{net} is positive and have a strength of moderate to very strong. However, there is a very strong correlation between COP_v and COP_{net} for both 30L and 30R sub-conditions (*30L :* .961 ± .047, *30R :* .983 ± .012). In overall, the correlation coefficient between COP_v and COP_{net} is significantly higher then the correlation between COP_c and COP_{net} for all subjects in both sub-conditions (*30L :* Z = -4.359, $p < .001$, *30R :* Z = -5.711, $p <.001$). The loading-unloading mechanism correlated more with the COP_{net} compared to the ankle mechanism. The activity of COP_{net} and two components (COP_c and COP_v) can be seen from a representative trial in Figure 4.9 and Figure 4.10.

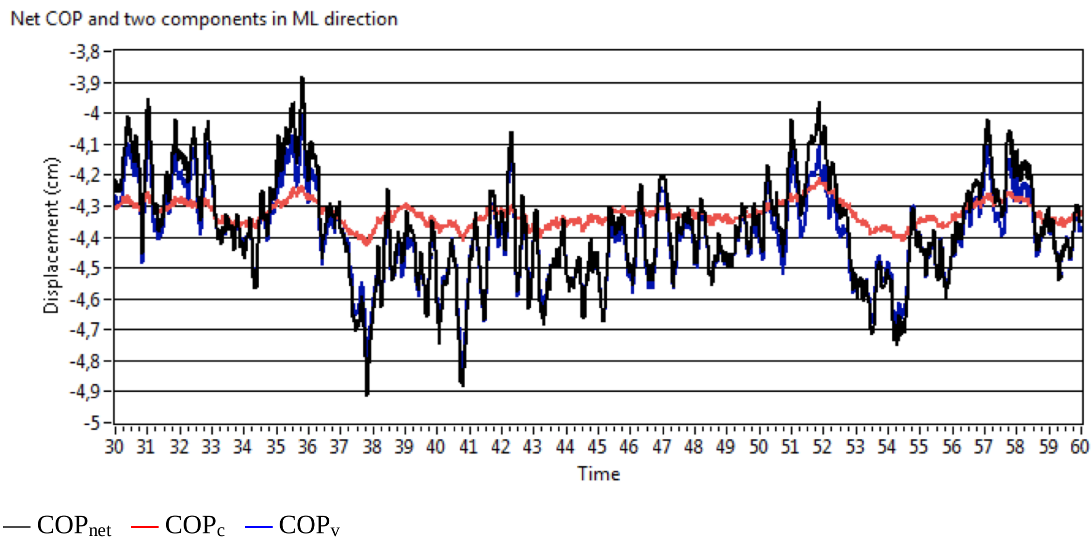

Figure 4.9. Net COP and two components in ML direction for 30L (S16)

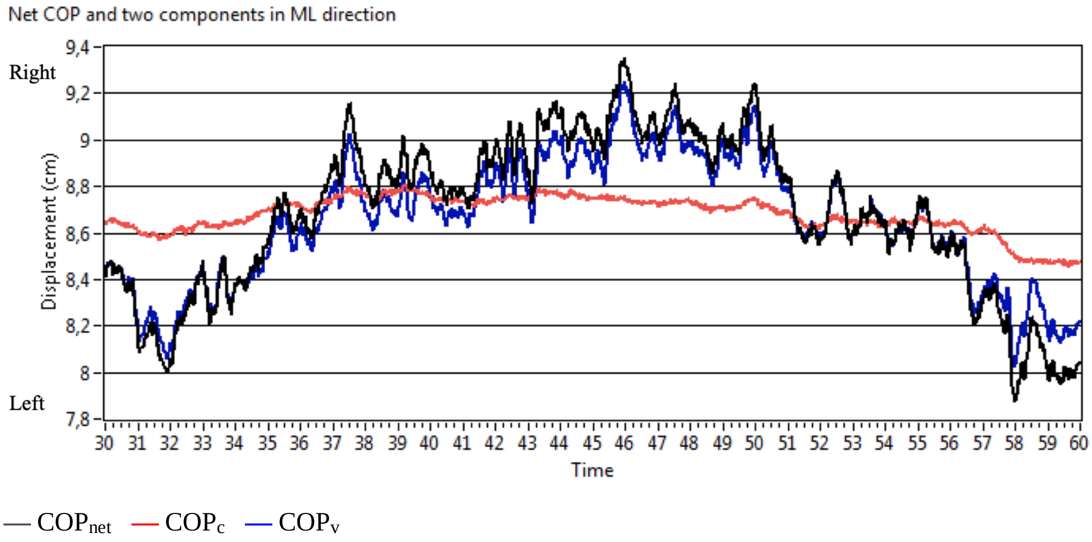

Figure 4.10. Net COP and two components in ML direction for 30R (S1)

Contribution coefficient analysis also supports the results above. The loading-unloading mechanism is significantly stronger than the ankle mechanism ($Contr_v$: *30L :* .752 ± .087, *30R :* .811 ± .056) in terms of amplitude. Figure 4.11 presents the mean contribution coefficiets and standard deviations. Unlike the variations seen in the correlation coefficients between COP_c and COP_{net} among subjects, almost no variation exists for amplitude contributions for both 30L and 30R sub-conditions. In ML direction the loading-unloading mechanism is dominant and it s the primary control mechanism for 30° stance.

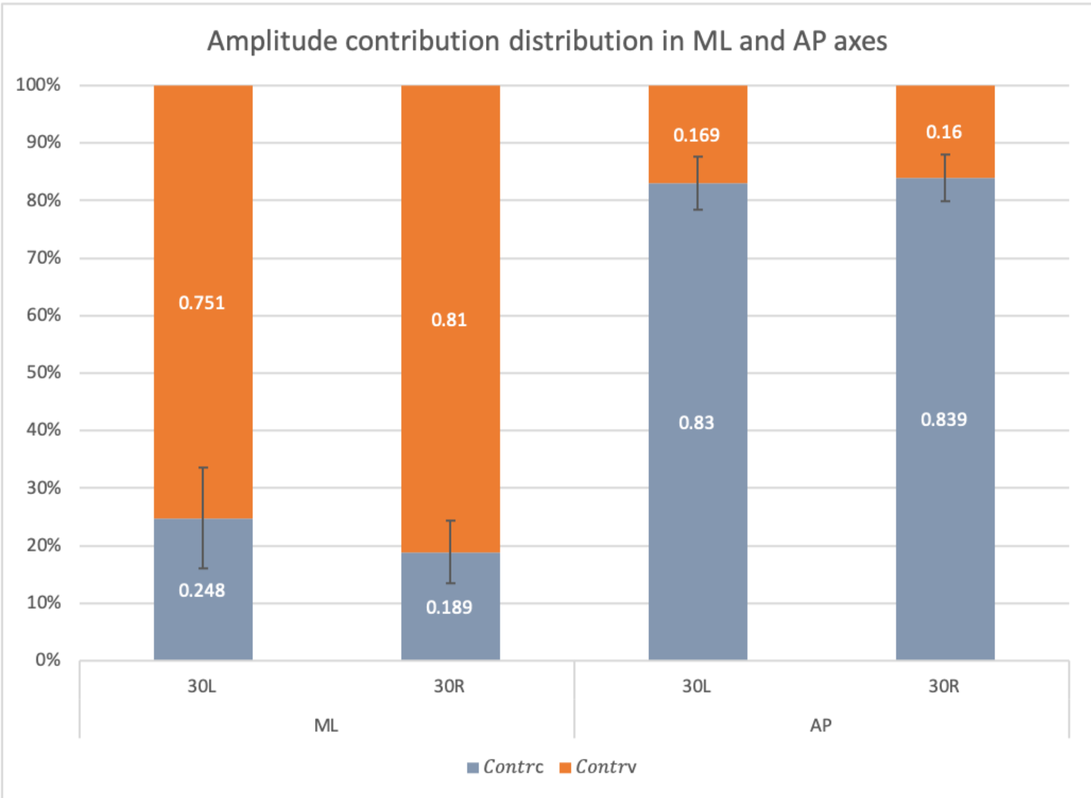

Figure 4.11. Mean amplitude contribution distribution and SD values of ankle and loading-unloading mechanisms in MLand AP direction for 30° stance

In the AP direction there is very strong correlation between COP_c and COP_{net} (*30L :* .981 ± .015, *30R :* .987 ± .010). The correlation between COP_v and COP_{net} is weak for 30L and moderate for 30R sub-condition (*30L :* -.378 ± .372, meanSD, *30R :* -.569 ± .307). The activity of COP_{net} and two components (COP_c and COP_v) can be seen from a representative trial in Figure 4.12 and Figure 4.13. The difference between correlations COP_c - COP_{net} and COP_v - COP_{net} is significant (*30L :* Z = 6.722, p < .001, *30R :* Z = 7.527, p < .001). For all subjects, correlation between COP_c and COP_{net} is very strong. On the other hand, for most of the subjects, correlation between COP_v and COP_{net} is negative and has a moderate to very strong strength. This means that the loading-unloading mechanism contributes to the ankle mechanism, but in a subtractive way.

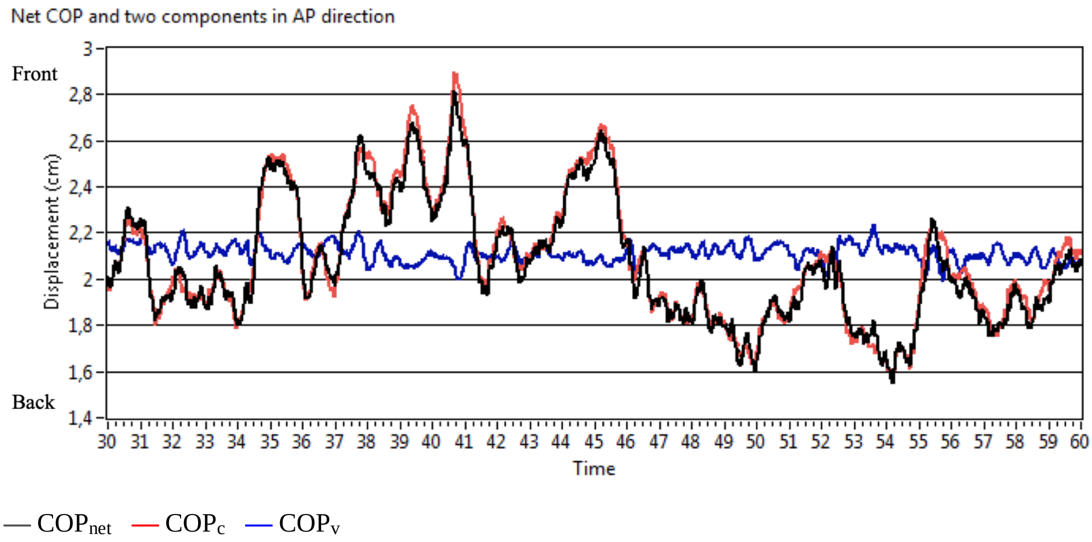

Figure 4.12. Net COP and two components in AP direction for 30L (S3)

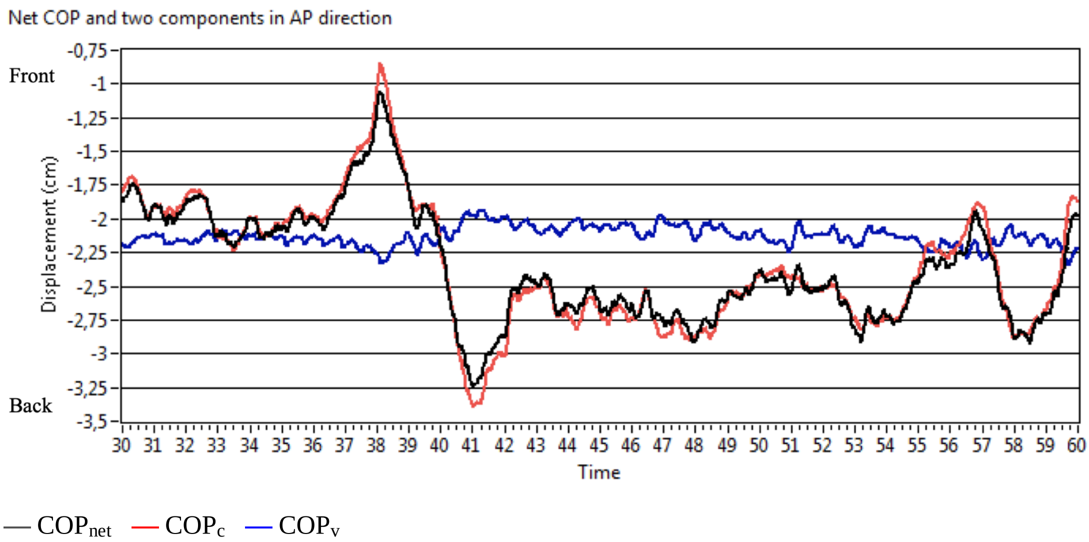

Figure 4.13. Net COP and two components in AP direction for 30R (S20)

Contribution coefficient analysis shows that the ankle mechanism is evidently stronger than the loading-unloading mechanism for both sub-conditions ($Contr_c$ for *30L :* .830 ± .046, *30R :* .839 ± .040). Figure 4.11 presents the mean contribution coefficiets and standard deviations. There are no significant variations among subjects. Combining the result of correlation and contribution coefficient analysis, ankle mechanism is both active and stronger than the loading-unloading mechanism and it is the dominant control mechanism in the 30° intermediate stance condition for AP direction.

Table 4.5. *Correlation values for all subjects in 30° stance*

	ML				**AP**			
	30L		**30R**		**30L**		**30R**	
	COP_{net} vs. COP_c	**COP_{net} vs. COP_v**	**COP_{net} vs. COP_c**	**COP_{net} vs. COP_v**	**COP_{net} vs. COP_c**	**COP_{net} vs. COP_v**	**COP_{net} vs. COP_c**	**COP_{net} vs. COP_v**
S1	0,945	0,991	0,770	0,971	0,996	-0,861	0,994	-0,781
S2	0,779	0,990	0,874	0,998	0,970	-0,291	0,961	-0,633
S3	0,766	0,946	0,531	0,967	0,990	-0,318	0,960	0,015
S4	-0,310	0,981	0,866	0,997	0,967	-0,466	0,985	-0,789
S5	-0,294	0,921	0,551	0,982	0,989	-0,800	0,994	-0,794
S6	0,336	0,991	0,455	0,982	0,933	0,256	0,983	-0,642
S7	0,492	0,961	0,858	0,995	0,982	0,262	0,990	-0,670
S8	0,900	0,980	0,797	0,975	0,992	-0,724	0,997	-0,687
S9	0,951	0,996	0,640	0,980	0,975	-0,637	0,993	-0,748
S10	0,375	0,911	0,328	0,985	0,988	0,004	0,976	-0,460
S11	0,185	0,996	0,893	0,997	0,961	-0,150	0,997	-0,884
S12	0,546	0,949	0,620	0,972	0,984	-0,101	0,980	-0,558
S13	0,661	0,987	0,896	0,984	0,968	-0,181	0,982	-0,684
S14	0,848	0,966	0,252	0,972	0,997	-0,885	0,997	-0,885
S15	0,571	0,884	0,618	0,970	0,970	0,103	0,987	-0,202
S16	0,618	0,974	0,611	0,978	0,984	-0,112	0,988	-0,234
S17	0,470	0,966	0,494	0,977	0,981	-0,594	0,989	-0,392
S18	0,751	0,966	0,729	0,996	0,993	-0,720	0,992	-0,793
S19	0,730	0,800	0,767	0,981	0,993	-0,170	0,995	-0,777
S20	0,888	0,994	0,595	0,995	0,998	-0,852	0,990	-0,419
S21	0,848	0,991	0,842	0,996	0,970	-0,293	0,995	-0,784
S22	0,739	0,988	-0,103	0,954	0,991	-0,783	0,984	0,300
Mean	0,5819	0,9608	0,6316	0,9825	0,9809	-0,3780	0,9873	-0,5686
SD	0,3524	0,0466	0,2455	0,0120	0,0153	0,3721	0,0102	0,3066

Table 4.6. *Contribution coefficient values for all subjects in 30° stance*

	ML				AP			
	30L		**30R**		**30L**		**30R**	
	$Contr_c$	$Contr_v$	$Contr_c$	$Contr_v$	$Contr_c$	$Contr_v$	$Contr_c$	$Contr_v$
S1	0.29	0.71	0.27	0.73	0.86	0.14	0.86	0.14
S2	0.18	0.82	0.11	0.89	0.80	0.20	0.74	0.26
S3	0.34	0.66	0.23	0.77	0.87	0.13	0.78	0.22
S4	0.17	0.83	0.12	0.88	0.77	0.23	0.78	0.22
S5	0.29	0.71	0.18	0.82	0.81	0.19	0.85	0.15
S6	0.12	0.88	0.17	0.83	0.73	0.27	0.81	0.19
S7	0.24	0.76	0.15	0.85	0.84	0.16	0.85	0.15
S8	0.32	0.68	0.27	0.73	0.85	0.15	0.91	0.09
S9	0.23	0.77	0.20	0.80	0.78	0.22	0.86	0.14
S10	0.31	0.69	0.15	0.85	0.87	0.13	0.81	0.19
S11	0.08	0.92	0.15	0.85	0.78	0.22	0.86	0.14
S12	0.27	0.73	0.23	0.77	0.85	0.15	0.81	0.19
S13	0.18	0.82	0.28	0.72	0.80	0.20	0.80	0.20
S14	0.33	0.67	0.19	0.81	0.91	0.09	0.86	0.14
S15	0.36	0.64	0.23	0.77	0.81	0.19	0.86	0.14
S16	0.22	0.78	0.21	0.79	0.85	0.15	0.87	0.13
S17	0.22	0.78	0.20	0.80	0.81	0.19	0.87	0.13
S18	0.28	0.72	0.10	0.90	0.86	0.14	0.83	0.17
S19	0.47	0.53	0.24	0.76	0.90	0.10	0.88	0.12
S20	0.19	0.81	0.10	0.90	0.89	0.11	0.87	0.13
S21	0.20	0.80	0.14	0.86	0.80	0.20	0.87	0.13
S22	0.18	0.82	0.23	0.77	0.83	0.17	0.84	0.16
Mean	0.248	0.751	0.189	0.81	0.83	0.169	0.839	0.16
SD	0.087	0.087	0.055	0.055	0.046	0.046	0.04	0.04

4.1.3.2. 45 ° Stance

In the ML direction, there is very strong correlation between COP_v and COP_{net} (*45L :* .974 ± .026, *45R :* .989 ± .010). There is a strong correlation between COP_c and COP_{net} (*45L :* .716 ± .239, *45R :* .784 ± .207). The activity of COP_{net} and two components (COP_c and COP_v) can be seen from a representative trial in Figure 4.14 and Figure 4.15. There is a statistically significant difference between the correlation relations of COP_c - COP_{net} and COP_v - COP_{net} (45L = Z = -4.662, $p < .001$, 45R = Z = -5.942, $p < .001$). Correlation coefficient between COP_v and COP_{net} has almost no variation and suggests very strong activity of the loading-unloading mechanism for all subjects where the variation among subjects is high for COP_c and COP_{net}. All of these suggests the comparibaly higher activity of the loading-unloading mechanism in the ML direction.

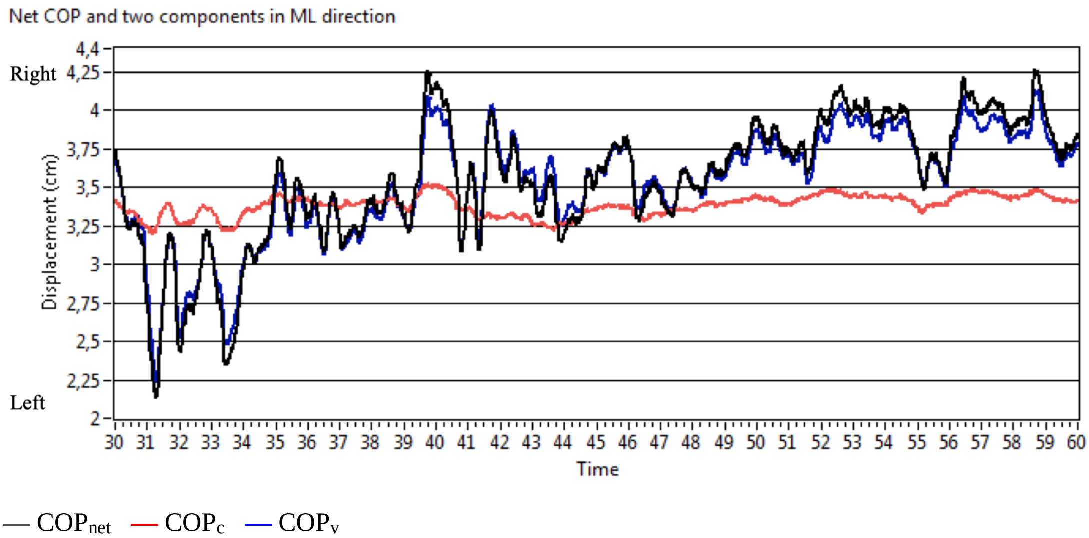

Figure 4.14. Net COP and two components in ML direction for 45L (S12)

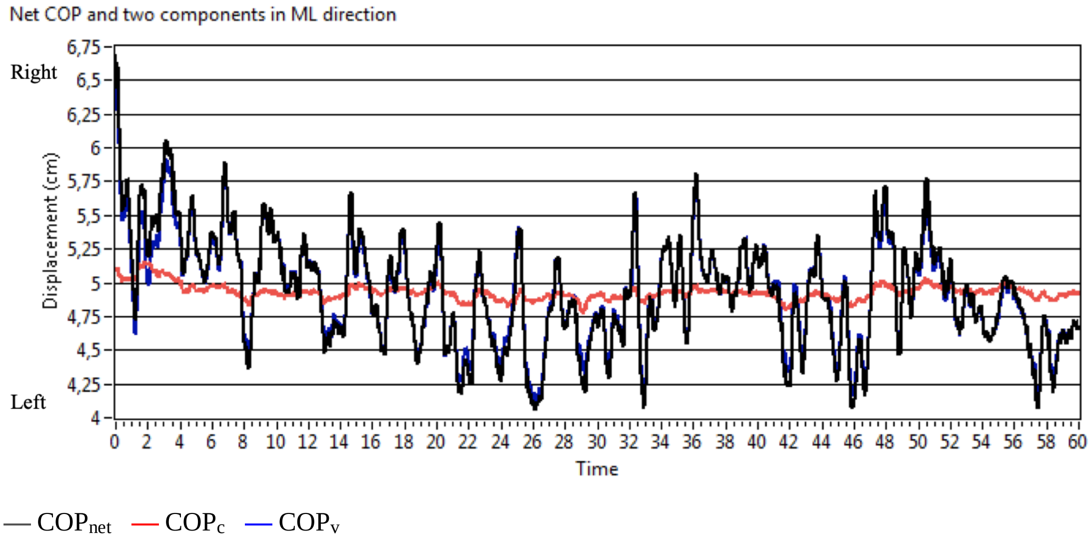

Figure 4.15. Net COP and two components in ML direction for 45R (S18)

For the contribution coefficients in the ML direction, the difference between $Contr_C$ and $Contr_V$ is also obvious. Figure 4.16 presents the mean contribution coefficiets and standard deviations. Amplitude of the loading-unloading mechanism is apparently higher compared to the ankle mechanism for both sub-conditions (for $Contr_V$ *45L :* .756 ± .080, meanSD, *45R :* .794 ± .054) and the variation among subjects is not high.

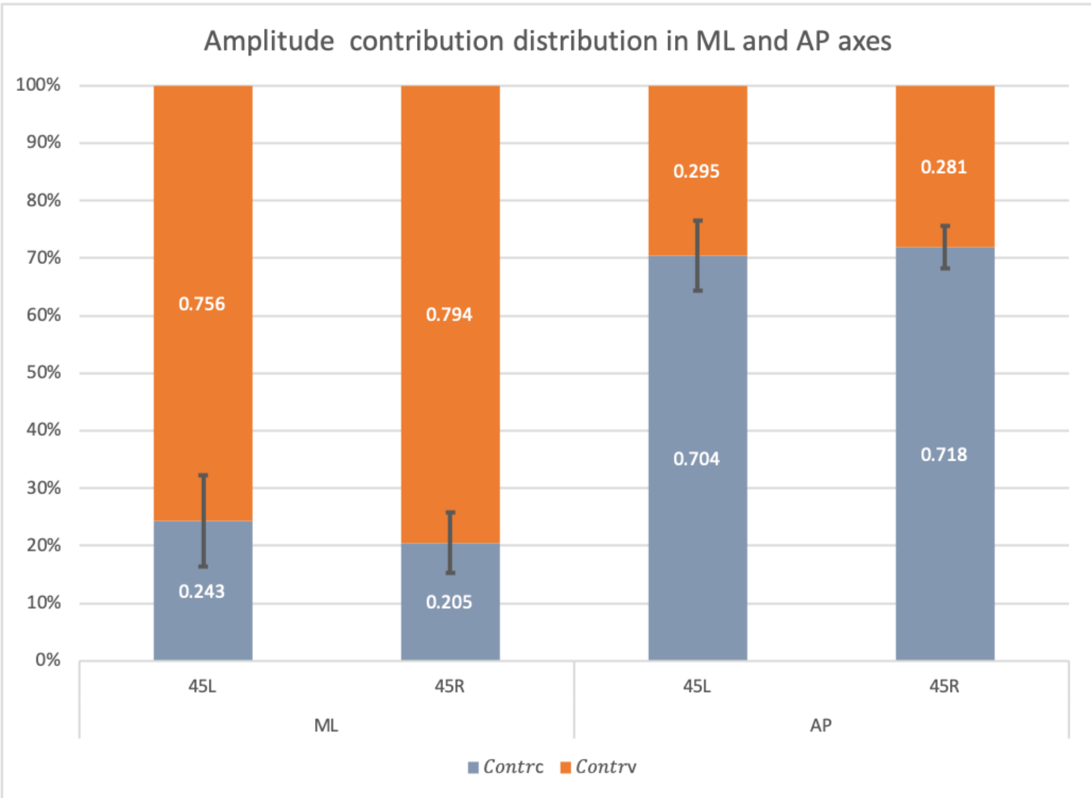

Figure 4.16. Mean amplitude contribution distribution and SD values of ankle and loading-unloading mechanisms in MLand AP direction for 45° stance

In the AP direction there is very strong correlation between COP_c and COP_{net} (*45L :* .908 ± .076, *45R :* .934 ± .043). There is a weak correlation between COP_v and COP_{net} and this correlation is negative (*45L :* -.316 ± .361, *45R :* -.479 ± .307). The activity of COP_{net} and two components (COP_c and COP_v) can be seen from a representative trial in Figure 4.17 and Figure 4.18. The difference between COP_c - COP_{net} and COP_v - COP_{net} correlations is statistically significant (*45L :* Z = 4.401, p < .001, *45R :* Z = 5.185, p < .001). Correlation coefficient analysis shows that the activity of the ankle mechanism is significantly higher than the loading-unloading mechanism. Note that, variation among subjects is high for COP_v vs. COP_{net}.

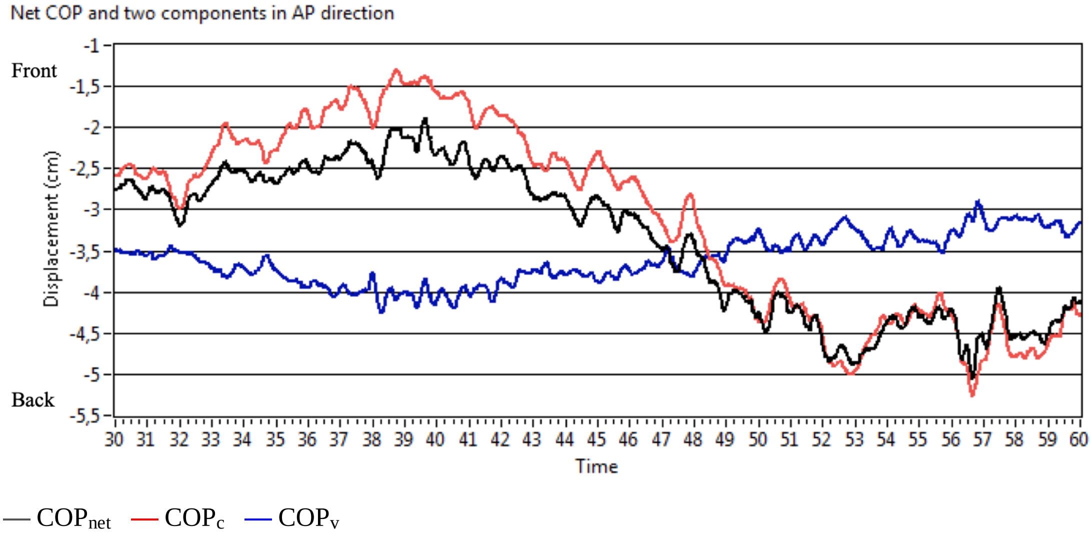

Figure 4.17. Net COP and two components in AP direction for 45L (S14)

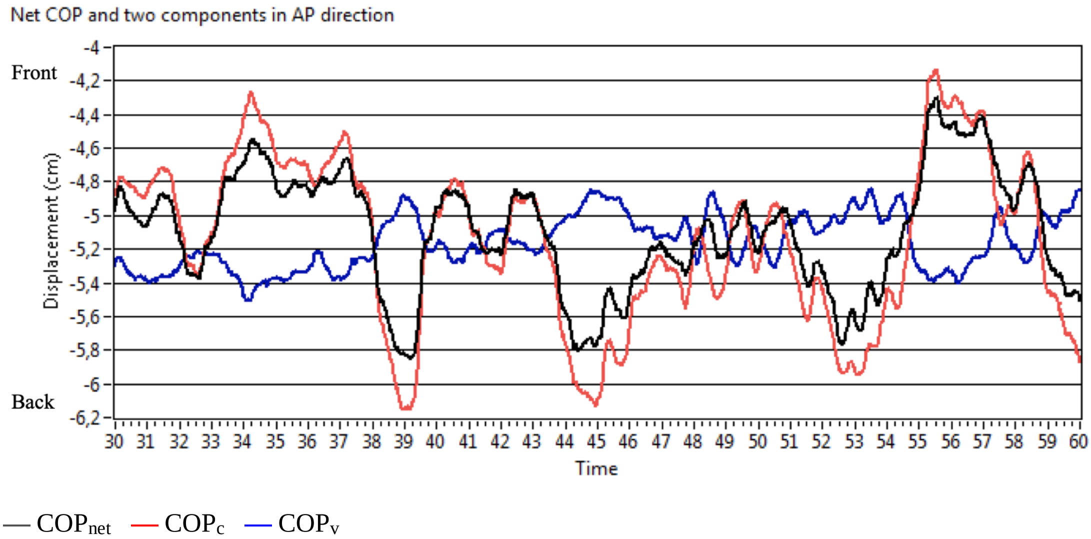

Figure 4.18. Net COP and two components in AP direction for 45R (S8)

Amplitude analysis reveals that the behaviour seen in the correlation relationship exists also in contribution coefficients. Figure 4.16 presents the mean contribution coefficiets and standard deviations. Ankle mechanism has a higher amplitude contribution to the AP balance (for $Contr_c$ *45L* = .705 ± .062, *45R* = .718 ± .038). Note that, the amplitude of the loading-unloading mechanism almost even though it is low, did not change among subjects. The loading-unloading mechanism has a weak contribution to this control with its low activity and amplitude. The results implicate that the primary control mechanism for the AP direction is ankle mechanism.

Table 4.7. *Correlation values for all subjects in 45° stance*

	ML				**AP**			
	45L		**45R**		**45L**		**45R**	
	COP_{net} vs. COP_c	**COP_{net} vs. COP_v**	**COP_{net} vs. COP_c**	**COP_{net} vs. COP_v**	**COP_{net} vs. COP_c**	**COP_{net} vs. COP_v**	**COP_{net} vs. COP_c**	**COP_{net} vs. COP_v**
S1	0,752	0,991	0,476	0,952	0,767	-0,196	0,894	-0,033
S2	0,937	0,992	0,857	0,994	0,965	-0,631	0,969	-0,712
S3	0,415	0,960	0,846	0,977	0,909	-0,041	0,940	-0,367
S4	0,615	0,913	0,815	0,983	0,959	-0,519	0,931	-0,537
S5	-0,001	0,914	0,658	0,983	0,922	0,124	0,909	-0,456
S6	0,745	0,991	0,841	0,979	0,815	-0,184	0,946	-0,563
S7	0,928	0,996	0,922	0,994	0,914	-0,496	0,971	-0,632
S8	0,887	0,968	0,951	0,995	0,973	-0,633	0,959	-0,577
S9	0,963	0,994	0,967	0,997	0,948	-0,671	0,991	-0,899
S10	0,574	0,934	0,294	0,981	0,850	0,137	0,891	0,271
S11	0,529	0,993	0,923	0,995	0,772	0,081	0,989	-0,892
S12	0,724	0,989	0,755	0,991	0,817	0,009	0,846	-0,128
S13	0,692	0,975	0,939	0,992	0,919	0,087	0,976	-0,702
S14	0,932	0,976	0,766	0,992	0,942	-0,459	0,893	-0,404
S15	0,534	0,943	0,939	0,991	0,925	0,447	0,965	-0,654
S16	0,921	0,987	0,888	0,996	0,975	-0,643	0,869	-0,212
S17	0,745	0,979	0,210	0,990	0,980	-0,471	0,886	0,003
S18	0,829	0,987	0,771	0,995	0,973	-0,771	0,957	-0,658
S19	0,894	0,958	0,727	0,980	0,972	-0,523	0,958	-0,623
S20	0,890	0,994	0,833	0,998	0,976	-0,800	0,892	-0,254
S21	0,867	0,997	0,961	0,996	0,752	-0,098	0,970	-0,785
S22	0,367	0,980	0,888	0,989	0,939	-0,698	0,977	-0,711
Mean	0,7158	0,9737	0,7837	0,9886	0,9079	-0,3162	0,9359	-0,4787
SD	0,2390	0,0258	0,2072	0,0103	0,0755	0,3616	0,0427	0,3074

Table 4.8. *Contribution coefficient values for all subjects in* 45° stance

	ML				**AP**			
	45L		**45R**		**45L**		**45R**	
	$Contr_c$	$Contr_v$	$Contr_c$	$Contr_v$	$Contr_c$	$Contr_v$	$Contr_c$	$Contr_v$
S1	0.16	0.84	0.26	0.74	0.60	0.40	0.70	0.30
S2	0.26	0.74	0.18	0.82	0.75	0.25	0.74	0.26
S3	0.23	0.77	0.28	0.72	0.71	0.29	0.73	0.27
S4	0.34	0.66	0.24	0.76	0.75	0.25	0.70	0.30
S5	0.29	0.71	0.19	0.81	0.72	0.28	0.68	0.32
S6	0.16	0.84	0.27	0.73	0.63	0.37	0.72	0.28
S7	0.18	0.82	0.22	0.78	0.68	0.32	0.77	0.23
S8	0.35	0.65	0.23	0.77	0.77	0.23	0.75	0.25
S9	0.28	0.72	0.21	0.79	0.70	0.30	0.77	0.23
S10	0.30	0.70	0.17	0.83	0.65	0.35	0.68	0.32
S11	0.11	0.89	0.20	0.80	0.61	0.39	0.76	0.24
S12	0.17	0.83	0.17	0.83	0.63	0.37	0.65	0.35
S13	0.23	0.77	0.26	0.74	0.72	0.28	0.77	0.23
S14	0.38	0.62	0.16	0.84	0.73	0.27	0.67	0.33
S15	0.28	0.72	0.27	0.73	0.70	0.30	0.75	0.25
S16	0.29	0.71	0.16	0.84	0.78	0.22	0.67	0.33
S17	0.23	0.77	0.12	0.88	0.82	0.18	0.68	0.32
S18	0.22	0.78	0.13	0.87	0.74	0.26	0.72	0.28
S19	0.39	0.61	0.23	0.77	0.79	0.21	0.73	0.27
S20	0.19	0.81	0.08	0.92	0.74	0.26	0.68	0.32
S21	0.12	0.88	0.24	0.76	0.61	0.39	0.72	0.28
S22	0.18	0.82	0.24	0.76	0.68	0.32	0.77	0.23
Mean	0.243	0.756	0.205	0.794	0.704	0.295	0.718	0.281
SD	0.08	0.08	0.053	0.053	0.061	0.061	0.037	0.037

4.1.3.3. 60 ° Stance

Correlation analysis in the ML direction shows a very strong correlation between COP_v and COP_{net} (*60L :* .984 ± .015, *60R :* .986 ± .013). The correlation between COP_c and COP_{net} is strong in 60L (.757 ± .350) and very strong in 60R sub-condition (.875 ± .094). Although both correlations (COP_c - COP_{net} and COP_v - COP_{net}) are high, the difference between COP_c - COP_{net} and COP_v - COP_{net} is statistically significant (*60L :* Z = -5.433, p < .001, *60R :* Z = -4.863, p < .001). The activity of COP_{net} and two components (COP_c and COP_v) can be seen from a representative trial in Figure 4.19 and Figure 4.20.

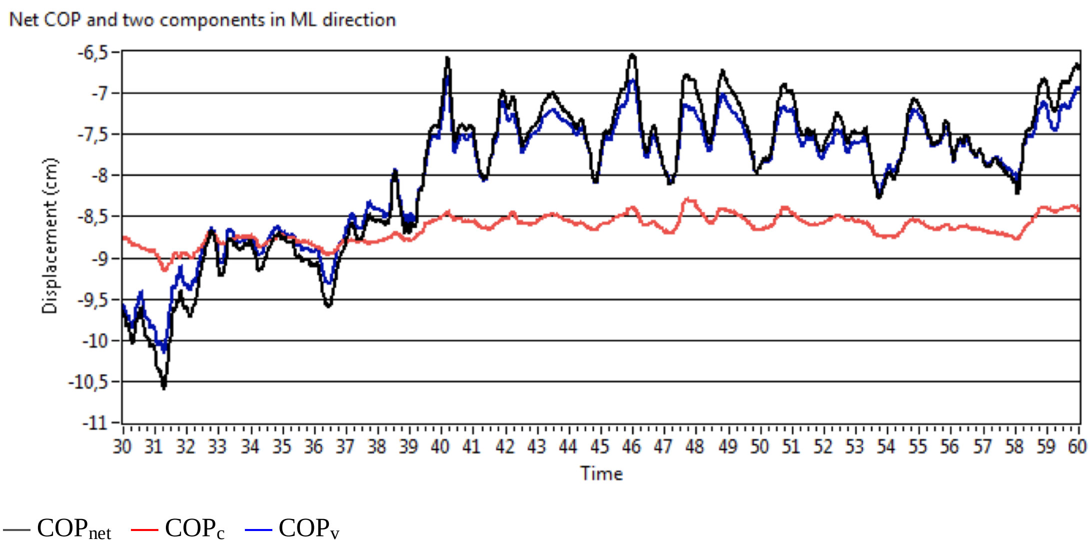

Figure 4.19. Net COP and two components in ML direction for 60L (S15)

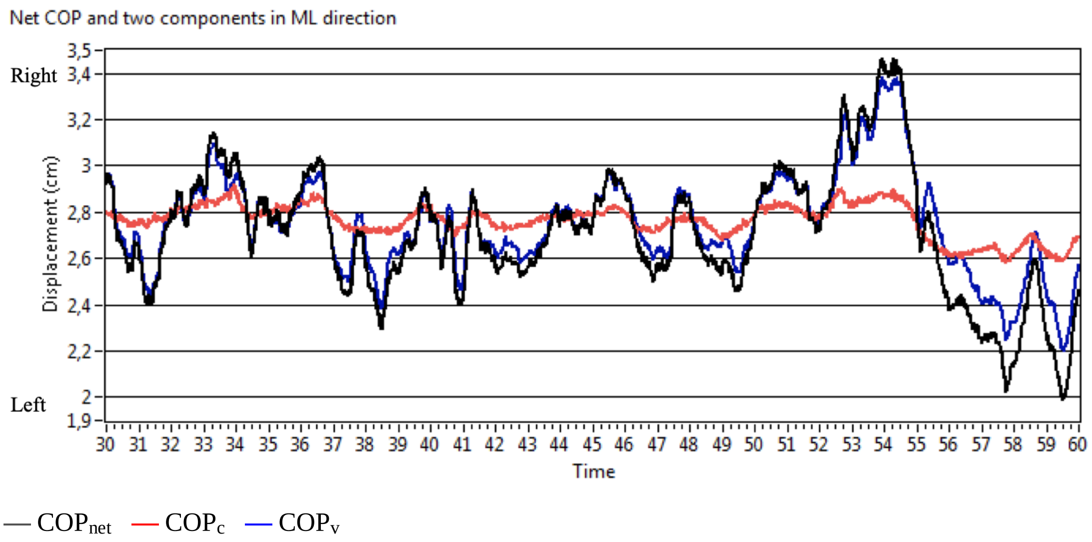

Figure 4.20. Net COP and two components in ML direction for 60R (S6)

Amplitude contribution of the loading-unloading mechanism is also higher than the amplitude contribution of ankle mehanism (for $Contr_V$ *60L :* .765 ± .075, *60R :* .741 ± .078). Ankle mechanism has a weak contribution to the ML balance. Figure 4.21 presents the mean contribution coefficiets and standard deviations. Finally, both amplitude and activity analysis reveals that the primary control mechanism for the 60° stance in the ML direction is the loading-unloading mechanism.

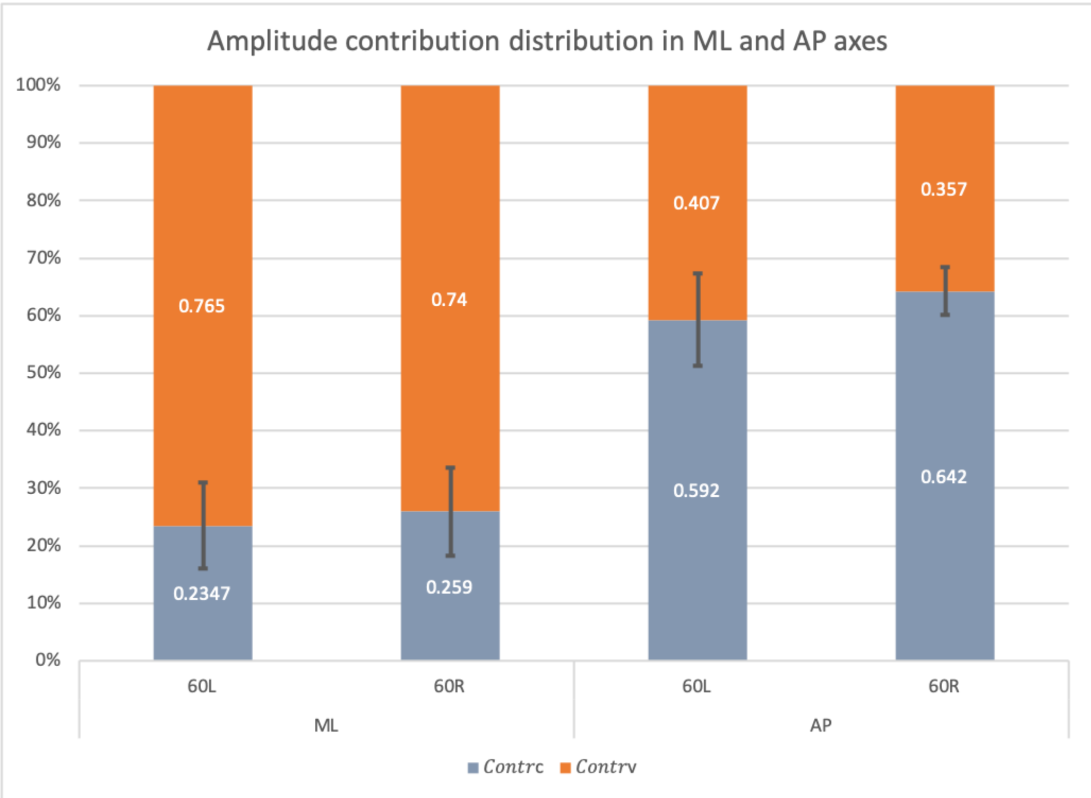

Figure 4.21. Mean amplitude contribution distribution and SD values of ankle and loading-unloading mechanisms in MLand AP direction for 60° stance

In the AP direction there is a strong correlation between COP_c and COP_{net} in 60L condition (.752 ± .172) and very strong correlation in 60R condition (.870 ± .072). The correlation between COP_v and COP_{net} on the other hand is weak and negative for 60L condition (-.158 ± .454) and moderate for 60R (-.455 ± .303). The difference between COP_c - COP_{net} and COP_v - COP_{net} was statistically significant (*60L :* Z = 2.672, p = 0.004, *60R :* Z = 4.171, p < .001). The activity of COP_{net} and two components (COP_c and COP_v) can be seen from a representative trial in Figure 4.22 and Figure 4.23.

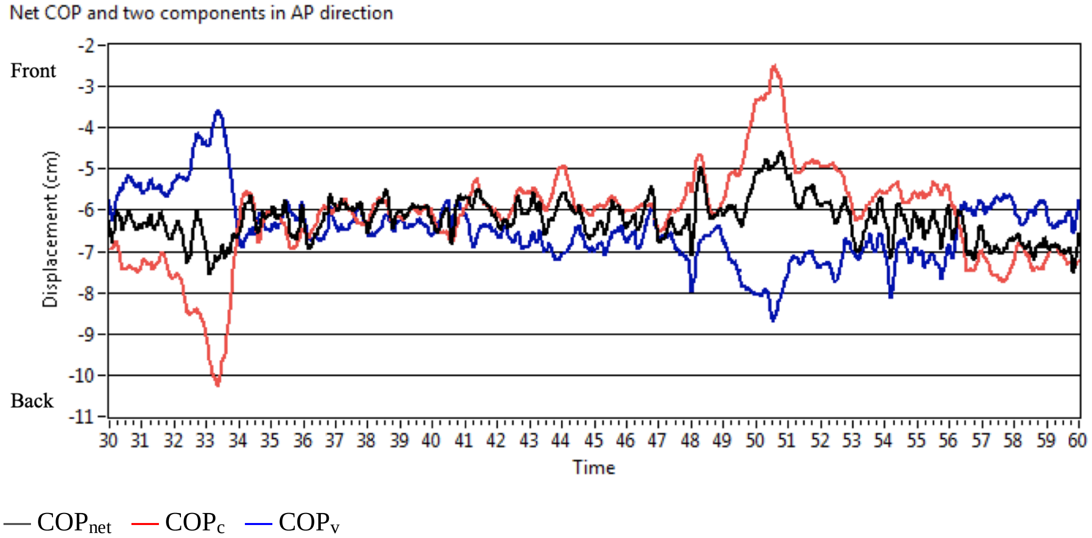

Figure 4.22. *Net COP and two components in AP direction for 60L (S13)*

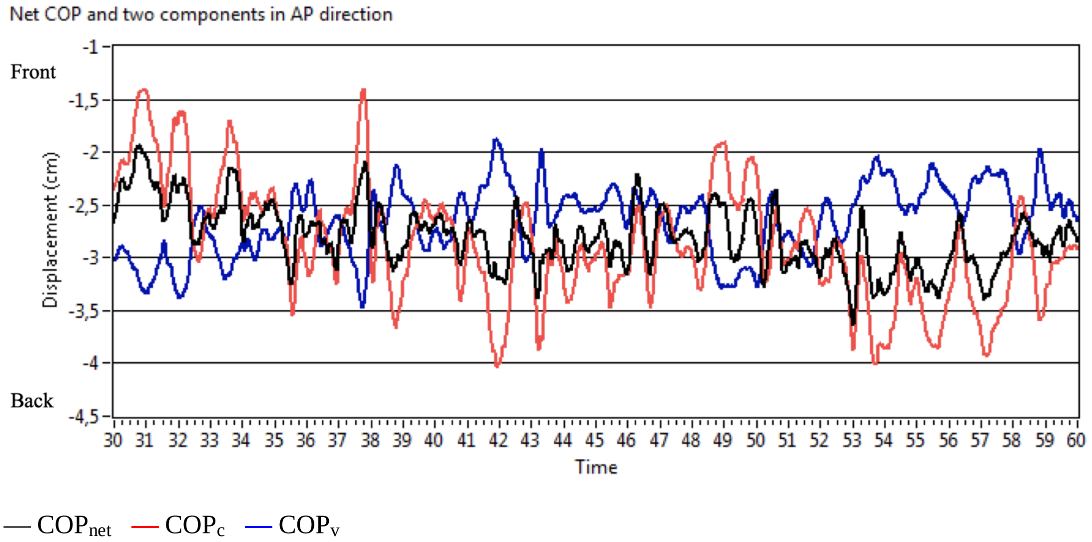

Figure 4.23. *Net COP and two components in AP direction for 60R (S22)*

Amplitude contributions of the ankle and loading-unloading mechanisms also suggests that ankle mechanism contributes more even though the difference is not that significant (Contr$_c$: *60L :* .593 ± .080, *60R :* .643 ± .042). Figure 4.21 presents the mean contribution coefficiets and standard deviations. The results of correlation and contribution coefficient analysis suggests that the primary control mechanism is ankle mechanism for AP direction.

Table 4.9. *Correlation values for all subjects in 60° stance*

	ML				AP			
	60L		60R		60L		60R	
	COP_{net} vs. COP_c	COP_{net} vs. COP_v	COP_{net} vs. COP_c	COP_{net} vs. COP_v	COP_{net} vs. COP_c	COP_{net} vs. COP_v	COP_{net} vs. COP_c	COP_{net} vs. COP_v
S1	0,892	0,995	0,681	0,951	0,868	-0,596	0,920	-0,343
S2	0,875	0,984	0,941	0,994	0,829	-0,001	0,950	-0,751
S3	0,783	0,942	0,874	0,978	0,869	-0,060	0,894	-0,472
S4	0,852	0,991	0,885	0,949	0,731	-0,294	0,904	-0,409
S5	-0,577	0,982	0,971	0,989	0,561	0,677	0,945	-0,759
S6	0,893	0,975	0,854	0,985	0,866	-0,355	0,920	-0,517
S7	0,886	0,995	0,918	0,988	0,831	-0,326	0,834	-0,196
S8	0,853	0,960	0,943	0,988	0,797	0,017	0,937	-0,645
S9	0,973	0,998	0,960	0,992	0,772	-0,408	0,962	-0,839
S10	0,182	0,963	0,887	0,996	0,557	-0,010	0,706	-0,304
S11	0,461	0,980	0,754	0,978	0,484	0,628	0,805	-0,203
S12	0,939	0,997	0,948	0,995	0,301	0,219	0,739	-0,441
S13	0,916	0,995	0,972	0,996	0,654	-0,151	0,841	-0,592
S14	0,961	0,995	0,982	0,996	0,901	-0,646	0,928	-0,744
S15	0,742	0,995	0,663	0,978	0,588	0,891	0,777	0,619
S16	0,649	0,967	0,954	0,991	0,830	0,042	0,795	-0,234
S17	0,905	0,977	0,836	0,992	0,899	-0,525	0,934	-0,684
S18	0,894	0,994	0,765	0,997	0,929	-0,661	0,849	-0,510
S19	0,946	0,983	0,920	0,969	0,855	-0,417	0,891	-0,374
S20	0,803	0,997	0,903	0,991	0,943	-0,805	0,877	-0,512
S21	0,938	0,987	0,839	0,993	0,891	-0,590	0,820	-0,501
S22	0,888	0,990	0,779	0,993	0,585	-0,096	0,895	-0,597
Mean	0,7574	0,9842	0,8745	0,9859	0,7522	-0,1578	0,8697	-0,4554
SD	0,3500	0,0146	0,0936	0,0134	0,1715	0,4541	0,0715	0,3031

Table 4.10. *Contribution coefficient values for all subjects in* 60° stance

	ML				**AP**			
	60L		**60R**		**60L**		**60R**	
	$Contr_{\mathrm{c}}$	$Contr_{\mathrm{v}}$	$Contr_{\mathrm{c}}$	$Contr_{\mathrm{v}}$	$Contr_{\mathrm{c}}$	$Contr_{\mathrm{v}}$	$Contr_{\mathrm{c}}$	$Contr_{\mathrm{v}}$
S1	0.17	0.83	0.30	0.70	0.62	0.38	0.71	0.29
S2	0.26	0.74	0.24	0.76	0.64	0.36	0.69	0.31
S3	0.35	0.65	0.30	0.70	0.67	0.33	0.66	0.34
S4	0.20	0.80	0.40	0.60	0.59	0.41	0.68	0.32
S5	0.18	0.82	0.38	0.62	0.47	0.53	0.67	0.33
S6	0.33	0.67	0.24	0.76	0.65	0.35	0.69	0.31
S7	0.17	0.83	0.28	0.72	0.63	0.37	0.64	0.36
S8	0.35	0.65	0.32	0.68	0.62	0.38	0.69	0.31
S9	0.21	0.79	0.30	0.70	0.59	0.41	0.67	0.33
S10	0.21	0.79	0.16	0.84	0.54	0.46	0.57	0.43
S11	0.18	0.82	0.24	0.76	0.47	0.53	0.62	0.38
S12	0.19	0.81	0.23	0.77	0.51	0.49	0.57	0.43
S13	0.19	0.81	0.25	0.75	0.57	0.43	0.60	0.40
S14	0.26	0.74	0.30	0.70	0.64	0.36	0.64	0.36
S15	0.13	0.87	0.22	0.78	0.36	0.64	0.56	0.44
S16	0.25	0.75	0.31	0.69	0.64	0.36	0.62	0.38
S17	0.33	0.67	0.19	0.81	0.66	0.34	0.68	0.32
S18	0.20	0.80	0.10	0.90	0.67	0.33	0.62	0.38
S19	0.36	0.64	0.39	0.61	0.63	0.37	0.67	0.33
S20	0.11	0.89	0.24	0.76	0.65	0.35	0.64	0.36
S21	0.31	0.69	0.17	0.83	0.65	0.35	0.60	0.40
S22	0.23	0.77	0.15	0.85	0.55	0.45	0.64	0.36
Mean	0.2347	0.765	0.259	0.74	0.592	0.407	0.642	0.357
SD	0.075	0.075	0.077	0.077	0.08	0.08	0.042	0.042

4.2. Effect of Stance Position on Contribution of the Mechanisms

Friedman Test was conducted to determine if there is an effect of stance position on contribution of mechanisms. In order to do so, right foot ahead sub-conditions and side-by-side stance and left foot ahead sub-conditions and side-by-side stance were paired as two groups, RFA and LFA, and for each group Friedman Test was applied for correlation coefficients and contribution coefficients. When an effect of stance position is identified, posthoc Wilcoxon Signed-Rank Test was conducted to identify which stance positions caused this effect.

4.2.1. ML Direction

For the activity of ankle mechanism, an effect of stance position is present (*LFA :* $\chi 2(4) = 59{,}891$, $p < .001$, *RFA :* $\chi 2(4) = 67{,}164$, $p < .001$). Posthoc analysis showed that there was no significant difference between 45L and 60L ($Z = -\ 0.061$, $p = .249$) and 45R and 60R ($Z = -1.899$, $p = .058$). All other conditions differed significantly from each other.

Table 4.11. *Test statistics for Wilcoxon signed-rank test for both LFA and RFA groups on correlation coefficients between COP_c and COP_{net} in ML direction*

	Z	Asymp. Sig. (2-tailed)		Z	Asymp. Sig. (2-tailed)
60L - TL	-4,107[b]	0.000	**60R - TR**	-4,107[b]	0.000
45L - TL	-4,107[b]	0.000	**45R - TR**	-4,107[b]	0.000
30L - TL	-4,107[b]	0.000	**30R - TR**	-4,107[b]	0.000
SbyS - TL	-4,107[b]	0.000	**SbyS - TR**	-4,107[b]	0.000
45L - 60L	-1,153[b]	0.249	**45R - 60R**	-1,899[b]	0.058
30L - 60L	-2,646[b]	0.008	**30R - 60R**	-3,523[b]	0.000
SbyS - 60L	-3,230[b]	0.001	**SbyS - 60R**	-3,977[b]	0.000
30L - 45L	-2,224[b]	0.026	**30R - 45R**	-2,613[b]	0.009
SbyS - 45L	-3,458[b]	0.0005	**SbyS - 45R**	-3,815[b]	0.000
SbyS - 30L	-1,997[b]	0.046	**SbyS - 30R**	-2,581[b]	0.010

a. Wilcoxon Signed Ranks Test
b. Based on positive ranks.

a. Wilcoxon Signed Ranks Test
b. Based on positive ranks.

For the activity of loading-unloading mechanism, again there is an effect of stance position adopted (*LFA :* $\chi2(4) = 44{,}836$, $p < .001$, *RFA :* $\chi2(4) = 44$, $p < .001$). Results of posthoc analysis is presented in Table 12. According to these results, the effect of stance position is arising mainly from tandem stance.

Table 4.12. *Test statistics for Wilcoxon signed-rank test for both LFA and RFA groups on correlation coefficients between COP_v and COP_{net} in ML direction*

	Z	Asymp. Sig. (2-tailed)		Z	Asymp. Sig. (2-tailed)
60L - TL	-4,107[b]	0.000	**60R - TR**	-4,107[b]	0.000
45L - TL	-4,107[b]	0.000	**45R - TR**	-4,107[b]	0.000
30L - TL	-4,107[b]	0.000	**30R - TR**	-4,107[b]	0.000
SbyS - TL	-3,652[b]	0.000	**SbyS - TR**	-3,490[b]	0.000
45L - 60L	-1,347[c]	0.178	**45R - 60R**	-,990[b]	0.322
30L - 60L	-2,062[c]	0.039	**30R - 60R**	-1,315[c]	0.189
SbyS - 60L	-1,899[c]	0.058	**SbyS - 60R**	-1,640[c]	0.101
30L - 45L	-1,737[c]	0.082	**30R - 45R**	-1,607[c]	0.108
SbyS - 45L	-,211[c]	0.833	**SbyS - 45R**	-1,964[c]	0.050
SbyS - 30L	-,438[b]	0.661	**SbyS - 30R**	-1,282[c]	0.200

a. Wilcoxon Signed Ranks Test
b. Based on negative ranks.
c. Based on positive ranks.

a. Wilcoxon Signed Ranks Test
b. Based on negative ranks.
c. Based on positive ranks.

There was an effect of stance position on the amplitude contribution of mechanisms (*LFA :* $\chi 2(4) = 49{,}709$, $p < .001$, *RFA :* $\chi 2(4) = 51{,}382$, $p < .001$) statistically significant difference in amplitude contribution of mechanisms contributions of mechanisms depending on which type of standing position is adopted,. Since there is a linear relationship between the contribution coefficients of ankle and loading unloading mechanisms (equation (3.17)), only contribution coefficient of ankle mechanism, i.e. $Contr_C$ is used in this analysis. Results of posthoc analysis revealed if there is a significant difference between each stance position to determine the origin of this stance effect, which are presented in Table 13.

Table 4.13. *Test statistics for Wilcoxon signed-rank test for both LFA and RFA groups on contribution coefficient $Contr_c$ in ML direction*

	Z	Asymp. Sig. (2-tailed)		Z	Asymp. Sig. (2-tailed)
60L - TL	$-4,107^b$	0.000	**60R - TR**	$-4,107^b$	0.000
45L - TL	$-4,107^b$	0.000	**45R - TR**	$-4,107^b$	0.000
30L - TL	$-4,107^b$	0.000	**30R - TR**	$-4,107^b$	0.000
SbyS - TL	$-4,107^b$	0.000	**SbyS - TR**	$-4,107^b$	0.000
45L - 60L	$-,730^c$	0.465	**45R - 60R**	$-2,516^b$	0.012
30L - 60L	$-,536^c$	0.592	**30R - 60R**	$-3,165^b$	0.002
SbyS - 60L	$-1,737^b$	0.082	**SbyS - 60R**	$-2,159^b$	0.031
30L - 45L	$-,438^c$	0.661	**30R - 45R**	$-1,120^b$	0.263
SbyS - 45L	$-1,997^b$	0.046	**SbyS - 45R**	$-,958^b$	0.338
SbyS - 30L	$-2,062^b$	0.039	**SbyS - 30R**	$-,016^b$	0.987

a. Wilcoxon Signed Ranks Test
b. Based on positive ranks.
c. Based on negative ranks.

a. Wilcoxon Signed Ranks Test
b. Based on positive ranks.

4.2.2. AP Direction

For the activity of ankle mechanism, an effect of stance position is evident (*LFA :* $\chi2(4) = 85,927$, $p < .001$, *RFA :* $\chi2(4) = 84,473$, $p < .001$). Posthoc analysis revealed that the difference between all stance positions are statistically significant, except 45R and 60R ($p = .002$).

Table 4.14. *Test statistics for Wilcoxon signed-rank test for both LFA and RFA groups on correlation coefficients between COP_c and COP_{net} in AP direction*

	Z	Asymp. Sig. (2-tailed)		Z	Asymp. Sig. (2-tailed)
60L - TL	-4,107[b]	0.000	**60R - TR**	-4,107[b]	0.000
45L - TL	-4,107[b]	0.000	**45R - TR**	-4,107[b]	0.000
30L - TL	-4,107[b]	0.000	**30R - TR**	-4,107[b]	0.000
SbyS - TL	-4,107[b]	0.000	**SbyS - TR**	-4,107[b]	0.000
45L - 60L	-3,328[b]	0.0009	**45R - 60R**	-3,068[b]	0.002
30L - 60L	-4,107[b]	0.000	**30R - 60R**	-4,107[b]	0.000
SbyS - 60L	-4,107[b]	0.000	**SbyS - 60R**	-4,107[b]	0.000
30L - 45L	-4,107[b]	0.000	**30R - 45R**	-3,945[b]	0.000
SbyS - 45L	-4,107[b]	0.000	**SbyS - 45R**	-4,107[b]	0.000
SbyS - 30L	-4,107[b]	0.000	**SbyS - 30R**	-4,107[b]	0.000

a. Wilcoxon Signed Ranks Test
b. Based on negative ranks.

a. Wilcoxon Signed Ranks Test
b. Based on negative ranks.

For the activity of loading-unloading mechanism, again there is an effect of stance position adopted (*LFA :* $\chi2(4) = 42{,}145$, $p < .001$, *RFA :* $\chi2(4) = 42{,}145$, $p < .001$). Results of posthoc analysis is presented in Table 15. According to these results, tandem stance and side-by-side stance were significantly different from almost all other stance positions. There were also significant differences between some intermediate stance positions as well (refer to Table 15).

Table 4.15. *Test statistics for Wilcoxon signed-rank test for both LFA and RFA groups on correlation coefficients between COP_v and COP_{net} in AP direction*

	Z	Asymp. Sig. (2-tailed)		Z	Asymp. Sig. (2-tailed)
60L - TL	-3,945[b]	0.000	**60R - TR**	-4,074[b]	0.000
45L - TL	-4,107[b]	0.000	**45R - TR**	-4,107[b]	0.000
30L - TL	-4,107[b]	0.000	**30R - TR**	-4,074[b]	0.000
SbyS - TL	-2,808[b]	0.005	**SbyS - TR**	-3,425[b]	0.0006
45L - 60L	-1,802[b]	0.072	**45R - 60R**	-,016[c]	0.987
30L - 60L	-2,224[b]	0.026	**30R - 60R**	-1,704[b]	0.088
SbyS - 60L	-1,185[c]	0.236	**SbyS - 60R**	-3,393[c]	0.0007
30L - 45L	-1,023[b]	0.306	**30R - 45R**	-1,607[b]	0.108
SbyS - 45L	-2,451[c]	0.014	**SbyS - 45R**	-3,036[c]	0.002
SbyS - 30L	-2,938[c]	0.003	**SbyS - 30R**	-3,328[c]	0.0009

a. Wilcoxon Signed Ranks Test
b. Based on positive ranks.
c. Based on negative ranks.

a. Wilcoxon Signed Ranks Test
b. Based on positive ranks.
c. Based on negative ranks.

There is an effect of stance position on the amplitude contribution of mechanisms (*LFA :* $\chi 2(4) = 83,382$, $p < .001$, *RFA :* $\chi 2(4) = 84,109$, $p < .001$). Again, due to the linear relationship between contribution coefficients of ankle and loading-unloading mechanism (equation (3.17)), only $Contr_c$ is used in this analysis. Posthoc analysis revealed that all stance positions are different in terms of amplitude contributions of the mechanisms and this difference is statistically significant.

Table 4.16. *Test statistics for Wilcoxon signed-rank test for both LFA and RFA groups on contribution coefficient $Contr_c$ in AP direction*

	Z	Asymp. Sig. (2-tailed)		Z	Asymp. Sig. (2-tailed)
60L - TL	-3,815[b]	0.000	**60R - TR**	-4,107[b]	0.000
45L - TL	-4,107[b]	0.000	**45R - TR**	-4,107[b]	0.000
30L - TL	-4,107[b]	0.000	**30R - TR**	-4,107[b]	0.000
SbyS - TL	-4,107[b]	0.000	**SbyS - TR**	-4,107[b]	0.000
45L - 60L	-3,880[b]	0.000	**45R - 60R**	-4,042[b]	0.000
30L - 60L	-4,107[b]	0.000	**30R - 60R**	-4,107[b]	0.000
SbyS - 60L	-4,107[b]	0.000	**SbyS - 60R**	-4,107[b]	0.000
30L - 45L	-4,074[b]	0.000	**30R - 45R**	-4,074[b]	0.000
SbyS - 45L	-4,107[b]	0.000	**SbyS - 45R**	-4,074[b]	0.000
SbyS - 30L	-4,074[b]	0.000	**SbyS - 30R**	-3,977[b]	0.000

a. Wilcoxon Signed Ranks Test
b. Based on negative ranks.

a. Wilcoxon Signed Ranks Test
b. Based on negative ranks.

4.3. Comparison of Right Foot Ahead and Left Foot Ahead Conditions

4.3.1. ML Direction

In terms of correlation coefficients both 30L and 30R, 45L and 45R sub-conditions are significantly different pairwise according to the correlation coefficients of loading-unloading mechanism, i.e. COP_V (30 ° Stance : Z = -2.224, p =.026; 45 ° Stance : Z = -1.217, p = .223). However this difference does not exist in the activity of the ankle mechanism.

In terms of contribution coefficients there is a statistically significant difference between 30L and 30R (Z = -2.678, p = .007) sub-conditions. All other sub-conditions were the same pairwise and no effect of left forward or right forward is seen. All the information together means that on 30 ° stance condition an effect of preferred support foot is evident. For this condition even though the results are in the same direction for 30L and 30R, these two sub-conditions are not same in terms of the postural control in ML direction.

Table 4.17. *Test statistics for Wilcoxon signed-rank test between right foot ahead and left foot ahead conditions, comparing COP_c, COP_v and $Contr_c$ values in ML dirction*

	COP_C		COP_V		$Contr_C$	
	Z	**Asymp. Sig. (2-tailed)**	**Z**	**Asymp. Sig. (2-tailed)**	**Z**	**Asymp. Sig. (2-tailed)**
TL - TR	-,828[b]	0.408	-,276[c]	0.783	-1,055[b]	0.291
60L - 60R	-,893[b]	0.372	-,698[b]	0.485	-1,055[b]	0.291
45L - 45R	-,860[b]	0.390	-2,841[b]	0.005	-1,899[c]	0.058
30L - 30R	-,308[b]	0.758	-2,224[b]	0.026	-2,678[c]	0.007

a. Wilcoxon Signed Ranks Test
b. Based on negative ranks.
c. Based on positive ranks.

4.3.2. AP Direction

In terms of correlation coefficients of the ankle mechanism, i.e. COP_c, TR and TL were found to be different ($Z = -4.107$, $p < .001$). However, this difference does not exist in loading-unloading mechanism. 60L and 60R sub-conditions were found different in terms of both ankle mechanism and loading-unloading mechanism activities according to the correlation coefficient comparison (COP_c : $Z = -3.003$, $p = .003$; COP_V : $Z = -2.971$, $p = .003$). 30L and 30R sub-conditions are different in terms of the activity of the loading-unloading mechanism, COP_V ($Z = -2.033$, $p = .042$) but this effect is not seen for ankle mechanism activity.

In terms of contribution coefficients, only the difference between 60L and 60R sub-conditions is significant (Z = -2.938, p = .003). All results together, it can be said that 60L and 60R sub-conditions are different experiences for the subjects in terms of postural control in AP direction.

Table 4.18. *Test statistics for Wilcoxon signed-rank test between right foot ahead and left foot ahead conditions, comparing COP_c, COP_v and $Contr_c$ values in AP dirction*

	COP_c				**COP_v**	
	Z	**Asymp. Sig. (2-tailed)**	**Z**	**Asymp. Sig. (2-tailed)**	**Z**	**Asymp. Sig. (2-tailed)**
TL - TR	-4,107[b]	0.000	-1,964[c]	0.050	-1,282[b]	0.200
60L - 60R	-3,003[c]	0.003	-2,971[b]	0.003	-2,938[b]	0.003
45L - 45R	-1,380[c]	0.168	-1,477[b]	0.140	-,828[b]	0.408
30L - 30R	-1,686[c]	0.092	-2,033[b]	0.042	-,925[b]	0.355

a. Wilcoxon Signed Ranks Test
b. Based on positive ranks.
c. Based on negative ranks.

4.4. Investigation of Effects of Dominant Extremity

Since the number of left handed/footed people are really small, this statistics does not have the power to imply anything significant on the effect of dominant extremity.

4.4.1. ML Direction

There is no difference between conditions for neither ankle mechanism nor loading-unloading mechanism in terms of correlation coefficients (p>.001).

A statistically significant difference between left handed and right handed people was observed in terms of contribution coefficients, only in TR sub-condition. It can be said that the effect of handedness is evident only in tandem stance condition and only in terms of contribution coefficients. However, since this effect is only observed for contribution coefficients and since participant number is small, such a conclusion is hard to make.

4.4.2. AP Direction

There was no difference between conditions for neither ankle mechanism nor loading-unloading mechanism in terms of both correlation and contribution coefficients ($p>.001$).

4.5. Investigation of Effects of Gender

Gender effect is investigated with Mann Whitney U Test for both ML and AP directions and for both correlation and contribution coefficients for all stance positions. An effect of gender is seen in side-by-side stance position in ML direction only for correlation coefficient between COP_v - COP_{net} ($p<.001$). This difference is not observed for correlation between COP_c - COP_{net} and contribution coefficients or in AP direction. Same as side-by-side stance, for 60L sub-condition also an effect of gender is evident only for correlation coefficient between COP_v - COP_{net} ($p<.001$) in ML direction. This difference is not observed for 60R sub-condition.

As in all investigations, in order to conclude that there is a significant effect of gender, it is necessary to have this difference in at least contribution coefficients as well, which is not the case. In general, it can be said that gender does not have an effect on the selected contribution mechanism.

CHAPTER 5

DISCUSSION AND CONCLUSION

The aim of this study is to discover the contribution of the ankle and loading-unloading mechanisms, the two mechanisms of upright stance, to the postural control while various stance positions are adopted by individuals.

The study is setup based on the expectation that there should be a relationship of the proportional contribution of each mechanism between similar stance conditions. Accordingly, hypotheses of this thesis were established as follows:

H1: Postural control mechanisms are affected from the emplacement of the feet and the seen effect is not arbitrary, rather it has a pattern between related conditions.

H2: The intermediate positions adopted by the participants will be controlled in such a way that the dominant control mechanism will be similar to the most lookalike border condition.

5 sub-hypotheses were also tested under the second hypothesis as listed below:

1. In side-by-side stance, the AP balance will be in the control of ankle mechanism and ML balance will be in the control of body weight distribution mechanism.
2. In tandem stance, the AP balance will be controlled by body weight distribution mechanism and ML balance will be controlled dominantly by ankle mechanism.

3. In the exact 45° intermediate stance position (the angle is defined as the angle between x-axis and the line passing through medial malleoli of the feet), both AP and ML plane postural controls will be achieved with the almost equal contribution of both mechanisms.
4. In the 30° intermediate position, the postural control will be under the dominance of ankle and body weight distribution mechanisms for AP and ML directions respectively. The behaviour will be in accordance with the most similar border condition side-by-side stance.
5. In the 60° intermediate position, the behaviour will be close to the most similar border condition, tandem stance. Dominance of body weight distribution mechanism in AP direction and dominance of ankle mechanism in ML direction are expected.

Control of posture in quiet stance has been investigated for different stance positions by analyzing the COP signals of 22 young adults. By collecting COP signals with two separate force platforms, two different mechanisms can be identified. Depending on the stance position taken many combinations of an ankle and a hip loading unloading mechanism are evident. In order to observe these contributions, several procedures were carried out. First, COP signals under each foot were used to extract hip and ankle components. Second, cross correlation analysis were carried out to present degree of activity of each mechanism by comparison to net COP signals. Meantime, contribution coefficients, i.e. amplitude contributions of each mechanism were calculated. In order to see the effects of different stance positions on activity and contribution of each mechanism, additional statistical analyses were carried out for comparison.

After all analyses were completed, it has been unveiled that original hypothesis were only partially true. As predicted, postural control mechanisms are affected from the emplacement of the feet and there really was a pattern among conditions however the combination of the mechanisms are not as simple as previously predicted. Rather than just a similarity between related conditions, a supportive mechanism has been revealed.

5.1. The Control Mechanisms of Each Stance Condition

For side-by-side stance, as predicted, ankle control has been found dominant in the control of AP balance and the contribution of hip activity was found negligible. The hip control over AP balance was found variable amongst subjects. Even though it is negligible for most of the subjects, a high standard deviation was calculated. The effect of hip loading-unloading mechanism is sometimes additive; sometimes subtractive. Some subjects show high activity and contribution of ankle mechanism and some show almost none. Even in subjects with high participation of hip control, ankle activity and contribution matches the total COP changes almost 100%.

ln ML direction, the hypothesis stating the prediction of hip loading unloading control being dominant is also correct. A small or negligible control was expected for the secondary mechanisms of side-by-side stance in both AP and ML directions, and analysis shows a weak contribution of ankle mechanism, i.e. the secondary mechanism in ML direction. According to the results, different than AP direction, many subjects used their ankle mechanism in order to keep standing still. For sure, even these subjects significantly used their hip loading-unloading mechanism over ankle but they used ankle mechanism as a supporting mechanism. Since ankle evertor invertor joints are not lined up, activity of ankle is necessarily a supportive activity to possibly prevent the over load-unload in a limb with eversion/inversion activity. Overall, sub-hypothesis for side-by-side stance is correct.

For tandem stance, it was expected that in contrary to side-by-side stance AP balance would be controlled by hip load-unload mechanism since the ankle dorsi/plantar flexor joints are not aligned anymore. In this condition it is much easier to transfer the body weight from one limb to another in order to maintain balance in the AP direction, the direction in which feet are aligned. Thus, it makes more sense to transfer the load from one limb to other instead of a toes to heel COP location control. Additionally, a toes to heel control is harder since feet are not aligned in the dorsal/plantar flexion direction. As expected, AP balance is controlled by hip load-unload mechanism. However, the difference between activity of the mechanisms were not found significant. Though, this was expected for sure since it was reported in the literature as well (Winter et al, 1996).

Contribution coefficient analysis yielded a similar result. Hip dominance over ankle is quite small (%51 - %49), yet above the threshold. Since most of the subjects show a significant hip load-unload control and negligible ankle control, it is possible to say that hip load-unload control is used more dominantly to control AP balance, as reported by the previous studies.

It should be noted that the variations are not small. Thus, it can be said that the control of tandem stance is a hard task to succeed and subjects might tend to use all their assets to fulfill this task. Consequently, contrary to side-by-side stance, these challenging tasks have a high standard deviation among subjects in terms of the preferred balance control solution. Even though all the subjects use high amounts of hip load-unload as a solution, the degree of activity and contribution of the secondary mechanism changes drastically amongst subjects.

This may lead one to think about the effects of novelty or difficulty of the task or else, difficulty as a result of the novelty of the task. Since tandem stance is an unusual task for all subjects, contrary to side-by-side stance the reaction of subjects are very different from each other in terms of secondary mechanism. Some subjects had to interfere the control of hip load-unload mechanism more to control balance in an additive or subtractive manner where some did not have to. These differences may be a result of a conscious control even though the experiment is designed in such a way to create an external focus of attention. This might also be a total automatic response created according to what body needs in that specific time of event. Unfortunately, with this study design we do not have information on the exact cause of these differences among subjects.

On the other hand, in ML direction for tandem stance, it was expected that ankle control will be dominant with a small or negligible contribution of hip control. As expected, ankle control dominates ML balance. Where hip load-unload mechanism is moderately active, its contribution is very weak when the amplitude contribution is investigated. The standard deviation of cross-correlation of COP_c and COP_{net} is very small. Variation between subjects is very small in terms of use of ankle control. However, the standard deviation is higher between cross-correlation coefficients of COP_v and COP_{net} yet, this does not have a counterpart in terms of amplitude contribution of the hip load-unload mechanism. This means that even though all subjects used ankle mechanism for ML balance control, the use of secondary mechanism again changes among subjects. Some are additive, some are subtractive or some random, even though the magnitude of the activity of hip load-unload mechanism is not high enough, the secondary mechanism is important for some subjects. This could be interpreted as there are individual differences and as a result there are different supportive behaviors in novel tasks.

For intermediate positions, it was assumed both AP and ML balances will be controlled by a mixed ankle and hip mechanism. For the 45º stance position, it was expected that neither ankle nor hip mechanism will be dominant, rather there will be almost equal contribution from both. For 30º and 60º on the other hand, a behaviour more similar to tandem and side-by-side stance respectively was expected. According to the results of this study, this hypothesis was not fully supported.

For 30º stance condition, AP balance is dominantly controlled by the ankle mechanism as expected. The activity and contribution of ankle mechanism is high. In addition, hip load-unload mechanism is also active. In fact, it is weak in overall but for many subjects its activity is moderate to strong. Yet, its strength is very weak.

Another important thing to note that hip mechanism is subtracting from the activity of ankle mechanism. This means that hip mechanism is used to diminish the activity of ankle. On the other hand, again there is a high variation among subjects for the secondary mechanism but not in degree of activity of primary mechanism, which may indicate that primary mechanisms are automatic responses and do not change amongst subjects and they are predictable, yet secondary mechanisms are some kind of a back-up, maybe not automatic or if it is automatic can change considerably among subjects. Even if this stance condition is the one which more looks like side-by-side stance compared to other conditions, at the end there are important differences. The key characteristic of side-by-side stance is that dorsal-plantar flexor joints are aligned. Since this alignment is broken by the new positioning of the feet, in this condition we see a higher activity and contribution of hip mechanism, and this contribution is in a way to diminish activity of ankle with the hip activity.

In the ML direction, as predicted, hip loading-unloading mechanism is dominant with a contribution of the ankle mechanism. Variation among subjects is higher for the ankle mechanism and very small for the loading-unloading mechanism. Since the degree of activity for the hip mechanism is almost the same for all subjects, it is clear that the hip mechanism for this position is the primary control mechanism of the ML balance. Like in all other conditions, variation among subjects highly increases for the secondary mechanism. This again indicates that the ankle mechanism is a support for the subjects and can be used as a backup. For most of the subjects, ankle mechanism is moderately to strongly active and in terms of magnitude it has an additive however small contribution. Thus even it does not look like as essential as the hip loading-unloading mechanism in this condition, it is still highly needed to support loading-unloading mechanism for most of the people which were healthy adults.

For 45º stance position, for both ML and AP directions the behaviour was not completely as expected. It was expected that for 45º position both mechanisms would almost equally contribute. In the AP direction the dominant control mechanism is the ankle. However, there is also a weak contribution from the hip mechanism. A negative weak activity from hip mechanism with again a weak magnitude exists. Hip mechanism is trying to cancel some of the effects of the ankle mechanism. Variation among subjects is similar to other conditions. Clearly, the primary mechanism for 45º condition in AP direction is the ankle mechanism but hip mechanism acts to reduce the effects of ankle mechanism. When compared to the hypothesis of both mechanisms will be active and contribute almost equally, turned out to be incorrect.

In the ML direction, hip mechanism is dominant. The correlation coefficient between COP_c and COP_{net} is also strong despite variations among subjects. In contrast with the high amount of activity of ankle mechanism, the amplitude contribution is small. Yet, it reinforces the hip mechanism. At least, it is the case for most of the subjects. Again, similar to other conditions, variation among subjects is high for the ankle mechanism, i.e. the secondary mechanism of this condition. Thus the reinforcement effect of ankle mechanism is probably evident in most people but it is not solid like the apparent control of hip mechanism.

For 60º intermediate stance, it was predicted to be similar to tandem stance, in other words, ankle mechanism would be dominant for ML direction and hip mechanism would be dominant for AP direction, with a small contribution of the other. Of course, both the activity and contribution of primary mechanism was predicted to be smaller and activity and contribution of the secondary mechanism was predicted to be higher than tandem stance. However, it comes out that this is not the case. Actually, it is almost the opposite of what was expected.

In AP direction, the dominant mechanism is the ankle. There is a negative and weak activity of the hip mechanism. Its contribution is weak yet higher than both 45º and 30º stance conditions. Variation in ankle control, primary control, is small, and variation in hip control is large as in all secondary mechanisms as it was seen through this chapter.

In ML direction, hip mechanism is the dominant control mechanism. Ankle mechanism also shows a strong to very strong activity. Variation of both ankle and hip mechanism are found to be small for 60R condition. In 60L condition ankle mechanism showed a high variation. Nevertheless there was a statistically significant difference between correlation coefficients of ankle and hip loading-unloading mechanisms in both conditions. In terms of amplitude contribution, again same result is achieved. Hip loading-unloading is the mechanism which controls the ML balance and ankle also has a weak contribution to the ML balance in 60º stance position.

5.2. The Effects of Distance Between Feet

Before further going into the details of this chapter, an important thing to be noted is the comparison of each condition with each other.

The expected result was finding significant differences between each stance condition based on the hypotheses. Unfortunately, this is not always the case. Particularly speaking, for ML direction, in terms of primary mechanism, i.e. hip loading-unloading mechanism, no significant difference was found between stance conditions, except tandem stance. Taking into account that for all of these conditions a very strong activity of hip mechanism was present, the result is probably related to the stance width, in other words, the distance between feet in horizontal direction. In the light of previous studies, it is known that stance width actually affects the activity of balance control mechanisms. Despite this, in a study by Kirby et al. (1987), while increasing the distance further, the effect of stance width on activity was not that visible after some point. This effect of course, is not definite since there are also studies suggesting the opposite. However, even though we changed the stance width in different stance conditions, the change in the activity of primary control mechanism not getting affected from this distance change in ML direction might be the reason of this result. Another thing to note is secondary mechanisms actually affected from the stance width since there was significant differences between stance conditions. At this point, it is also important to note that the study carried by Kirby et al. (1987) does not take into account that there is a secondary mechanism.

On the other hand, in the AP direction, the degree of activity of primary control mechanisms were significantly different from each other for all stance conditions. For secondary mechanisms, there were some non-significant results for some conditions, however these effects were not visible for both right foot ahead and left foot ahead conditions.

The results at the end, does not exactly in accordance with the hypothesis. But actually, they are in the same direction with the results presented in some previous studies. The reason might be that hypotheses were not properly stated. In order to give all of the results presented a meaning by means of the hypotheses, we should discuss the results of intermediate conditions in detail starting from 45º and 60º conditions because they were the ones which clearly do not satisfy the expectations. Investigating stance conditions together with a closer look will make it easier to understand the actual pattern between all conditions and to compare them with previous studies and reveal the actual contribution of the results of this study.

5.3. The Relationship Among Different Stance Conditions

For all intermediate conditions, what was expected was participation of both mechanisms in the control of stance rather than an obvious lead without the other. Even though there is definitely a dominant mechanism, we can say that for all intermediate conditions, compared to side-by-side stance and tandem stances we can say that both mechanisms participated and without one another the balance of the subject would be struggled. This is hard to say for especially side-by-side condition since the effect of the secondary mechanisms were so small in general.

However if we take a look at the intermediate conditions, starting from 30º condition, we can clearly say that balance in these conditions is not possible without the secondary mechanism at all. This statement can be more pronounced for AP direction.

In the AP direction ankle mechanism is clearly the dominant control mechanism of postural control as it is in side-by-side stance which was anticipated due to the reason that 30º stance position is the more similar condition to side-by-side stance. A decrease in the amount of activity and amplitude contribution of ankle mechanism, increase in the activity of hip loading-unloading mechanism was also expected. However, the important part is how the secondary mechanism contibutes to this control.

Different than predicted, ankle mechanism is over activated in this situation. Insomuch that, there is an opposing activity of hip loading-unloading mechanism in order to cancel the over displacement of COP caused by ankle dorsi/plantar flexor activity. Without this negative contribution of loading unloading mechanism, COP_{net} would be out of optimal range which would lead to an instability or to an excessive oscillation.

Another important point to note here is the degree of activity of hip loading-unloading mechanism. Even though the amplitude contribution is very small, degree of activity is between weak to moderate. This amount of activity is actually quite big if we take into account that in side-by-side stance there is almost no activity of the hip for the ML direction. Thinking that, this particular condition, one foot being a little ahead of the other or if we express differently, breaking the ankle plantar/dorsal flexor alignment, totally changes the preferred postural control behavior. Yet, this much of activity of hip is not surprising, it being in a subtractive manner was not estimated.

When we look at the results of ML balance in 30° stance condition we see that, as predicted the control of ML balance is under the influence of hip loading-unloading mechanism. The activity of ankle mechanism is also quite high compared to the side-by-side-stance condition. This shows us that, as hypothesized even though the hip loading-unloading mechanism is still the dominant control mechanism with very strong activity, a support from the ankle mechanism is also needed to keep the COP_{net} in optimum position.

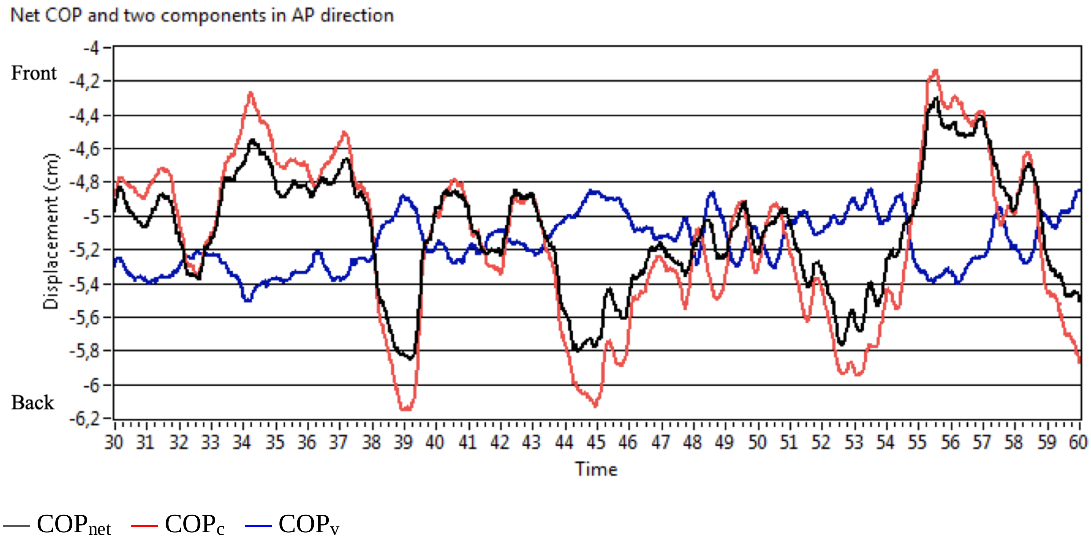

Figure 5.1. A general look to the subtractive behavior of intermediate conditions.

If we follow through the 45° stance position in the light of the findings on 30° condition, different than expected we see a behavior similar to 30° condition. The hypothesis was to have involvement of both mechanisms for sure, but almost in the same amount of activity was predicted.

In the AP direction ankle mechanism was very active and also showing high amplitude contributions which makes it the primary control mechanism for this stance position. Net COP change however, is not a sum of both mechanisms but it is rather a subtraction of these mechanisms as we saw already in the control of 30º stance position. With the over activity of ankle dorsal/plantar flexors, ankle mechanism exerts a high amount of change in the displacement where loading-unloading mechanism causes an opposing displacement change to keep the COP_{net} at a proper position. This subtractive cooperation between the two mechanisms actually shows us the importance of the secondary mechanism in a more visible, obvious way than the additive behaviour we see in ML control. This effect is also seen in 60º condition which will be detailed in the following paragraphs.

Another essential point, the degree of activity of the secondary mechanism, i.e. hip loading-unloading mechanism decreases compared to 30º condition. The reason behind this is actually directly related to the activity of ankle mechanism. Decrease in the activity of ankle mechanism from 30º to 45º results with a decrease in the activity of hip loading-unloading mechanism as well. If there is not an excessive amount of displacement caused by the over activity of ankle dorsal/plantar flexor muscles, apparently there is no need for more hip activity.

When we look at the results for ML direction, it can be said that the results were supporting our hypothesis. Because both mechanisms showed strong correlations and they were both additive and reinforcing the COP_{net} even though the dominant mechanism was contributing more in terms of amplitude, the resulting COP_{net} was sum of both.

Again for this condition there was a clear relationship between 30º condition as well. In 30º condition, the activity of the secondary mechanism, i.e. ankle mechanism, was lower than 45º condition. When we increased the angle between feet and move further away from side-by-side stance, the secondary mechanism needed to support hip loading-unloading mechanism more than before both in terms of amplitude and activity contributions.

For 60º stance condition in the AP direction we see a decrease in the degree of activity of ankle mechanism. By looking at the hypothesis, we were expecting to see more involved hip loading-unloading mechanism. Actually, this is not the case. Ankle mechanism is still over activated as we see in 45º condition. Consequently, this results with an opposite response from the hip loading-unloading mechanism to keep the COP_{net} in position as we see in 45º condition. Even though the degree of activity of ankle mechanism is more than enough to keep the COP_{net} in a steady position, it is lower than the activity we see in 45º condition. As a result, contrary to expectations of the hypothesis of seeing higher activity at hip as we approach tandem stance, we see an activity even less than 45º condition. This result is completely logical since the activity of ankle mechanism is less than in 45º position but still large for a steady stance, hip loading-unloading mechanism needs to subtract from that but this time the amount of activity needed is less than the previous 45º condition.

This may lead one to think that even if we further increase the angle between feet, we may not see hip loading-unloading mechanism as an absolute primary mechanism. Actually even in tandem stance position we do not see an obvious dominant hip loading-unloading mechanism, which will be detailed further while discussing the results of tandem stance.

In the ML direction, again we see strong, positive correlations for both mechanisms which means they both are very active in the control of ML balance and they act in an additive manner. This was expected for an intermediate condition. However the unexpected part is the role of hip loading-unloading mechanism in the control of ML balance.

The results reveal that ML balance is primarily controlled by hip loading-unloading mechanism, which was not predicted. Although the participation of ankle mechanism is also very high in this case, since this stance condition is the closer one to tandem stance, ankle mechanism was expected to be more dominant in this situation. However results are similar to 45º degree condition. It can be inferred that this stance condition is similar to 45º stance rather than tandem stance. Yet, more visible change towards tandem stance was anticipated. Maybe using an intermediate position like a 75º stance position in this case which is a more similar stance to tandem stance compared to 60º stance condition. This was the limitation of the set-up with the embedded force plates, which will be explained later in this chapter. Still, even in that case we might be getting similar results as we got in 60º stance case. The reason behind this thought is emerged out of the results of similar behaviors between all intermediate stance conditions.

In this stance position we are seeing that still the hip loading-unloading mechanism is the primary control mechanism of ML balance instead of ankle mechanism. Even though there is not a decrease in terms of the degree of activity of hip loading-unloading mechanism, there is an increase in the degree of activity of the ankle mechanism from 45º to 60º. This possibly suggests that, the closer the position to tandem stance, surely the amount of contribution of the ankle mechanism will increase even if this increase is not statistically significant. These kind of small changes may be interpreted as even if the angle between feet is further increased, the results would not be changing dramatically. Yet, it is still possible to come to conclusion that ankle mechanism would increase its activity as the position gets closer and closer to tandem stance.

Again, it is also important to state that as already mentioned in the previous paragraph for ML direction, actually in both AP and ML directions, 45º and 60º stance positions did not show significant differences even though we see a change in the amount of use of secondary mechanisms by increasing the angle between feet. This result is also related to the experimental setup, which we are going to take a look later in this chapter.

All the discussion made so far may imply that, maybe the relationship expected in this study does not only come from the similarity between conditions but also a more potent change which has a huge impact is needed. Which is actually the alignment of joints which can be seen from the results of tandem stance in ML direction.

For tandem stance, the balance control in the ML direction is clearly under the control of the ankle mechanism both in terms of activity and amplitude contributions. Hip loading-unloading mechanism also contibutes to this control in an additive way. Though the degree of activity is moderate, the amplitude of the contribution is almost negligible. If these results were compared with the control of balance in the ML direction for intermediate stance conditions, up until this point, i.e. tandem stance, what is observed was hip mechanism being dominant all the way up to tandem stance without even decreasing its activity at all. On the other hand, as the stance condition gets closer to tandem stance, ankle mechanism slowly increases its activity. However, only after aligning the feet in the vertical direction, the ML control completely governed by ankle mechanism both in terms of activity and amplitude contributions.

At the end there is a change implying that results are getting closer to the tandem position by increasing the angle between feet from 45º to 60º, even from 30º to 60º which was possible to read from the increasing participation of secondary mechanisms. However, the transition between intermediate conditions and tandem stance was not as smooth and gradual as it was predicted.

To elaborate, it looks like aligning ankle invertor evertor joints in tandem stance has stronger impact on the selection of controlling mechanisms than the gradual changes in the stance position, which is understandable. Aligning feet results with a clearer dominant mechanism identification in both AP and ML directions. By aligning feet in eversion/inversion direction, the gradual increase in the participation of ankle mechanism and dominance of hip mechanism totally changes and hip mechanism steps out and gives the control to the ankle mechanism completely.

Discussing the results of AP direction, in tandem stance there is not a significant dominant mechanism, but the hip load-unload mechanism is more active. For this case, comparing it with the changes in the results from 30º to 60º, in intermediate conditions, the decrease in the participation of ankle gradually both in terms of activity and amplitude contributions is seen, but the hip activity does not actually increase, rather decreases together with the decrease in the ankle activity. Yet, a gradual increase in the amplitude contribution of hip mechanism is present.

When the angle is further increased from 60º to tandem, even though it is observed that ankle mechanism leaves the stage and gives its place to hip mechanism, it does not happen gradually. Indeed the change from the behavior used in intermediate conditions to tandem is completely different. Tandem stance is controlled by both mechanisms with the higher contribution of loading-unloading mechanism in terms of both activity and amplitude contributions in an additive manner while these mechanisms subtract each other in intermediate conditions. By looking at this behavioral change, transition between intermediate conditions and tandem stance is more gradual in postural control of ML balance.

Significant difference between the activity contribution of two mechanisms was expected in this condition in alignment with the previous studies instead of seeing hip contribution and activity higher but not more significant than ankle. In any case, according to this result it can be said that, not being in alignment of neither ankle nor hip joints in AP direction might be the reason prevented one of the mechanism to take the lead.

Another reason might be that, since the stance surface was in its smaller state in tandem for ML direction, the joint activities were more focused to control this struggling condition rather than spending the energy in comparably easier direction which is AP direction with two feet are one in front of the other which creates a quite big base of support (even though not the biggest in all conditions). This approach is more convenient because the control of balance under a particular stance condition is not two separate tasks in AP and ML directions in different times but rather a simultaneous task for the body to handle. In this case, tandem stance is probably the newest and most challenging tasks in all conditions which is not similar to any daily life condition. It can be interpreted as the body did spent enough energy to keep the AP balance and worked in full force to handle ML balance.

Again, going back to the comparison of the results of tandem stance with intermediate stance conditions in the AP direction, the complete change in how body uses the two mechanisms not being a smooth transition between mechanisms may be explained. Since, it is more like a sudden change of behavior on how two mechanisms work together rather than one giving its place to another as position switches to tandem stance, it might be related to the extreme placement of feet in tandem stance.

At the end there are two possibilities in this case. Either increasing the angle between feet, closer to tandem stance will at some point show an additive contribution of both mechanisms before reaching tandem position, or this subtractive behavior of two mechanisms is a behavior for all intermediate stance positions and unless both feet are aligned there is no need for the hip loading-unloading mechanism to be dominant. Without the necessary data it is not certain to reach a conclusion, yet the similarities and non-significance of changes between two intermediate positions in question hints at, it is more possible that the alignment of two feet makes a more impactful change in the selected approach of balance control. Additionally, 30º stance condition being controlled by a subtractive participation of both mechanisms and the gradual changes through three intermediate stance conditions implicates that, even though there is no data for another intermediate condition closer to tandem stance, it is more likely that seeing a subtractive participation of two mechanisms would be the result anyways.

Another proof that this subtractive behavior would continue when the angle further increased is the variation among subjects in tandem stance for AP balance control. At the end, hip loading-unloading mechanism is more active and showing more amplitude contributions, yet the reason behind this mechanism is not significantly dominant is the behavior of ankle mechanism in this condition. For some subjects, ankle mechanism was out of phase with COP_{net}. This means that it was subtracting from the activity of hip loading-unloading mechanism for some subjects. An example to this behavior in Figure 4.9 belonging to Subject 16.

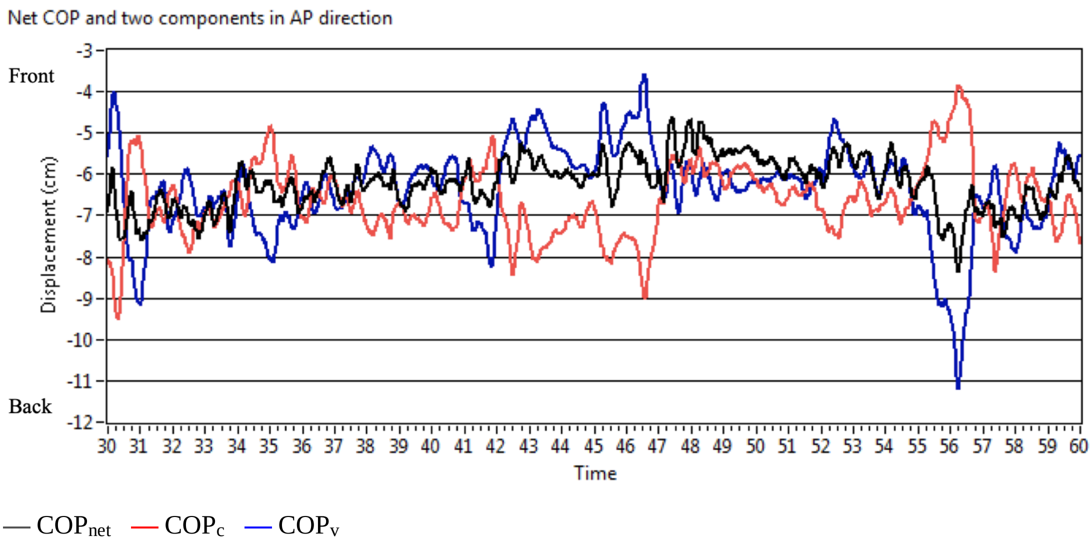

Figure 5.2. Out of phase behavior of COP_c in Tandem Stance

However, this subtractive behavior in AP direction is not exactly the behavior observed in intermediate conditions. In those cases, the dominant mechanism was always the ankle mechanism and the hip loading-mechanism was the controlling mechanism. In the case of tandem stance for subjects showing a subtractive behavior, this time the overactive mechanism is ankle mechanism instead of hip loading-unloading mechanism. This behavior only supports the idea that further increasing the angle between feet from 60º to some position closer to tandem stance would be similarly showing a subtractive relationship in which again the dominant mechanism is ankle mechanism. Because it looks like whether both mechanisms act in an additive or a subtractive manner, in any case the resulting selected control behavior is completely different than the behavior identified in intermediate conditions. This reveals that tandem stance is a novel condition for all subjects with its way of balance control approach and variation among the behavior of subjects.

It is more likely that when feet are aligned, the stance condition is taken as a completely new situation compared to the previous ones, which needs a completely new approach. As predicted, hip loading-unloading is taking the control in this case. However, hip being dominant is not because of a similarity with the similar conditions but rather as a result of adapting to a more appropriate new response for that condition.

One thing is for sure, further increasing the angle between feet from side-by-side stance condition did not lead equal amount of contributions at 45º condition and not a gradual increase of the use of hip loading-unloading mechanism while approaching to tandem stance as hypothesized. Instead, the way of participation of two mechanisms in the control of balance completely changed from additive to subtractive behavior in intermediate conditions and to an additive behavior again when the feet were aligned one more time for tandem stance.

Finally, reviewing the hypotheses to compare them with the results obtained, it can be said that in general the hypotheses presented for each condition held only for some of the conditions. A comparison of hypotheses and obtained results can be followed from Table 5.1. Mainly, the assumption that there would be a pattern of behavior on the relationship between each stance condition is true. Still, this pattern is not exactly as expected. Moving from side-by-side stance to tandem stance, although there is a gradual change, it is not as expected for particularly intermediate stance conditions. Additionally, the transition between intermediate stance and tandem stance is not smooth. It rather reveals that tandem stance is a completely different case requiring a different balance control strategy rather than being similar to extremes of intermediate stance conditions. Even though the gradual change is in the same direction with our hypothesis, the way of participation of both mechanisms for the control of stance is different than hypothesized.

Table 5.1. *Comparison of hypotheses and obtained results for dominant mechanism. Note that, for AP direction hip mechanism is not significantly dominant.*

	Hypothesis		**Results**	
	AP	**ML**	**AP**	**ML**
Side-by-side	Ankle	Hip	Ankle	Hip
30°	Ankle	Hip	Ankle	Hip
45°	Equal		Ankle	Hip
60°	Hip	Ankle	Ankle	Hip
Tandem	Hip	Ankle	Hip	Ankle

Another essential point is the important role the secondary mechanisms take while adapting to new stance conditions. For sure, almost for all conditions we revealed a dominant control mechanism as a primary mechanism. On the other hand, we also observed that even in cases where the activity of dominant control mechanism does not change much through the conditions, e.g. the activity of hip loading-unloading mechanism in ML direction, the secondary mechanism is definitely affected more from the stance position change. Of course, there is a preferred primary strategy for balance control in every stance condition. Still, the body needs to adapt to the new position by changing its balance control behavior somehow, which is made by changing the contribution of the secondary mechanism in this case.

This approach was also present in the AP direction for intermediate conditions, where hip loading-unloading mechanism activity was arranged in line with ankle mechanism to cancel its overmuch activity. Moreover, for AP balance in tandem stance, it is seen that high variations between subjects in terms of using the secondary mechanism in a subtractive or additive manner but not such behavior on primary mechanism. Keeping the balance under this highly new stance condition was only possible for subjects by modifying the secondary mechanism to this unusual condition accordingly.

This demonstrates the vital role of secondary mechanisms in the control of balance under new conditions. Insomuch that, it is possible to say for new stance conditions, secondary mechanism is as crucial as the primary mechanism as it acts as a regulator of the primary mechanism in some cases or supporter for some other. Even in conditions where the primary mechanism is clearly taking almost all the control, the loss of assistance from secondary mechanism due to injuries in patients may cause instabilities and even serious balance problems.

In conclusion, it would be appropriate to say that body tends to use the mechanism which is more familiar, particularly, more familiar to the side-by-side stance which is the only, essential, basic stance condition for a human, for both AP and ML directions in possibly new stance conditions such as the intermediate stance conditions. However, when two joints are aligned in a stance position, as in tandem stance, at the end this joint alignment is crucial for the preferred balance control strategy selection. This is something body can benefit during balance control which results with body to leave the familiar control strategy and adapt a new one which if more favorable for that case.

Regarding the implications of this study, it can be said that there are two aspects. In the biomechanics field, especially the studies on postural control, to the best of author's knowledge, this study is the first study to investigate the postural control behavior in different intermediate stance positions and to examine the relationship between the postural control mechanisms of similar stance positions. On the other hand the results of this study can be used in the health field as well. Different balance problems cause different posture and walking patterns which require individualized treatments. Even though this study is conducted in static conditions, the implication is not restricted to the static balance. Improving static balance will clearly have a positive effect on the dynamic balance as well. Having the knowledge on postural control mechanisms of different stance positions may help the health professionals in the field to plan the treatment accordingly. Standing balance and walking patterns can be improved by using the knowledge obtained from the study by focusing on the dominant control mechanism of that particular stance. For a patient having mediolateral instability while walking, working on abductor and adductor muscles would have more benefits then improving ankle stability for both standing and walking balance.

Future studies should be focused on the effects of different stance conditions in patients with ankle and hip injuries. These kind of subject groups may actually bring out the effect of secondary mechanisms in the control of quiet stance. Furthermore, they may reveal new strategies adapted to control balance when one or both mechanisms are lacking proper motor responses.

Another aspect that can be investigated in the future is regarding the method of the study. In this study time domain analysis is used to better compare the results with the literature. Frequency domain analysis may reveal other information including the frequency content of the postural control mechanisms.

Additionally, investigating these control mechanisms under perturbed conditions can also make it possible to understand if selection of the primary strategy is mostly a result of stance position adapted or external effects can completely change the approach of selected mechanism in order to control dynamic balance.

5.4. Limitations of the Study

The findings of this study should be considered in light of some limitations. The most important limitation concerns the method of the study, specifically speaking, experimental setup. In order to collect the necessary data for this study, two force platforms were needed and this was the most important part of the study design. During experiments, participants were supposed to place their feet on separate platforms, while the distance between feet is subjected to a change in each stance condition. Since the two force platforms are embedded to the ground, even though an extremely high effort is spent to use maximum area out of two force plates by changing the axis of force plates for each condition to use them in diagonal directions, for some conditions optimal distances between feet in either horizontal or vertical direction could not be achieved. It is believed that the differences between some intermediate conditions not being significant in one or both directions in some cases is actually under the effect of the similar stance widths because of this limitation. Due to same reason, only three different intermediate conditions were possible to evaluate and selected for investigation.

The second limitation is related to the data collection from participants which is used to analyze some additional information on intermediate stance conditions. The information on dominant extremities (both upper and lower extremity) of the participants were determined by asking them their dominant extremities in a questionnaire. For this question, most of the female subjects declared that they are not sure which foot is their dominant foot. Even though at the beginning of the study this was foreseen and a test to obtain dominant foot was searched for, a standard test is not found. By relying on the information provided by the subjects, the dominant extremity differences are investigated anyway. However, the significant differences between left foot ahead - right foot ahead conditions might be related to lower extremity dominance.

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APPENDIX A: ETHICAL APPROVAL

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04 TEMMUZ 2017

Konu: Değerlendirme Sonucu

Gönderen: ODTÜ İnsan Araştırmaları Etik Kurulu (İAEK)

İlgi: İnsan Araştırmaları Etik Kurulu Başvurusu

Sayın Doç. Dr. Ergin TÖNÜK ;

Danışmanlığını yaptığınız yüksek lisans öğrencisi Ezgi SÜMBÜL'ün ***"Mediolateral ve Anteroposterior Motor Cevapların Farklı Duruş Pozisyonları için Genellenmesi"*** başlıklı araştırması İnsan Araştırmaları Etik Kurulu tarafından uygun görülerek gerekli onay **2017-FEN-033** protokol numarası ile **17.07.2017 – 31.05.2019** tarihleri arasında geçerli olmak üzere verilmiştir.

Bilgilerinize saygılarımla sunarım.

Prof. Dr. Ş. Halil TURAN
Başkan V

Prof. Dr. Ayhan SOL
Üye

Prof. Dr. Ayhan Gürbüz DEMİR
Üye

Doç. Dr. Yaşar KONDAKÇI
Üye

Doç. Dr. Zana ÇITAK
Üye

Yrd. Doç. Dr. Pınar KAYGAN
Üye

Yrd. Doç. Dr. Emre SELÇUK
Üye

APPENDIX B: EXPERIMENT QUESTIONNAIRE

Katılımcı Bilgi Formu

Ad:

Soyad:

Doğum Tarihi:

Boy:

Kilo:

Ayakkabı Numarası:

Dominant Üst Ekstremite (Hangi elinizi kullanmayı tercih ediyorsunuz?) **:**

Dominant Alt Ekstremite (Bir topa vuracak olsanız hangi ayağınızla vurmayı tercih edersiniz?) **:**