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ABSTRACT

HAND CONTROL OF BIPEDAL BALANCE IN QUIET STANDING: IMPLEMENTATIONS FOR LOWER EXTREMITY EXOSKELETON

by
Ala'a Al-Rashdan

Maintaining stable posture is important for humans, even though it is challenging because of our bipedal structure. One of the main balance related disorders is paraplegia due to spinal cord injury. People with a complete spinal cord injury have motor and sensory impairment that greatly reduces the ability to move their lower extremities. In recent years, lower extremity exoskeletons that apply torques generated by motors to the joints of the person have helped to them stand and walk.

This research is a part of an extended project to build a new exoskeleton for use by individuals with paraplegia due to motor complete spinal cord injury. The goal of the project is to develop a device with an intuitive control mechanism capable of generating real time gait and balance. Commercial exoskeletons have achieved great steps regarding restoring ambulation. On the other hand, most of them do not actively support bipedal balance. In addition, commercially available exoskeletons except the REX need crutches to balance for people with motor complete paraplegia. The NJIT TREKKER, our laboratory's research exoskeleton, suggests a novel, human-robot interface strategy that allows users to completely control and feel the trajectories of their exoskeleton-assisted feet, and be able to walk with considerably greater independence. The first study to develop TREKKER was performed before where a trekking pole was attached to each foot of a biped robot. Subjects controlled the trajectory of the foot of the biped by applying

small forces to the trekking poles. The study proved that hands can produce trajectories similar to human foot trajectories when provided with haptic and visual feedback.

If the hands and arms are effective surrogates for expressing ambulation, **can they also be surrogates for natural balance in quiet standing**? This is the main question that this dissertation answers. Importantly, this dissertation considers the ability of the arms and hands to make rapid adjustments to the center of pressure (COP) that will follow the center of mass (COM) and allow the person to retain balance to achieve this aim a perturbing system was constructed to study human body response to perturbations. Special shoes with small blocks attached to their soles were designed to study the capability of human body to adapt to base of support (BOS) reduction, and two special platforms with shoes on Pivots and two trekking poles attached to them were designed to study the effectiveness of using trekking poles. The pivots were used to eliminate the use of ankle strategy to retain balance by non-disabled subjects. In this study, subjects were asked to stand in front of the perturbing system and within the motion capture system's field of view, then they were perturbed with at seven different forces with and without visual feedback in three different experiments: using regular shoes, the shoes with small blocks attached to their soles, and the shoes with pivots and trekking poles. Biomechanical parameters were studied to assess balance in A/P plane in each of the three experiments. The results suggest that the use of trekking poles is a viable approach to maintain balance during quiet standing.

The main conclusion of this study is that using trekking poles is a good approach to maintain balance in quiet standing and as a response to small perturbations. Statistical analysis of SI, error signal peaks, and correlations comparing Pivots experiment to Regular experiment support this hypothesis. In addition, the high correlation coefficients between

COM and COP of quiet standing on Pivots and in Pivots experiment with perturbations, and the high correlation coefficients of the correlation between COP and the trekking poles trajectories indicates that the trekking poles are working as a surrogate to the ankle joint. It is concluded that using the trekking poles, though the response to perturbations does not match the biological response, is good enough to maintain balance in quiet standing and perturbed quiet standing especially for small perturbations.

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by
Ala'a Al-Rashdan

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APPROVAL PAGE

HAND CONTROL OF BIPEDAL BALANCE IN QUIET STANDING: IMPLEMENTATIONS FOR LOWER EXTREMITY EXOSKELETON

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{الْحَمْدُ لِلَّهِ الَّذِي هَدَانَا لِهَٰذَا وَمَا كُنَّا لِنَهْتَدِيَ لَوْلَا أَنْ هَدَانَا اللَّهُ}

This dissertation is lovingly dedicated to

My Dad's Soul

"Abdelqader Alrashdan"

I wouldn't be where I am without you.

My Mother

"Khawla Almomani"

Who showed her love and support in the best way.

My brothers and sisters

"Tabarak, Shorouq, Hamzah, Batool and Omar"

And my little angles

"Sanad and Leen"

For all their affection and encouragement.

All my friends

Who supported me throughout these years, and to my extended family for all their affection and encouragement.

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CHAPTER 1

BACKGROUND AND SIGNIFICANCE

1.1 Introduction

Maintaining stable posture is important for humans, even though it is challenging because of our bipedal structure. Approximately two thirds of our body mass including some delicate organs are located two thirds of body height from the ground over the legs. This provides a narrow base of support, which is why the control system is complex in humans. Balance is critically important to move safely. When the balance control system deteriorates with age, or due to pathological reasons, the results can be disastrous. Falls in elderly people have been identified as a major health problem and a common reason of death.

This research is a part of an extended project to build an exoskeleton for use by individuals with paraplegia due to complete spinal cord injury. The goal of the project is to develop a lower extremity exoskeleton with an intuitive control mechanism capable of generating real time gait and balance, while also providing proprioceptive feedback. This control mechanism allows other articulators to express the neural encoding of the desired trajectory instead of employing simple switch mechanisms or as yet unproven brain computer interface (BCI).

This dissertation examines the potential of these alternative articulators to maintain bipedal balance when individuals are perturbed in the sagittal plane. Since the user's arms and hands are employed as the alternative actuators for expressing gait, their typical use with current exoskeletons for maintaining balance with crutches is severely limited, thus adding to the importance of this investigation.

This study focuses on the potential of using the arms and hands, to provide an effective and intuitive way to match the user's center of pressure with his/her center of mass projection and maintain safe balance.

1.2 Basic Definitions

Following are some basic definitions that are important to study posture and balance:

Posture can be defined as the orientation of body segments relative to the gravitational vector. It is an angular measure from the vertical [1]. Another interesting definition is the geometric relation between two or more body segments, where the relation is expressed in terms of joint angles between segments in addition to the relation of the body to the surrounding environment (e.g. body relative to ground and/or other supporting surfaces) [2].

Winter defines **Balance** as the "dynamics of body posture to prevent falling" [1]. While Balasubramaniam defines it as the "equilibrium resulting from the matching of torques", which can be organized in, anticipation of, or as a reaction to the effects of postural perturbation [2]. In addition, balance can be technically defined as the ability to maintain the center-of-gravity (COG) of an object within its base-of-support (BOS).

Winter classified posture and balance into four classes: maintenance of a static unperturbed posture, static posture control under the presence of perturbations, balance control during the voluntary execution of a movement, and balance control during movement in the presence of perturbations [3].

The **Base of Support** (BOS) can be defined as a convex polygon beneath an object or person bounded by the perimeter of the contact that the object or person makes with the supporting surface (ex. ground). The points of contact can be body parts such as feet or hands, or mobility aid such as crutches or a chair a person uses to sit. The BOS is an important concept to understand human balance, as balance is defined as the ability to maintain the center of gravity within the BOS. As mentioned before this can be the practical definition of balance [4].

In quiet standing, the BOS is the area defined by the boundary surrounding the ground contact of the two feet. During normal gait it is defined as the "horizontal stride width during the double-support phase when both feet are in contact with the ground and the whole-body center of gravity remains within the BOS". In impaired walking, aids such as a walker, crutches, cane, etc. provides more stability by extending the BOS [4].

Center of Pressure (COP) is the main parameter that has been used to study quiet standing. Force plates are commonly used to record COP in both anteroposterior (A/P) and mediolateral (M/L) directions. It can be defined as the projection of the vertical ground reaction force vector that represents a weighted average of the pressures over the contact surface (both feet) with the ground. Its units are in meters [1].

Center of Mass (COM) is another important parameter in studying balance. It is defined as the equivalent point of the total body mass in the global reference system (GRS). Practically it can be defined as the weighted average of the COM of each body segment. Center of Gravity (COG) is the vertical projection of the COM onto the ground. Its units are meters (m) [1].

1.3 Sensory Feedback for Balance

Three major **Sensory Systems** are involved in balance and posture:

1. Visual system is mainly involved in locomotion planning and avoiding obstacles.

2. The vestibular system works as a gyroscope that senses linear and angular accelerations.

3. The proprioception system includes many sensors that are responsible for sensing entire body's position and velocity using each segment's position and velocity and their contact with external environment, in addition to the orientation of gravity [2, 5]. Winter claims that the entire somatosensory system is involved in providing feedback for balance [3, 6].

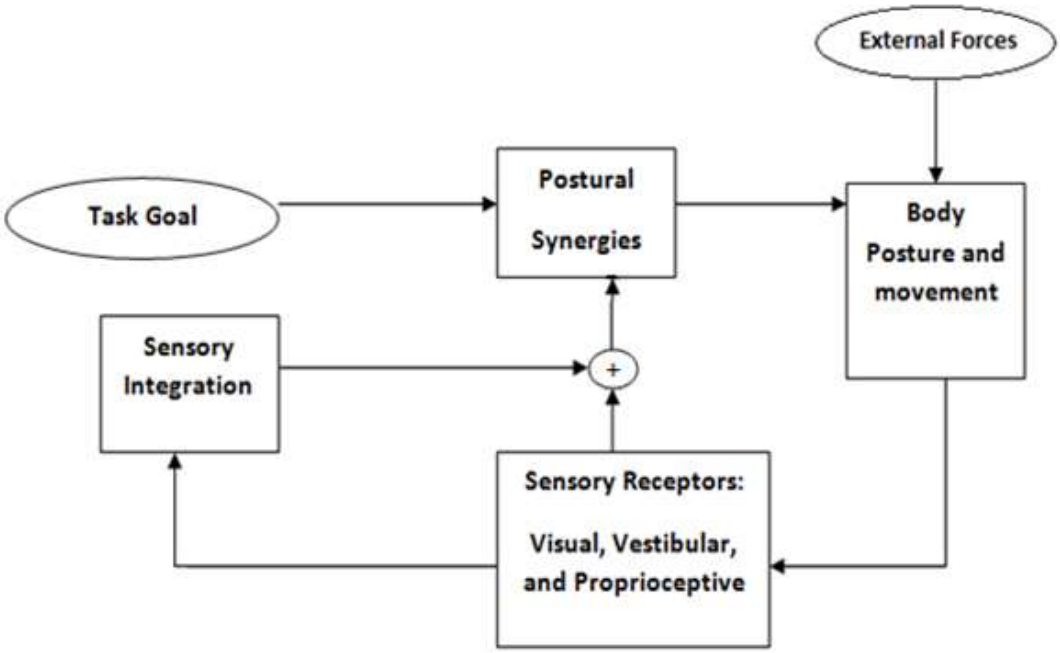

Figure 1.1 A conceptual schematic diagram of the postural control system [3].

Neurophysiologists have suggested and established a wide range of experiments to demonstrate the contribution of each of these systems and further confuse each system by

providing conflicting or false sensory inputs to understand the contribution of each system to the posture and balance control [6].

1.4 Muscle Synergies Supporting Balance

Several studies defined muscle synergies (M-modes) as a “low-dimensional modules formed by muscles activated in synchrony or with fixed time delays may be used by the nervous system as building Blocks for construction motor output patterns during both locomotor and postural tasks” [7]. Another possible definition for a muscle synergy is a group of muscles activated in synchrony with fixed relative gains (activation patterns) with consistent spatial characteristics [8].

It was hypothesized in 1985 that there are two discrete strategies to maintain balance as a response to disturbances: the ankle strategy and the hip strategy. The ankle strategy works as a **single-segment inverted pendulum** by applying torques to the ankle joint of each leg, while the hip strategy works as a **double-segment inverted pendulum** with counter-phase motion at the ankle and the hip of each leg [8, 9]. These strategies can be used either separately or can be used together by the central nervous system to control the horizontal position of the center of mass.

Previous experimental observations showed that ankle strategy was used to respond to translations in A/P plane during stance on a flat support surface. On the other hand, hip strategy was observed during responses to backward translations during stance on a narrow support surfaces(an example is a 10 cm beam) [9].

Strategies to Maintain/Restore Balance:

1. Ankle strategy: Ankle strategy is used when perturbation is slow, low amplitude and when the contact surface is firm, wide and longer than foot. Muscles are recruited distal to proximal. Also, head movement is in-phase with hips.

2. Hip strategy: Hip strategy is used in response to fast or larger amplitude perturbations, or when the surface is unstable or shorter than feet. Muscles get recruited proximal-to-distal and head movement is out-of-phase with hips.

3. Stepping strategy: Stepping strategy is used to prevent falling when perturbations are fast or of large amplitude or when other strategies fail to maintain balance. Base-of-support (BOS) moves in some way to keep COG and COP within the BOS.

4. Suspensory strategy: Suspensory strategy is a forward bend of trunk with hip/knee flexion and may progress to a squatting position. Thus COM is lowered, thus reducing the excursion for the COG and reducing the destabilizing moment arm [1, 2, 6, 10, 11].

It is interesting to know that biomechanical optimization models have suggested a mixed hip-ankle strategy in the anteroposterior direction as a "Unified Theory [11]", instead of a pure ankle strategy to correct postural disturbances of any speed if the main objective of optimization is a minimal 'neural effort'. They relied on the limited effectiveness of small ankle torques to correct disturbances due to the larger moment of inertia of the human body and on the hardness of independent control of ankle and hip postural mechanisms. The application of these models to maintain balance, however, is still far from clear [8, 10].

Many people have measured the M/L location of the COP but the first to identify its motor mechanisms were Day et al. (1993), who suggested that M/L movement of COP could be a result of the hip abductors/adductors in addition to the ankle invertors/evertors [11].

In addition, Collins and DeLuca inferred the same control mechanism in both M/L and A/P directions in their "Free motion" analysis of COP changes. This new approach was used instead of the old approach that calculate statistic scores such as time-domain distance, area and hybrid measures, and frequency domain measures which lacks Sensitivity [12] and

do not consider the dynamic properties [13]. In the new approach, COP is modeled as fractional Brownian motion with a dual-control model based on postugraphic analyses of COP trajectories in quiet standing. It uses the mean square displacement to plot stabilograms, and demonstrates two control models: short-term mechanism that represents an open-loop control schemes that are used by the postural control system, and a long-term mechanism that represents a closed-loop control mechanisms [14].

Winter studied quiet stance with feet side by side with each foot on a separate force plate. He found that the motor responses in the M/L direction were totally dominated by a hip load/unload strategy. Since the hip strategy is perpendicular to the A/P control, it is seen to be completely independent of the ankle strategy.

Winter studied the relative role of each of the motor control strategies when subjects adopt foot positions other than standing such as: side-by-side stance, tandem stance, or an intermediate position halfway between the two (called 45° position). This intermediate position is a critical rebalancing phase during the gait cycle. Based on his previous findings, it was hypothesized that since the invertor/evertor ankle joint axes are lined up in tandem stance, the M/L control will be dominated by an ankle strategy (inverter/evertor). On the other hand, the A/P balance could be a combination of two strategies: ankle control and hip load/unload mechanism. In the intermediate 45° position, neither of the ankle joints are in line, so the control is expected to be exerted by both the ankle and the hip load/unload mechanisms [11].

It is important to note that in the literature there is a major misuse of the COP when it is referred to as ‘sway’, thereby inferring that it is the same as the COG. Unfortunately some researchers even refer to the COP directly as the COG [8].

COP is the point location of the vertical ground reaction force vector and totally independent of the COM which is a point equivalent of the total body mass in the global reference system [1, 3, 6].

The (COP-COM) signal is directly related to the horizontal acceleration of the COM, hence it can be considered as the error signal sensed by the balance control system. The magnitude and frequency of this error signal is of importance in the interpretation of the balance control system [1, 3, 6].

1.5 Center of Pressure

Upright posture is commonly investigated to provide information about posture and balance. Posture is often analyzed using a force platform, which provides center of pressure data; referred to as 'statokinesigram'. It is one of the several methods that have been proposed in the past for the calculation of posturographic parameters from COP trajectories. Many previous studies were limited to the summary of statistic scores, such as time-domain distance, area and hybrid measures, and frequency domain measures which lacks sensitivity [12]. Sensitivity (also called the true positive rate, or the recall in some fields) measures the proportion of positives that are correctly identified (e.g., the percentage of sick people who are correctly identified as having the condition) and do not consider the dynamic properties [13].

On the other hand, different models that consider dynamic stability of COP motion over the base of support were established; assuming posture as the dynamic stability of a continuously swaying body. Yamada described the fluctuation of the COP as a chaotic process [15], in contrast, Collins and De Luca described it as a stochastic process [14]. The

stochastic hypothesis has been popular since it has an explanation for the role assumed by physiological noise in static posturography [13].

1.6 Center of Mass and Balance

To examine stability, many studies have used the vertical projection of the whole body center of mass motion (COG) and its relative position to the center of pressure (COP) of the supporting foot to examine gait stability to establish a relationship between them.

Whole body COM position data are calculated as the weighted sum of all body segments, with 13 segments representing the whole body including head and neck, trunk, pelvis, two upper arms, two forearms (with hands), two thighs, two shanks, and two feet [15].

The body's center of mass is a key factor in the analysis of human gait, as it reflects the motion of the whole body. The knowledge of the three-dimensional movements to the center of mass is prerequisite for the calculation of walking parameters and design of artificial joints. Alteration to the trajectory of the center of mass may indicate a clinical manifestation of an underlying pathology or only a means of maintaining stability in gait. In balance control studies the body center of mass can be related to the ground reaction force or to the base of support, which is quite small in biped. Since small displacements of the body center of mass are important in balance control studies, it is essential to obtain valid estimates of the body center of mass. The center of mass can offer useful parameters for the total evaluation of walking, and, in combination with other kinematic and kinetic data, will give a more precise analysis such that practical application is possible [16].

A person has to confine projection of the COM within the base of support (BOS) in order for the body to remain balanced while standing [17].

Practically there are three ways to find COM:

1. Kinematic method which is directly based on the definition of the COM [3, 18].
2. Zero-point-to-zero-point double integration, which depends on the idea of that when the net of the horizontal ground reaction forces equal zero, the COP and the vertical projection of the COM on the ground are at the same point [18].
3. COP low-pass filter method uses a low-pass filter which is defined by the relation of the COP and COM in the frequency domain, since that the relative magnitude of the COM with respect to the COP is a function of the frequency of oscillation. The vertical projection of the COM is computed by applying a low-pass filter, which is related to the inertial characteristics of the subject and the COP frequency content [18].

1.7 Exoskeletons

Exoskeletons are external structural mechanisms with joints and links corresponding to human body joints and links, they are worn by people with disability. They provide torques generated by motors via the exoskeleton's links to the joints of the person. Different algorithms are used to control the exoskeleton depending on its application [19]. Below are four main application fields for such a device:

1. Physiotherapy: Active or passive modes can be used depending on the instructions of the physio therapist considering the case of the patient. The exoskeleton can be used to augment intended movement or to impede unwanted movement according to therapeutic goals [20].
2. Assistive device: Used in situations where impairment is permanent and there is significant gait limitation. Most of the load is supported by the exoskeleton [21].
3. Haptic device: There is a physical interaction between the patient and objects, torques are generated by the feedback due to that interaction and that is how the exoskeleton transport many characteristics of the object such as: shape, stiffness, texture etc. [22].

4. Master device: The operator replaces the virtual environment with a real robot, where the exoskeleton is used to control the robot in a master slave mode. The slave robot interacts with environment and according to that interaction the exoskeleton makes the subject move [23].

Many medical conditions can cause loss of motor control. Of concern in this dissertation is motor complete spinal cord injury. In addition, exoskeletons are being employed with individuals with motor incomplete spinal cord injury as well as neurological injuries such as hemiparesis from stroke [24], multiple sclerosis [25], and cerebral palsy [26].

Orthotic devices have been used and developed for a long time to help people with paraplegia to get some degree of locomotion and to reduce secondary complications due to lack of motion. Passive orthoses were usually used, but because of the high metabolic energy expenditure that makes people very tired, disappointed, and mostly with less tendency to go on with rehabilitation, researchers started developing active exoskeletons in the sixties of the twentieth century. Rapid developments have been achieved in the previous decades, and many wearable exoskeletons were built mainly to restore walking for spinal cord injury people [27].

1.8 Lower Extremity Exoskeletons

Many lower extremity exoskeletons have recently been developed and commercialized, among them is the EKSO from the US Company **Ekso Bionics** [28]. In the last few years, additional exoskeletons have caught the attention of the media. Examples: the **ReWalk** system from the Israeli company Argo [29], **HAL** from the Japanese company Cyberdyne[30], **INDEGO** from the US Company Parker Hannifin [31] and **REX** from the New- Zealand Company RexBionics [32]. All these exoskeletons except REX require the

people with **motor complete spinal cord injury** to use crutches to achieve balance. The exoskeletons have only two motors per leg and have no active balancing mechanisms. The REX ensures stability without crutches, but with very low speed and quasi static properties, with a high mass and low center of mass that contribute to stability. Figures 1.2-1.6 show pictures of the commercially available exoskeletons, Figure 1.7 shows MINDWALKER, an exoskeleton under research in The University of Twente.

Figure 1.2 Ekso Exoskeleton of EksoBionics [28].

Figure 1.3 Rewalk Exoskeleton of Argo [29].

Figure 1.4 HAL Exoskeleton of Cyberdyne [30].

Figure 1.5 INDEGO exoskeleton of Parker Hannifin [31].

Figure 1.6 REX Exoskeleton of RexBionics [32].

Figure 1.7 MINDWALKER of University of Twente [33].

1.9 Assessing Balance

Since control of balance is an essential component of human movements, it is critical to find a way to asses it. Because balance is not yet well understood, it is difficult to find a single quantitative criterion for assessing a person's ability to control balance. Experiments to study balance are difficult to establish, as well. This is because there are many movements to choose from which accomplish the same task, and because of the wide variability in how different people do the same task (due to anthropometric differences, and CNS differences) and even the variability within the subject to do the same task and respond to the same perturbation. All of this leads to the challenge to get a good model that controls the exoskeleton's balance.

Until now, the clinical evaluations of human body balance in physical rehabilitation are mostly centered on identifying regional limitations of a joint or a segment (For example, joint range of motion assessment, muscle strength assessment, sensation assessment). Conversely, this type of assessments does not provide clinicians with a global description of a patient's capability to perform daily tasks. Adding regional constraints to global motion can give a reasonable quantitative evaluation to human balance and their ability to fall. The best way can be to predict the feasible COM movements that a patient can perform relative to the COP. Previous studies have investigated COP [12, 14], COM-COP relations [3, 6, 17], joints angles and torques [7, 8, 11].

Two main points need to be considered during balance control design: The first one is not to override human control in order to avoid unwanted maladaptation of the humans. Users should not be discouraged from using existing human balance capabilities. And, the

exoskeleton and human body control systems must operate synergistically in a tight closed loop, to avoid errors that leads to compromised stability [34].

The MINDWALKER is a really interesting exoskeleton regarding balance control. The control system of the early model of MINDWALKER was used to ensure the overall system's (user and exoskeleton overall structure) balance , depending on proprioceptive sensors and a high level controller that makes use of exteroceptive sensors: laser range-finder, time-of-flight camera and Kinect, to obtain a short term 3D model of the frontward environment that allows checking and, inhibiting or adapting the original control requests if necessary [33, 35-38]. A unique approach used to balance MINDWALKER is step-width adaptation algorithm. It aims to enhance balance in M/L plane. It depends on the results of the mathematical modeling that showed that even though it retains stability in the sagittal plane, passive bipedal walking is laterally unstable and so foot placement has to get actively controlled. And, since people with paraplegia have lost or impaired control of their legs, self-balanced walking exoskeleton requires powered hip abduction/adduction. That is why an online step-width adaptation (SWA) algorithm is presented was created to solve this problem. The algorithm is based on a concept which has been successfully applied to analyze human body balance using extrapolated COM concept. It is a spatial variable used to formulate a stability condition in static and active situations based on COM displacement and velocity using linear inverted pendulum model [38].

Another unique approach to control exoskeletons balance involves robotic device generating open-loop assistance, which is triggered only when loss of balance is detected. It calculates the feed-forward trajectory using a model of the falling human, and the design gives model assumptions are true when the person is in upright position [39, 40]. For fall

detection, a balancing aid consisting of multiple parallel-mounted control moment gyroscopes (CMGs) contained within a backpack-like orthopedic corset is used (Figure 1.8). CMG is an attitude control device consisting of a spinning rotor and motorized gimbals that tilt the rotor's angular momentum. The angular momentum changes when the rotor tilt causes a gyroscopic torque [41] that is used to control balance. A critical disadvantage of using conventional CMG control techniques is geometric singularities that results in a number of performance problems, such as not planned oscillations or freezing of the gimbals, such problems are usually solved by adding redundant actuators or by allowing errors in the generated moment, though, a new methodology is required because of the limitations of the size and weight of the design and focus on accurate moment tracking. A new control scheme is proposed to solve this problem depending on the directional singularity-robust control law [42].

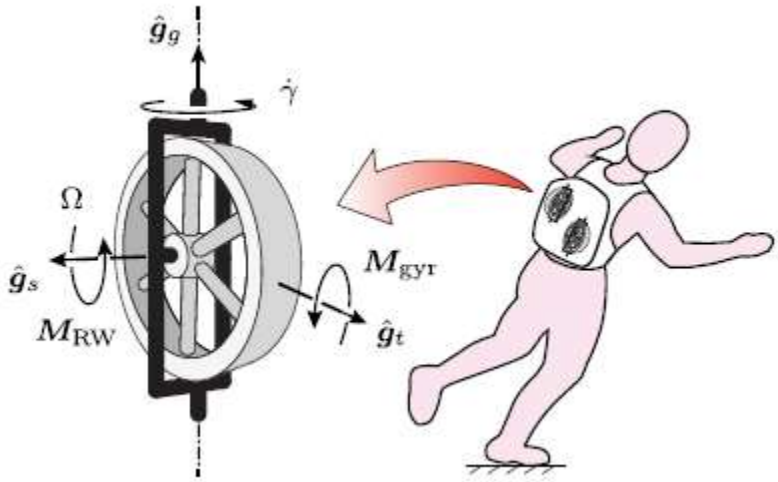

Figure 1.8 Schematic of a single-gimbal CMG showing the orientations of the gyroscopic moment (Mgyr) and the reaction wheel moment (MRW) exerted on the device in the case of variable flywheel speed [42].

1.10 NJIT TREKKER

Commercial exoskeletons have achieved great strides regarding restoring ambulation. On the other hand, much remains to be done. They perform with only two or three powered degrees of freedom (DOF) per leg, which is less than the number of DOF in a biological leg. Consequently, they are constrained to walking in a straight path on flat surfaces. In addition, most of them do not actively support bipedal balance. All commercially available units except REX needs crutches to balance for people with motor complete paraplegia, thus much more can be done in terms of balance.

The NJIT TREKKER, our laboratory's research exoskeleton, suggests a novel, human-robot interface strategy that allows users to completely control and feel the trajectories of their exoskeleton-assisted feet, and be able to walk with considerably greater independence. Our approach allows users to generate correct neutral motor plans for leg movement and express those actions by making walking-like movements using their ipsilateral hands. The TREKKER will use admittance control paradigm to command the exoskeleton to track the movement of the hands. This method allows the configuration of apparent inertia, damping, and stiffness of each exoskeleton leg, providing an intuitive way to physically interact the exoskeleton for the user. Figure 1.9 shows the full-scale TREKKER one-leg prototype with a trekking pole used to control foot trajectory [43].

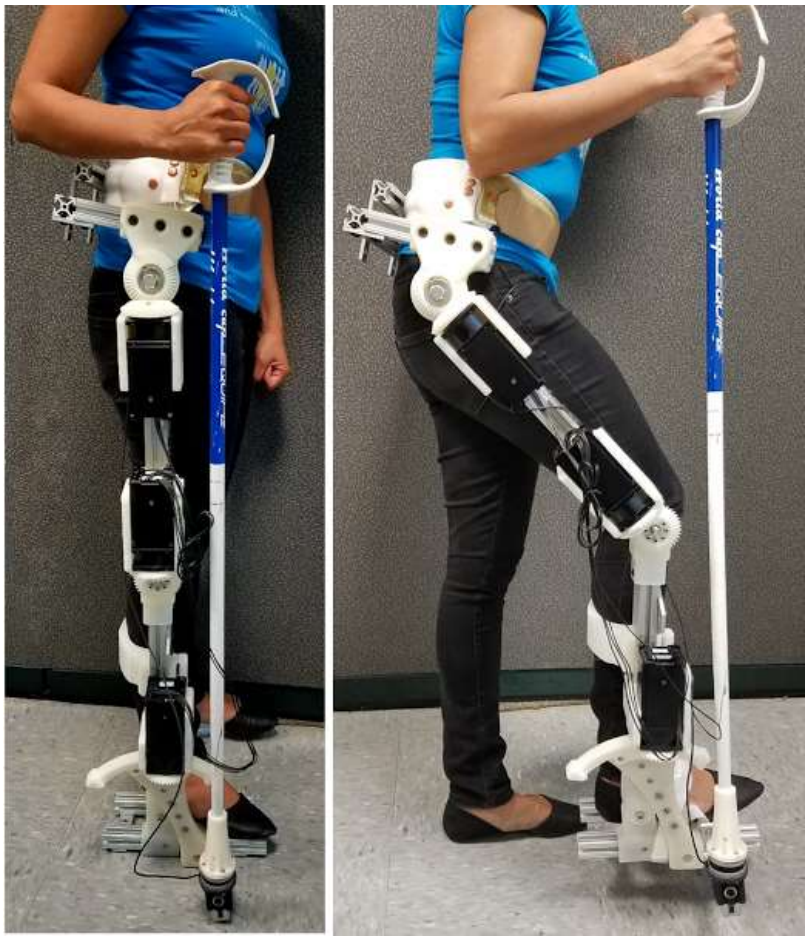

Figure 1.9 Full-scale TREKKER one-leg prototype with a trekking pole used to control foot trajectory.

The first study to develop TREKKER was performed by Kiran Karunakaran. She built a 10 DOF half scale prototype to investigate the capability of using hand-trajectory as a substitute for foot trajectory. A trekking pole was attached to each foot of the biped. The subjects controlled the trajectory of the foot of the biped by applying small forces to the trekking poles in real time in the intended direction. The efficiency of the control mechanism was evaluated by comparing the gait of the biped with normal human walking. Figure 1.10 represents the comparison of the ankle trajectories in the sagittal plane of the robot foot (left)

and human foot (right) when walking on the treadmill. The shapes are quite similar, with a vertical/horizontal excursion ratio of approximately 0.275 in each case [44, 45].

The study proved that hands can produce trajectories similar to human foot trajectories when provided with haptic and visual feedback.

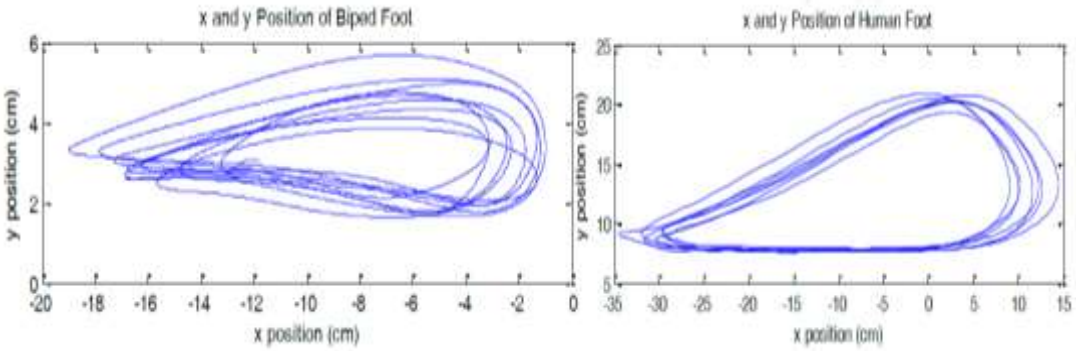

Figure 1.10 Comparison of ankle trajectories in the sagittal plane of the robot foot (left) and human foot (right) while walking on the treadmill. Direction of walking is to the right [45].

If the hands and arms are effective surrogates for expressing ambulation, **can they also be surrogates for natural balance in quiet standing**? Since the trekking poles convey the user's intentions for movement and provide sensory feedback (during the swing phase of walking), is it possible that during the stance phase, the same trekking poles can be used by the hands and arms to provide an alternative to the user's ankle strategy that is lost due to their disability. Importantly, this dissertation considers the ability of the arms and hands to make rapid adjustments to the COP that will follow the COM and allow the person to remain balanced.

1.11COP-COM Relationship

1.11.1 Base of Support and COM velocity

The feasible movements for the control of balance are usually described relative to the horizontal position of the body's COM to keep balance while standing. A person has to confine the projection of the COM (COG vector) within the base of support (BOS). Recently, it was found that this condition alone is not sufficient to guarantee that the standing posture will be sustained. It has been proposed that the horizontal velocity of the COM should also be considered because it controls the destiny of the horizontal position of the COM over the BOS. This implies that the time necessary to execute a step response may introduce instability if the step is triggered when the COP reaches the boundary of the BOS in its pursuit of the COM. The CNS may use the velocities of the COM and COP to predict the error between COM and COP. The error will increase as the COM proceeds beyond the BOS limit, while the COP is constrained to that limit. Stepping may be triggered before the COP travel is stopped in order to allow time for a corrective step to extend the boundaries of the BOS.

When a sufficiently large horizontal velocity exists, even though the horizontal COM is currently located inside the BOS, standing balance will not be maintained. On the other hand, it is possible for the COM to be located outside the BOS, such as in walking termination, and still be able to achieve balanced upright standing, such that horizontal COM velocity direction is toward the BOS.

Thus, in addition to the horizontal location of the COM with respect to the BOS, the magnitude and the direction of its velocity may also provide information about the ability to control balance. In addition, much research assesses a person's ability to maintain a standing

posture in two cases: unperturbed and perturbed conditions. Clinical observations indicate that falls occur most frequently when moving with larger displacements of the COM, such as walking, stepping up or down, or standing up [17].

1.11.2 COP-COG Relationship in Quiet Standing

In most studies, the quiet standing position involved subjects standing with their feet side by side and the ankle strategy employed in sagittal plane was analyzed. In studies about quiet standing, the major measure that has been recorded is COP from a single force plate. The movements of the COP in both A/P and M/L directions have been reported. However, Winter suggested that it is important to see the relationship between COP and COG during quiet standing.

To demonstrate the difference between the COG and COP and at the same time introduce an inverted pendulum model of balance in the A/P direction. Figure 1.11 shows a subject swaying back and forth in the sagittal plane while standing on a force plate. This shows a different posture at five different points.

Time 1, the body's COG is in front of COP, and the angular velocity w is assumed to be clockwise. Body weight W is equal and opposite to the vertical reaction force R, and these forces acts at distances g and p respectively from the ankle joint. W and R will remain constant during quiet standing. Assuming the body to be an inverted pendulum, pivoting about the ankle, a counterclockwise moment equal to R*p and a clockwise moment equal to W*s will be acting.

$$R*p - W*s = I*\alpha \tag{1.1}$$

Where:

I: is the moment of inertia of the total body about the ankle joint (kg.m).

α: is the angular acceleration of the inverted pendulum (rad.s^{-2}).

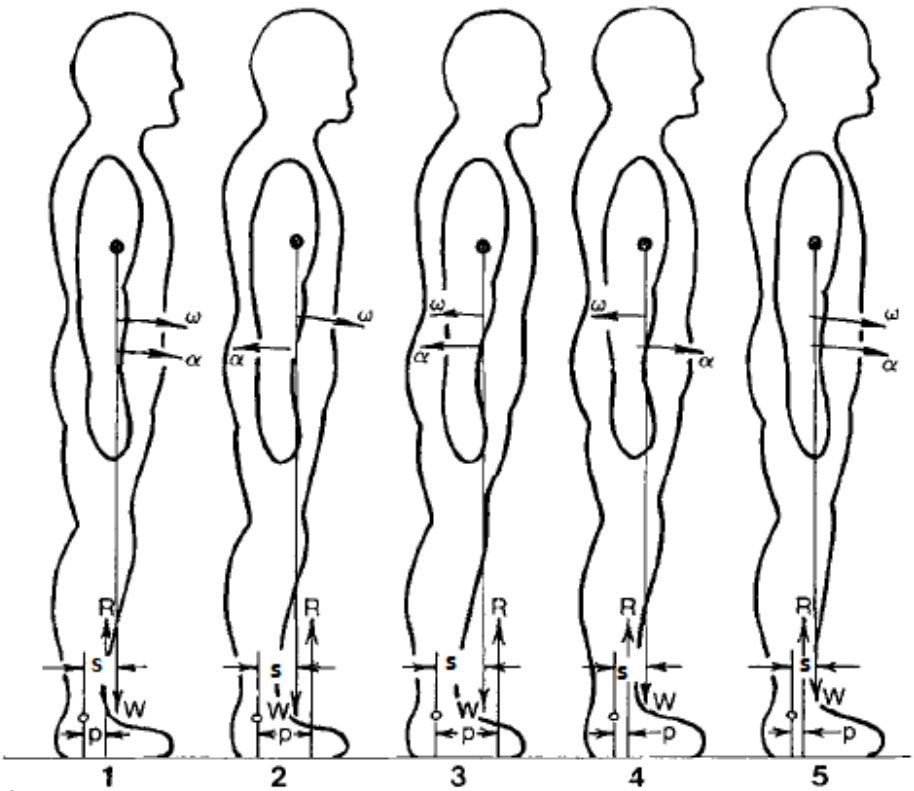

Figure 1.11 A subject swaying back and forth while standing quietly on a force platform. Five different points in time are described, showing the center of gravity (s) and the center of pressure (p) locations along with the associated angular accelerations (a) and angular velocities (w) [3].

If W*s > R*p, the body will experience a clockwise angular acceleration and to correct this forward tilting, the subject will move his COP forward (by increasing plantarflexion activation) such that at **time 2**, the COP will be slightly anterior to the COG. Now R*p > W*s, so α will reverse and will start to decrease ω until **time 3**, the time integral of α will result in a reversal of ω. Now both ω and α are counterclockwise and the body is

experiencing a backward tilt. As soon as the CNS senses that this posterior shift of COG needs correcting, COP is moved proportionally (by decreased plantarflexor activation) until it lies posterior to COG. Thus α will reverse to become clockwise again at **time 4**, and after a period of time, α will again decrease and reverse, so the body will get back to the original conditions, as seen at **time 5**. From this sequence of COG and COP conditions it can be seen that the plantarflexors- dorsiflexors control the net ankle moment to regulate body's COG. Apparently, the dynamic range of the COP must somewhat be greater than that of the COG: the COP must be continuously moving anteriorly and posteriorly with respect to COG. In contrast, if COG moves within a few centimeters of the toes, it is possible that a corrective movement of the COP to end of BOS would not be adequate to reverse ω. The subject may adopt hip strategy to alter the anterior movement of the COG, or if the COG anterior velocity is sufficiently high, the individual makes a step forward, thus changing the base of support and allowing the COP to move forward.

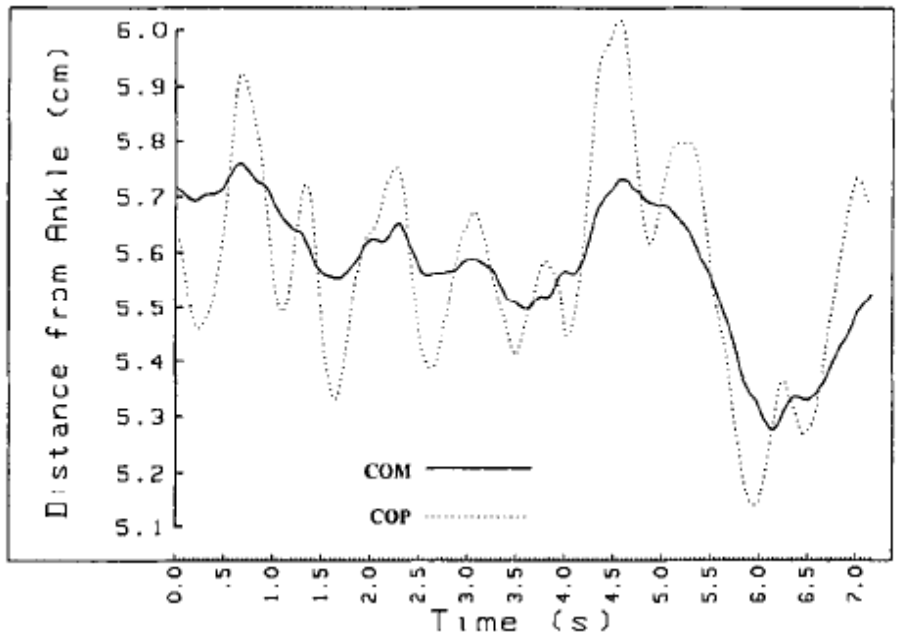

Figure 1.12 A 7 seconds record showing simultaneous center of gravity and center-of-pressure fluctuations for a subject in quiet stance. Centre-of-pressure excursions oscillate on either side of the center of gravity and have a higher frequency and greater amplitude [3].

Figure 1.12 is a record of COP versus COG in a subject performing quiet standing on a force platform. The previous sequence of events is repeated many times during this data collection period. In an inverted pendulum we can estimate the horizontal linear acceleration x" of the COM from the relationship:

$$\alpha = x''/d \tag{1.2}$$

Where,

d: the distance from the ankle joints to the total body COM (m).

And, substituting Equation (1.1) in Equation (1.2), we get:

$$p - s = \frac{I\ddot{x}}{Wd} = K\ddot{x} \tag{1.3}$$

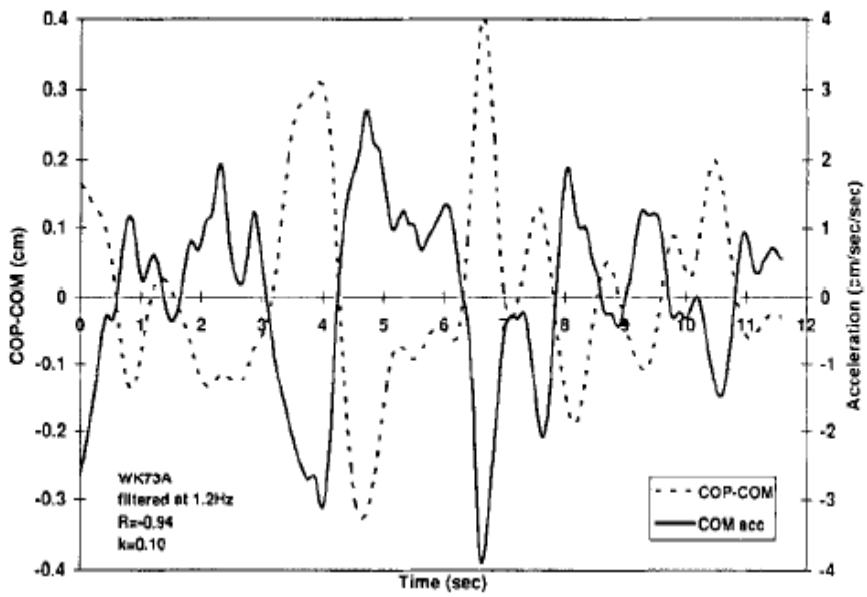

Figure 1.13 COP –COM and the horizontal acceleration of COM in the A/P direction.

Based on Equation (1.2), the inverted pendulum model predicts a high correlation between COP-COM and the horizontal acceleration of COM in the A/P direction as shown in Figure 1.13. In quiet standing, the correlation for this subject while standing quietly was 0.94. In large voluntary sways correlations exceed 0.99, giving credence to the validity of the inverted pendulum model in all standing situations [3]. Thus the difference between the COP and COM is proportional to the horizontal acceleration of the COM.

We can consider the difference between COP and COM as the ‘error’ signal in the balance control system which is causing the COM’s horizontal acceleration. The horizontal acceleration described here is in the sagittal plane; though; the same applies to the M/L

acceleration with some differences. Figure 1.13 illustrates the fundamental relationship given by the Equation (1.2).

COP and COM were measured for a subject standing quietly for 12 s record. COP-COM was then plotted against the COM acceleration in the frontal plane. It can be inferred from Figure 1.13 that there is a very high negative correlation between COP-COM and the acceleration. When COP is in front of COM in A/P plane when the subject is in quiet standing the acceleration is backward and vice versa when COP is behind COM. In M/L direction similar correlation is evident.

1.11.3 COP-COM Relationship in Gait

Center of mass (COM) motion and its relative position to the center of pressure (COP) of the supporting foot have been used in many studies to examine gait stability. Dynamic stability during movement has been assessed using COM momentum, and a more lateral momentum was identified in elder people with balance troubles.

Recent research also explained that linear measures of COM motion in the frontal plane during obstacle navigation could better distinguish elderly subjects with balance disorders from healthy aged matched subjects. It is important to keep in mind magnitudes of these linear measures of the COM motion and the COM-COP separation distance may, however, be different for different subjects with different height and body shape. Biomechanics measures of gait stability that can give information on instantaneous coordination between the COM and COP and exclude inter subject variability are still needed. “Instantaneous orientation of the line connecting COP and COM can characterize whole body position with respect to the supporting foot during gait” [15]. One of the studies used this line in reference to the vertical line passing through the COP in A/P plane and

mediolateral M/L inclination angles to account for both the instantaneous COM height and horizontal distance between the COM and COP, and a biomechanical relation between the COM and COP during standing has been established. Similar angles have been used to quantify postural sway during standing and were found to be similar for people of various heights. It was found that elderly people demonstrated a significantly greater medial, but a significantly smaller anterior, inclination angle than their matched controls during both unobstructed and obstructed gait. The medial COM-COP inclination angle was not affected by the gait velocity in the healthy subjects [15].

1.12 Stability Index

Since the Sensory Organization Test [50] that is usually used to assess balance does not account for some key biomechanical aspects of postural stability, such as weight, ankle moment, and shear force, a new measure of A/P postural stability called the Postural Stability Index (SI) was proposed. SI is defined as "the percentage ratio of the destabilizing torque due to gravity and the stabilizing torque due to the ankle muscles" [51]. Four stability zones were found using the COP, i.e., high preference, low preference, undesirable, and unstable. Ellipses were used to model the boundaries of the stability zones to capture their two-dimensional form and orientation BOS. However, physicians find that quickly identifying these stability zones to assess postural stability is practically very difficult. Since having a single number representing postural stability is important; a single measure defining postural stability, PSI, based on the physics of standing was established [51-53].

To evaluate postural stability, the effort needed to maintain stability across an entire test of dynamic balance has to be considered when perturbation occurs. The total value of the

stabilizing torque to counteract the destabilizing torque due to gravity in quiet standing represents SI [52].

For this experiment the stability index is defined as:

$$\text{Stability Index (SI)} = \frac{\text{Destabilizing Torques}}{\text{Stabilizing Torques}} \tag{1.4}$$

$$\text{SI} = \frac{(m*g*L1*\sin\theta + F*L2)}{GRF*L} \tag{1.5}$$

Where,

m: total body mass (kg).

g: gravitational acceleration ($m.s^{-2}$).

θ: sway angle (degrees).

Fz: force of perturbation (N).

GRF: Ground reaction force (N).

L1: vertical distance between COM and ankle joint (m).

L2: vertical distance from the perturbation level to ankle joint (m).

L: horizontal distance between COP and ankle joint (m).

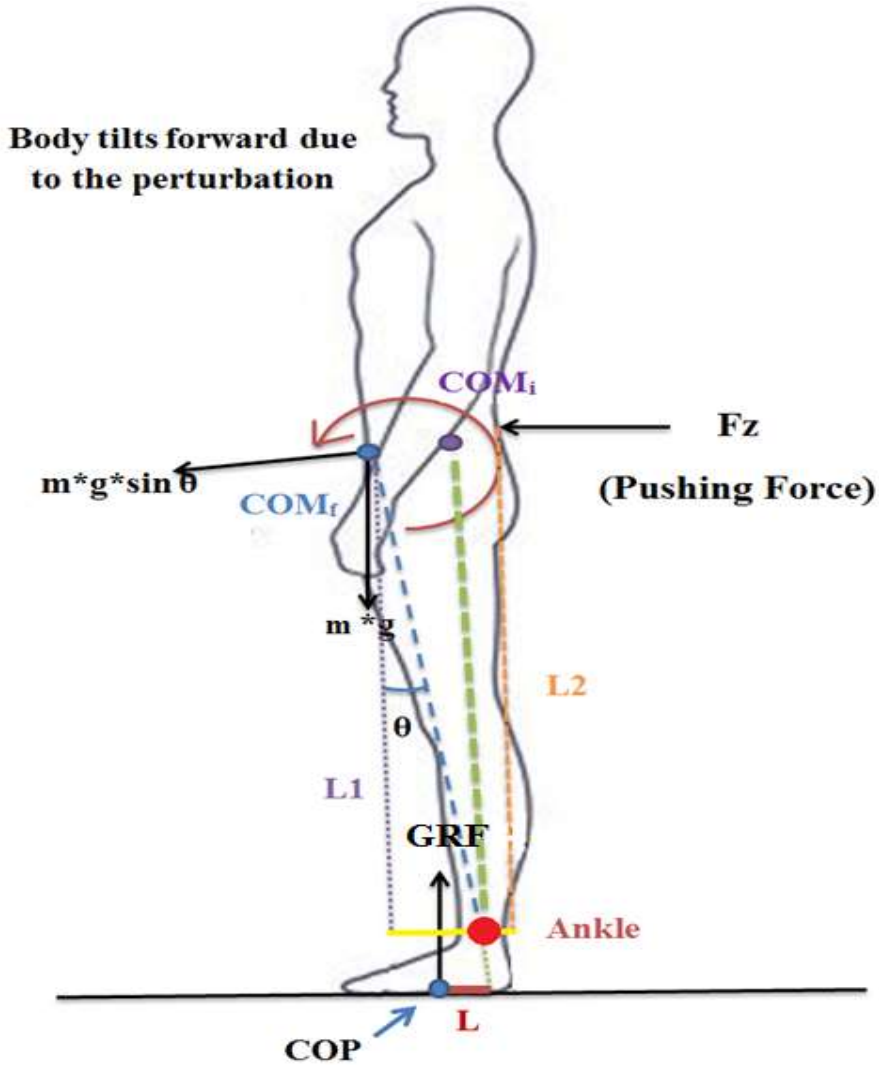

Figure 1.14 Free body diagram of body. Ankle is small open circle. θ= sway angle, m=mass, g= acceleration due to gravity. Fz= perturbing force, and GRF = vertical ground reaction force acting at the COP, COM_i =the initial position of center of mass before the perturbation, COM_f = the final position of COM due to the perturbation, COP is the center of pressure, L1 is the vertical distance between COM and ankle joint, L2 the vertical distance from the perturbation level to ankle joint, and L is the horizontal distance between COP and ankle joint.

CHAPTER 2

EXPERIMENTAL DESIGN AND METHODS

2.1 Overview

This study is the first of its kind to address possible control of exoskeleton balance by controlling the location of the COP with input from the arms and hands. To achieve a tight focus, this investigation has been designed with specific limitations.

Experiments involve only unperturbed and perturbed quiet standing. It is expected that subjects may employ ankle, hip, stepping and suspensory strategies. Studies will not include consideration of balance during walking. The latter is left for later investigation that will be stimulated by the findings of this dissertation.

The study also is limited to balance in the anterior/posterior direction and does not consider M/L direction.

In consideration of safety, only non-disabled subjects are involved in this study. While this work is intended to enhance the development of a new control strategy for exoskeleton walking and balance, no exoskeleton is used. Instead, custom fabricated passive devices are used to study subjects' ability to accommodate altered range of COP movement and to prevent non-disabled subjects from using their knee and ankle strategies and substituting hand and arm strategies for balance. This assumes that individuals with motor complete SCI cannot generate ankle torque necessary for a biological ankle strategy.

As discussed earlier, sensory feedback associated with balance includes vision, vestibular sensation and proprioceptive and somatosensory cues. The use of the shoes alters

the somatosensory cues and eliminates proprioceptive ankle and knee information. Conditions of eyes opened and eyes closed alter visual feedback. Only vestibular feedback is unaltered in this study.

There are four specific Aims/goals of this research:

1) Building a perturbation system using a linear actuator fixed on a height-adjustable stand. The height can be adjusted to set the actuator a little above the level of the pelvis. A force sensor in line with actuators tip to detect the force exerted on each push.

    a) Synchronizing the actuator and the force sensor with the laboratory's Motion capture system (camera system and force plates).

    b) Building shoe platforms with small Blocks attached to their base, with shoes attached on the top of the Blocks for the subjects to wear to be used in Aim 3 thus limit subject's range of their center of pressure.

    c) Building platforms with Pivots that have one platform on ground (fixed), and the other below subject's shoes (moving) connected by a pivot in between. These platforms prevent nondisabled subjects from adjusting their center of pressure using ankle torque and will be used in Aim4.

    d) Computing COM of human body based on the kinematics using markers trajectories of the camera system.

2) Studying normal human body response to perturbations with different forces with and without visual feedback, in A/P plane.
   Assessing human balance under external perturbation; by pushing subjects using an automated actuator at different ascending seven forces, and studying their reaction by investigating the following criteria: lower limb joint angles, COP, COM, COP-COM error signal, inclination angle between COP and COG vector, Stability index (SI). In addition to which strategy to use and why.

    - Sub-Aim 2.1 Studying normal human body response to perturbations of different forces while there is visual feedback (eyes opened) in A/P plane.

    - Sub-Aim 2.2 Studying normal human body response to perturbations of different forces while there is no visual feedback (eyes closed), in A/P plane.

3) Investigation of the human capability to adjust to a reduction in COP range with and without visual feedback, while perturbed with different forces, in A/P plane.

Using shoes designed with small Blocks attached to their soles, subjects stand on the platform, and when perturbed the COP moves forward to the end of the block and the subjects take a step (transition from ankle strategy to stepping strategy). The effect of minimizing the range of COP on balance will be studied.

- Sub-Aim 3.1 Investigation of the human capacity to adjust to a reduction in COP range with eyes opened (visual feedback available), in A/P plane.

- Sub-Aim 3.2 Investigation of the human capacity to adjust to a reduction in COP range with eyes closed (no visual feedback), in A/P plane.

4) Investigation of the capability of the hands and arms to accommodate for the confined COP range with and without visual feedback, while perturbed with different forces in A/P plane.

   This Aim is directly related to the exoskeleton project. Individuals with paraplegia due to spinal cord injury cannot adjust the torque of their ankles to control their center of pressure. Due to safety issues, we have chosen not to use subjects with SCI. Instead the specially constructed pivot shoes provide vertical support of standing but remove the ability of non-disabled subjects to control their center of pressure. Standing on Pivots makes retaining balance impossible. Two trekking poles that are used in our exoskeleton project design will be used to allow the use of the arms to control COP. This will increase the base of support and lets the subjects go back to full range of COP balance.

   - Sub-Aim 4.1 Investigation of the human capacity to adjust to a reduction in COP range with eyes opened (visual feedback available) in A/P plane.

   - Sub-Aim 4.1 Investigation of the human capacity to adjust to a reduction in COP range with eyes closed (no visual feedback) in A/P plane.

2.2 Subject Selection

Twelve healthy subjects, ages (18-40) were randomly selected from the NJIT community to be subjects for this experiment. They were assigned to two groups randomly:

1. Non – Randomized group – six subjects to experience perturbing forces arranged in ascending order.

2. Randomized group – six subjects to experience perturbing forces arranged in a random order.

Exclusion criteria included the following:

- Balance Disorder
- Inner Ear Problems
- Impairment of Gait
- Fear of Falling
- Leg Weakness
- Hip, knee, or Ankle Injuries
- Ringing In Ears
- Back Problems
- Whip Lash

2.3 Experimental Procedure for Each Aim

2.3.1 Aim 1: Assembling the Apparatus for a Perturbation/Motion Capture System

The Perturbation/Motion Capture System consists of:

- **Force Plates**

Force platforms are measuring instruments that measure the ground reaction forces generated by a standing or moving body across them. They are used for many purposes such as to quantify balance, gait and other parameters of biomechanics. The simplest force platform is a plate with a single pedestal, instrumented as a load cell which is a transducer that is used to create an electrical signal whose magnitude is directly proportional to the force being

measured. Force plates that we have are advanced; they measure the three-dimensional components of the single equivalent force applied to the surface and its point of application (COP) [46].

The measurements from a force platform can be either studied in isolation, or combined with other data, such as limb kinematics such as in this research to understand the principles of locomotion. Motion capture measurements of leg joint angles and force plate output can allow the determination of torque, work and power at each joint using a method called inverse dynamics [1-4, 6, 12, 14, 46-49].

Force plates used in this experiment are OR6-7-2000 force plates from AMTI [50]. They have been synchronized with laboratory's motion capture system using custom MATLAB software. In these experiments, data are collected at 1000 frames per second.

- **Motion Capture System**

Motion capture is the process of tracking the movement of objects or people. In motion capture sessions, movements of moving objects (in our case human body) are sampled in a specific sampling frequency. Images from multiple cameras are used to calculate 3D positions to record only the movements of the actor, regardless of his or her visual appearance [46].

The laboratory's 12 camera Naturalpoint OptiTrack captures the 3-D positon of passive markers at 100 frames/second with sub-millimeter accuracy. Its Motive software allows calibration, data capture, and marker trajectory correction (e.g, accommodating marker occlusion and ambiguity) [51]. A software generated synchronization pulse time syncs the motion capture data with force plate data and EMG.

- **Perturbation Actuator**

A 4" Stroke Firgelli Automation 12VDC 35lb-force linear actuator is used to provide the horizontal perturbation to subjects [52].

To adjust the height of the actuator, since subjects have different anthropometrics, and to ensure that all subjects are perturbed at the height of their COM, adjustable-height stand was built using two height adjustable camera tripods carrying a horizontal piece of 80/20 extrusion, one of the 80/20 ends is in contract with a rigid wall. The Firgelli actuator with attached force sensor is mounted at the other end of the extrusion using a custom designed coupling 3D printed ABS. On the distal end of the force sensor a soft pad printed in thermoplastic elastomer (TPE) protects the user at the point of actuator contact. The 80/20 can move forward and backward to adjust the actuator position in the sagittal plane.

An Arduino UNO microcontroller is used to control the actuator. Arduino code detects actuation signals sent from the controlling PC and generates a pulse-width modulated signal that extends and retracts the linear actuator [53]. A high current motor shield interfaces the Arduino to the actuator [54].

- **Force sensor**

The force sensor used in this research is Optoforce- OMD-45-FE-1000N, which is a 3D sensor that measures the magnitude and the direction of Fx, Fy, and Fz forces based on optical principles. Only Fz is used in this study to detect the time-varying force applied to the back of the subjects. Custom MATLAB code reads the sensor at 100 frames per second. Force data are synchronized with the motion capture and force plate data.[55].

- **EMG**

The EMG recording equipment used in this project is the Delsys Bagnoli 8channel system. This uses parallel-bar EMG sensors and includes an array of features designed to make EMG recordings effortless and consistent. Collection is done via a custom MATLAB function that is integrated with force plate data acquisition[56].

- **MATLAB**

MATLAB Software is used to integrate the equipment (triggering the camera system, force plates, Arduino-Monster Moto shield, and force sensor), synchronization, collection of the data, and post-processing the results.

- **SPSS**

SPSS (Statistical Package for the Social Sciences) was used to find the statistical results that are presented in Chapter 3.

- **Experimental Shoes**

Three types of shoes were used in these experiments and are referred to as Regular, Blocks and Pivots, shown in Figure 2.1.

Two sizes of running shoes were purchased to accommodate male and female subjects. In the study of Regular balance, Aim 2, subjects wore these shoes as they would their own shoes.

For Aim two, subjects wore modified versions of the running shoes. A Delran block whose length is less than that of the sole of the shoe is rigidly attached immediately below the user's COM. This has the effect of dramatically reducing the possible range of the COP.

A third pair of running shoes (both men's and women's sizes) were attached to a one degree of freedom pivot, which was mounted on a Delran plate whose length was the same

as the original shoe. The pivot joint was placed immediately below the A/P location of the COM. The ball bearing in the pivot prevents the use of ankle torque to control the location of the COP.

Attached to the Delran plate below the pivot joint of each shoe is a vertical trekking pole that allows the user to apply torques to the platform beneath the Pivots. This torque is used by subject to provide torque that modifies the position of the COP.

A **B** **C**

Figure 2.1 A: The commercial running shoes used in Aims 2-4. B: Blocks shoes with the range of COP limited. C: Pivots shoes, Pivots located below the expected COP.

2.3.2 Aim 2: Studying Normal Human Body Response to Perturbations of Different Forces (Regular), in A/P Plane

Assessing human balance to external perturbation by studying the response to pushing subjects using an automated actuator at seven different forces, and studying their reaction by investigating the following parameters: lower limb joint angles, COP, COM, COP-COM error signal, Stability index (SI), EMGFOR Lateral Gastrocnemius and Tibialis anterior muscles , stepping strategy (number of steps and total forward displacement). In addition to characterizing which strategy to use and why.

- Sub-Aim 2.1 Studying normal human body response to perturbations of different forces with visual feedback available (eyes opened), in A/P plane.

The following procedure was performed on each subject:

1) The subjects were asked to stand on the force plates in front of the actuator. The actuator will be adjusted to perturb the user in the lower back region at belly button level. A calibration was performed at the beginning of each perturbation using the Optoforce force sensor to ensure the user returns to the same starting position for every trial.

2) The position of the tip of the shoes will be marked. And the subject will be asked to go back to the same position every time he/she steps.

3) The users will be instructed to close or open their eyes after the calibration.

4) The perturbation system will be activated when the user is relaxed and in position. The actuator will Push the subject seven times with an ascending no load speeds of (0.1121, 0.1255, 0.1434, 0.1614, 0.1793, 0.1972, 0.2286 (cm/s)) which are proportions of the maximum no load speed of the actuator which is nine inch/s (0.2286 m/s). Each speed will result in increasing perturbation forces.

5) The synchronized data from the optitrack system, force plates, force sensor, and EMG were collected during each trial.

6) The above procedure will repeated seven times with increasing force during each trial.

This procedure was repeated three times, the time estimated for each block of seven perturbations is estimated to be 220 sec.

- Sub-Aim 2.2 Studying normal human body response to perturbations of different forces while there is no visual feedback (eyes closed), in A/P plane.

  Experimental procedure same as Sub – Aim 2.1 but the subjects performed three trials with seven perturbations with eyes closed (with no visual feedback).

For each experiment there are two protocols:

1) With visual feedback (eyes opened (EO)).

2) Without visual feedback (eyes closed (EC)).

All subjects were included in the two protocols, with and without visual feedback protocol. Both protocols of each experiment were performed in a separate day after each other. Order of performing the two protocols was randomized (half of the subjects performed eyes opened first while the other half performed eyes closed first). Data from Sub – Aims 2.1 and 2.2 were used as a **control**.

Figure 2.2 A subject standing in front of the perturbing system.

2.3.3 Aim 3: Investigation of the human capability to adjust to reduction in COP range, while perturbed with different forces

Subject wear the shoes with Blocks and when perturbed this moves COP forward to the end of the Blocks and then to take a step (transition from ankle strategy to stepping strategy). The effect of minimizing the range of COP on balance was studied.

- Sub-Aim 3.1 Investigation of the human capability to adjust to a reduction in COP range, using the shoes with small Blocks, in A/P plane.

This sub-Aim follows the same procedure of sub-Aim 2.1.

- Sub-Aim 3.2 Investigation of the human capability to adjust to a reduction in COP range, using the shoes with Blocks while there is no visual feedback, in A/P plane.

  This sub-Aim follows the same procedure of sub-Aim 2.2.

Figure 2.3 The above Figure shows the reduction COP range of shoe with Blocks compared to Regular shoe.

2.3.4 Aim 4: Investigation of the human capability to use hands with trekking poles to accommodate for the confined COP range, while perturbed with different forces

A specially constructed pivot shoes provide vertical support of standing but remove the ability of non-disabled subjects to control their center of pressure. Standing on Pivots makes retaining balance impossible. Two trekking poles that are used in our exoskeleton project design will be used to allow the use of the arms to control COP. This will increase the base of support and let subjects go back to full range of COP balance.

- Sub-Aim 4.1 Investigation of the human capability to use hands with trekking poles to accommodate for the confined COP range, in A/P plane.

  This sub-Aim follows the same procedure of sub-Aim 2.1.

- Sub-Aim 4.2 Investigation of the human capability to use hands with trekking poles to accommodate for the confined COP range while there is no visual feedback, in A/P plane.

  This sub-Aim follows the same procedure of sub-Aim 2.2.

Subjects were asked to wear specially designed shoes that prevent the use of ankle torques to alter the COP. The subject will hold two trekking poles that are attached to the base of the shoes. Rather than use ankle strategy to control COP changes, this experiment will examine the ability of the subject to use his/her hands to alter the COP position by applying torques via the trekking poles.

Moreover, subjects were asked to wear ankle braces to limit dorsiflexion and plantar flexion of the feet to restrict the ability of tipping on toes or heels.

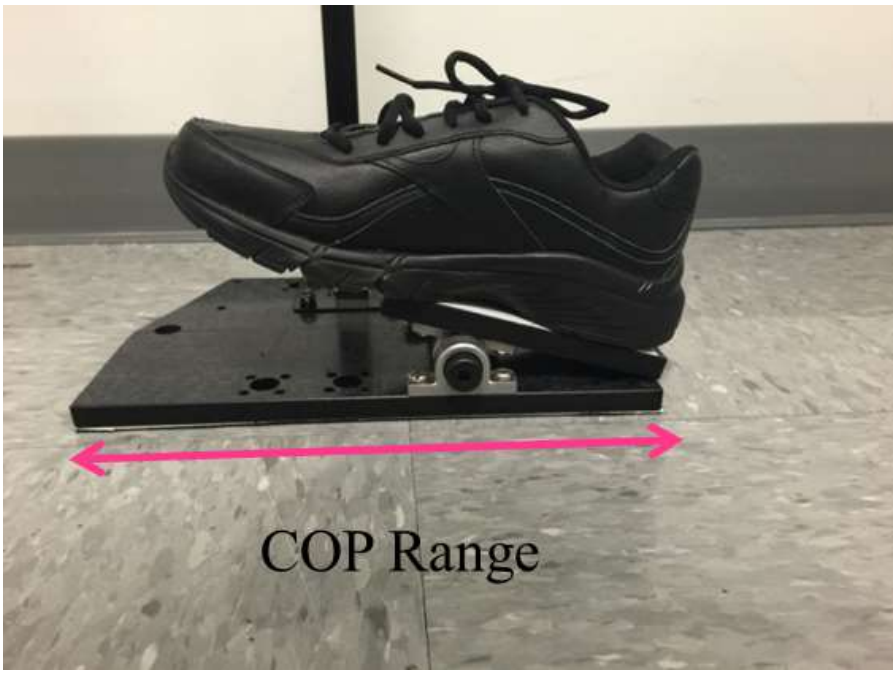

Figure 2.4 Shoe with a pivot and trekking pole to study the ability of the hands to compensate for the confining COP in A/P plane.

For safety, an air mattress will be put in front of subjects in case they can't accommodate with the push in the three previous experiments. In addition, observers will stand beside the subject to prevent any fall. In addition a 3-D printed pad made of TPE rubbery material has been evaluated, tried and results in a comfortable perturbation punching was noticed.

Figure 2.4 shows one of the shoes with a pivot and trekking pole to study the ability of the hands to compensate for the confined COP in A/P plane. Figure 2.5 shows a subject wearing the shoes with Pivots and catching the trekking poles in front of the perturbation system.

Figure 2.5 A subject wearing the shoes with Pivots and catching the trekking poles, in front of the perturbation system.

Table 2.1 Experimental Design Summary

Non-random group	Random group
	Regular (Ascending perturbing forces order) EO/EC
Regular (Ascending perturbing forces order) EO/EC	Regular (Random perturbing forces order) EO/EC
Blocks (Ascending perturbing forces order) EO/EC	Blocks (Random perturbing forces order) EO/EC
Pivots (Ascending perturbing forces order) EO/EC	Pivots (Random perturbing forces order) EO/EC

Table 2.1 shows a summary of the experimental design for both Non-Randomized and Randomized groups. It is noticed that Regular (Ascending perturbing forces order) experiment was added to the Randomized group in purpose to do within subjects statistical comparison for the studied parameters.

- **Stepping Strategy Evaluation**

Two parameters were studied to statistically assess **Stepping Strategy**:

1) Total forward displacement: the total forward displacement of the last step with respect to the original position.

2) Number of Steps.

The parameters that were studied to statistically analyze the differences between the three experiments (Regular, Blocks, and Pivots):

a. Stability index: the stability index values that were used are the stability index signal peaks that are locally aligned with the first peak of the force signal of the actuator.

b. Error signal peaks: where the error signal is the difference between COP and COM.

c. The correlation between COM and COP for the whole signal and for each perturbation.

- **Lower limbs joint angles**

Ankle angle was computed as the angle between the vector formed by ankle joint marker (that is placed on the ankle marker) and toe marker (that is placed on the big toe of the foot) and the horizontal plane. While knee angle was computed as the angle between the vector formed by ankle joint marker and knee marker (that is placed at the lateral side of the knee) and the horizontal plane. While hip angle was computed as the angle between the vector formed by (knee marker and the hip marker that is placed on the ispina illiaca anteroposterior of the hip) and the horizontal plane.

- **Stability Index**

The SI was calculated only when the both the foot of the subjects were aligned in the plane. Hence the gaps in the SI data are when subject's feet were not aligned either due to step or when they were returning to their position. For statistical analysis purposes the SI values that have been used are the SI peaks that are locally aligned to the perturbing force peaks.

In addition, when applying the stability index equation on Pivots experiment's data, the model was modified to match the fact that the linear inverted pendulum model in this case is oscillating about the Pivots instead of ankle joints.

- **Statistical Model**

SPSS was used to run the statistical analysis. The statistical model that is used for data analysis consists of three main comparisons:

1) Non-Randomized and Randomized comparisons.

The difference between Non-Randomized and randomized data of the randomized group for the first experiment was analyzed, applying paired T-Test or Wilcoxon test

(non-parametric) in case the data failed to pass the normality test (significance level (α) = 0.05) to study the difference between the following parameters:

a) Total forward displacement (Regular, Non-Randomized vs Randomized, EO and EC).

b) Number of steps (Regular, Non-Randomized vs Randomized, EO and EC).

c) Stability index (Regular, Non-Randomized vs Randomized, EO and EC).

d) Error signal peaks (Regular, Non-Randomized vs Randomized, EO and EC).

e) Correlation between COM and COP for (Regular, Non-Randomized vs Randomized, EO and EC).

2) First (Regular), second (Blocks) and third (Pivots) experiments comparison.

Here the difference between the different experiments, the first (Regular), second (Blocks), and third (Pivots) experiments is studied for each perturbation separately except for the correlation between COM and COP, that is done both for the entire signal and for each perturbation too. Applying paired T-Test or Wilcoxon test (non-parametric) in case if the data fails normality when comparing parameters applicable for only Regular and Blocks experiments (significance level (α) = 0.05). And repeated measures ANOVA (rmANOVA) and then paired T-Test to compare all possible comparisons whenever rmANOVA gives a significant difference when studying the difference between the three experiments (α = 0.1667, Bonferroni correction is used when comparing three different conditions). The differences between the following parameters:

a) Total forward displacement (Regular vs Blocks, EO and EC).

b) Number of steps (Regular vs Blocks, EO and EC).

c) Stability index (Non-Randomized, Regular vs Blocks vs Pivots, EO and EC).

d) Stability index (Randomized, Regular vs Blocks vs Pivots, EO and EC).

e) Error signal peaks (Non-Randomized, Regular vs Blocks vs Pivots, EO and EC).

f) Error signal peaks (Randomized, Regular vs Blocks vs Pivots, EO and EC).

g) Correlation between COM and COP (Non-Randomized, Regular vs Blocks vs Pivots, EO and EC).

h) Correlation between COM and COP (Randomized, Regular vs Blocks vs Pivots, EO and EC).

3) Eyes opened (EO) vs Eyes closed (EC).

On here, the effect of the visual feedback on perturbation during quiet standing is studied. Comparisons between eyes opened and eyes closed trials within each experiment is done for all parameters (significance level (α) = 0.05). The following comparisons were studied:

a) Total forward displacement (EO vs EC, Regular, and Blocks).

b) Number of steps (EO vs EC, Regular and Blocks).

c) Stability index (Non-Randomized, EO vs EC, Regular, Blocks and Pivots).

d) Stability index (Randomized, EO vs EC, Regular, Blocks and Pivots).

e) Error signal peaks (Non-Randomized, EO vs EC, Regular, Blocks and Pivots).

f) Error signal peaks (Randomized, EO vs EC, Regular, Blocks and Pivots).

g) Correlation between COM and COP (Non-Randomized, EO vs EC, Regular, Blocks and Pivots).

h) Correlation between COM and COP (Randomized, EO vs EC, Regular, Blocks and Pivots).

All of the statistical analysis in this study is done within subjects.

Repeated measures ANOVA test was used to find if there is any possible signifiicant difference among the three possible combinations of the data (Regular vs Blocks, Blocks vs Pivots, and Regular vs Pivots). Table 4.11 shows a summary of the statistical results. Bonefroni correction is used ($\alpha = (0.05/3) = 0.01667$), therefore, it is harder to get probability. Sphericity test results were taken in consideration, if data do not pass sphericity test, Greenhouse-Geisser test is used instead of Regular rmANOVA. Post-hoc.

NJIT IRB approval was issued for the proposed project (IRB Protocol Number: F299-17, 12-6-2016) .

CHAPTER 3

PILOT STUDIES

3.1 Center of Mass

In this experiment the kinematic method was used to find COM. Winter's anthropometric data Table 3.1 was used in this experiment for being well known because of their accuracy and since they are frequently used in research. Fifteen Markers (Figure 3.1) were attached on specific proximal and distal bony landmarks of several segments. The anthropometric model is composed of ten segments two (shanks and feet), two thighs, trunk, head, two (lower arms and hands) and two forearms.

Table 3.1 Winter's Anthropometric data [3]

Segment	Definition	Segment Weight/ Total Body Weight	Center of Mass/ Segment Length		Radius of Gyration/ Segment Length			Density
			Proximal	Distal	C of G	Proximal	Distal	
Hand	Wrist axis/knuckle II middle finger	0.006 M	0.506	0.494 P	0.297	0.587	0.577 M	1.16
Forearm	Elbow axis/ulnar styloid	0.016 M	0.430	0.570 P	0.303	0.526	0.647 M	1.13
Upper arm	Glenohumeral axis/elbow axis	0.028 M	0.436	0.564 P	0.322	0.542	0.645 M	1.07
Forearm and hand	Elbow axis/ulnar styloid	0.022 M	0.682	0.318 P	0.468	0.827	0.565 P	1.14
Total arm	Glenohumeral joint/ulnar styloid	0.050 M	0.530	0.470 P	0.368	0.645	0.596 P	1.11
Foot	Lateral malleolus/head metatarsal II	0.0145 M	0.50	0.50 P	0.475	0.690	0.690 P	1.10
Leg	Femoral condyles/medial malleolus	0.0465 M	0.433	0.567 P	0.302	0.528	0.643 M	1.09
Thigh	Greater trochanter/femoral condyles	0.100 M	0.433	0.567 P	0.323	0.540	0.653 M	1.05
Foot and leg	Femoral condyles/medial mafleolus	0.061 M	0.606	0.394 P	0.416	0.735	0.572 P	1.09
Total leg	Greater trochanter/medial malleolus	0.161 M	0.447	0.553 P	0.326	0.560	0.650 P	1.06
Head and neck	**C7-T1 and 1st rib/ear canal**	**0.081 M**	**1.000**	**— PC**	**0.495**	**1.116**	**— PC**	**1.11**
Shoulder mass	Sternoclavicular joint/ glenohumeral axis	—	0.712	0.288	—	—	—	1.04
Thorax	C7-T1/T12-L1 and diaphragm*	0.216 PC	0.82	0.18	—	—	—	0.92
Abdomen	T12-L1/L4-L5*	0.139 LC	0.44	0.56	—	—	—	—
Pelvis	L4-L5/greater trochanter*	0.142 LC	0.105	0.895	—	—	—	—
Thorax and abdomen	C7-T1/L4-L5*	0.355 LC	0.63	0.37	—	—	—	—
Abdomen and pelvis	T12-L1/greater trochanter*	0.281 PC	0.27	0.73	—	—	—	1.01
Trunk	Greater trochanter/ glenohumeral joint*	0.497 M	0.50	0.50	—	—	—	1.03
Trunk head neck	Greater trochanter/ glenohumeral joint*	0.578 MC	0.66	0.34 P	0.503	0.830	0.607 M	—
HAT	Greater trochanter/ glenohumeral joint*	0.678 MC	0.626	0.374 PC	0.496	0.798	0.621 PC	—
HAT	Greater trochanter/mid rib	0.678	1.142	—	0.903	1.456	—	—

*NOTE: These segments are presented relative to the length between the greater trochanter and the glenohumeral joint.

Source Codes: M, Dempster via Miller and Nelson; *Biomechanics of Sport*, Lea and Febiger, Philadelphia, 1973. P, Dempster via Plagenhoef; *Patterns of Human Motion*, Prentice- Hall, Inc. Englewood Cliffs, N.J., 1971. L, Dempster via Plagenhoef from living subjects; *Patterns of Human Motion*, Prentice-Hall, Inc., Englewood Cliffs, N.J., 1971. C, Calculated.

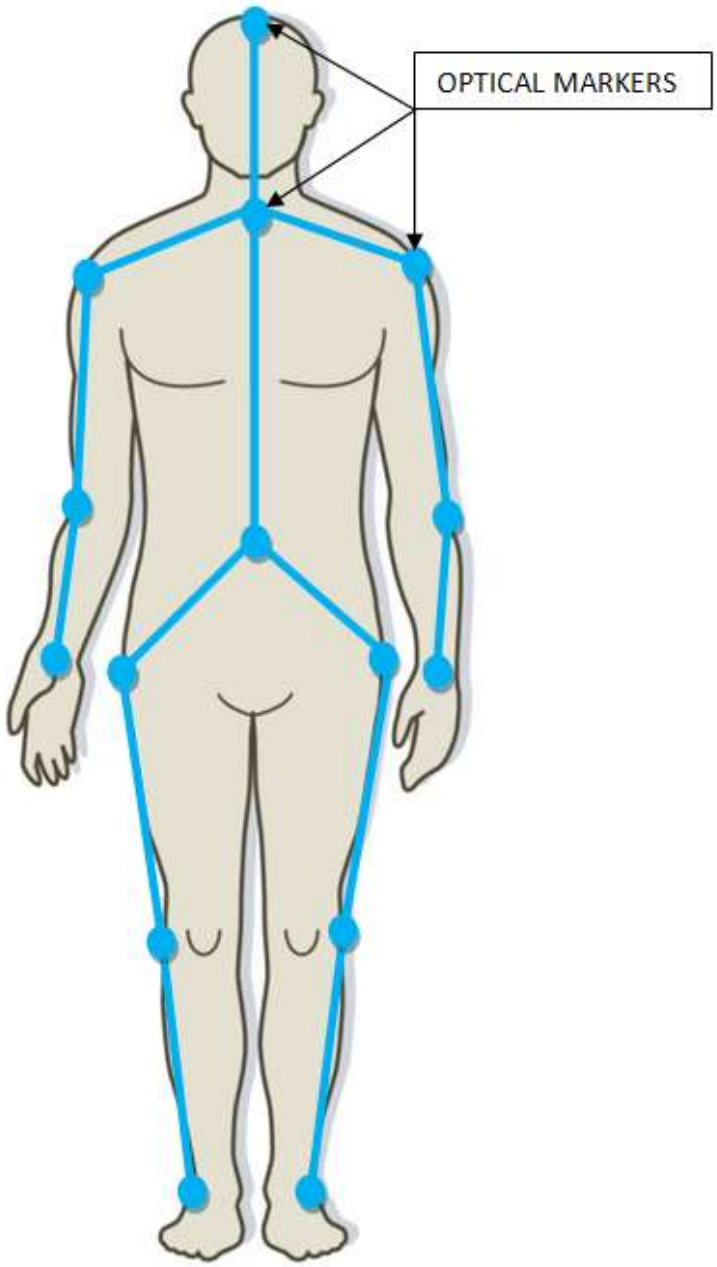

Figure 3.1 A free body diagram of human body with optical markers positions and main body segments used to find COM.

"It is important to notice that, the accuracy of the COM location is related to the validity of the mass inertia parameters providing the COM position and mass fraction of each segment of the model" [18].

Twelve OPTITRAK cameras were used to record markers displacement during the experiment at a sampling frequency of 100 Hz. The COM location in a given direction is calculated as follows:

$$COM = \Sigma\ (com_i \times m_i)\ /\ N \qquad (3.1)$$

Where,

M is the total body mass, mi the mass of ith segment, com_i is the coordinate of ith segment and N the number of segments defining the body COM.

One subject was asked to walk within the area of the motion capture system. The subject's markers trajectories were recorded using the camera system. A MATLAB code was established based on the kinematic method to compute COM, using trajectories data COM was computed for about 3 seconds (Figure 3.2) to confirm that the code works in a right way.

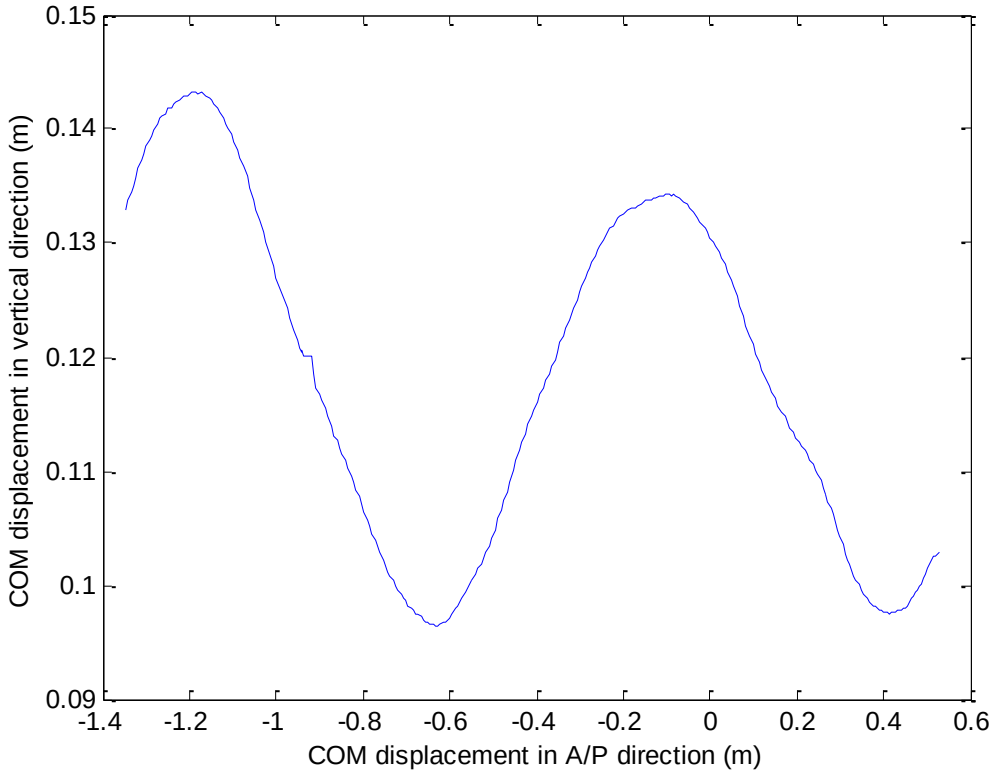

Figure 3.2 COM of a walking subject in the sagittal plane using my COM code in A/P plane.

3.2 COP-COM Relationship (Self-Perturbation)

The aim of this study is to find the relation between the COP and COM signals in A/P plane. In this experiment, the correlation coefficient is used to prove the strong relation between COM and COP.

One subject was asked to stand on one force plate with markers. Then he was asked to tilt forward and backward around his ankles (using ankle strategy), COM was calculated, then the correlation between COP and COM was found using SPSS. Figure 3.3 shows COM and COP in A/P plane.

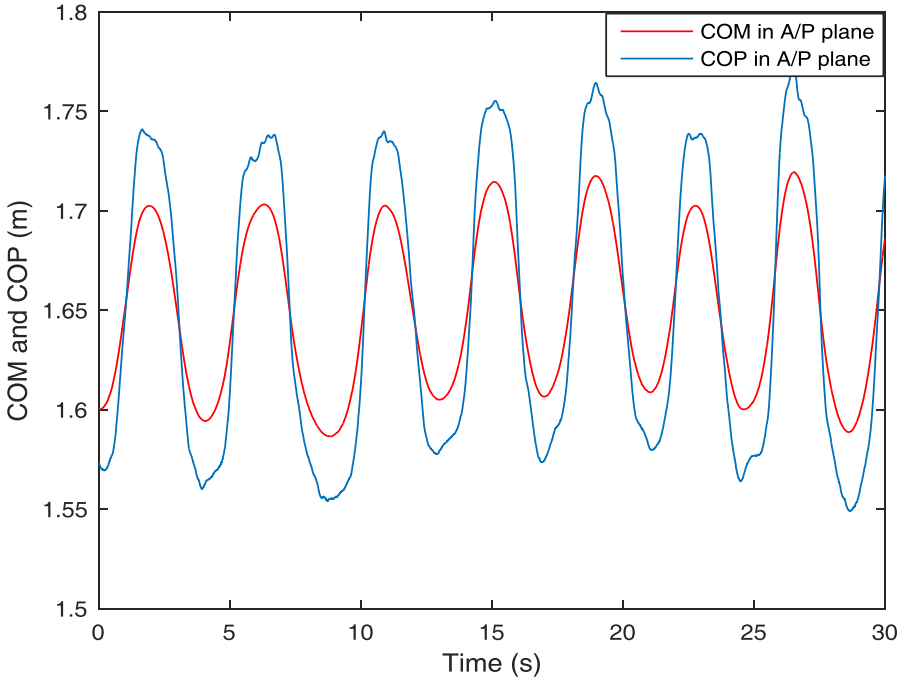

Figure 3.3 COM and COP of a subject tilting forward and backward.

SPSS was used to find the correlation between COM and COP in A/P plane. COP and COM are normally distributed.

Figure 3.4 shows the correlation between COM and COP. COM and COP are significantly correlated with a two-tailed 0.01 level with a pearson's Correlation of *(r=0.992).*

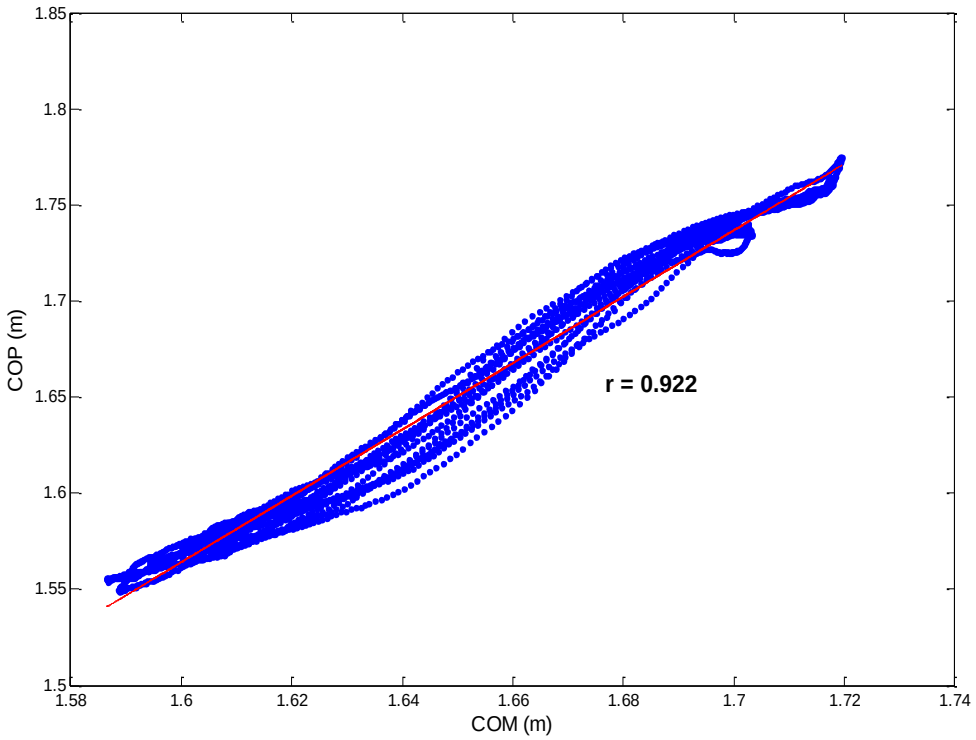

Figure 3.4 The correlation between COM- COP in A/P plane.

3.3 Perturbation Trial

The pilot data for Aim 2 is shown below with one subject. The methodology for the protocol is described as part of Aim 2 and was followed for the pilot data.

The reaction of COP and COM in sagittal plane is shown below. Before perturbation COM and COP are aligned. Pushing leads to COM movement forward, COP follows COM in order to retain balance. After a while, COP catches with COM, goes a little further than COM, peaks and goes back to its original position achieving balance.

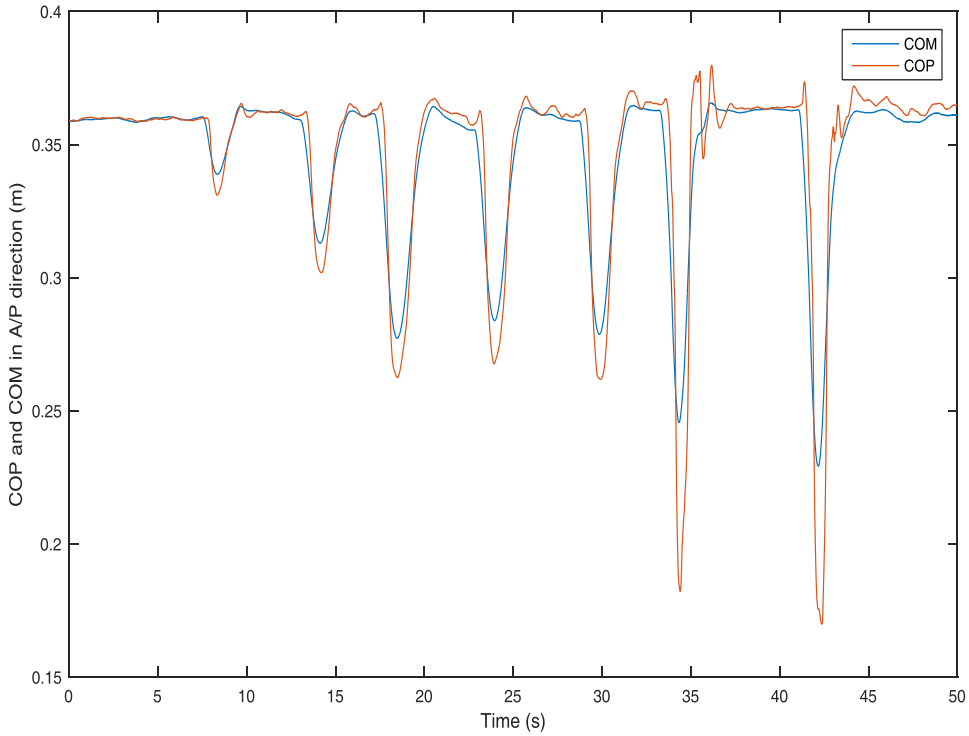

Figure 3.5 COM and COP response to multiple perturbations with different forces.

SPSS was used to find the correlation between COP and COM in A/P plane. COP and COM are normally distributed. Figure 3.6 shows the correlation between COM and COP. COM and COP are significantly correlated with a two-tailed 0.01 level with a pearson's Correlation of *(r=0.942).*

Using this correlation it could be possible to use COP to predict COM.

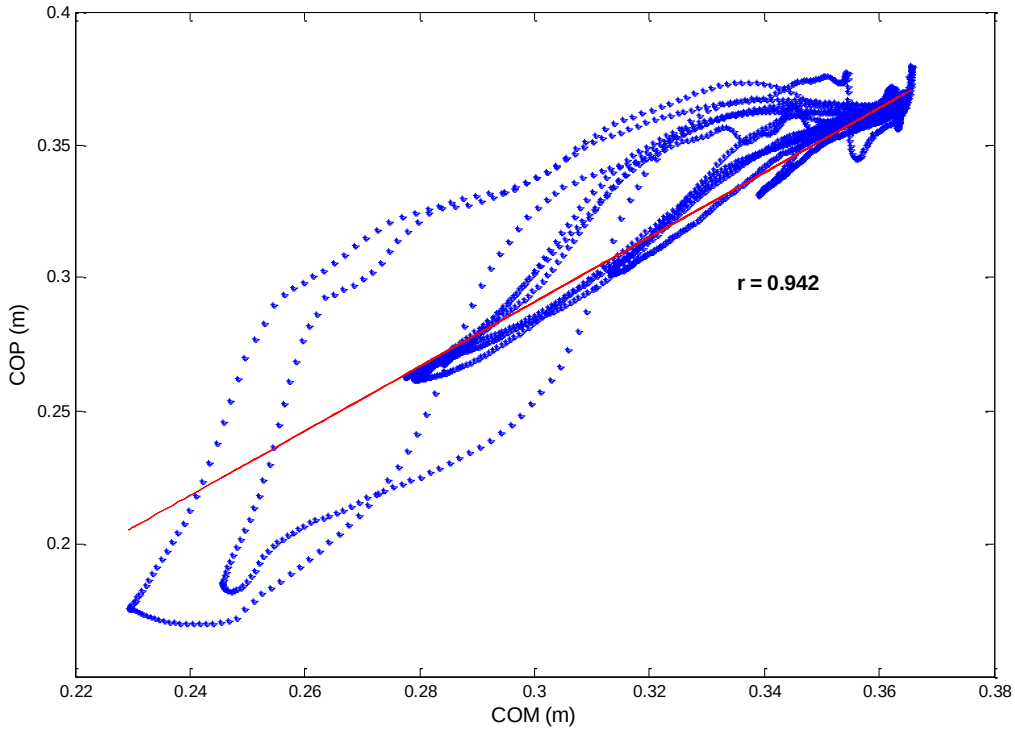

Figure 3.6 The correlation between COM- COP in A/P plane.

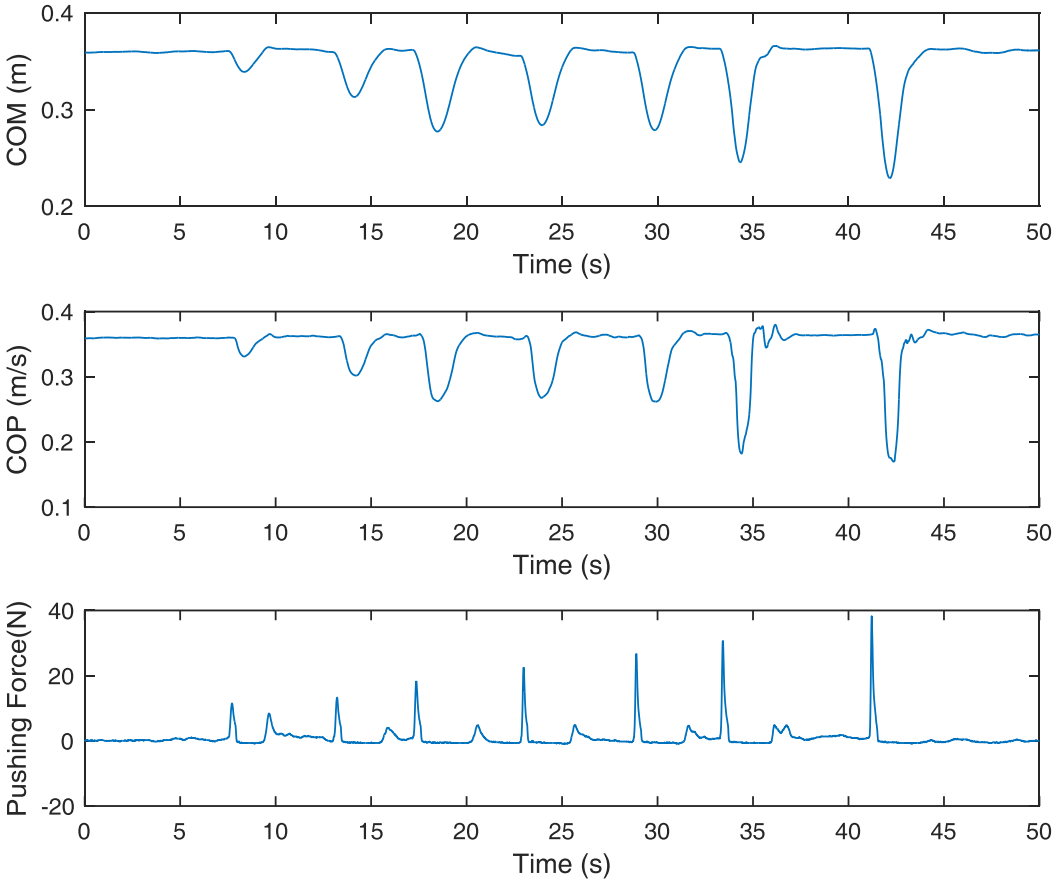

Figure 3.7 COM and COP (m) in frontal plane, corresponding to pushes forces (N).

The Figure 3.7 shows COM and COP changes in sagittal plane, corresponding to perturbation forces in Newton. As can be observed, COM and COP increases as the pushing force increase.

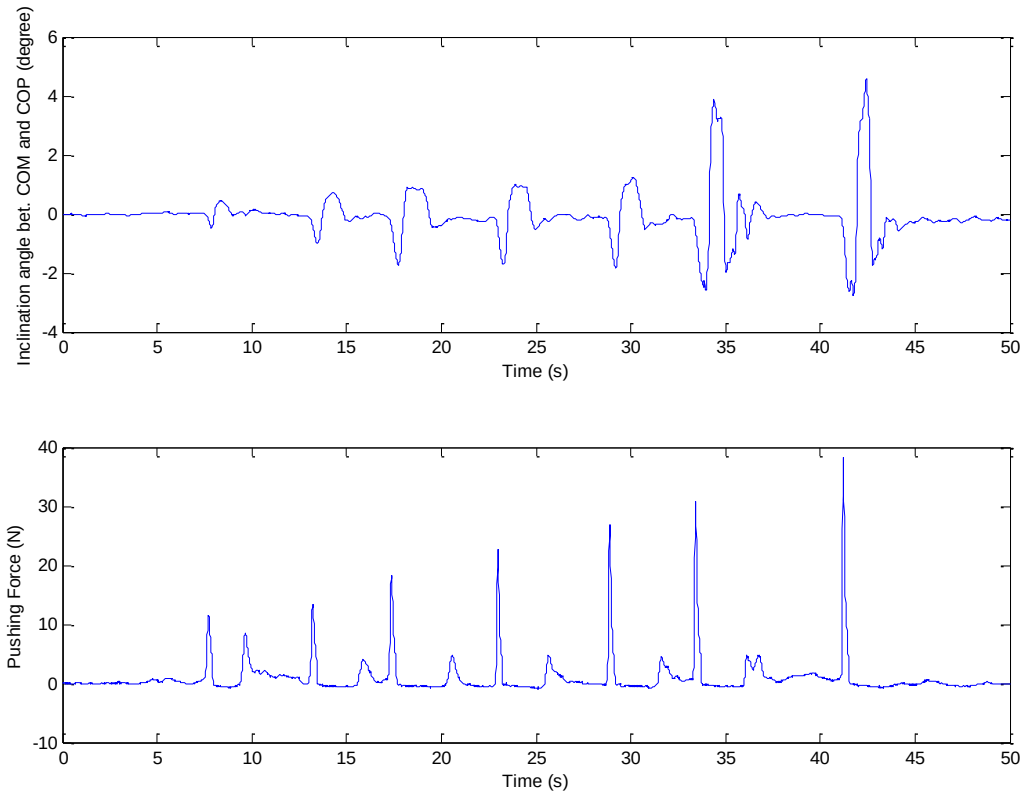

Figure 3.8 Inclination angle between COM and COP in response to multiple pushes with different forces.

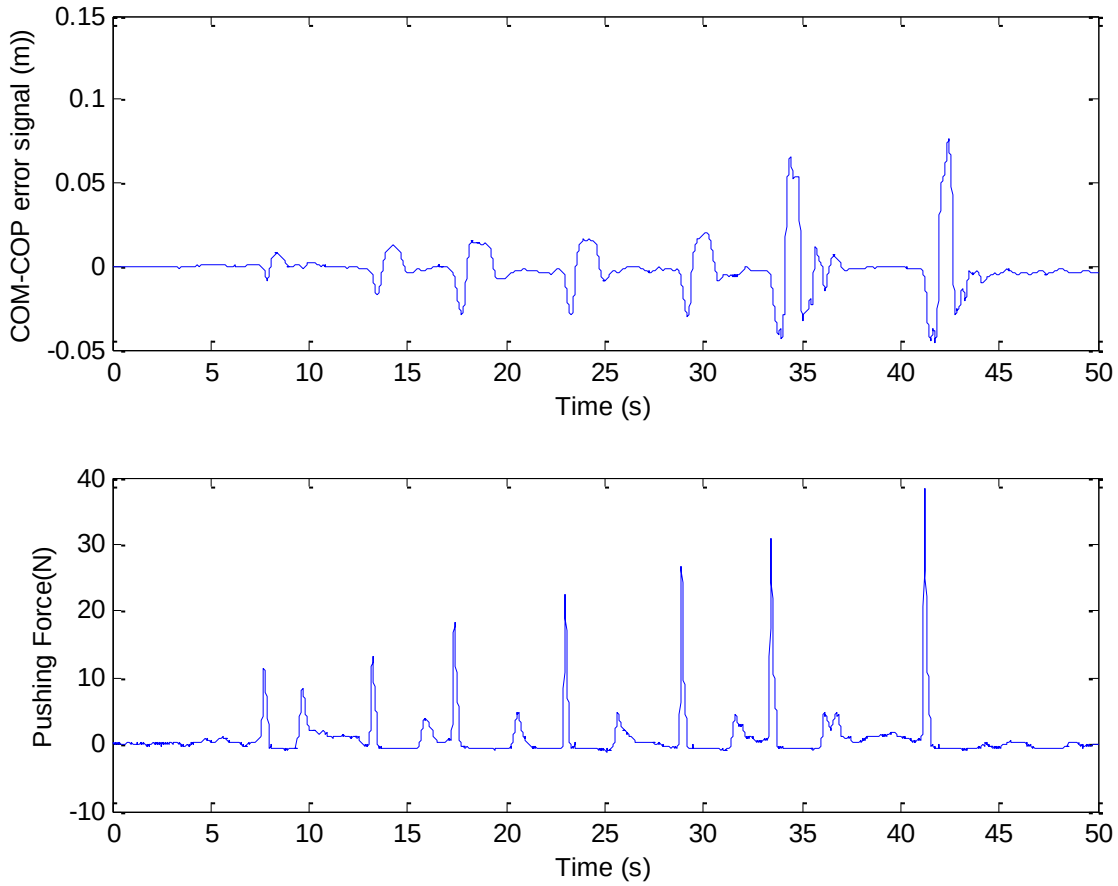

Figure 3.9 Error signal (The difference between COM and COP) in response to multiple pushes with different forces.

In Figures 3.8 and 3.9 it can be seen that inclination angle signal (which is the angle between COP and COG vector in A/P plane) and the error signal (which is the difference between COM and COP in A/P plane) looks like each other, and this is predictable since inclination angles are the inverse tan of the error signal over the change of COM in the vertical direction which is small.

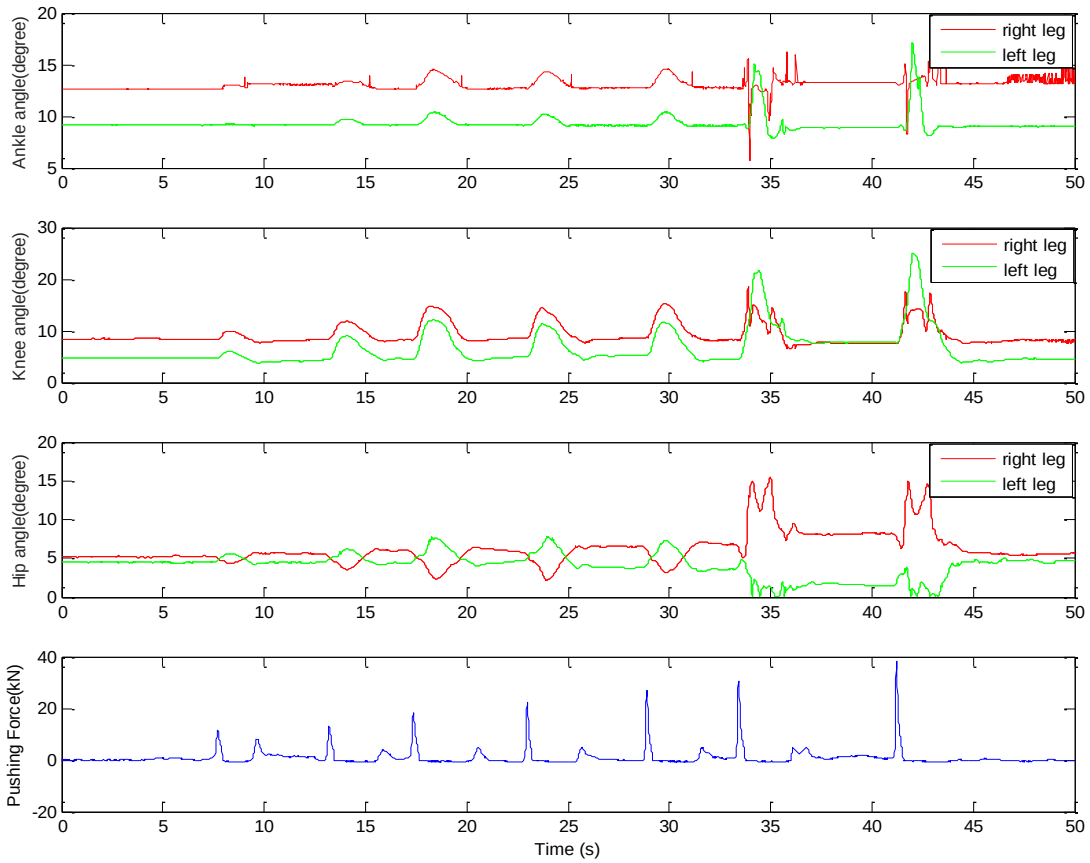

Figure 3.10 Lower limb joint (ankle, knee, and hip) angles in A/P plane in response to multiple pushes with different forces.

Figure 3.10 shows lower limbs joint angles in A/P plane. Lower extremity joint angles change rapidly after each perturbation indicating lower limb segments movement to retain balance. Ankle angle changes before stepping represents ankle strategy, while a hip angle change represents hip strategy. It can be concluded that knee angle changes more than the ankle and hip angles, even though it is not classified as a strategy in itself. Drastic changes in joints angles are observed when the subject takes a step (last two perturbations).

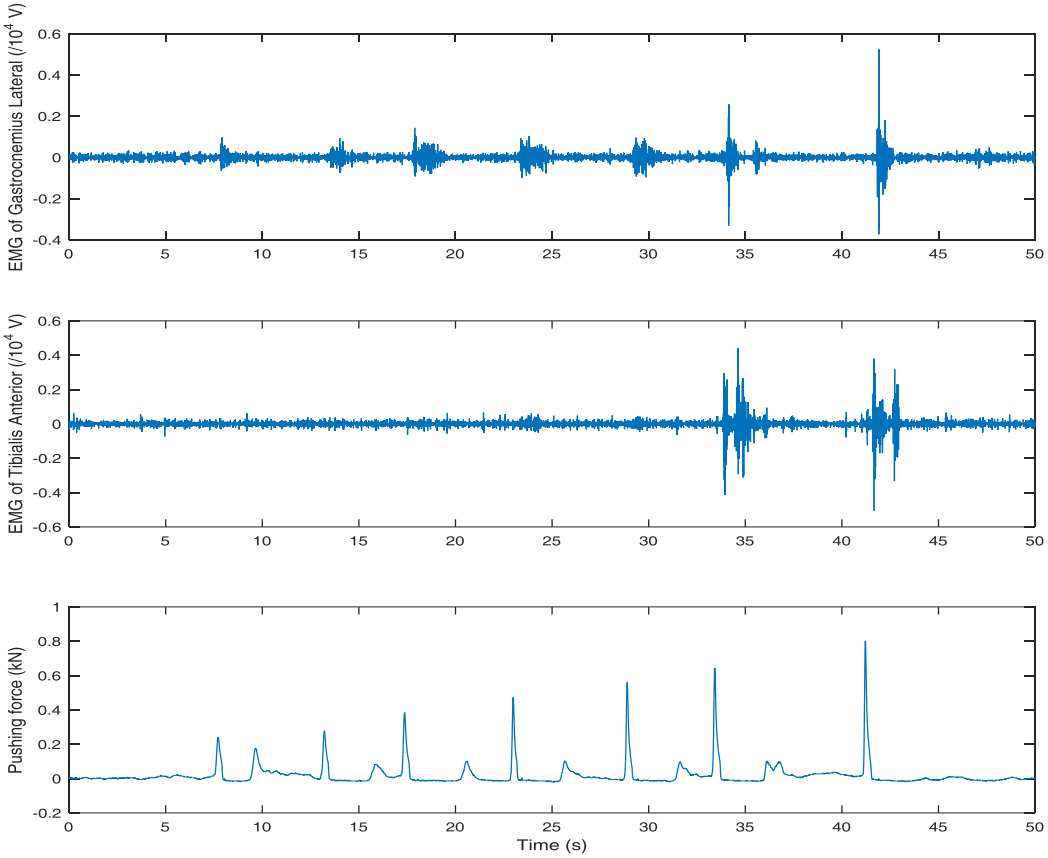

Figure 3.11 EMG of Tibialis anterior and Lateral Gastrocnemius of the dominant leg-stepping in response to multiple perturbations with different forces.

Two muscle EMG data were recorded: Tibialis anterior and Lateral Gastrocnemius of the right leg (dominant for this subject).

Table 3.2 shows the onset frame number of each push, COM, COP, EMG of Lateral Gastrocnemius muscle, and time difference between each push onset (Fz onset) and each corresponding COM movement onset, each push onset and each corresponding COP movement onset, each push onset and each corresponding Lateral Gastrocnemius muscle EMG onset. EMG onset was detected depending on a 3*STD level threshold, while other data onset was detected using a 2*STD threshold. EMG, threshold was calculated as the

mean of at least 100 or more baseline of quiet standing data+/- 3* STD of the same set of data. On the other hand, threshold for the rest of the other parameters was calculated as the mean of at least 100 baseline samples of quiet standing data+ 2* STD of the same set of data.

Table 3.2 The Onset Frame Number of Each Perturbation, and COM, COP, EMG of Lateral Gastrocnemius Muscle after each Perturbation. In Addition to the Time Difference between each Perturbation Onset (Fz onset) and each Corresponding COM, COP and Lateral Gastrocnemius Muscle EMG onset

Push number	No load Speed (m/s)	Fz onset frame number	COM onset frame number	COP onset frame number	EMG onset frame number	COM Time Shift (s)	COP Time Shift (s)	EMG Time shift (s)
1	0.1121	747	766	787	787	0.1900	0.4000	0.4000
2	0.1255	1304	1310	1345	1356	0.0600	0.4100	0.5200
3	0.1434	1708	1732	1771	1784	0.2400	0.6300	0.7600
4	0.1614	2287	2291	2324	2338	0.0400	0.3700	0.5100
5	0.1793	2876	2877	2915	2922	0.0100	0.3900	0.4600
6	0.1972	3322	3338	3376	3405	0.1600	0.5400	0.8300
7	0.2286	4100	4118	4154	4183	0.1800	0.5400	0.8300

Table 3.3 shows the frame number of peaks of: each push, and COM, COP displacement in response to each push, in addition to the maximum displacement of COM and COP (m) after each perturbation.

Table 3.3 The Frame Number of Peaks of Each Perturbation, COM, and COP. In Addition to the Maximum Displacement of COM and COP (m) after Each Perturbation

Perturbation number	Peak force frame number	Peak force (N)	Peak COM frame number	Peak COM displacement (m)	Peak COP frame number	Peak COP Displacement (m)
1	771	240.8000	834	0.0206	834	0.0287
2	1321	277.8000	1410	0.0465	1416	0.0576
3	1737	384.3000	1847	0.0822	1848	0.0917
4	2299	472.3000	2396	0.0756	2393	0.0920
5	2889	560.3000	2985	0.0808	2986	0.0975
6	3342	643.6000	3433	0.1139	3440	0.1775
7	4121	801.1000	4217	0.1303	4224	0.1851

Table 3.4 shows data related to stepping strategy (last two pushes). Fz onset, right toe off onset, error convex shape peak, error convex shape beginning, no load speed, COM onset frame, COP onset frame, EMG onset of ch1 (Tibialis anterior) and ch2 (Lateral Gastrocnemius), in addition to COM and COP time to start moving forward after each perturbation.

Table 3.4: Stepping Strategy Data

Fz onset frame number	rtoe off frame number	error convex peak	Error convex beginning	No load Speed (m/s)	COM onset frame number	COP onset frame number
3322	3379	3389	3381	0.1972	3338	3376
4100	4176	4165	4154	0.2286	4118	4154
EMG onset frame number ch1	EMG onset frame number ch2	COM Time Shift (s)	COP Time Shift (s)	EMG Time shift (s)		
3391	3357	0.1600	0.5400	0.8300		
4183	4131	0.1800	0.5400	0.8300		

The convex shape of the error signal (COM-COP) in A/P plane was found that starts almost at the same time when right toe off starts and it is a good indicator of the occurrence of stepping strategy (Figures 3.12 and 3.13). The possible explanation is that it occurs when COP cannot follow COM anymore (when COP is located at the maximum possible range of COP). The same pattern is observed in the inclination angles signal.

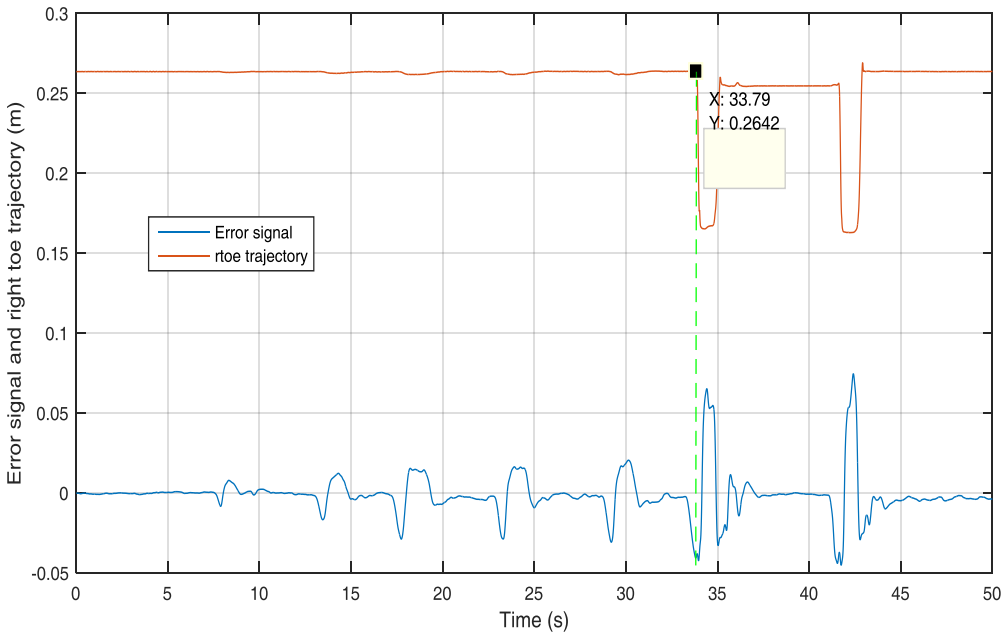

Figure 3.12 Error signal (COM-COP) and right toe trajectory in meters. The green dashed line shows rtoe onset regarding the convex of the error signal.

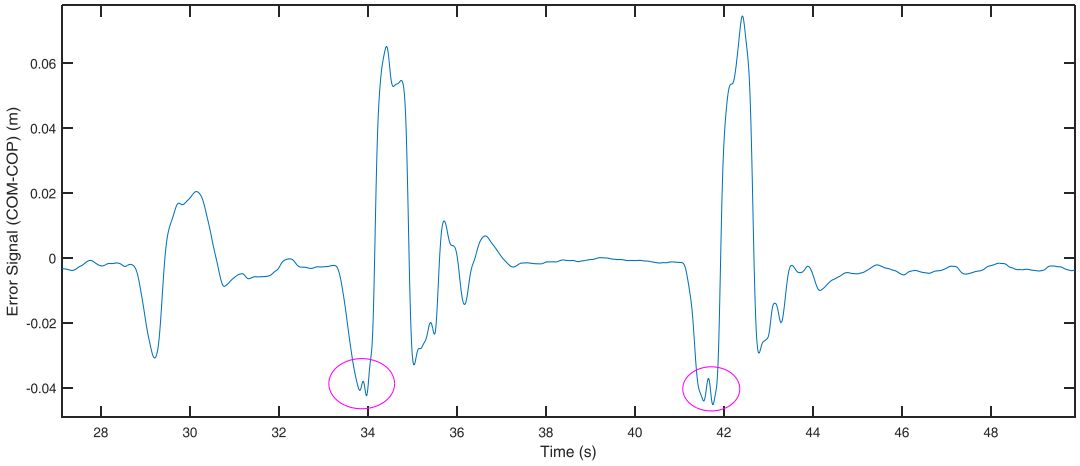

Figure 3.13 Error signal (COM-COP) of the last two pushes (stepping strategy) and the two convex shapes corresponding each step are clear.

3.4 Shoes With Blocks (Aim 3)

In Aim 3 we are studying the effect of minimizing COP range on the human body response to perturbation. One subject was asked to wear the shoes with Blocks, perturbed using the perturbation system seven times, and his data was recorded. Figure 3.14 shows the COP with respect to the trajectory of the rear edge of the left block (left foot) which is the stable foot, and the trajectory of the front edge of the right block (right foot) which is the foot that the subject steps with, and it is clear that COP is located within this range.

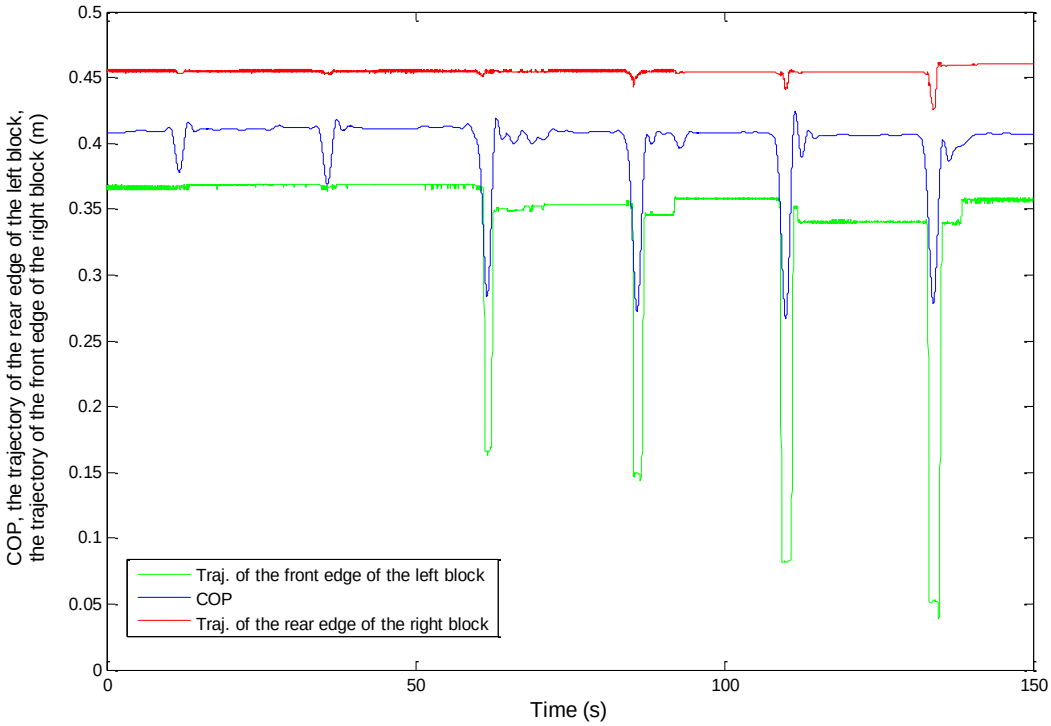

Figure 3.14 COP of a subject wearing these shoes with respect to the trajectory of the rear edge of the right block (right foot), and the trajectory of the front edge of the right block (right foot).

3.5 Platform With Pivot (Aim 4)

In this pilot study the subject was asked to stand on the platforms over Pivots, and use the poles to balance as shown in Figure 3.15. Total experimental duration was 90 seconds. In the first 30 seconds, subject was asked to stand still, and for the following 30 seconds the subject was asked to tilt forward and hold the position, and for the last 30 second she was asked to tilt further forward and hold the position. Looking at Figure 3.15, we observe that there is a small gap between COM and COP in the first 30 seconds during quiet standing, while this gap gets bigger in the next two intervals which indicates that the subject is leaning on the trekking poles rather than using them to balance. To solve this problem the platform will be modified by moving the trekking poles closer to the subject's feet.

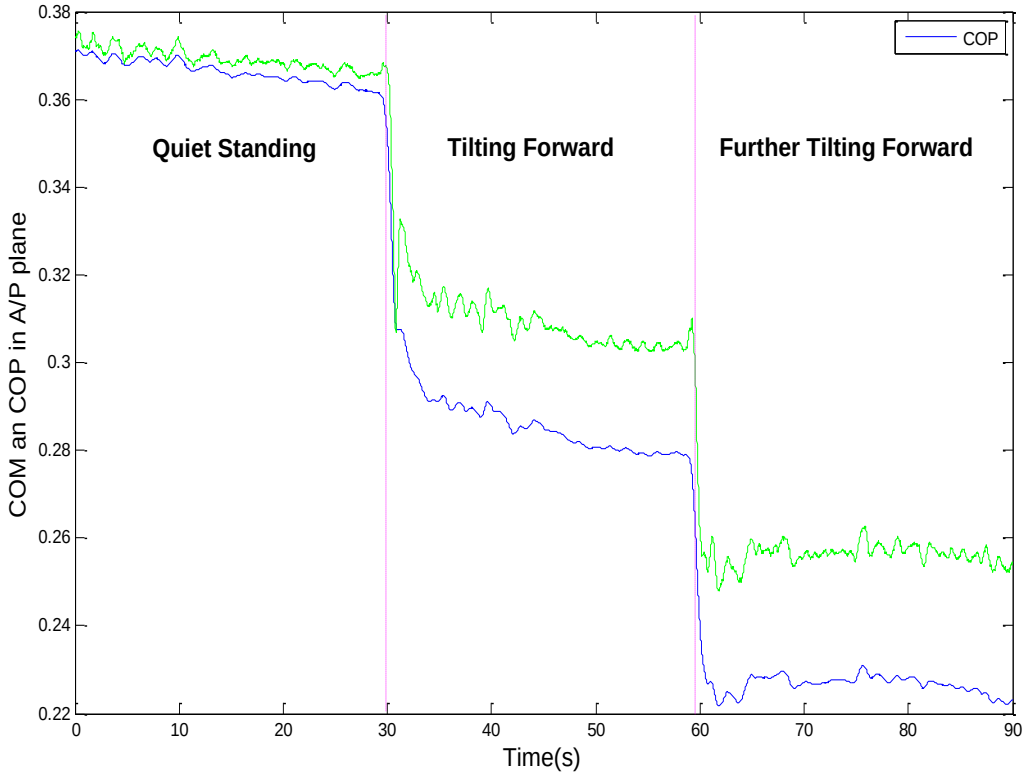

Figure 3.15 COM and COP in A/P plane for a subject standing on the platforms over Pivots, using trekking poles to control balance. The subject was asked to stand there for 90 seconds, during the first 30 seconds stand still, during the second 30 seconds she was asked to tilt forward, during the third 30 seconds she was asked to tilt further forward.

CHAPTER 4

RESULTS AND DISCUSSION

4.1 Overview of Results

This chapter begins with a visual comparison between the behaviors of the COM, COP, error signal, stability index, EMG and correlation between COP and COM of one representative subject when perturbed and during the period of quiet standing between perturbations. It should be noted that all other subjects had similar signals patterns. Figures 4.1 through 4.21 relate to this individual.

The statistical results presented later in this chapter are based on data from all 12 subjects and allow comparison of three major sets of conditions: shoe (Regular, block and pivot), visual feedback (eyes opened and eyes closed) and order of perturbations (Randomized and Non-Randomized). The Non-Randomized perturbations were presented in ascending order of force.

As was discussed at the end of Chapter 3, the pilot study determined that while subjects used the stepping strategy when their COM exceeded the anterior boundary of the BOS with the Regular and block shoes. The Pivots experiments was performed on a couple of subjects who experienced difficulty taking a step, thus, for safety, the experiments with the pivot shoes were limited to only six perturbations so as not to require a step. In this chapter, statistical results are presented only for six perturbations when all three shoes were compared. Furthermore, in the results pertaining to number of steps only Regular and block shoes were considered.

4.2 Comparison of COM and COP Behavior by Experimental Conditions

Figures 4.1-4.3 provide a very interesting overview of one subject's ability to maintain balance by controlling the COP to match the COM. They contain periods of quiet standing between the perturbations during the use of the three types of shoes (ascending order of perturbing forces. The original position of the COM and COP were set to zero for easier comparison of their movements. The dashed line in each Figure displays the forward (anterior) boundary of the BOS in quiet standing (Regular 15 cm, Block 6.7 cm, and Pivots 17 cm with respect to the original COM/COP).

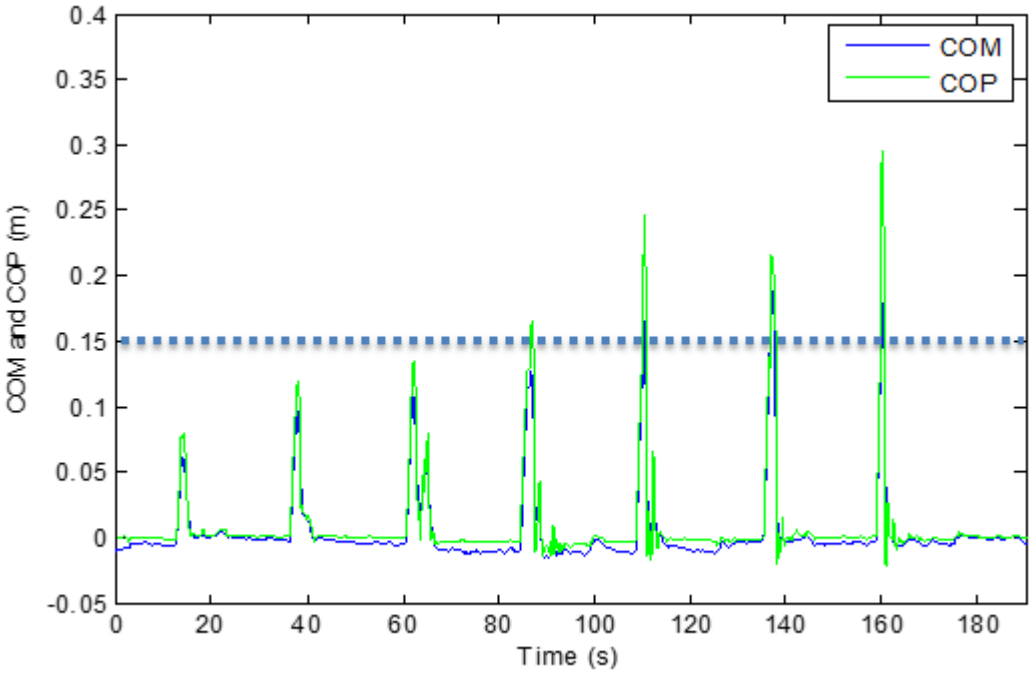

Figure 4.1 COM and COP in A/P plane for a subject while wearing Regular shoes when perturbed with seven different forces presented in ascending order. The dashed line displays the forward (anterior) boundary of the BOS.

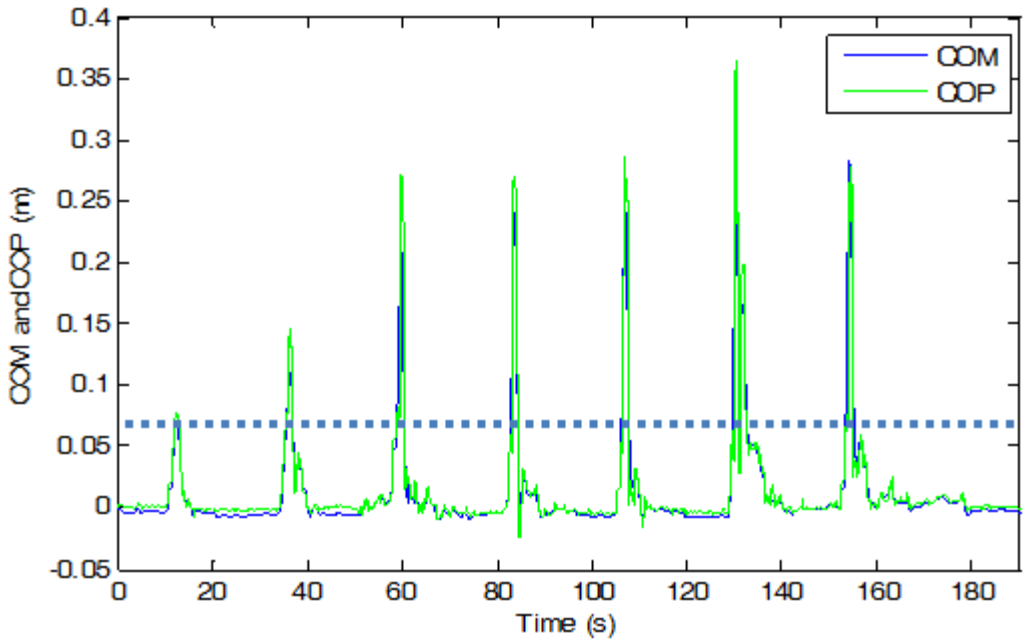

Figure 4.2 shows COM and COP in A/P plane for the same subject perturbed with seven perturbations of ascending force while wearing the shoes with Blocks. The dashed line displays the forward (anterior) boundary of the BOS.

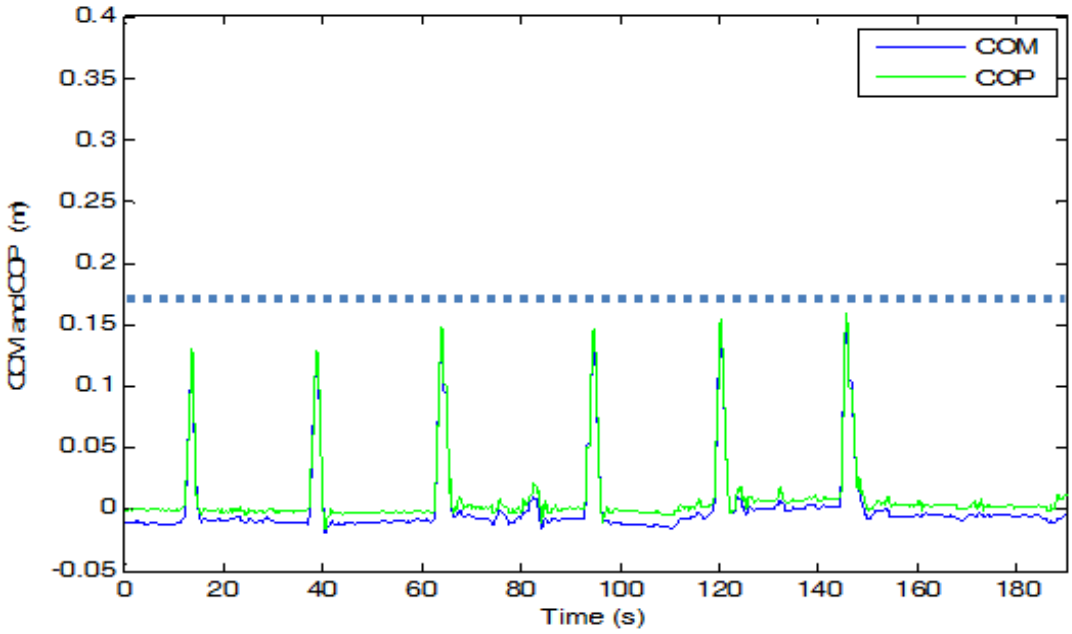

Figure 4.3 COM and COP in A/P plane for the same subject perturbed with six forces perturbations of ascending force while wearing the shoes with Pivots.

Figures 4.1 and 4.2 clearly show that the subject uses the stepping strategy when the COM exceeds the anterior boundary of the BOS. The step allows the COP to closely follow the COM and retain the individual's balance. In Figure 4.3, it is clear that the total forward displacement for both COM and COP when using the pivot shoes is much less than the Regular and block shoes since COM and COP do not exceed the front perimeter of the BOS, hence stepping strategy was not used. It is important to recall that the pivot shoes completely eliminate the subject's ability to use the ankle and hip strategies, since no torque can be applied about the pivot. The very smooth adjustment of the COP is provided exclusively by providing torque via the hands and trekking poles.

The inclusion of the block shoes show that subjects who have had a learning time to accommodate to their normal range of their COP can readily adjust to the greatly shortened COP range of the Blocks and employ the stepping strategy earlier than when normal COP range exists. The observation that subjects could not as easily initiate a step in the pivot shoes even with their normal length range of COP is interesting and requires further study.

Figures 4.4 and 4.5 present the COM and COP for the three shoe conditions: Regular, Blocks and Pivots respectively. It is observed that COM and COP forward displacement is almost the same for all perturbations regardless of their forces. Subjects lean forward (shifting their COM forward) to generate torque with the trekking poles and appear to make the same compensatory response to all perturbations. The largest movement of the COM and COP occur with the Blocks as subjects on Blocks are less stable (BOS is smaller) so they use stepping strategy in response to smaller forces compared to using Regular shoes, they take more steps with larger total forward displacement.

COM and COP values for the different perturbations in the third experiment (Pivots) are close to each other, unlike Regular experiment, this could be a result of the fact that the spring and damping factors of Pivots and trekking poles are different than ankle joints, in addition to that the use of the trekking poles that are attached to the outer frontal edge of each force plate make it a quadriceps process, it is easier to tilt further forward around Pivots, and the interaction of the trekking poles to maintain balance is different too. It is observed that despite that the fact that trekking poles were able to control COP, the reaction to perturbations is different than normal biological reaction (Regular experiment).

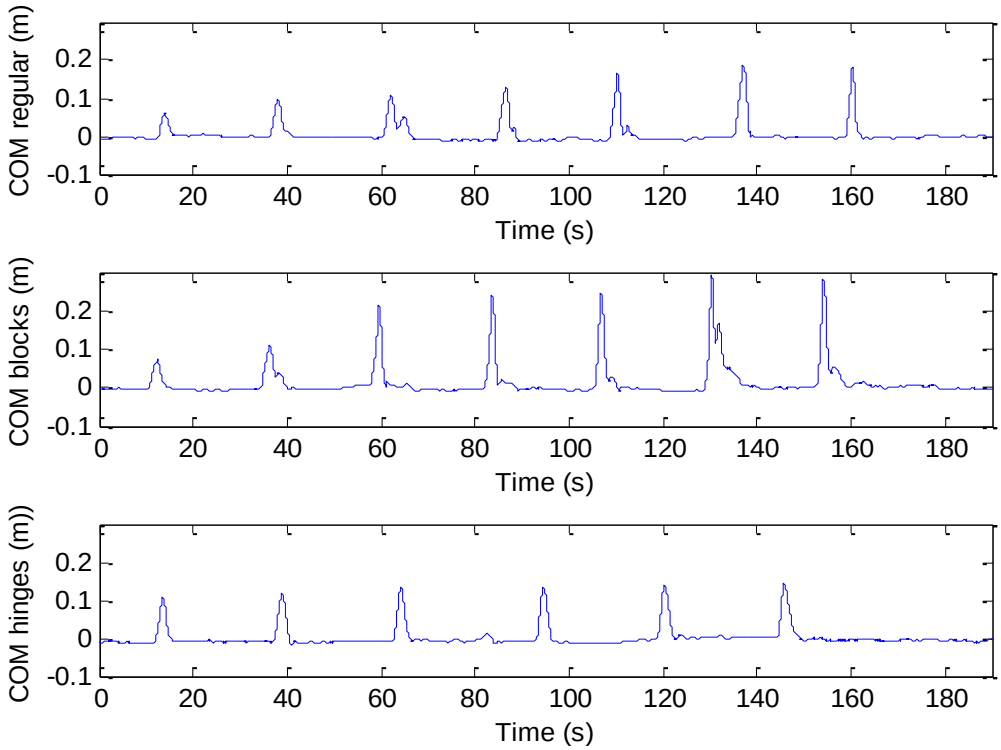

Figure 4.4 COM for the three experiments: Regular, Blocks, and Pivots.

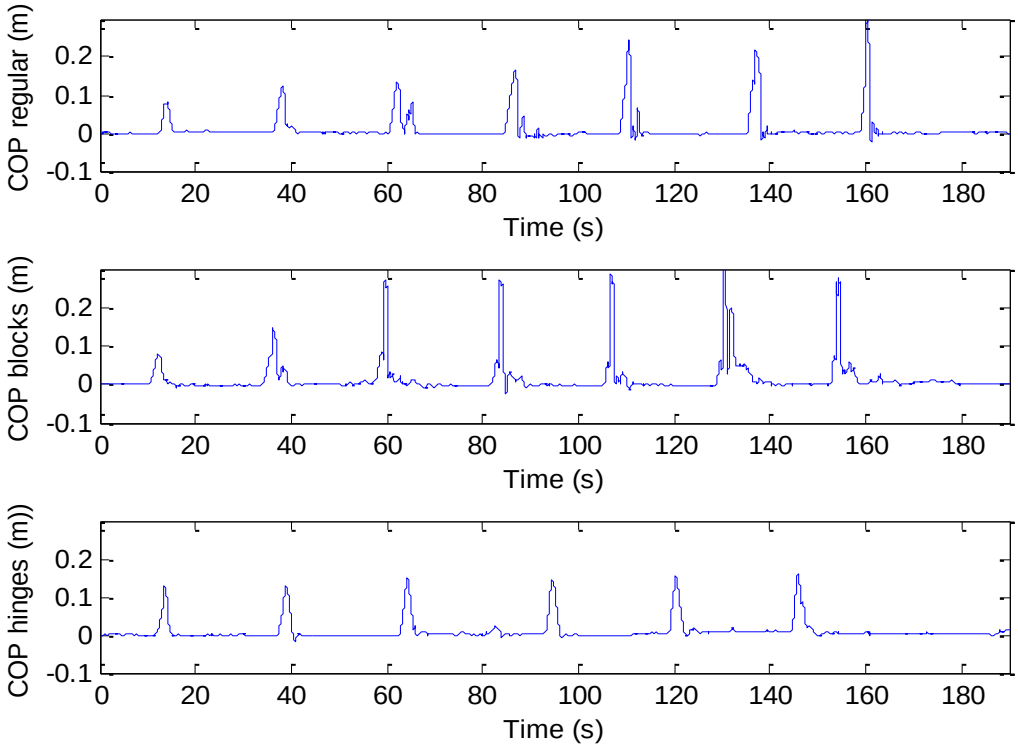

Figure 4.5 COP for the three experiments: Regular, Blocks, and Pivots.

4.3 Comparison of Error Signal Behavior by Experimental Conditions

Figure 4.6 shows the error signal that identifies the difference between COM and COP in the A/P plane. There is a clear difference between error signal peaks among the three experiments. The error signal peaks are the largest for Blocks experiments where the subjects are less stable and use the stepping strategy, and the smallest for Pivots experiment because stepping strategy was not used at all in Pivots experiment. The error between the COM and COP is momentarily exaggerated during the step (forward and backward) with Regular and block shoes. As the COM passes the anterior BOS boundary and continues to move forward, resulting in a step where the swing foot lifts off the ground leaving the stance foot to provide a fixed COP. When the swing foot contacts the ground in its new position, the COP jumps to

a location closer to the COM though mostly preceding it. During the swing of the step, the COM continues to move while the COP remains constant, creating a larger error.

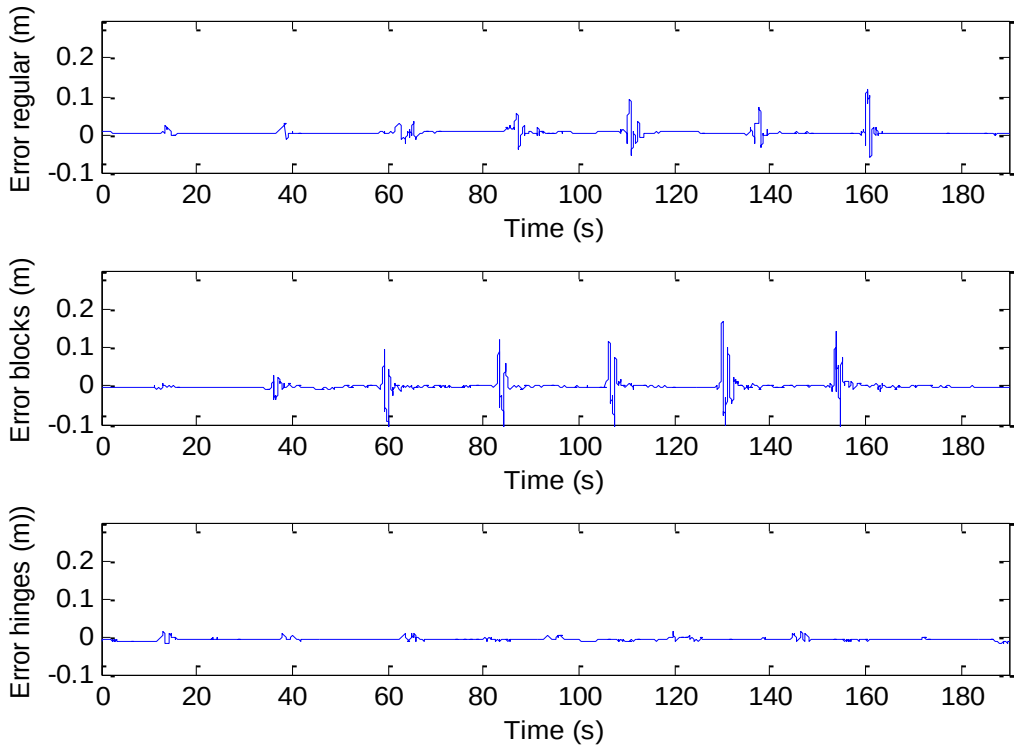

Figure 4.6 Error signals for the three experiments: Regular, Blocks, and Pivots.

4.4 Comparison of Joint Angle by Experimental Conditions

Figures 4.7-4.9 show ankle, knee, and hip joint angles for the right and left legs in **A/P plane with the horizontal plane** for the first experiment (Regular), the second experiment (Blocks), and the third experiment (Pivots).

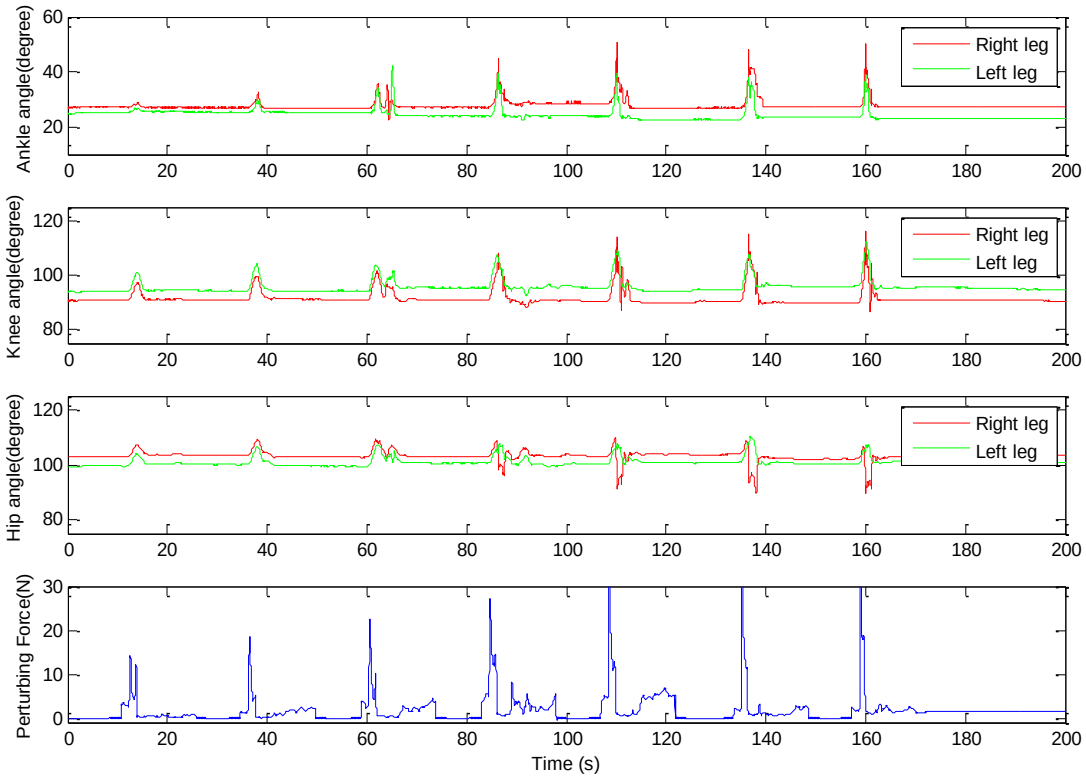

Figure 4.7 Ankle, knee, and hip joint angles with horizontal plane for Regular.

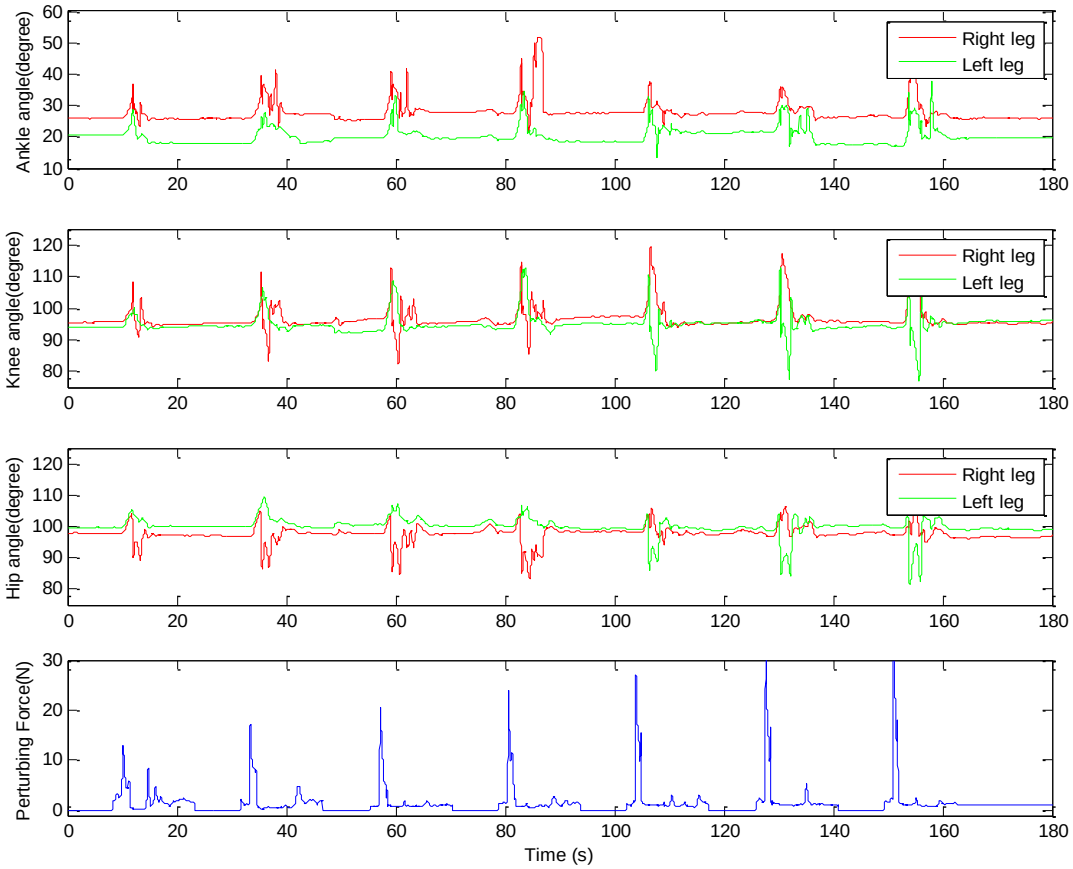

Figure 4.8 Ankle, knee, and hip joint angles with horizontal plane for Blocks.

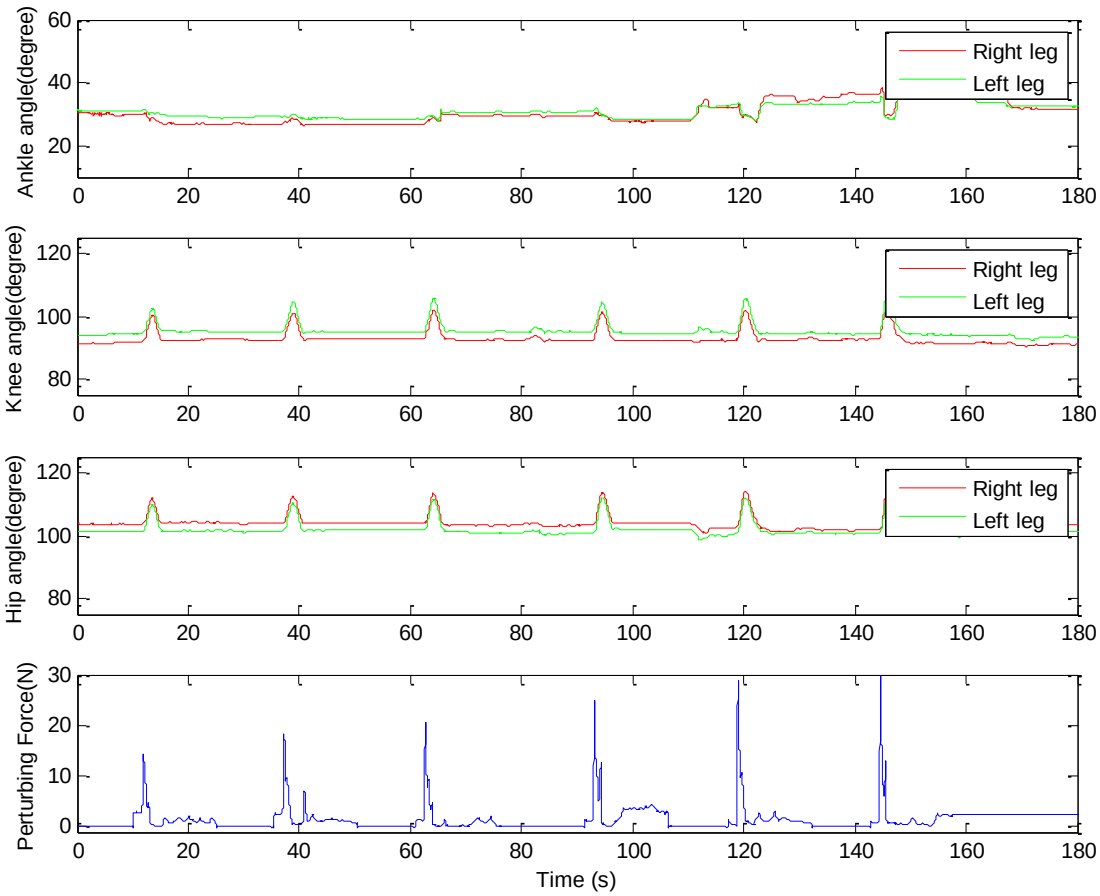

Figure 4.9 Ankle, knee, and hip angle joints with horizontal plane for Pivots.

The change in ankle, knee and hip angles in the second experiment (Blocks) is larger than the change in the first experiment (Regular). The joint angle differences observed could be attributed to the fact that the number of steps and the total forward displacement taken in perturbations with smaller forces in the second experiment compared to first experiment is larger.

There is a very small change in ankle angle resulting in no distinct observable pattern in the third experiment. The ankle brace used in the third experiment restricted the movement of ankle resulting in the smaller joint angles at the ankle as is observed. The change in the knee angle in the third experiment is less than the first and second experiments

especially for big forces perturbations as steps were taken by the subjects in first two experiments.

Hip and knee joint angle change directions when a step is taken as can be observed in the Figure 4.7 (perturbations four-seven) and Figure 4.8 (all perturbations) while no change in the joint angle direction is observed when no step is taken in all experiments.

4.5 Comparison of Stability Index (SI) by Experimental Conditions

Figure 4.10 shows stability index for the first (Regular), the second (Blocks), and the third (Pivots) experiments. It shows that stability index is slightly larger for the second experiment compared to the first one. This is predictable since the subjects in the second experiment were wearing the shoes with Blocks to shorten the BOS which results in subjects being less stable. It also shows that SI fluctuates much more in the case of Blocks even during quiet standing between perturbations. This reflects the instability of subjects on Blocks even in quiet standing. For the third experiment (Pivots) the SI looks close to SI of Regular, this is a good indication that subjects are stable enough when on Pivots and that trekking poles are controlling stability in a good way.

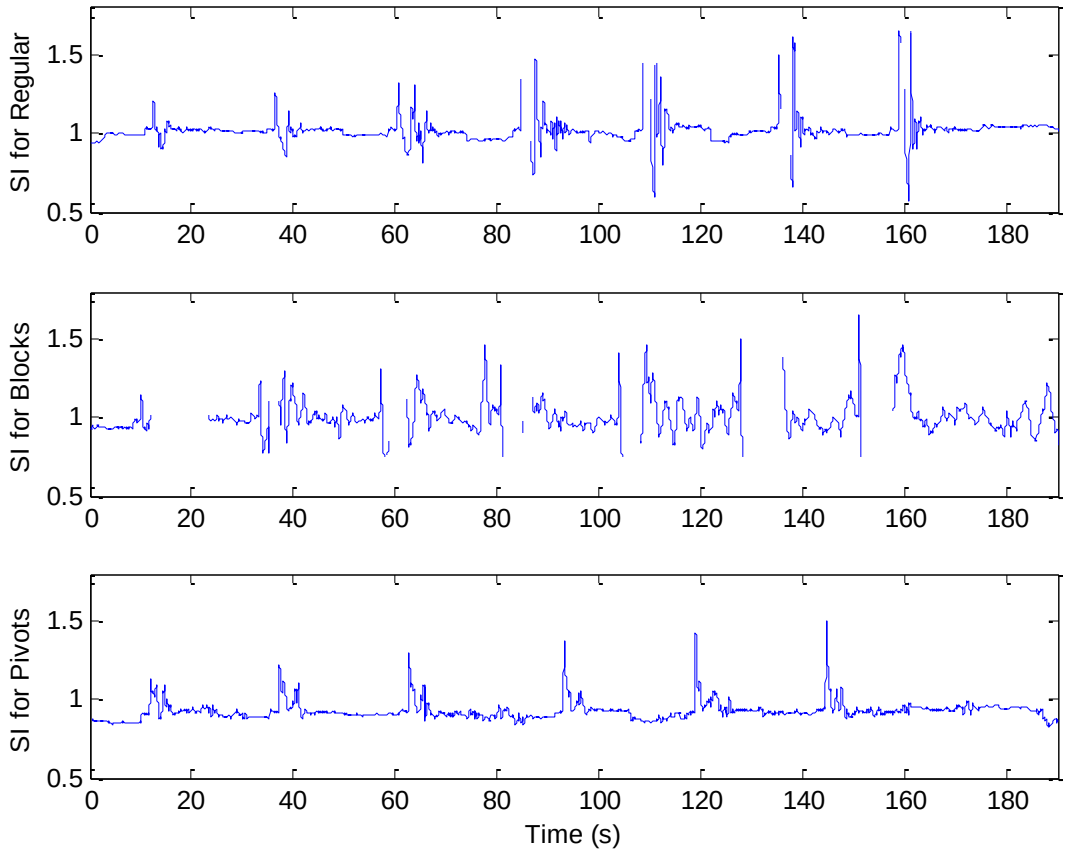

Figure 4.10 Stability index for the three experiments: Regular, Blocks, and Pivots.

4.6 Comparison of EMG by Experimental Conditions

Figures 4.11-4.13 show EMG of Lateral Gastrocnemius and Tibialis Anterior muscles of the right leg for the first experiment (Regular), the second experiment (Blocks), and the third experiment (Pivots).

For the first experiment (Regular) Lateral Gastrocnemius is active during quiet standing, and that there is no recognizable activity for Tibialis Anterior when there is no stepping. In addition, both muscles get activated when a step is taken.

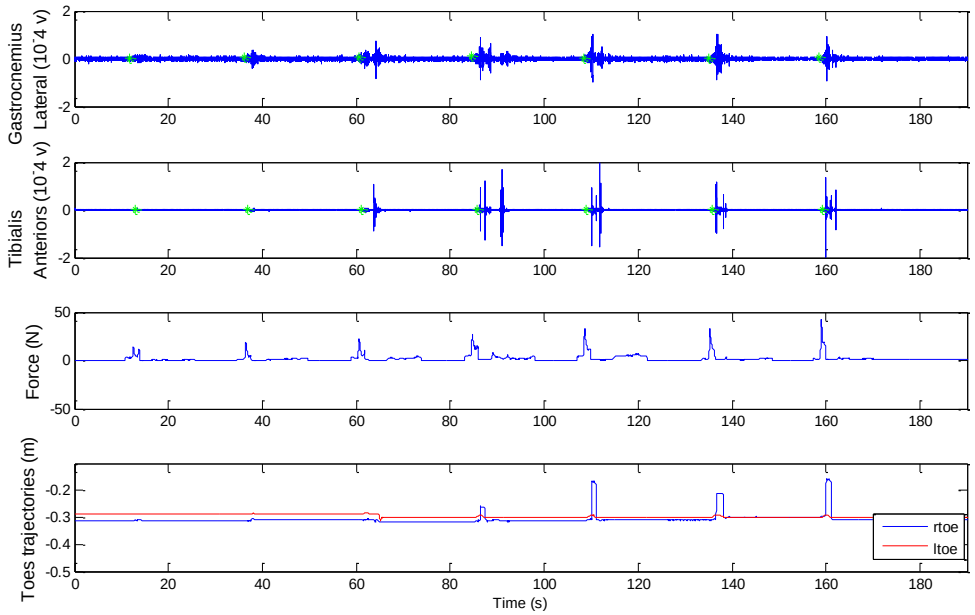

Figure 4.11 EMG of Lateral Gastrocnemius and Tibialis Anterior muscles of the right leg for the first experiment (Regular), corresponding to the perturbing forces and the trajectories of the right and left toes. Where the onset of each activation is shown as a green star.

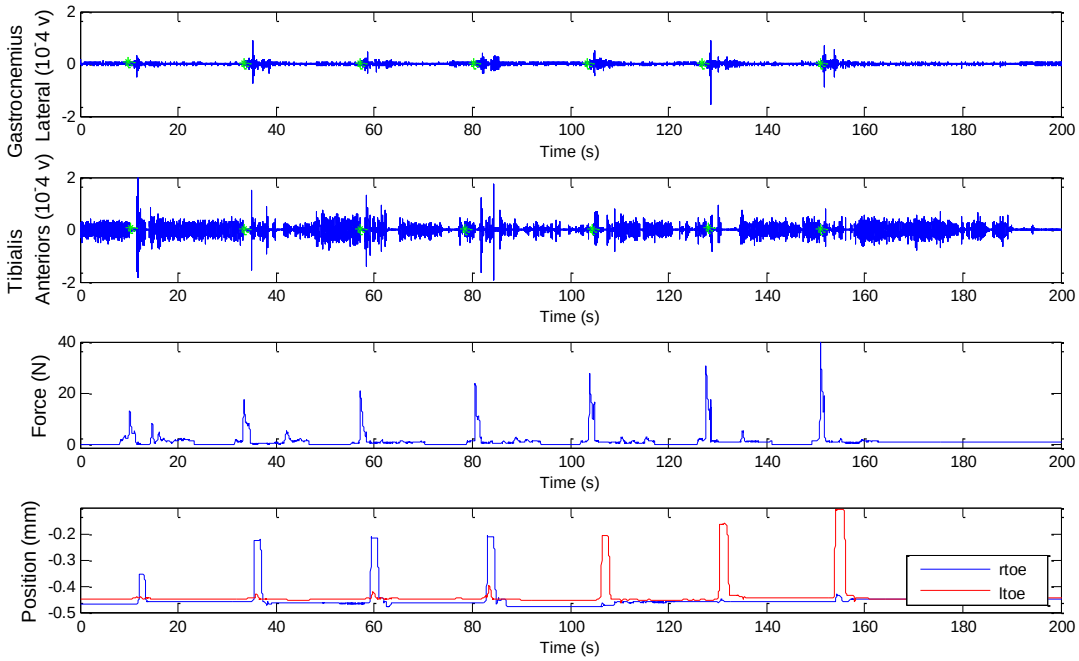

Figure 4.12 EMG of Lateral Gastrocnemius and Tibialis Anterior muscles of the right leg for the second experiment (Blocks), corresponding to the perturbing forces and the trajectories of the right and left toes. Where the onset of each activation is shown as a green star.

In the second experiment, we can observe that Tibialis Anterior shows more activity during quiet standing compared to the first experiment. Also, when a step is taken, Lateral Gastrocnemius gets activated while Tibialis Anterior gets deactivated, followed by the deactivation of Lateral Gastrocnemius and activation of Tibialis Anterior.

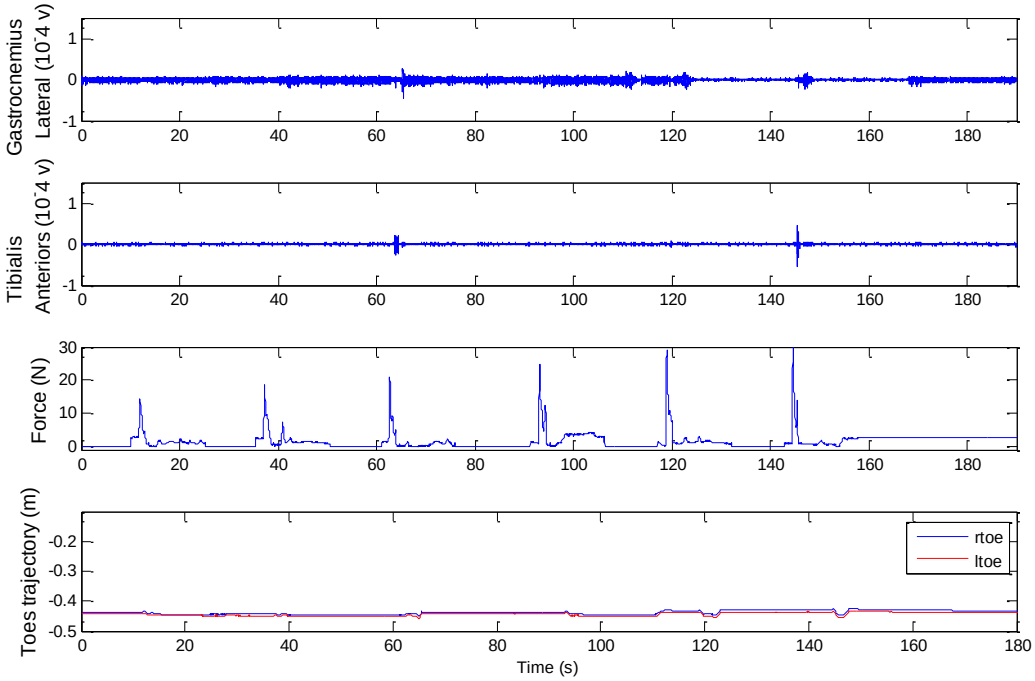

Figure 4.13 EMG of Lateral Gastrocnemius and Tibialis Anterior muscles of the right leg for the third experiment (Pivots), corresponding to the perturbing forces and the trajectories of the right and left toes.

Figure 4.13 shows EMG of Lateral Gastrocnemius and Tibialis Anterior muscles of the right leg for the third experiment (Pivots). It is observed that the activity of the two muscles is very low, this is expected since the ankle joints were tightly clamped with braces. Although subjects may have tried to use their muscles to balance, they were not able to do so since the Pivots prevent the use of ankle torque to assist in balance. Other than that Lateral Gastrocnemius muscle was active most of the time, there is no specific pattern observed with perturbations.

4.7 Comparison of COM and COP with Joints Trajectories during Quiet Standing (Third Experiment (Pivots))

Figure 4.14 shows COM, COP, right trekking pole, right hip, right knee, and right ankle trajectories for a quiet standing trial on Pivots. Figure 4.15 shows COM, COP, right trekking pole, right hip, right knee, and right ankle trajectories for the third experiment (Pivots). During quiet standing, there are some fluctuations in the trajectories hence the subject is trying to blanace on Pivots, but the fluctuations are pretty small. In Figure 4.15 it is observable that ankles are not moving much because of the braces that are used to make the ankle immobile, even the knee is not moving much while the hip and the trekking pole are moving further. It is noticed that the upper body is moving more and the trkking poles are compensating. It is also shown that when the subject gets perturbed with different forces right pole (rpole) trajectory has the same pattern as COP, this demonstrate the role of the trekking poles in controlling COP. The right hip (rhip) moves as the whole body tilts forward, the change in the trajectory of the right knee (rknee) is less even though there is a pattern. The ankle trajectory is almost not changing and there is no pattern at all, this can be explained by the fact that the ankle brace prevents ankle joint's movement.

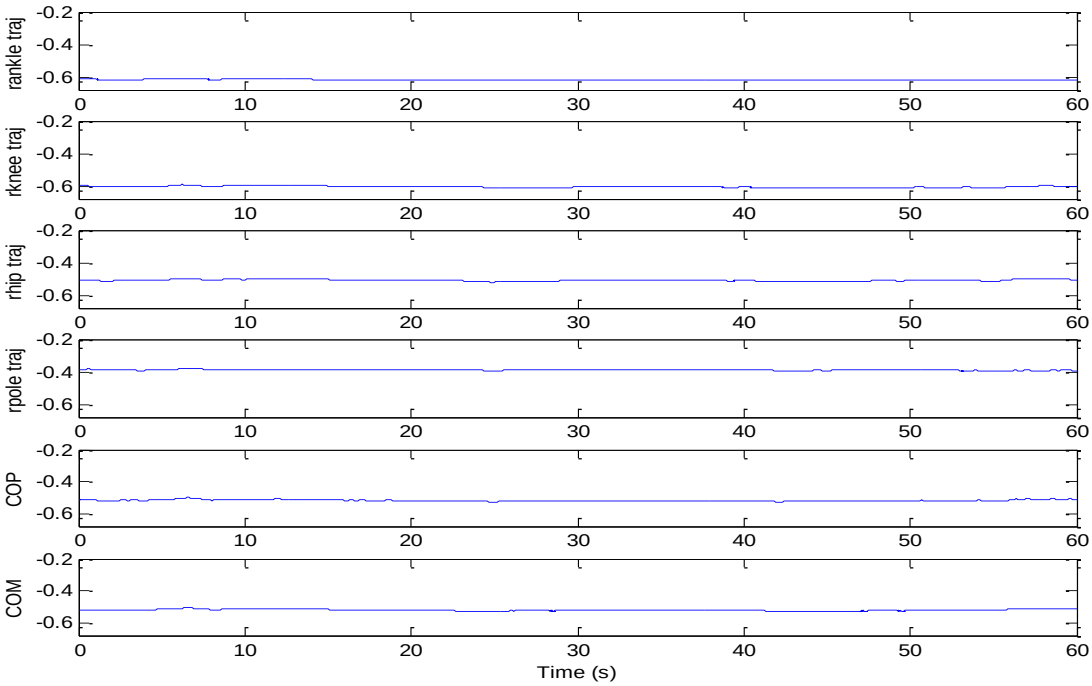

Figure 4.14 COM, COP, right trekking pole, right hip, right knee, and right ankle trajectories for quiet standing trial on Pivots.

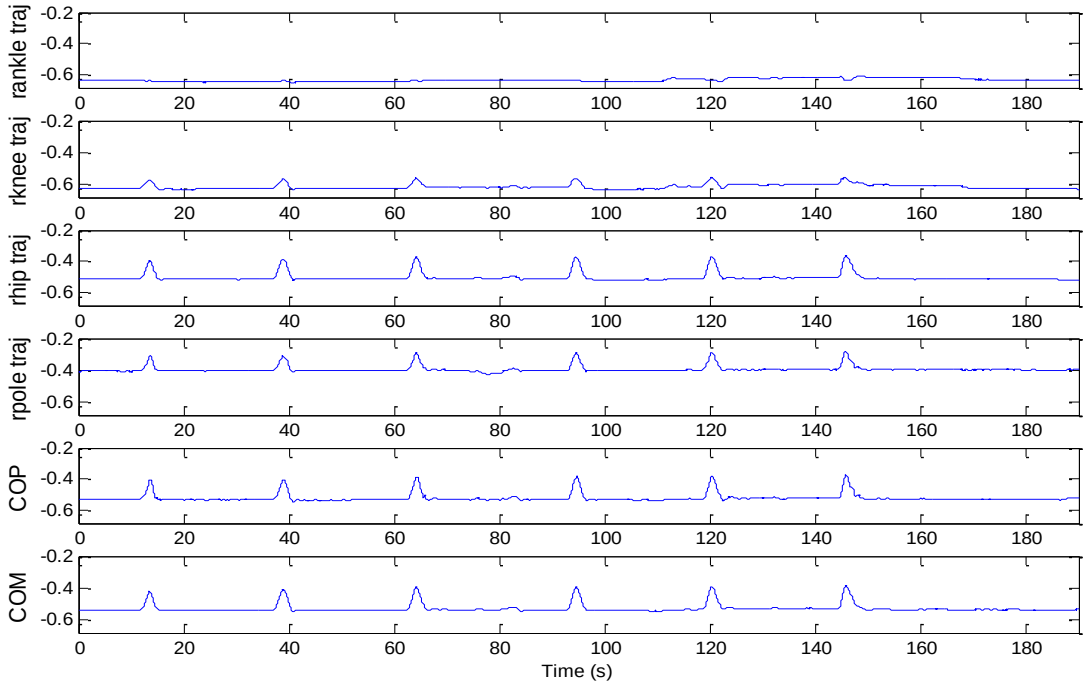

Figure 4.15 COM, COP, right trekking pole, right hip, right knee, and right ankle trajectories for the third experiment (Pivots).

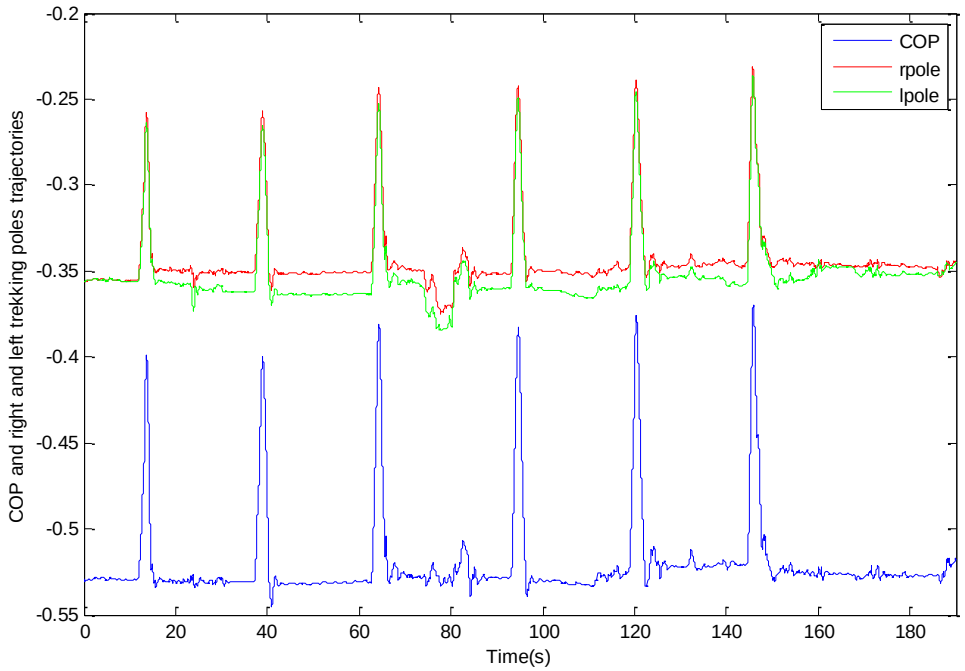

Figure 4.16 COP in addition to right and left trekking poles trajectories for a subject performing Pivots experiment (third experiment).

Figure 4.16 shows COP in addition to right and left trekking poles trajectories for a subject performing Pivots experiment. The COP trajectory looks the same as the trekking pole trajectories. To show the strong relation between them the correlation between COP and the average trajectory of the right and left trekking poles was computed. Figure 4.17 shows the correlation between them. They are highly correlated with a pearson's Correlation of (r = 0.978).

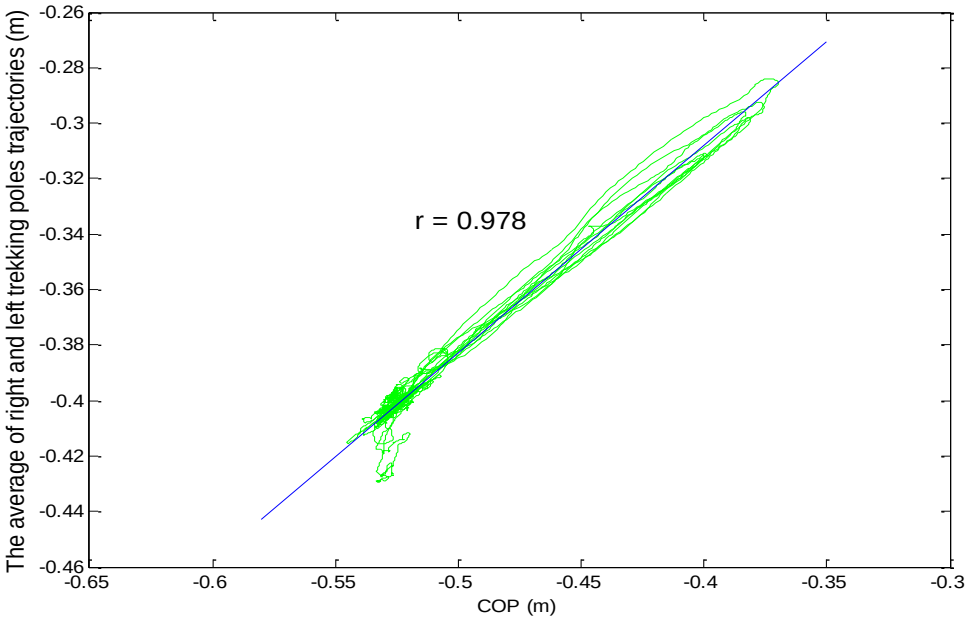

Figure 4.17 The correlation between COP and the average of right and left trekking poles trajectories for the third experiment (Pivots).

To investigate if the trekking poles play the same role in quiet standing on Pivots the same analysis was performed on its data. Figure 4.18 shows COP in addition to right and left trekking poles trajectories for a subject performing Pivots experiment. The COP trajectory looks the same as the trekking pole trajectories. To show the strong relation between them the correlation between COP and the average trajectory of the right and left trekking poles was computed using SPSS. Figure 4.19 shows the correlation between them. They are highly correlated with a pearson's Correlation of (r = 0.932). The high correlation between COP and the average trajectory of the right and left trekking poles in the case of perturbations and for quiet standing on Pivots shows that the trekking poles are a good approach to control COP and maintain balance.

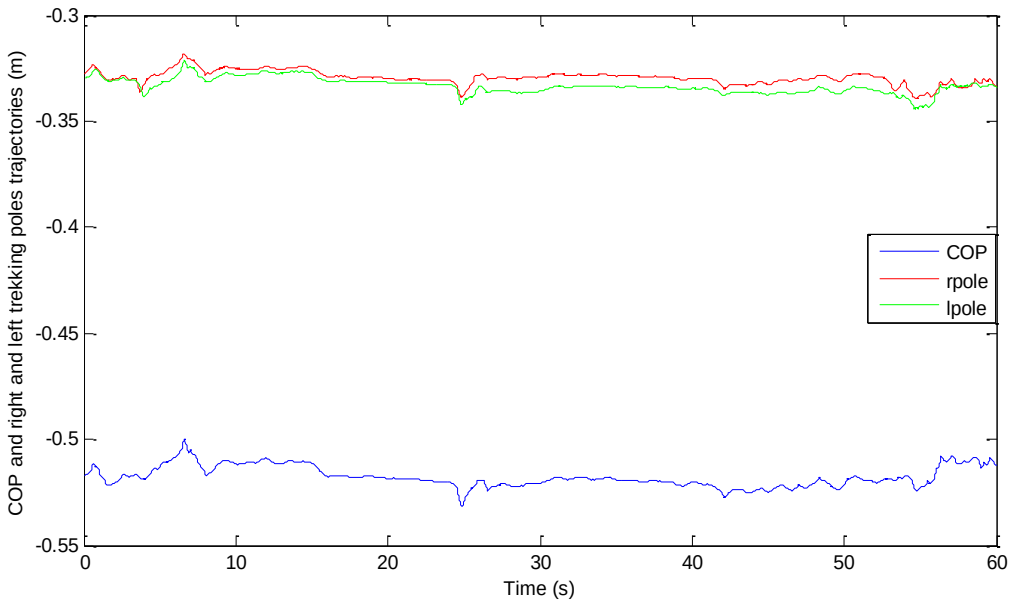

Figure 4.18 COP in addition to right and left trekking poles trajectories for a subject performing quiet standing on Pivots.

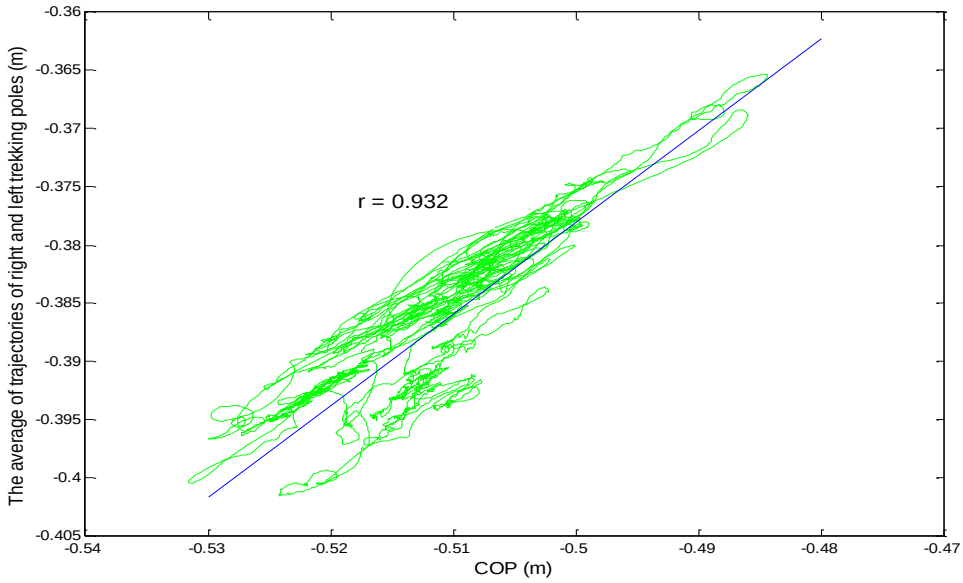

Figure 4.19 The correlation between COP and the average of right and left trekking poles trajectories for quiet standing on Pivots with the y-axis zoomed since the trekking pole movements are much smaller in quiet standing.

4.8 COM Prediction

COM is Predicted using COP data. Prediction equation is the result of regression between COM and COP for quiet standing data, SPSS is used to find all correlations and regressions. Then predicted COM is filtered using a low pass filter with a low cut off frequency to get rid of the high frequency content of COP (the independent variable of the regression equation).

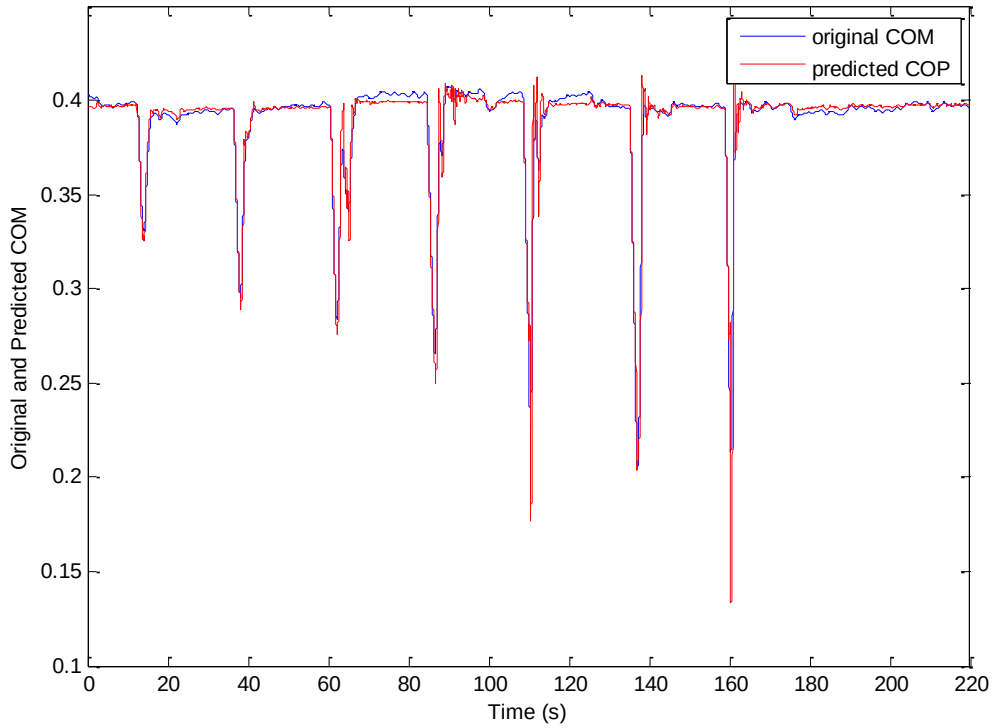

Figure 4.20 Original and predicted COM. Predicted COM is filtered using a low pass filter with a cut off frequency of 6 Hz.

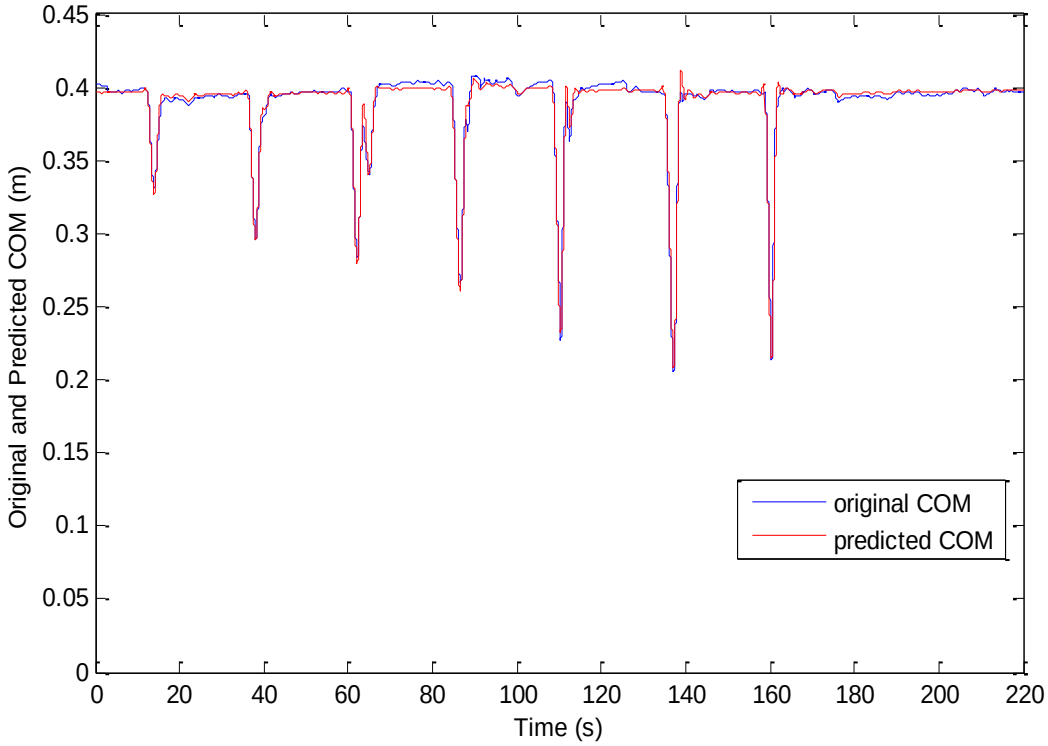

Figure 4.21 Original and predicted COM. Predicted COM is filtered using a low pass filter with a cut off frequency of 0.5 Hz.

Figure 4.20 shows the original and the predicted COM where the predicted COM is filtered using a low pass filter with a cut off frequency of 6 Hz. Figure 4.21 shows the original and predicted COM while the predicted COM is filtered using a low pass filter with a cut off frequency of 0.5 Hz. It is perceived that the predicted COM that is filtered with a cut off frequency of 0.5 Hz matches the original COM better, since the low cut off frequency filter removed the unwanted overshoots of predicted COM.

The **root-mean-square error (RMSE)** is a frequently used measure of the differences between values predicted by a model or an estimator and the values actually observed [57]. The equation below shows how to find RMSE:

$$RMSE = \sqrt{\frac{\sum_{i=1}^{n}(\hat{y}_i - y_i)^2}{n}} \quad (4.1)$$

$\hat{y}_i$: predicted value.

y_i: original value.

n=number of samples.

i=1: n

Table 4.1 RMSE Values for Predicted COM for a Cut off Frequency of 6Hz and 0.5Hz Compared to the Original COM and the Normalized RMSE of each of them

Subject	RMSE1 (fc=6 Hz)	Normalized value of RMSE1	RMSE2 (fc=0.5 Hz)	Normalized value of RMSE2
1	0.0083	0.04028021	0.0055	0.0052
2	0.018	0.210157618	0.0176	0.2157
3	0.0292	0.406304729	0.029	0.4139
4	0.0097	0.064798599	0.0073	0.0365
5	0.006	0	0.0056	0.007
6	0.0125	0.113835377	0.0086	0.0591
7	0.01	0.070052539	0.0061	0.0157
8	0.0084	0.042031524	0.0054	0.0035
9	0.0631	1	0.0627	1
10	0.0101	0.071803853	0.0098	0.08
11	0.008	0.03502627	0.0077	0.0435
12	0.0081	0.036777583	0.0052	0

Table 4.1 shows RMSE values for predicted COM for a cut off frequency of 6Hz and 0.5Hz, compared to the original COM and the normalized RMSE of each of them. It is observed that RMSE values are quiet small, especially RMSE2 for the data that are filtered with f_c = 0.5 Hz. To check further, normalized RMSE values were computed and we notice that normalized RMSE2 values are closer to 0; this is a good indication that RMSE is low and this procedure of predicting COM gives close enough COM to the original one.

4.9 Statistical Analysis

4.9.1 Statistical Model

Table 4.2 The Statistical Model Summary

COPMPARISON		Trial/Experiment	Parameters	Test	Pass/Fail	Test	
Regular vs random		EO	SI, error signal peaks, correlation, number of steps, total forward displacement	Normality	Yes	Paired T-Test	
					No	Wilcoxon	
		EC	SI, error signal peaks, correlation, number of steps, total forward displacement	Normality	Yes	Paired T-Test	
					No	Wilcoxon	
Between Experiments	Regular vs Blocks vs Pivots	EO	SI, Error signal peaks, correlation	Sphericity	Yes	rmANOVA	Paired T-Test
					No	Green house	Paired T-Test
		EC	SI, Error signal peaks, correlation	Sphericity	Yes	rmANOVA	Paired T-Test
					No	Green house	Paired T-Test
	Regular vs Blocks	EO	number of steps, total forward displacement	Normality	Yes	Paired T-Test	
					No	Wilcoxon	
		EC	number of steps, total forward displacement	Normality	Yes	Paired T-Test	
					No	Wilcoxon	
EO vs EC		Regular	SI, error signal peaks, correlation, number of steps, total forward displacement	Normality	Yes	Paired T-Test	
					No	Wilcoxon	
		Blocks	SI, error signal peaks, correlation, number of steps, total forward displacement	Normality	Yes	Paired T-Test	
					No	Wilcoxon	
		Pivots	SI, error signal peaks, correlation, number of steps, total forward displacement	Normality	Yes	Paired T-Test	
					No	Wilcoxon	

4.9.2 Non-Randomized vs Randomized (Regular Experiment)

Non-Randomized vs randomized comparisons are done to know if there is a significant effect of perturbing force prediction on the reaction to the perturbation. In addition, since there are two groups (Non-Randomized (where ascending forces arrangement was applied to perturb each subject in this group) and randomized (where randomized forces arrangement was applied to perturb each subject in this group)), it was used to decide to use randomized and Non-Randomized groups data as one or separate data sets between experiments (Regular, Blocks, and Pivots) and EO vs EC comparisons. Using both groups data sets as one data set whenever there is no significant difference between them increases sample size and so increases the opportunity to find Probabilities whenever it is there.

1) **Total forward displacement (Regular, Non-Randomized vs Randomized, EO and EC).**

In the beginning normality test was run to check if total forward displacement of Non-Randomized and randomized experiments for each perturbation is normal or not, then the appropriate statistical test was performed to compare between them. Table 4.3 shows a statistical analysis summary for the comparison of total forward displacement between Non-Randomized and randomized for the first experiment for both EO and EC trials. These results indicate that there is no significant difference between them for any perturbation.

Table 4.3 Statistical Analysis Summary for the Comparison of Total Forward Displacement between Non-Randomized and Randomized for the First Experiment (Regular, EO and EC)

	Eyes opened		Eyes Closed	
Perturbation	Normality	Probability	Normality	Probability
1	Yes (Paired T-Test)	1.000	Yes (Paired T-Test)	1.000
2	Yes (Paired T-Test)	1.000	Yes (Paired T-Test)	1.000
3	Yes (Paired T-Test)	0.850	No (Wilcoxon)	0.317
4	No (Wilcoxon)	1.000	No (Wilcoxon)	0.465
5	No (Wilcoxon)	0.465	No (Wilcoxon)	0.144
6	Yes (Paired T-Test)	0.062	Yes (Paired T-Test)	0.212
7	Yes (Paired T-Test)	0.053	No (Wilcoxon)	0.500

Figures 4.22 and 4.23 show the total forward displacement and standard error bars for all perturbations of Non-Randomized and randomized experiments during eyes opened and eyes closed conditions respectively. It is noticeable that the total forward displacement of Non-Randomized and Randomized are close to each other.

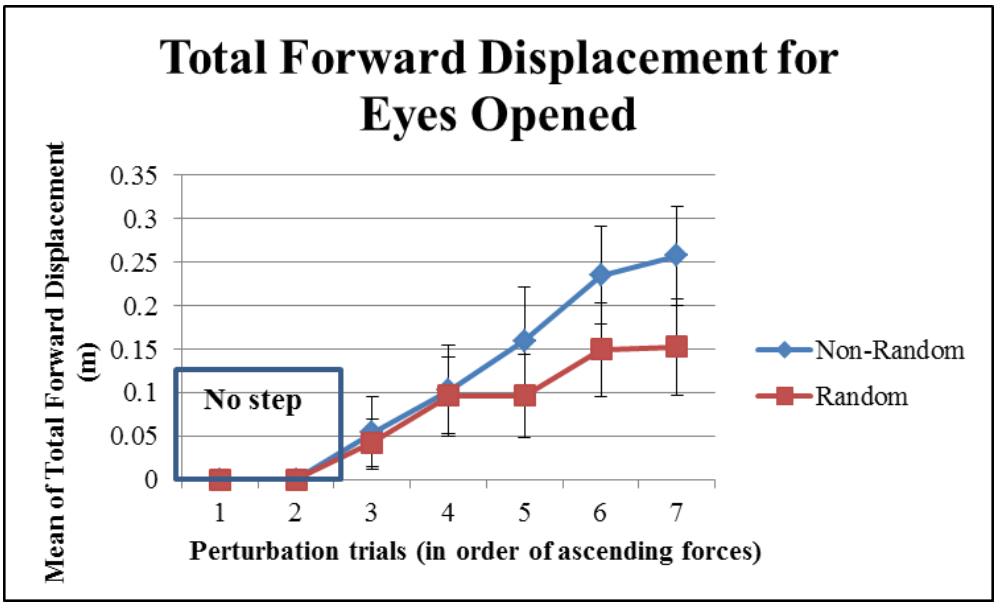

Figure 4.22 Total forward displacement means and standard error bars for each perturbation of Non-Randomized and randomized experiments (Regular, EO).

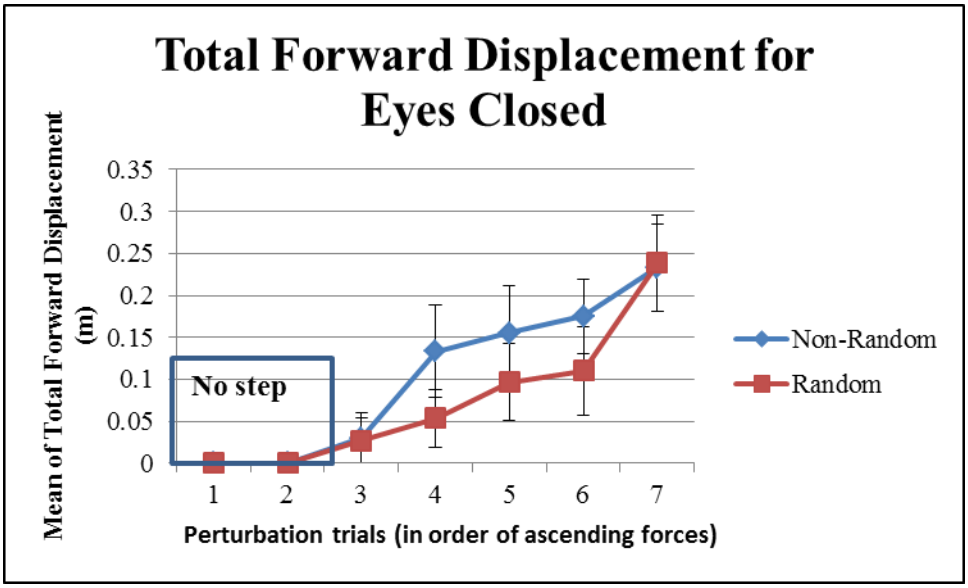

Figure 4.23 Total forward displacement means and standard error bars for each perturbation of Non-Randomized and randomized experiments when there is no visual feedback (Regular, EC).

2) Number of steps (Regular, Non-Randomized vs Randomized, EO and EC).

Table 4.4 shows a statistical analysis summary for the comparison of total forward displacement between Non-Randomized and randomized for the first experiment for both EO and EC trials. These results indicate that there is no significant difference between them for any perturbation. Since no one of the subjects used stepping strategy in response to the first two perturbations. Table 4.4 below shows that there is no significant difference between the total number of steps of Non-Randomized and Randomized experiments.

Table 4.4 Statistical Analysis Summary for the Comparison of Number of Steps between Non-Randomized and Randomized for the First Experiment (Regular, EO and EC)

	Eyes opened		Eyes Closed	
Perturbation	Normality	Probability	Normality	Probability
1	Yes (Paired T-Test)	1	Yes (Paired T-Test)	1
2	Yes (Paired T-Test)	1	Yes (Paired T-Test)	1
3	Yes (Paired T-Test)	1	Yes (Paired T-Test)	1
4	No (Wilcoxon)	0.317	Yes (Paired T-Test)	0.661
5	No (Wilcoxon)	0.157	No (Wilcoxon)	0.317
6	No (Wilcoxon)	0.157	No (Wilcoxon)	0.157
7	No (Wilcoxon)	0.317	No (Wilcoxon)	0.317

Figures 4.24 and 4.25 show the number of steps and standard error bars for all perturbations of Non-Randomized and randomized experiments during eyes opened and eyes closed conditions respectively. It is observed that the number of steps in both experiments is close to each other.

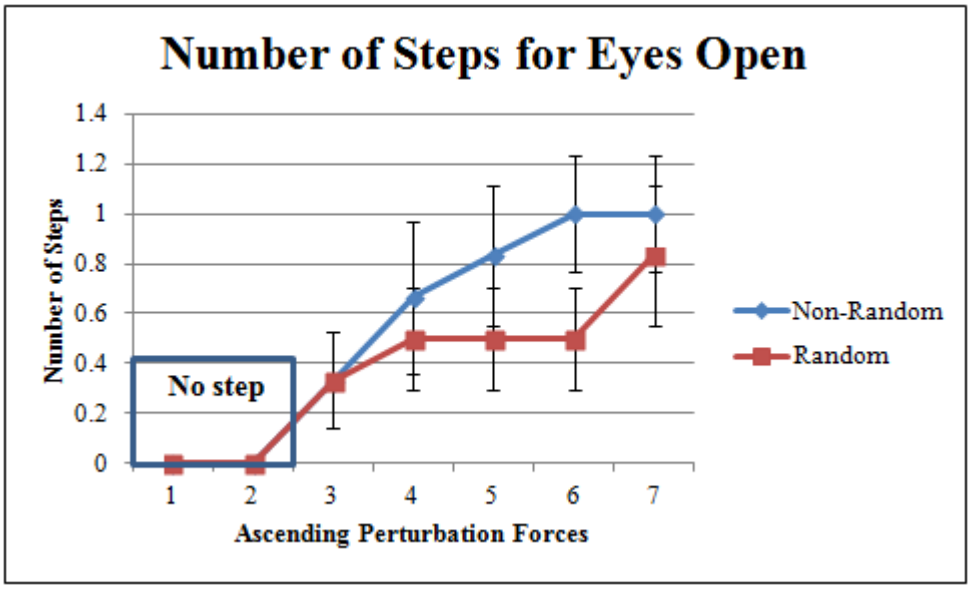

Figure 4.24 Number of steps means and standard error bars for each perturbation of Non-Randomized and randomized experiments for the first experiment (Regular, EO).

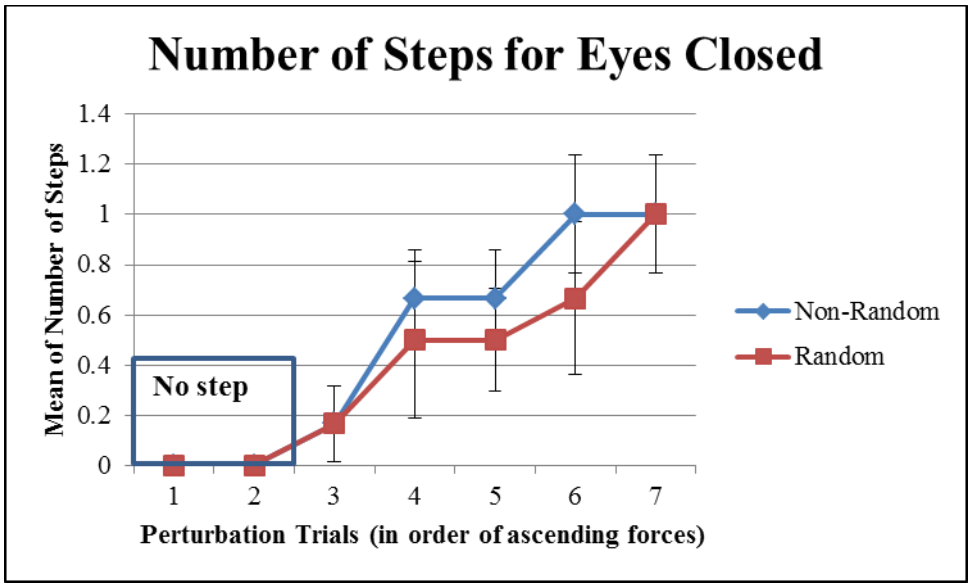

Figure 4.25 Number of steps means and standard error bars for each perturbation of Non-Randomized and randomized experiments for the first experiment (Regular, EC).

3) Stability index (Regular, Non-Randomized vs Randomized, EO and EC).

Table 4.5 below shows a summary of the statistical results of EO and EC trials. There is significant difference between the stability index of Non-Randomized and randomized experiment only for the second perturbation when there is and when there is no visual feedback.

Table 4.5 Statistical Analysis Summary for the Comparison of Stability Index between Non-Randomized and Randomized for the First Experiment (Regular, EO and EC), (* indicates significance, $\alpha = 0.05$)

	Eyes opened		Eyes Closed	
Perturbation	Normality	Probability	Normality	Probability
1	No (Wilcoxon)	0.116	Yes (Paired T-Test)	0.071
2	Yes (Paired T-Test)	0.028 *	Yes (Paired T-Test)	0.016 *
3	Yes (Paired T-Test)	0.888	Yes (Paired T-Test)	0.615
4	Yes (Paired T-Test)	0.064	Yes (Paired T-Test)	0.248
5	Yes (Paired T-Test)	0.689	Yes (Paired T-Test)	0.720
6	Yes (Paired T-Test)	0.808	Yes (Paired T-Test)	0.144
7	Yes (Paired T-Test)	0.192	Yes (Paired T-Test)	0.165

Figure 4.26 shows the stability index with standard error bars for all perturbations of Non-Randomized and randomized experiments when there is visual feedback. While Figure 4.27 shows the stability index with standard error bars for all perturbations of Non-Randomized and Randomized experiments when there is no visual feedback. Though there is no significant difference between them except for the second's perturbation response, It is clear that stability index is larger for Non-Randomized than randomized, and since the stability index is defined as the destabilizing torques over the stabilizing torques, we can say that larger stability index indicates less stability, since randomized group subjects do not have any idea about the coming perturbation force amount, they are less stable.

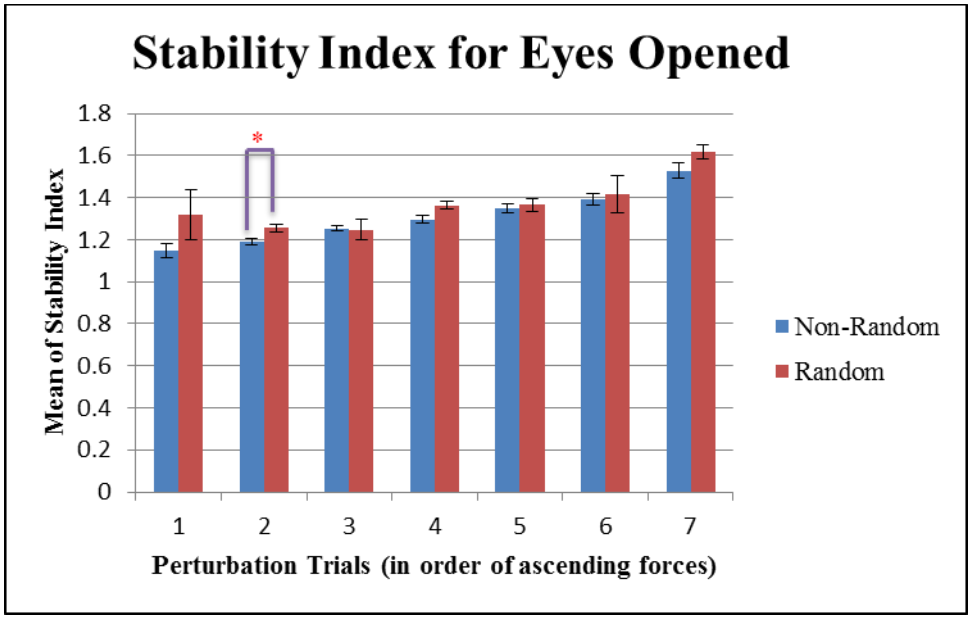

Figure 4.26 SI means and standard error bars for each perturbation of Non-Randomized and randomized experiments for the first experiment (Regular, EO), (* indicates significance, α = 0.05).

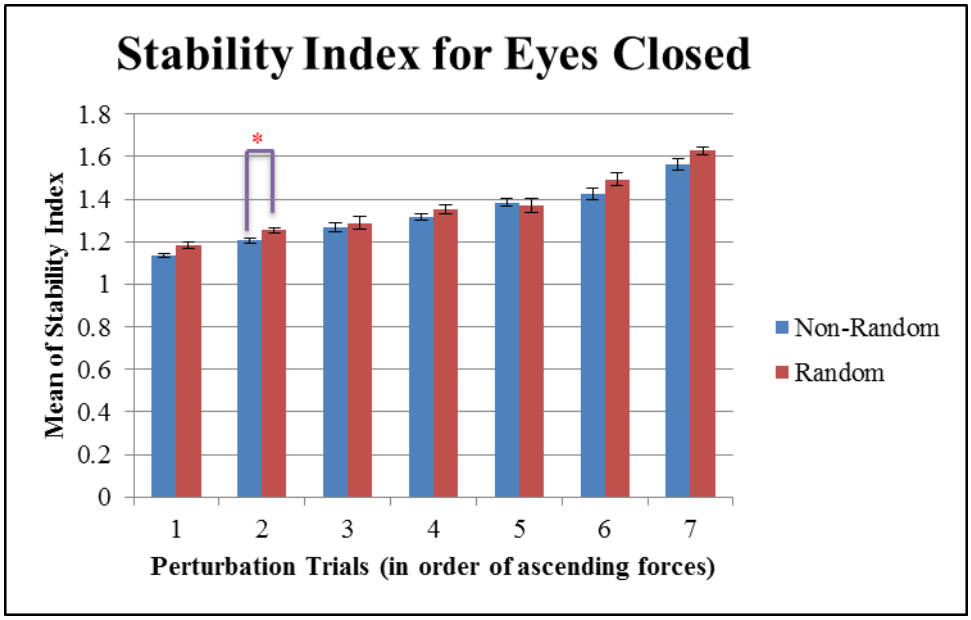

Figure 4.27 SI means and standard error bars for all perturbations of Non-Randomized and Randomized experiments for the first experiment (Regular, EC), (* indicates significance, α = 0.05).

4) Error signal peaks (Regular, Non-Randomized vs Randomized, EO and EC).

Only the fourth perturbation data of the EO trial did not pass the normality test. Table 4.6 below shows a summary of the statistical results. There was significant difference between the error signal peak of Non-Randomized and Randomized experiment for perturbation four.

Table 4.6 Statistical Analysis Summary for the Comparison of Error Signal Peaks between Non-Randomized and Randomized for the First Experiment (Regular, EO and EC), (* indicates significance, α = 0.05)

	Eyes opened		Eyes Closed	
Perturbation	Normality	Probability	Normality	Probability
1	Yes (Paired T-Test)	0.836	Yes (Paired T-Test)	0.626
2	Yes (Paired T-Test)	0.218	Yes (Paired T-Test)	0.234
3	Yes (Paired T-Test)	0.919	No (Wilcoxon)	0.600
4	No (Wilcoxon)	0.028 *	Yes (Paired T-Test)	0.211
5	Yes (Paired T-Test)	0.970	Yes (Paired T-Test)	0.395
6	Yes (Paired T-Test)	0.635	Yes (Paired T-Test)	0.171
7	Yes (Paired T-Test)	0.283	Yes (Paired T-Test)	0.202

Figures 4.28 and 4.29 show the error signal peaks with standard error bars for all perturbations of Non-Randomized and Randomized experiments during eyes opened and eyes closed conditions respectively.

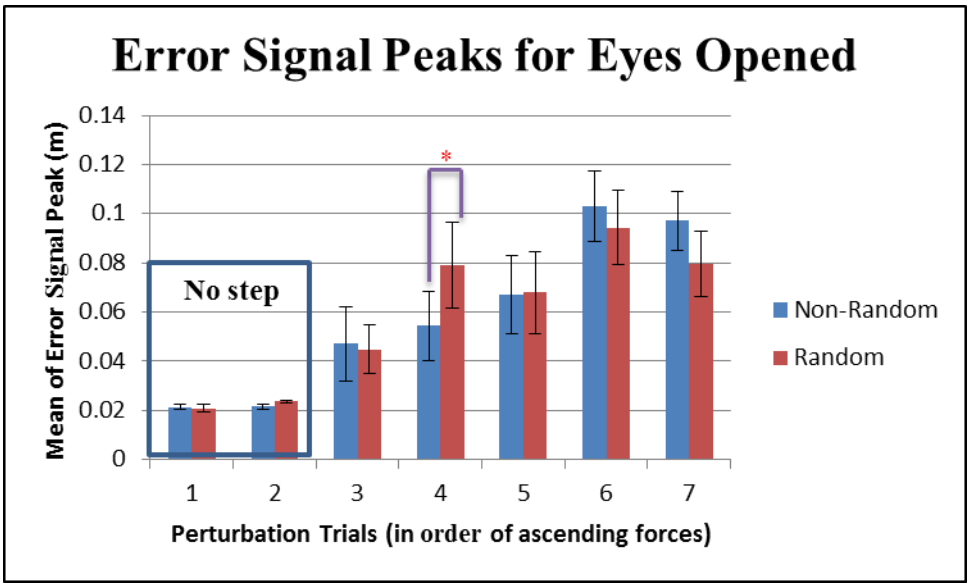

Figure 4.28 Error signal peaks means and standard error bars for Non-Randomized and Randomized experiments for the first experiment (Regular, EO), (* indicates significance, $\alpha = 0.05$).

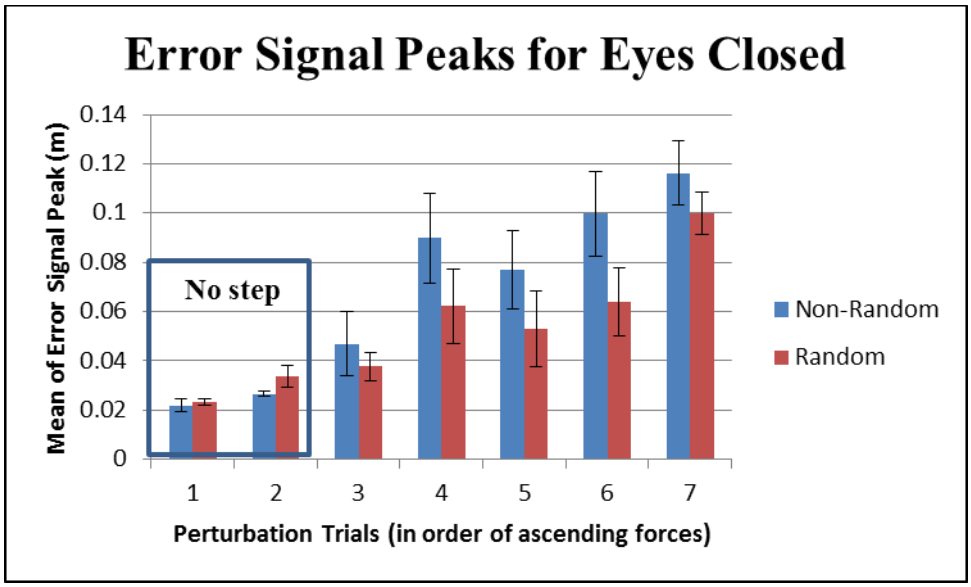

Figure 4.29 Error signal peaks means and standard error bars for Non-Randomized and Randomized experiments for the first experiment (Regular, EO), (* indicates significance, α = 0.05).

5) Correlation between COM and COP for (Regular, Non-Randomized vs Randomized, EO and EC).

The correlations between COM and COP for Non-Randomized and Randomized trials were found. Table 4.7 shows subjects correlation coefficients for Non-Randomized and Randomized trials. Table 4.8 shows a summary of the statistical analysis of correlation coefficients for Non-Randomized and Randomized data for all of the perturbations.

Table 4.7 The Correlation Coefficients for Non-Randomized and Randomized

	Eyes opened		Eyes Closed	
subject	Non-Randomized	Randomized	Non-Randomized	Randomized
1	0.973	0.972	0.968	0.972
2	0.977	0.976	0.968	0.98
3	0.96	0.959	0.964	0.977
4	0.95	0.971	0.956	0.955
5	0.99	0.985	0.985	0.982
6	0.961	0.974	0.985	0.875

Table 4.8 Summary of the Statistical Analysis of the Correlation between COM and COP For Each Perturbation (Regular, Non-Randomized vs Randomized, EO), (* indicates significance, α = 0.05)

	Eyes opened		Eyes Closed	
Perturbation	Normality	Probability	Normality	Probability
1	Yes (Paired T-Test)	0.287	Yes (Paired T-Test)	0.045 *
2	Yes (Paired T-Test)	0.286	Yes (Paired T-Test)	0.061
3	Yes (Paired T-Test)	0.376	Yes (Paired T-Test)	0.167
4	Yes (Paired T-Test)	0.577	Yes (Paired T-Test)	0.910
5	Yes (Paired T-Test)	0.026 *	Yes (Paired T-Test)	0.564
6	Yes (Paired T-Test)	0.178	Yes (Paired T-Test)	0.401
7	Yes (Paired T-Test)	0.013 *	Yes (Paired T-Test)	0.697

There is no significant difference between Non-Randomized and Randomized except for perturbations five and seven for EO and the first perturbation of EC. Figures 4.30 and 4.31 show the correlations between COM and COP with standard error bars for Non-Randomized and Randomized experiments during eyes opened and eyes closed conditions respectively.

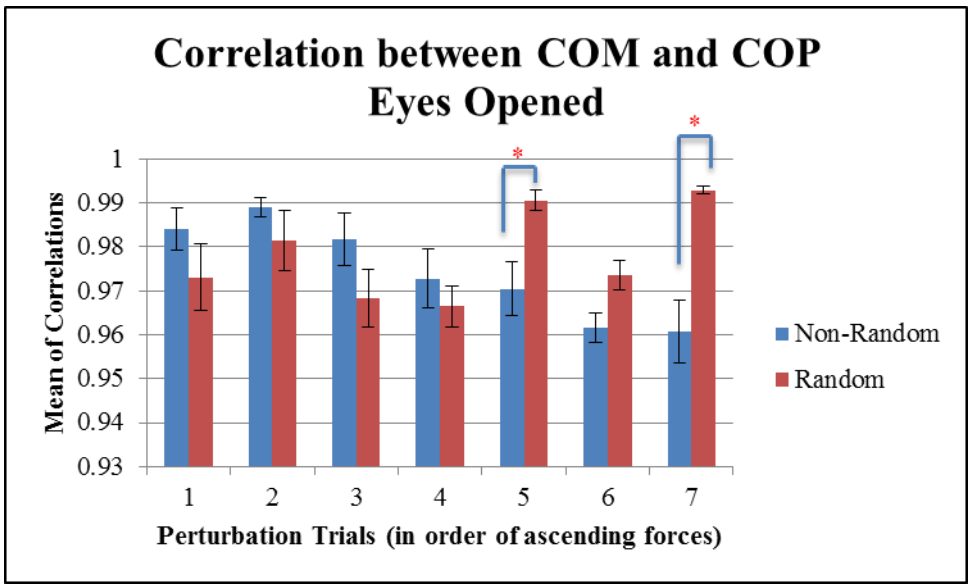

Fig 4.30 The correlation between COM and COP for Non-Randomized and Randomized Regular experiment, EO, (* indicates significance, α = 0.05).

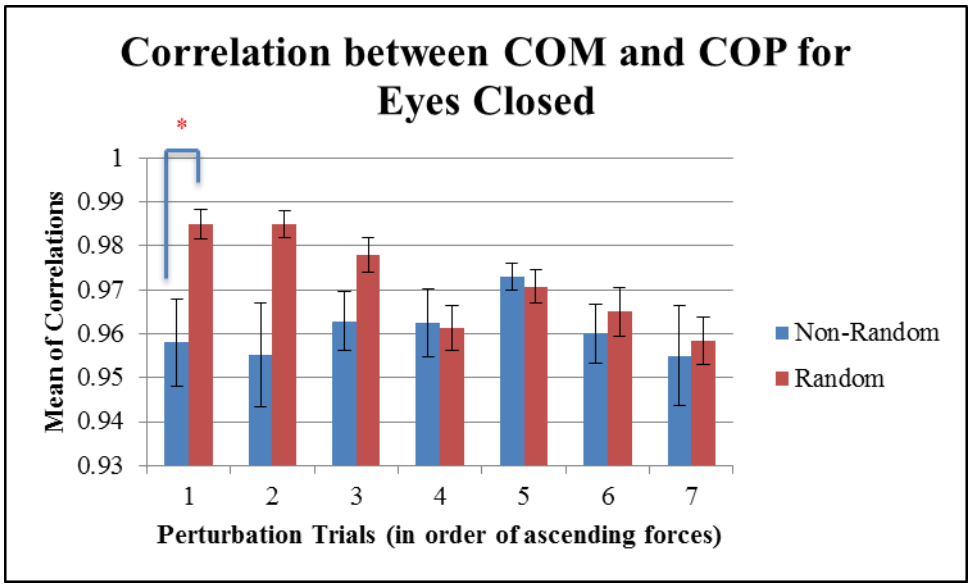

Fig 4.31 The correlation between COM and COP for Non-Randomized and Randomized Regular experiment, EC, (* indicates significance, $\alpha = 0.05$).

Generally we can say that randomizing the order of perturbations forces does not affect the reaction to the perturbations. Since there is no significant difference in total forward displacement and number of steps between Non-Randomized and Randomized experiments, Non-Randomized and Randomized groups data was combined when statistical analysis was performed for these two parameters. For the rest of the parameters there is a significant difference between Non-Randomized and Randomized experiments for at least one perturbation for EO and/or EC so when statistically analyzing these parameters they were analyzed for Non-Randomized and Randomized separately.

4.9.3 Regular and Blocks Comparison Results

1) Total forward displacement (Regular vs Blocks, EO and EC).

Normality test was performed to check if the total forward displacement of Regular and Blocks experiments for each perturbation is normal or not, then the appropriate statistical test was implemented to compare between them. Table 4.9 below shows a summary of the statistical results for both EO nd EC trials. There is a significant difference between the total forward displacement of Regular and Blocks experiments for perturbations two-seven, considering that perturbations two-five show a high significnt difference for EO trial. And there is a significant difference between the total forward displacement of Regular and Blocks experiments for perturbations one-six, considering that perturbations one-six show a high significnt difference for EC trial.

Table 4.9 Statistical Analysis Summary for Total Forward Displacement for Non-Randomized and Randomized (Regular experiment, EO and EC), (* indicates significance, α = 0.05)

	Eyes opened		Eyes Closed	
Perturbation	Normality	Probability	Normality	Probability
1	No (Wilcoxon)	0.680	Yes (Paired T-Test)	0.003 **
2	Yes (Paired T-Test)	0.000 **	Yes (Paired T-Test)	0.006 **
3	Yes (Paired T-Test)	0.000 **	Yes (Paired T-Test)	0.000 **
4	Yes (Paired T-Test)	0.003 **	Yes (Paired T-Test)	0.004 **
5	Yes (Paired T-Test)	0.002 **	Yes (Paired T-Test)	0.009 **
6	Yes (Paired T-Test)	0.011 *	Yes (Paired T-Test)	0.005 **
7	Yes (Paired T-Test)	0.017 *	Yes (Paired T-Test)	0.170

Figures 4.32 and 4.33 show the total forward displacement with standard error bars for all perturbations of Regular and Blocks experiments during eyes opened and eyes closed conditions respectively. The significant difference in the total forward displacement for Regular and Blocks is clear from the Figures 4.32 and 4.33, where the total forward

displacement for Blocks experiment is larger than for Regular experiment when eyes closed similar to eyes opened trial. This is expected since wearing the shoes with Blocks makes the BOS smaller and so makes subjects less stable, to compensate subjects step further forward compared to the first experiment.

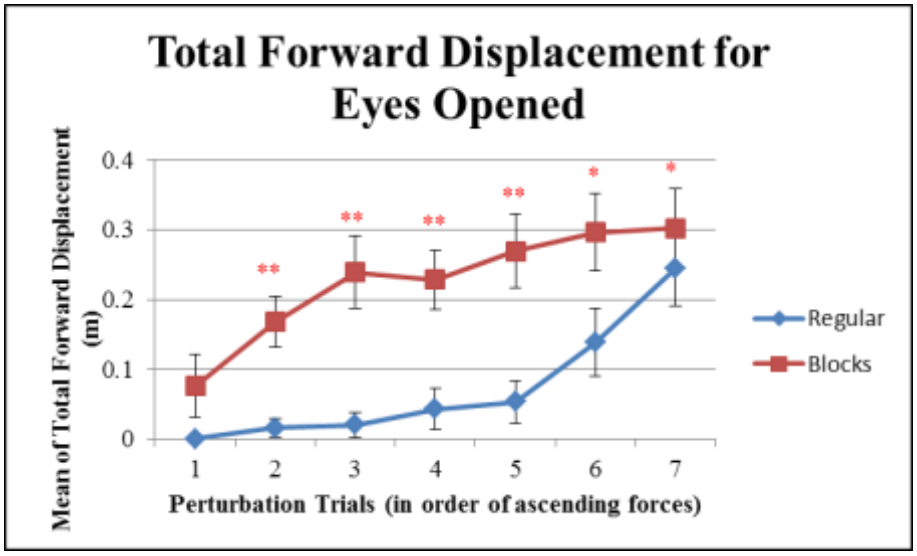

Figure 4.32 The means of the total forward displacement and standard error bars for Non-Randomized and Randomized for the first experiment (Regular, EO), (* indicates significance, $\alpha = 0.05$).

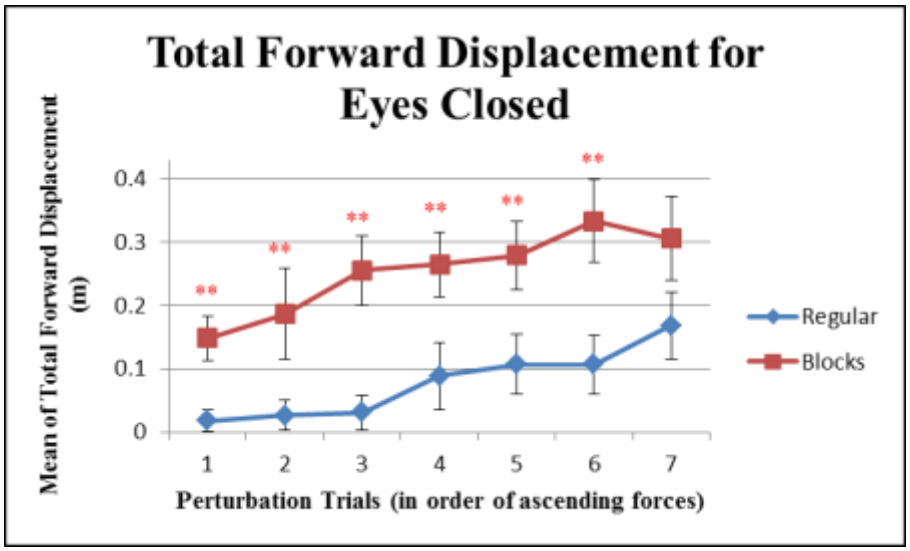

Figure 4.33 The means of the total forward displacement and standard error bars for Non-Randomized and Randomized for the first experiment (Regular, EC), (* indicates significance, $\alpha = 0.05$).

2) Number of steps (Regular vs Blocks, EO and EC).

Table 4.10 below shows a summary of the statistical results for both EO nd EC trials. For EO there is a significant difference between the number of steps of Regular and Blocks experiments for perturbations one-three and six-seven, considering that perturbations two and three show a high significnt difference. While for EC there is a significant difference between the number of steps of Regular and Blocks experiments for perturbations one-six, considering that data regarding to perturbations two and three show a high significant difference such as for eyes opened trials before.

Table 4.10 Statistical Analysis Summary of Number of Steps for Non-Randomized and Randomized for the First Experiment (Regular, EO and EC), (* indicates significance, α = 0.05)

	Eyes opened		Eyes Closed	
Perturbation	Normality	Probability	Normality	Probability
1	No (Wilcoxon)	0.034 *	No (Wilcoxon)	0.020 *
2	No (Wilcoxon)	0.007 **	Yes (Paired T-Test)	0.005 **
3	Yes (Paired T-Test)	0.005 **	Yes (Paired T-Test)	0.001 **
4	No (Wilcoxon)	0.058	No (Wilcoxon)	0.024 *
5	Yes (Paired T-Test)	0.054	No (Wilcoxon)	0.024 *
6	No (Wilcoxon)	0.034 *	No (Wilcoxon)	0.038 *
7	No (Wilcoxon)	0.046 *	No (Wilcoxon)	0.279

Figure 4.34 shows the number of steps with standard error bars for all perturbations of Regular and Blocks experiments with visual feedback. Figure 4.35 shows the number of steps with standard error bars for all perturbations of Regular and Blocks experiments without visual feedback. According to the Figures 4.34 and 4.35, the number of steps for Blocks experiment is more than for Regular experiment for eyes closed such as in eyes opened trials, this is expected since wearing the shoes with Blocks makes the BOS smaller and so makes subjects less stable, to compensate subjects take more steps forward compared to the first experiment.

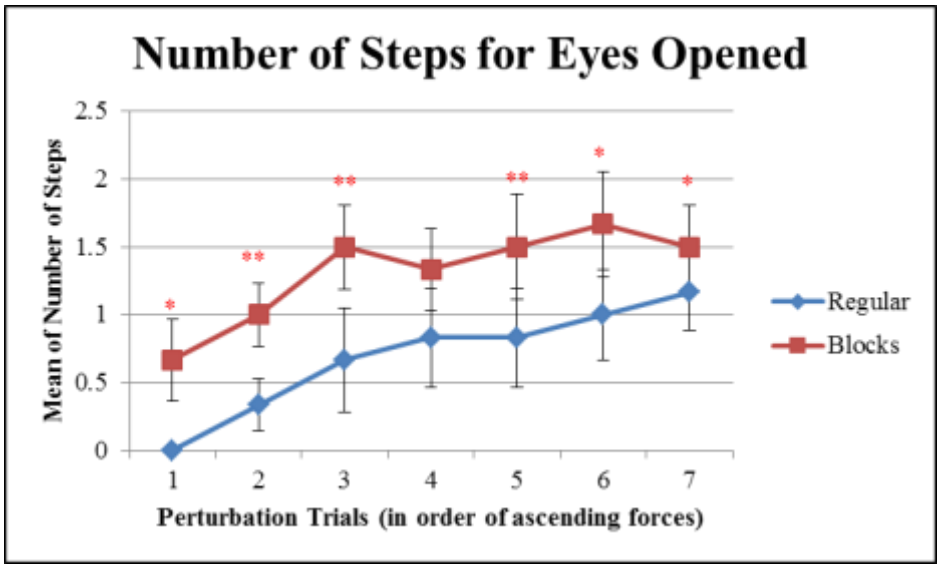

Figure 4.34 The means of the total forward displacement and standard error bars for Non-Randomized and Randomized for the first experiment (Regular, EO), (* indicates significance, $\alpha = 0.05$).

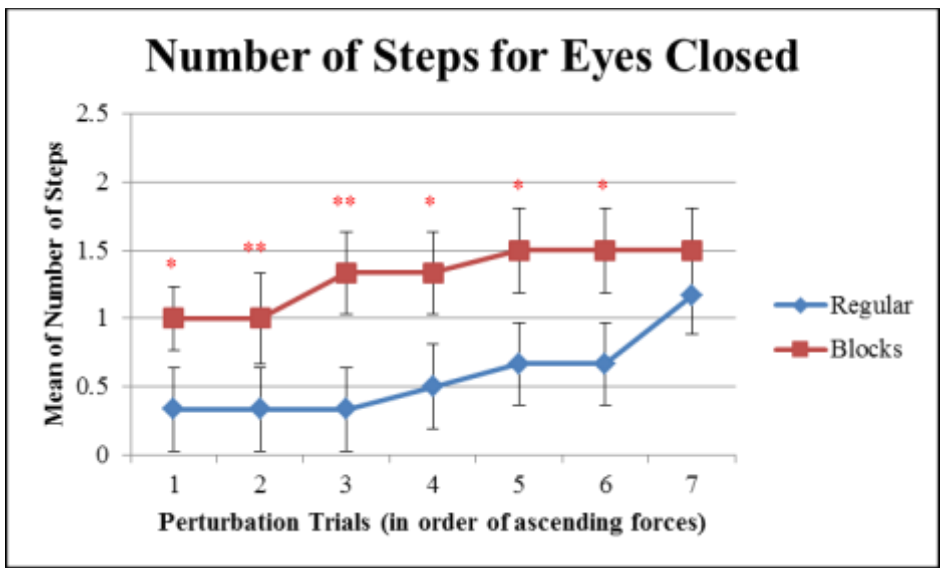

Figure 4.35 The mean of the total forward displacement and standard error bars for Non-Randomized and Randomized for the first experiment (Regular, EC), (* indicates significance, $\alpha = 0.05$).

4.9.4 Regular, Blocks, and Pivots Comparison Results

1) Stability index (Non-Randomized, Regular vs Blocks vs Pivots, EO and EC).

Repeated measures ANOVA test was used to find if there is any possible signifiicant difference among the three possible combinations of the data (Regular vs Blocks, Blocks vs Pivots, and Regular vs Pivots), and a post-hoc test was performed the significant difference is within which combination. Table 4.11 shows a summary of the statistical results. Bonefroni correction is used (α = (0.05/3) = 0.01667). There is no significant difference amongst the stability index of the three experiments EO. On the other hand, there is a significant difference in SI between Regular and Pivots for perturbations three,five, and six.In addition to a significant difference between Blocks and Pivots Corresponding to the sixth perturbation.

Table 4.11 Repeated Measurements ANOVA Summary for Stability Index of First, Second, and Third Experiments Six Perturbations (**Non-Randomized,** EO and EC), (* indicates significance, $\alpha = 0.01667$)

Non Randomized					
	Eyes opened		Eyes Closed		
Perturb -ation	Mauchly's Test Of Sphericity	Probability	Mauchly's Test Of Sphericity	Probability	
1	Yes	0.211	Yes	0.048	
2	Yes	0.545	Yes	0.132	
3	Yes	0.511	Yes	0.01 *	
				Regular-Blocks	0.935
				Blocks-Pivots	0.050
				Regular-Pivots	0.006 *
4	Yes	0.673	No (Greenhouse-Geisser)	0.026	
5	Yes	0.877	Yes	0.01 *	
				Regular-Blocks	0.293
				Blocks-Pivots	0.018
				Regular-Pivots	0.014 *
6	No (Green house-Geisser)	0.425	Yes	0.001 **	
				Regular_Blocks	0.608
				Blocks-Pivots	0.015 *
				Regular-Pivots	0.005 *

Figures 4.36 and 4.37 shows the stability index for all perturbations among the first, second and third experiment of the Non-Randomized group when there is visual feedback and there is no visual feedback. From the Figure 4.37, we notice that whenever there is a significant difference between them, SI of Pivots experiment is lower, which indicates that subjects on Pivots are stable. Furthermore, the stability index for Pivots experiment is very close to Regular. This indicates that using trekking poles to balance gives a good stability.

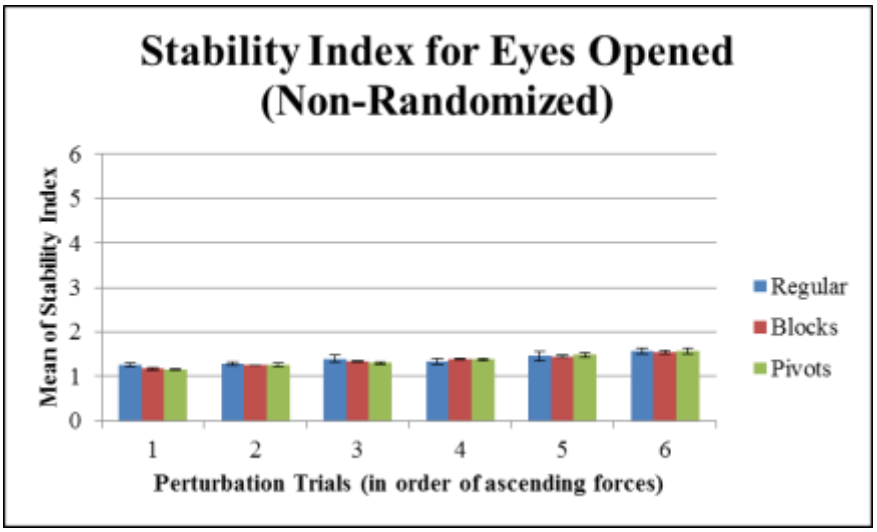

Figure 4.36 The means of the stability index and standard error bars for all perturbations of the first, second, and third experiment (Non-Randomized, EO).

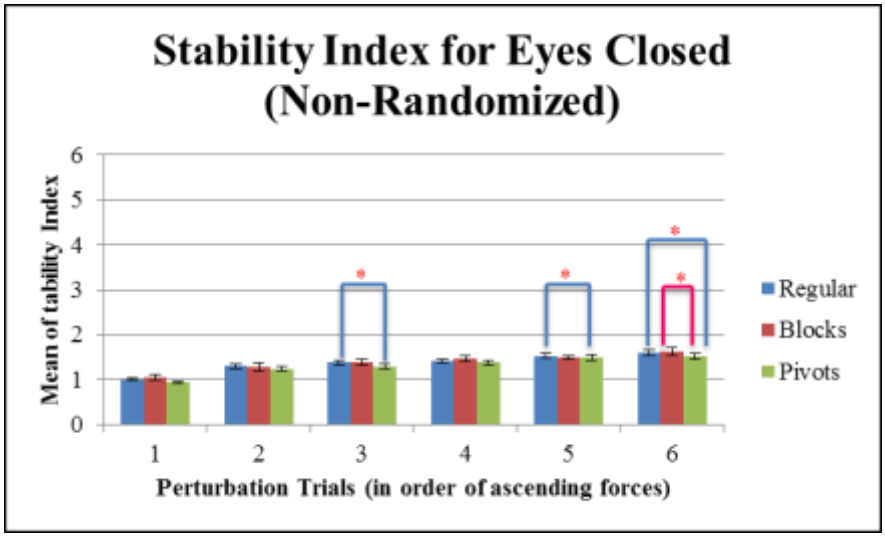

Figure 4.37 The means of the stability index and standard error bars for the first, second, and third experiment (Non-Randomized, EC), (* indicates significance, $\alpha = 0.01667$).

2) Stability index (Randomized, Regular vs Blocks vs Pivots, EO and EC).

Table 4.12 shows a summary of the statistical results of the rmANOVA and post-hoc test compairing SI among Regular, Blocks, and Pivots experiments. There is a significant difference amongst the stability index of the three experiments EO between Regular and Pivots regarding perturbations four and five, and a high significant difference between Regular and Blocks regarding perturbations three and four. When there is visual feedback there is a significant difference in SI between Regular and Pivots for perturbations four, between Regular and Blocks for perturbations one, two and four whereas for perturbations one and two the difference is highly significant.A significant difference between Blocks and Pivots for first perturbation was also observed.

Table 4.12 Repeated Measures ANOVA Results Summary for Stability Index of First, Second and Third Experiments (Randomized**,** EO and EC), (* indicates significance, α = 0.01667)

Randomized						
	Eyes opened			Eyes Closed		
Perturbation	Mauchly's Test Of Sphericity	Probability		Mauchly's Test Of Sphericity	Probability	
1	Yes	0.242		Yes	0.01 *	
					Regular-Blocks	0.002 **
					Blocks-Pivots	0.009 *
					Regular-Pivots	0.486
2	Yes	0.013 *		Yes	0.014 *	
		Regular-Blocks	0.023		Regular- Blocks	0.003
		Blocks-Pivots	0.486		Blocks-Pivots	0.272
		Regular-Pivots	0.022		Regular-Pivots	0.106
3	Yes	0.008 *		Yes	0.024	
		Regular -Blocks	0.003 **			
		Blocks-Pivots	0.676			
		Regular-Pivots	0.034			
4	No (Greenhouse-Geisser)	0.01 *		Yes	0.01 *	
		Regular- Blocks	0.000 **		Regular_Blocks	0.015 *
		Blocks-Pivots	0.946		Blocks-Pivots	0.978
		Regular-Pivots	0.011 *		Regular-Pivots	0.006 *
5	Yes	0.008 *		Yes	0.098	
		Regular-Blocks	0.049			
		Blocks-Pivots	0.238			
		Regular-Pivots	0.005 *			
6	No (Greenhouse-Geisser)	0.323		No (Greenhouse-Geisser)	0.196	

Figure 4.38 shows the stability index and standard error bars for all perturbations of the first, second, and third experiment (Randomized, EO). And Figure 4.39 shows the stability index and standard error bars for all perturbations of the first, second, and third experiment (Randomized, EC). According to the Figures, we notice that even though there is statistical difference in stability index between Regular and Pivots experiment corresponding many perturbations, they are still very close to each other, particularly for the first two perturbations.

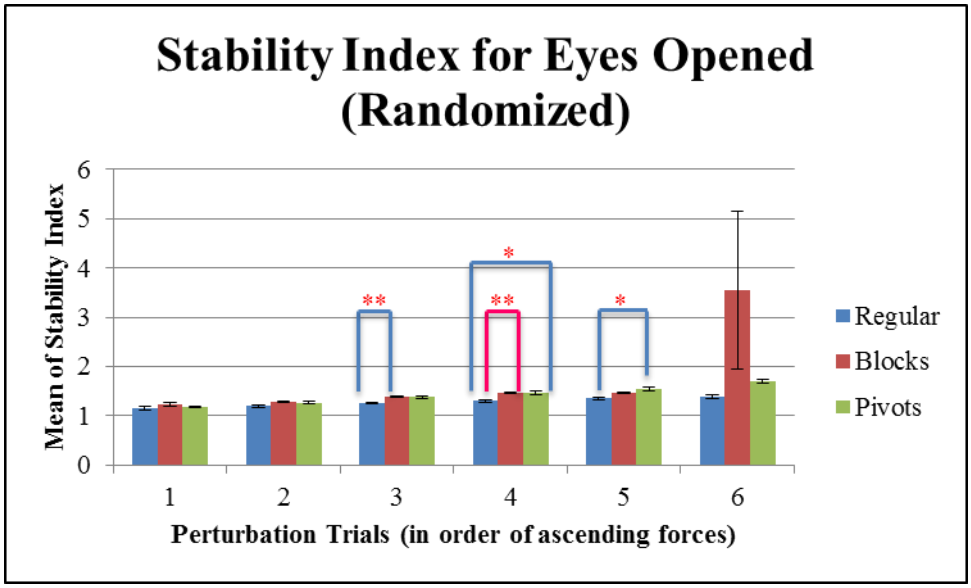

Figure 4.38 The means of the stability index and standard error bars for the first, second, and third experiment (Randomized, EO), (* indicates significance, α = 0.01667).

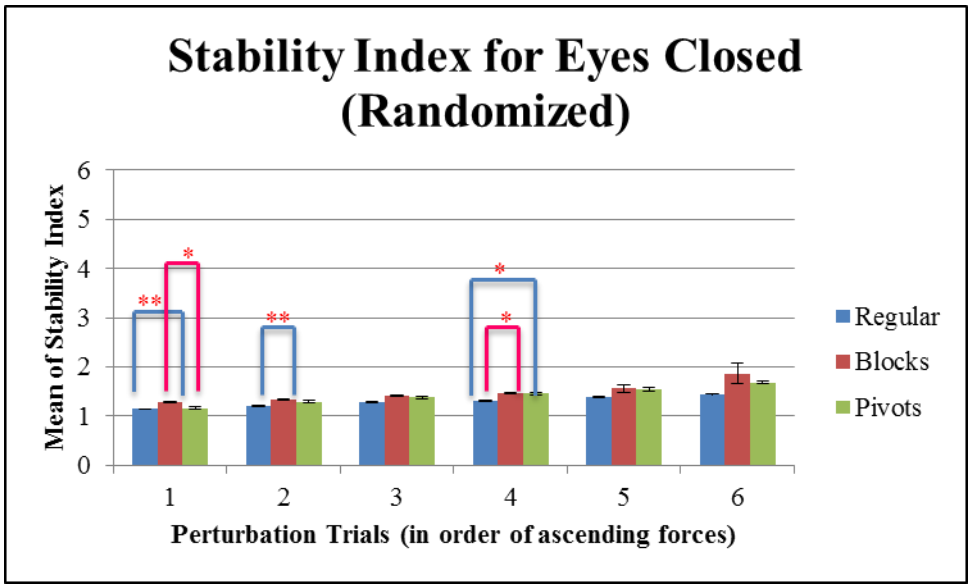

Figure 4.39 The means of the stability index and standard error bars for the first, second, and third experiment (Randomized, EC), (* indicates significance, α = 0.01667).

3) Error signal peaks (Non-Randomized, Regular vs Blocks vs Pivots, EO and EC).

Table 4.13 shows a summary of the statistical results. There is no significant difference amongst the error signal peaks of the three experiments for EO trial. It is interisting that there

is no significant difference in SI among the three experiments for the same trial additionally (Non-Randoized, EO). On the other hand, there is a significant difference in error signal peaks only between Regular and Pivots for the sixth perturbation for EC trial.

Table 4.13 Repeated Measures ANOVA Results Summary for Error Signal Peaks of First, Second, and Third Experiments (Non-Randomized, EO and EC), (* indicates significance, α = 0.01667)

Non Randomized					
	Eyes opened		Eyes Closed		
Perturb-ation	Mauchly's Test Of Sphericity	Probability	Mauchly's Test Of Sphericity	Probability	
1	Yes	0.058	Yes	0.022	
2	No (Greenhouse-Geisser)	0.035	Yes	0.131	
3	No (Greenhouse-Geisser)	0.160	Yes	0.023	
4	No (Greenhouse-Geisser)	0.092	Yes	0.009 *	
				Regular_Blocks	0.097
				Blocks_Pivots	0.027
				Regular_Pivots	0.039
5	Yes	0.070	Yes	0.022	
6	Yes	0.434	Yes	0.011 *	
				Regular_Blocks	0.295
				Blocks_Pivots	0.025
				Regular_Pivots	0.006*
7	Yes	0.058	Yes	0.022	

Figures 4.40 and 4.41 show the error signal peaks for all perturbations of the first, second, and third experiment of Non-Randomized groups, EO and EC trials. Though there is a significant difference for one combination of the sixth perturbation of EC trial. From Figures 4.40 and 4.41, it is obvious that there is difference between error signals peaks of Regular, Blocks and Pivots. Blocks error signal peaks is the highest while Pivots is the lowest and close to Regular experiment error signal peaks. The fact that error signal peaks are close to each other corresponding to the first two perturbations, when subjects did not use stepping strategy enhances the hypothesis that trekking poles are good to maintain balance.

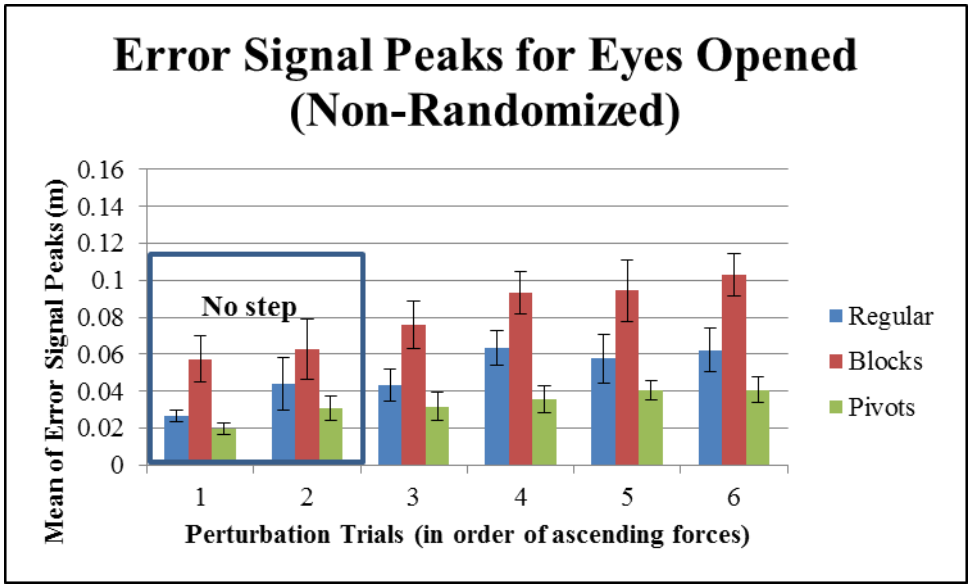

Figure 4.40 The means of the error signal peaks and standard error bars for all perturbations of the first, second, and third experiment (Non-Randomized, EO).

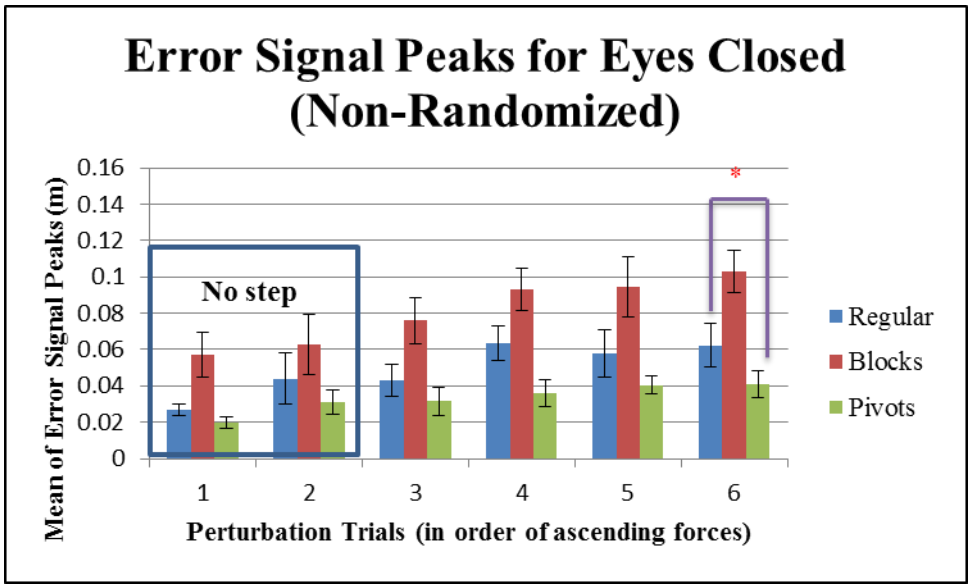

Figure 4.41 The means of the stability index and standard error bars for all perturbations of the first, second, and third experiment (Non-Randomized, EC), (* indicates significance, $\alpha = 0.01667$).

4) Error signal peaks (Randomized, Regular vs Blocks vs Pivots, EO and EC).

Table 4.14 shows a summary of the statistical analysis results. There is a significant difference in the error signal peaks in EO trial between Blocks and Pivots in the sixth perturbation. Significant difference was observed in the fourth and sixth perturbation in EC trial between Regular and Pivots experiments.

Table 4.14 Repeated Measures ANOVA Results Summary for Error Signal Peaks of First, Second, and Third Experiments (Randomized, EO and EC), (* indicates significance, α = 0.01667)

Randomized					
	Eyes opened			Eyes Closed	
Perturb-ation	Mauchly's Test Of Sphericity	Probability		Mauchly's Test Of Sphericity	Probability
1	Yes	0.047		Yes	0.470
2	Yes	0.012 *		Yes	0.059
		Regular-Blocks	0.039		
		Blocks-Pivots	0.063		
		Regular-Pivots	0.022		
3	Yes	0.061		Yes	0.082
4	Yes	0.035		Yes	0.010 *
					Regular-Blocks 0.110
					Blocks-Pivots 0.128
					Regular-Pivots 0.010 *
5	Yes	0.069		Yes	0.055
6	Yes	0.005 *		Yes	0.008 *
		Regular-Blocks	0.018		Regular-Blocks 0.011 *
		Blocks-Pivots	0.006 *		Blocks-Pivots 0.035
		Regular-Pivots	0.936		Regular-Pivots 0.192
7	Yes	0.047		Yes	0.470

Figures 4.42 and 4.43 show the error signal peaks for all perturbations in the first, second, and third experiment of the randomized group for both EO and EO trials. It is

observed that the error signal peaks of Regular and Pivots experiments are close to each other, especially for the first two perturbations, when subjects did not use stepping strategy.

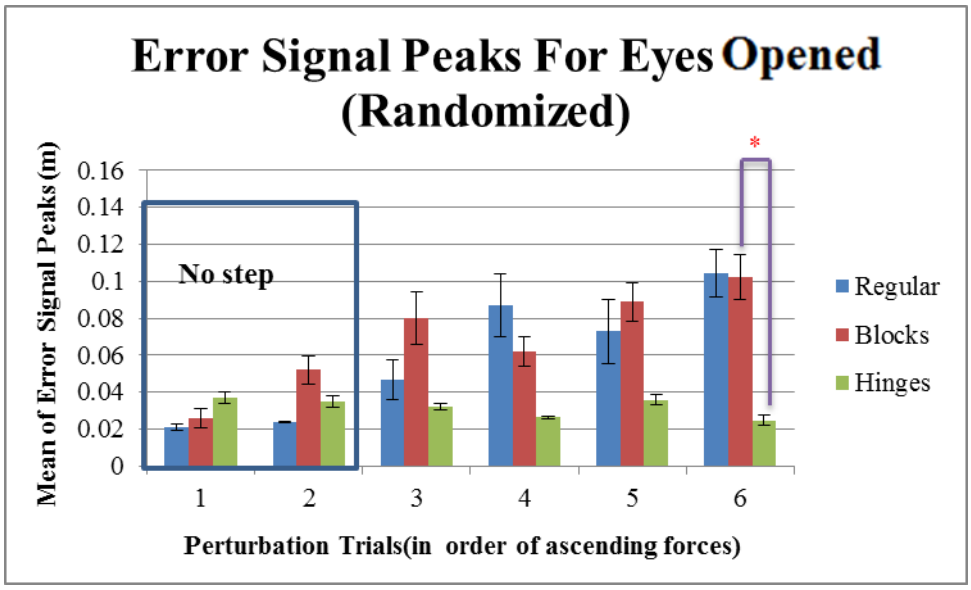

Figure 4.42 The means of the stability index and standard error bars for the first, second, and third experiment (Randomized, EO), (* indicates significance, α = 0.01667).

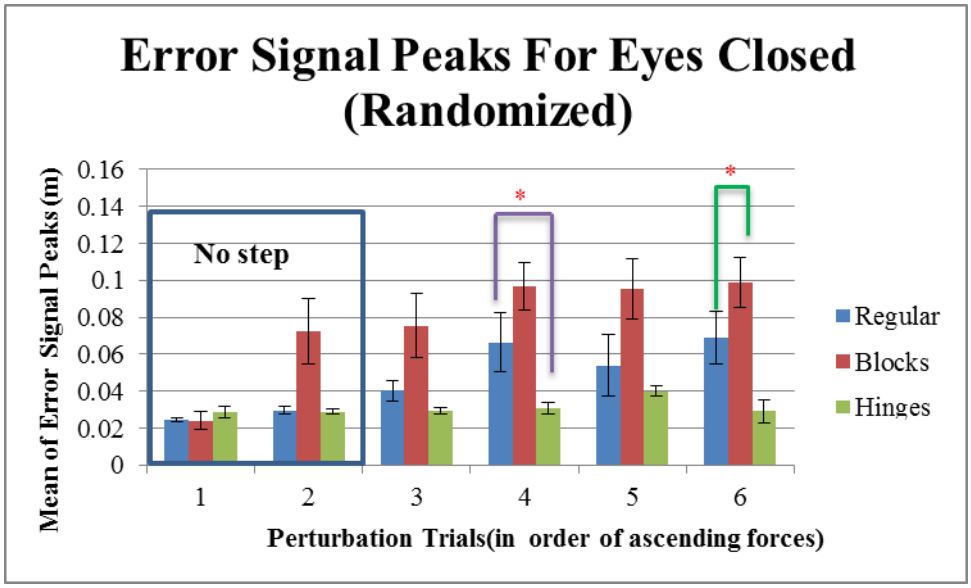

Figure 4.43 The means of the stability index and standard error bars for the first, second, and third experiment (Randomized, EC), (* indicates significance, α = 0.01667).

5) **Correlation between COM and COP for each perturbation (Regular vs Blocks vs Pivots, Non-Randomized, EO and EC).**

Table 4.15 below displays the correlation coefficients of all Non-Randomized subjects and table 4.16 displays a summary of the statistical results. There is a significant difference in the correlations between COM and COP between Blocks and Pivots and a high significant difference between Regular and Blocks in the first perturbation for EO trial, and no significant difference between any combinations for the EC trial.

Table 4.15 The Correlation Coefficients of Non-Randomized Subjects for the Three Experiments: Regular, Blocks and Pivots (Non-Randomized, EO and EC)

	Eyes opened			Eyes Closed		
Subject	Regular	Blocks	Hinge	Regular	Blocks	Pivots
1	0.97	0.738	0.992	0.9690	0.9590	0.9740
2	0.934	0.967	0.967	0.9240	0.9710	0.9720
3	0.934	0.982	0.971	0.9690	0.9800	0.9810
4	0.978	0.968	0.987	0.9800	0.9710	0.9910
5	0.95	0.968	0.989	0.9800	0.9750	0.9730
6	0.975	0.975	0.965	0.9530	0.9770	0.9880

Table 4.16 Repeated Measures ANOVA Results Summary for Correlation between COM and COP Among the First, Second, and Third Experiments (Non_Randomized, EO and EC), (* indicates significance, α = 0.01667)

	Non Randomized				
	Eyes opened			Eyes Closed	
Perturbation	Mauchly's Test Of Sphericity	Probability		Mauchly's Test Of Sphericity	Probability
1	Yes	0.001 *		No Greenhouse-Geisser	0.263
		Regular-Blocks	0.0 **		
		Blocks_Pivots	0.01 *		
		Regular_Pivots	0.834		
2	Yes	0.087		No Greenhouse-Geisser	0.340
3	No Greenhouse-Geisser	0.336		No Greenhouse-Geisser	0.322
4	No Greenhouse-Geisser	0.347		No Greenhouse-Geisser	0.312
5	No Greenhouse-Geisser	0.361		No Greenhouse-Geisser	0.367
6	No Greenhouse-Geisser	0.370		No Greenhouse-Geisser	0.352

From Figures 4.44 and 4.45 it is observed that the correlation for Blocks is the lowest and for Pivots is the highest and is close to Regular. This enhances the hypothesis that trekking poles are a good approach to maintain balance.

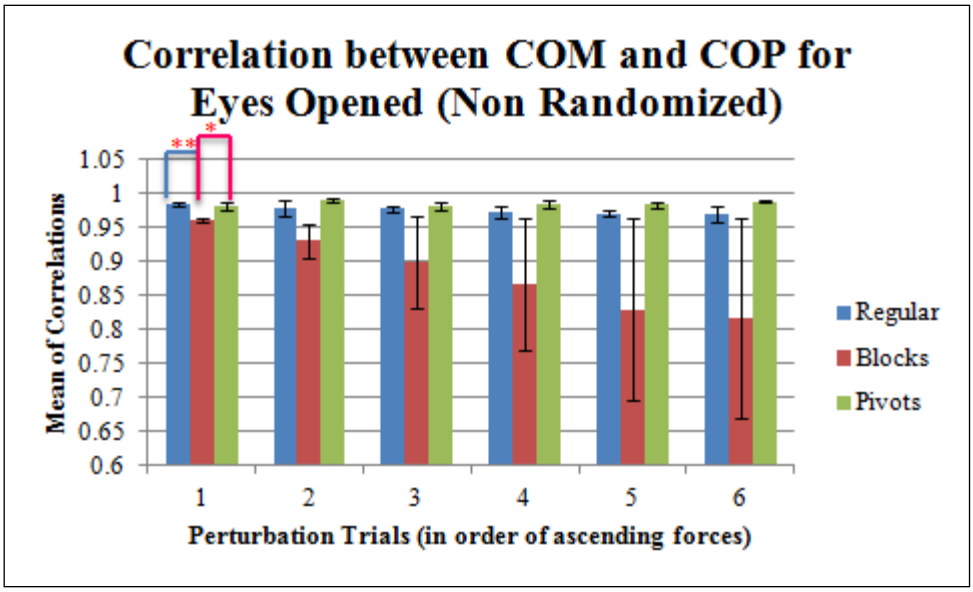

Figure 4.44 The means of the correlation between COM and COP and standard error bars for the first, second, and third experiment (Non-Randomized, EO) , (* indicates significance, $\alpha = 0.01667$) .

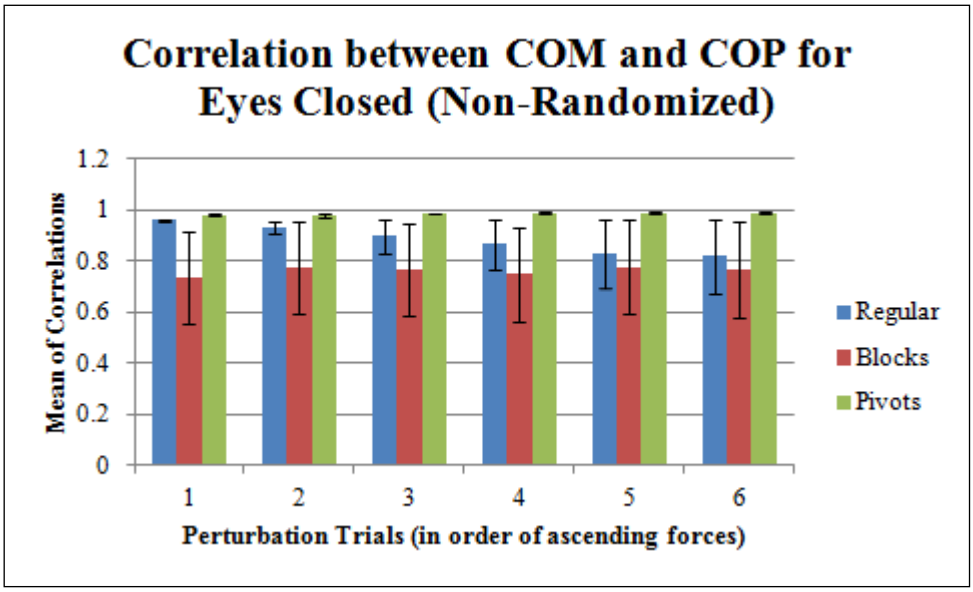

Figure 4.45 The means of the correlation between COM and COP and standard error bars for the first, second, and third experiment (Non-Randomized, EC).

6) Correlation between COM and COP (Regular vs Blocks vs Pivots, Randomized, EO and EC), for each perturbation.

Table 4.17 below displays the correlation coefficients of all Non-Randomized subjects and table 4.18 displays a summary of the statistical results. There is a significant difference in the correlations between COM and COP between Blocks and Pivots of the sixth perturbation and a high significant difference between Regular and Blocks regarding for the fifth perturbation for EO trial, and a high significant difference between Blocks and Pivots of the third perturbation for EC trial.

Table 4.17 The Correlation Coefficients of Non-Randomized Subjects for the Three Experiments: Regular, Blocks and Pivots (Non-Randomized, EO and EC)

	Eyes opened			Eyes Closed		
Subject	Regular	Blocks	Hinge	Regular	Blocks	Pivots
1	0.972	0.966	0.982	0.9720	0.9540	0.9820
2	0.976	0.98	0.982	0.9800	0.9800	0.9920
3	0.959	0.976	0.982	0.9770	0.9750	0.9890
4	0.971	0.974	0.921	0.9550	0.8920	0.9800
5	0.985	0.962	0.99	0.9820	0.9650	0.9850
6	0.974	0.956	0.989	0.8750	0.9280	0.9880

Table 4.18 Repeated Measures ANOVA Results Summary for Correlation between COM And COP Among the First, Second, and Third Experiments (Randomized, EO and EC), (* indicates significance, α = 0.01667)

	Randomized					
	Eyes opened			Eyes Closed		
Perturbation	Mauchly's Test Of Sphericity	Probability		Mauchly's Test Of Sphericity	Probability	
1	No	0.397		Yes	0.023	
2	Yes	0.086		Yes	0.007 *	
					Regular-Blocks	0.026
					Blocks-Pivots	0.035
					Regular-Pivots	0.480
3	Yes	0.072		Yes	0.001 **	
					Regular-Blocks	0.037
					Blocks-Pivots	0.002 **
					Regular-Pivots	0.042
4	Yes	0.027		No Greenhouse-Geisser	0.033	
5	Yes	0.003 *		Yes	0.133	
		Regular-Blocks	0.002 **			
		Blocks-Pivots	0.027			
		Regular-Pivots	0.932			
6	Yes	0.010 *		No Greenhouse-Geisser	0.151	
		Regular-Blocks	0.241			
		Blocks-Pivots	0.011 *			
		Regular-Pivots	0.017			

Figures 4.44 and 4.45 show that the correlation for Blocks is the lowest and for Pivots is the highest and is close to Regular. This is a good indication that COM and COP follow each other smoothly using trekking poles. On the other hand, this can be a result of standing on Pivots or due to the high elasticity of the trekking poles.

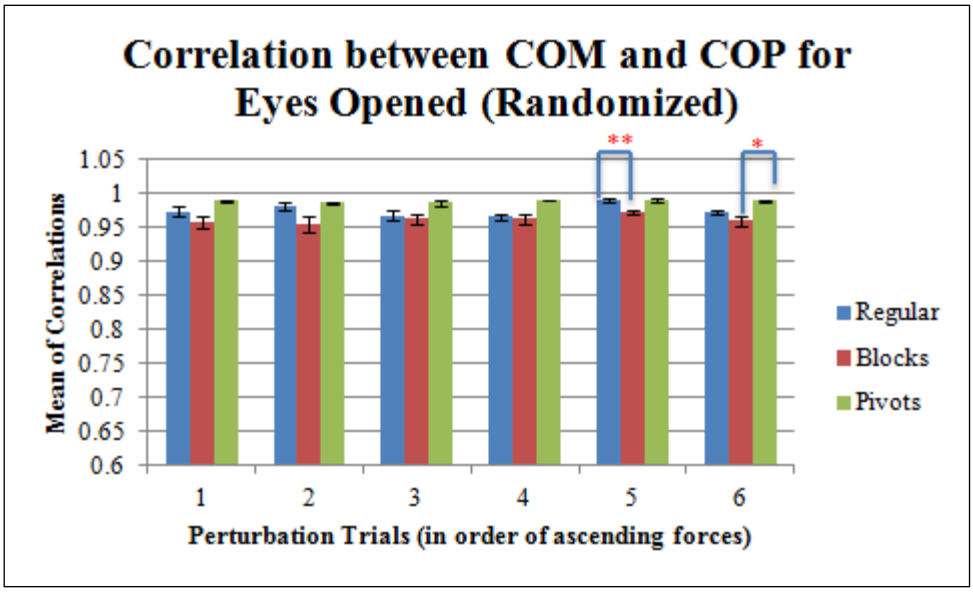

Figure 4.46 The means of the correlation between COM and COP and standard error bars for the first, second, and third experiment (Randomized, EO), (* indicates significance, α = 0.01667).

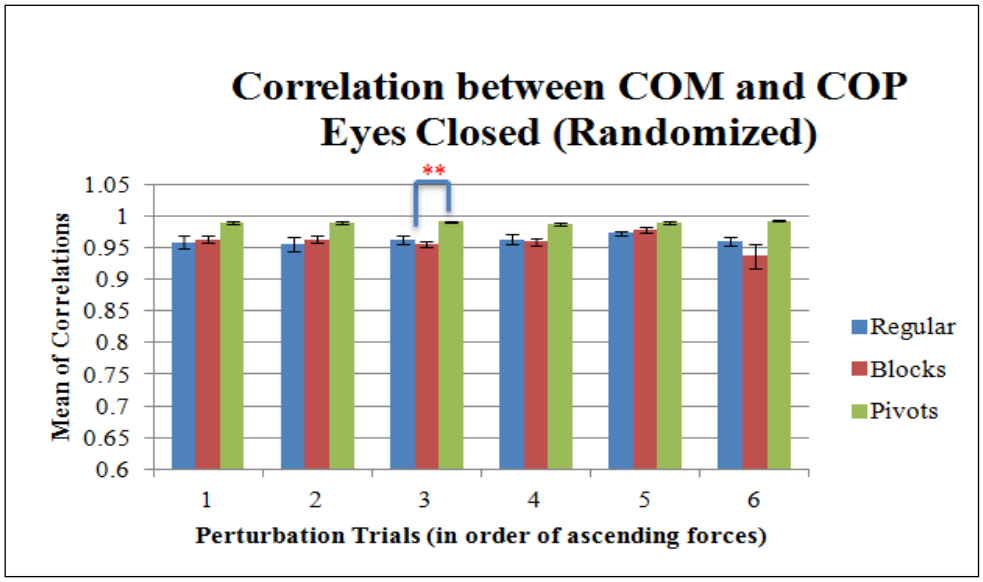

Figure 4.47 The means of the correlation between COM and COP and standard error bars for the first, second, and third experiment (Randomized, EC), (* indicates significance, α = 0.01667).

4.3.5 Eyes opened – Eyes Closed Comparisons

1) Total forward displacement (EO vs EC, Regular and Blocks).

For Eo vs EC comparisons the comparison is between two parameters so depinding on the normality test results either Wilcoxon or Paired T-Test is used.There is no significant difference in the total forward displacement metric between EOand EC for Regular and Blocks in any of the perturbations. Table 4.19 below shows a summary of the statistical results.From Figures 4.48 and 4.49 it is observed that the total forward displacement for EO and EC of both Regular and Blocks are similar to each other.

Table 4.19 Eyes opened and Eyes Closed Statistical Analysis Summary for Total Forward Displacement of Eyes opened and Eyes Closed Trials of Regular and Blocks Experiments

	Regular		Blocks	
Perturbation	Normality	Probability	Normality	Probability
1	No (Wilcoxon)	0.317	Yes (Paired T-Test)	0.670
2	No (Wilcoxon)	0.317	No (Wilcoxon)	0.374
3	No (Wilcoxon)	0.593	Yes (Paired T-Test)	0.754
4	No (Wilcoxon)	0.866	Yes (Paired T-Test)	0.108
5	No (Wilcoxon)	0.176	No (Wilcoxon)	0.169
6	Yes (Paired T-Test)	0.051	Yes (Paired T-Test)	0.321
7	Yes (Paired T-Test)	0.984	Yes (Paired T-Test)	0.415

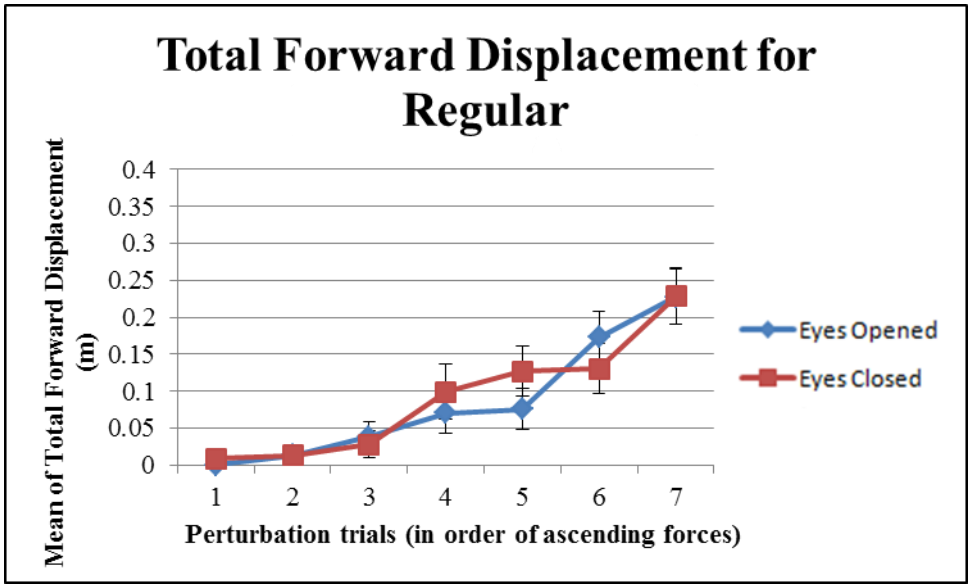

Figure 4.48 The means of the total forward displacement and standard error bars for eyes opened and eyes closed trials of the first experiment (Regular).

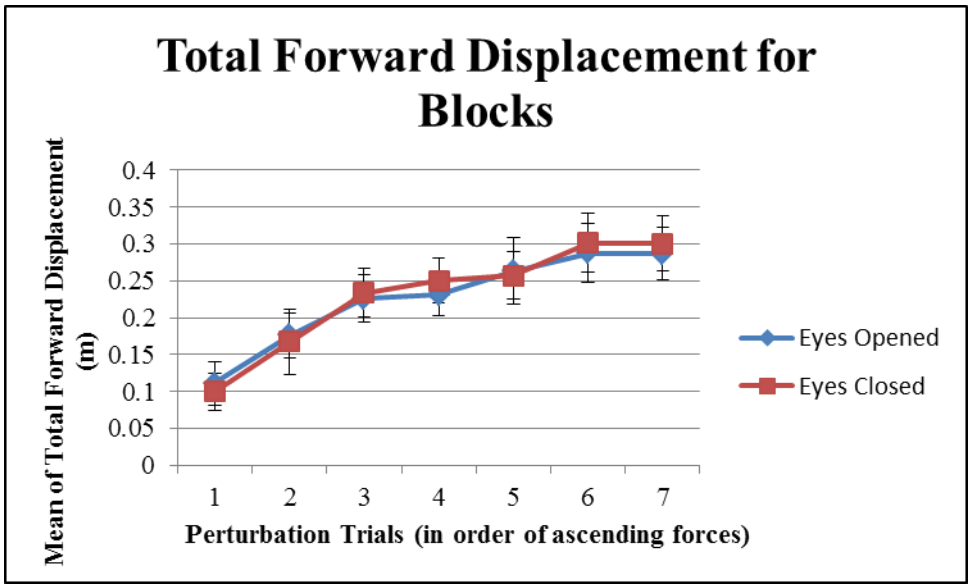

Figure 4.49 The means of the total forward displacement and standard error bars for eyes opened and eyes closed trials of the first experiment (Blocks).

2) Number of steps (EO vs EC, Regular and Blocks).

There is no significant difference in the number of steps metric between EOand EC for Regular and Blocks in any of the perturbations. Table 4.20 below shows a summary of the

statistical results. From Figures 4.50 and 4.51 it is observed that the total forward displacement for EO and EC of both Regular and Blocks are close to each other.

Table 4.20 Eyes opened and Eyes Closed Statistical Analysis Summary for Number of Steps of Eyes opened and Eyes Closed Trials of the First Experiment (Regular)

	Regular		Blocks	
Perturbation	Normality	Probability	Normality	Probability
1	No (Wilcoxon)	0.083	No (Wilcoxon)	0.655
2	No (Wilcoxon)	1.000	No (Wilcoxon)	0.564
3	No (Wilcoxon)	0.317	No (Wilcoxon)	0.317
4	No (Wilcoxon)	0.655	No (Wilcoxon)	0.564
5	No (Wilcoxon)	0.317	No (Wilcoxon)	0.317
6	No (Wilcoxon)	1.000	No (Wilcoxon)	1.000
7	No (Wilcoxon)	0.317	No (Wilcoxon)	0.317

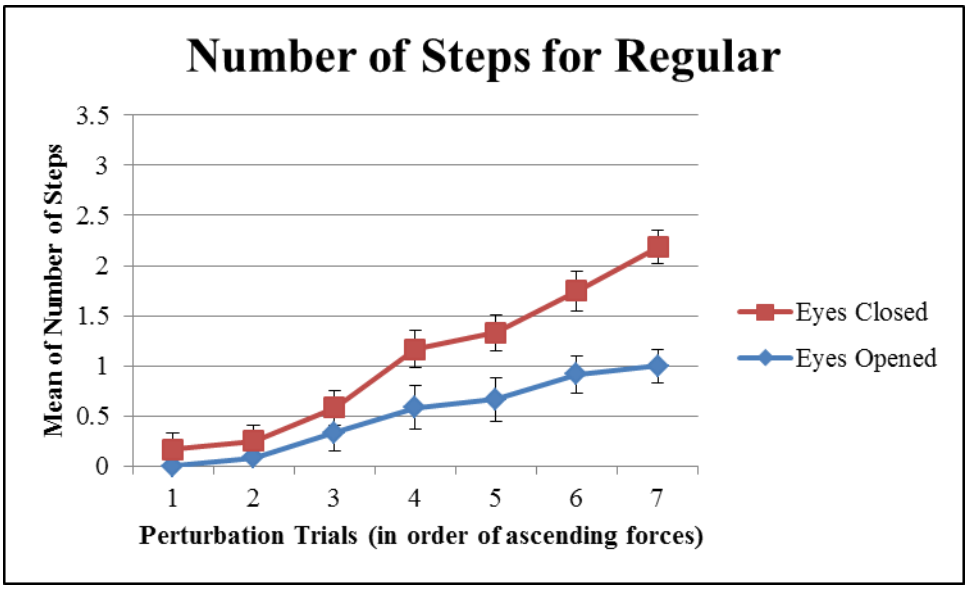

Figure 4.50 The means of the number of steps and standard error bars for eyes opened and eyes closed trials of the first experiment (Regular).

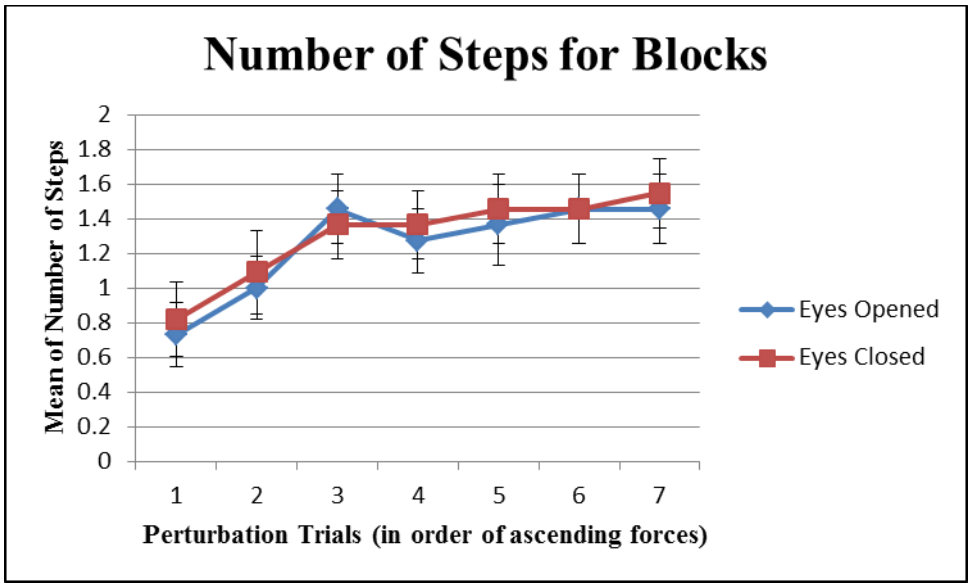

Figure 4.51 The means of the number of steps and standard error bars for eyes opened and eyes closed trials of the first experiment (Blocks).

3) Stability index (Non-Randomized, EO vs EC, Regular, Blocks, and Pivots).

Stability index comparison between EO vs EC was performed among the three experiments: Regular, Blocks and Pivots. Statistical analysis resulted in no significant difference in SI between EO and EC for any experiment. In addition, Figures 4.52, 4.53, and 4.54 display that SI scores for EO and EC are similar for the three experiments.

Table 4.21 Eyes opened and Eyes Closed Comparison Statistical Analysis Summary for Number of Steps of Eyes opened and Eyes Closed Trials of the First Experiment (Regular, Blocks and Pivots, Non-Randomized)

Perturbation	Non- Randomized					
	Regular		Blocks		Pivots	
	Normality	Probability	Normality	Probability	Normality	Probability
1	Yes	0.270	No	0.917	Yes	0.349
2	Yes	0.975	No	0.463	No	0.345
3	No	0.917	Yes	0.278	Yes	0.765
4	Yes	0.915	No	0.917	Yes	0.797
5	Yes	0.842	No	0.463	Yes	0.787
6	Yes	0.981	No	0.917	Yes	0.155
7	No	0.753	Yes	0.398		

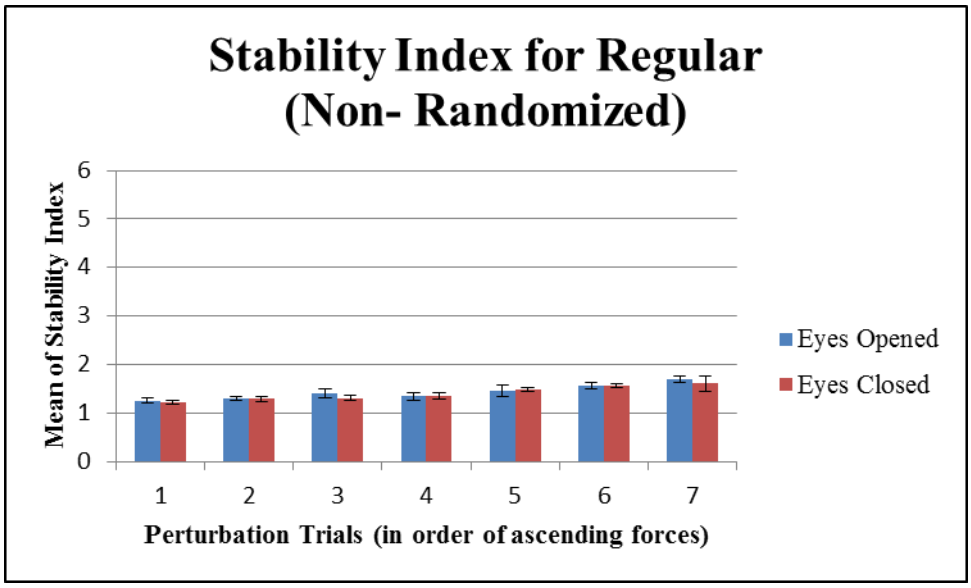

Figure 4.52 The means of the stability index and standard error bars for eyes opened and eyes closed trials of the first experiment (Regular (Non-Randomized)).

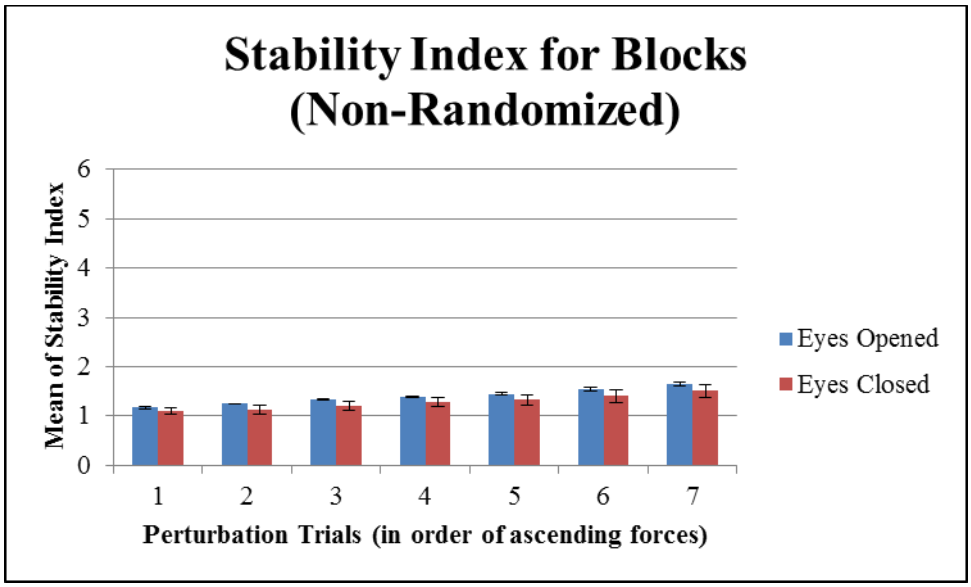

Figure 4.53 The means of the stability index and standard error bars for eyes opened and eyes closed trials of the second experiment (Blocks, Non-Randomized).

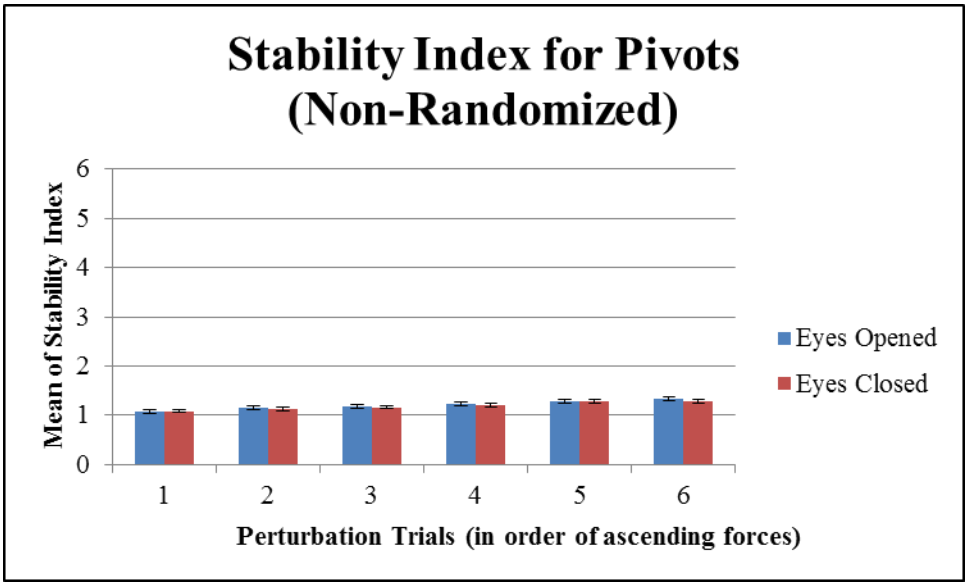

Figure 4.54 The means of the stability index and standard error bars for eyes opened and eyes closed trials of the third experiment (Pivots, Non-Randomized).

4) Stability index (Randomized, EO vs EC, Regular, Blocks, and Pivots).

Table 4.22 shows the statistical analysis results for the comparison of SI between EO and EC trials of Regular, Blocks and Pivots experiments of the Randomized group. There is no

significant difference in SI between EO and EC except for the sixth perturbation of Regular experiment, and the second perturbation of Blocks experiment.

Table 4.22 Eyes opened and Eyes Closed Comparison Statistical Analysis Summary for Number of Steps of Eyes opened and Eyes Closed Trials of the First Experiment (Regular, Blocks and Pivots, Randomized), (* indicates significance, $\alpha = 0.05$)

Perturbation	Randomized					
	Regular		Blocks		Pivots	
	Normality	Probability	Normality	Probability	Normality	Probability
1	No	0.500	Yes	0.090	Yes	0.469
2	No	0.104	Yes	0.028 *	Yes	0.625
3	Yes	0.133	Yes	0.487	Yes	0.937
4	No	0.080	Yes	0.895	Yes	0.958
5	Yes	0.098	Yes	0.289	Yes	0.423
6	Yes	0.018 *	No	0.686	Yes	0.562
7	Yes	0.151	Yes	0.151		

Figures 4.55-4.57 do not show a clear difference in SI between EO and EC except for the sixth perturbation of Blocks where SI of EO is too high and mostly is an outlier.

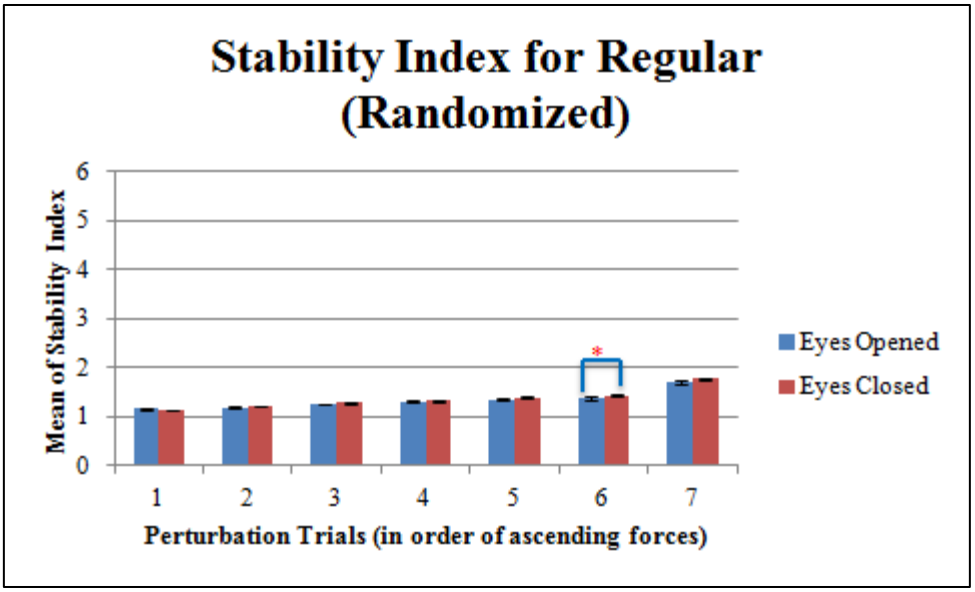

Figure 4.55 The means of the stability index and standard error bars for eyes opened and eyes closed trials of the first experiment (Regular, Randomized), (* indicates significance, α = 0.05).

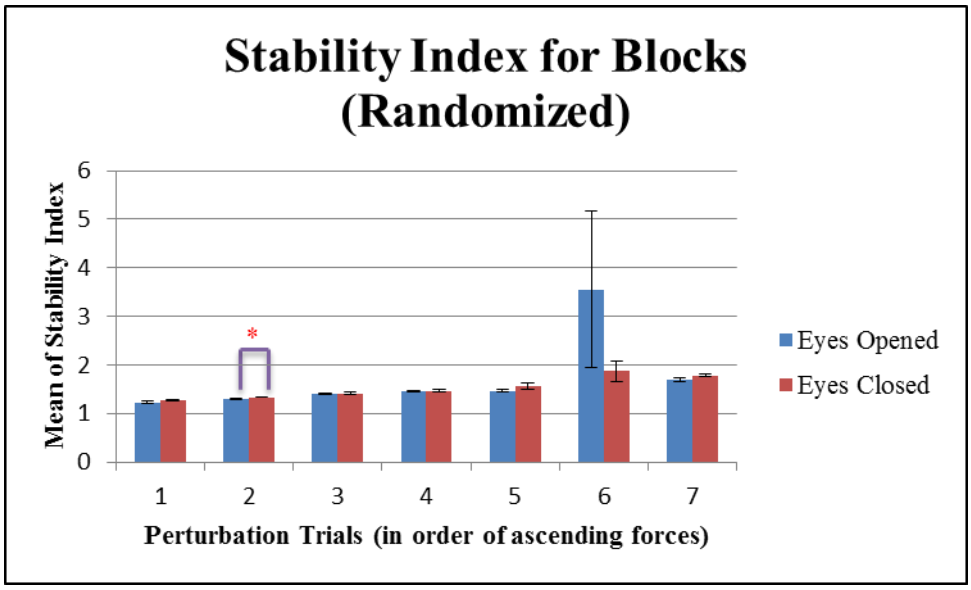

Figure 4.56 The means of the stability index and standard error bars for eyes opened and eyes closed trials of the second experiment (Blocks, Randomized), (* indicates significance, α = 0.05).

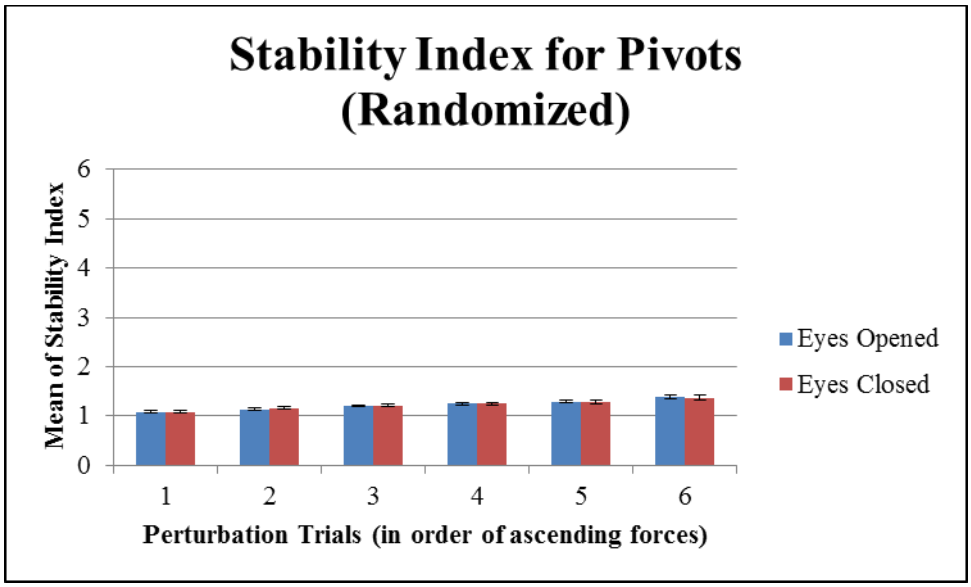

Figure 4.57 The means of the stability index and standard error bars for eyes opened and eyes closed trials of the third experiment (Pivots, Randomized).

5) Error signal peaks (Non-Randomized, EO vs EC, Regular, Blocks, and Pivots).

Table 4.23 shows the statistical analysis results for the comparison of the error signal peaks between EO and EC trials of Regular, Blocks and Pivots experiments of the Non-Randomized group. There is no other significant difference for error signal peaks in the three experiments except of the second perturbation for Pivots experiment,.

Table 4.23 Eyes opened and Eyes Closed Comparison Statistical Analysis Summary for Error Signal Peaks of Eyes opened and Eyes Closed Trials of the First Experiment (Regular, Blocks and Pivots, Non-Randomized), (* indicates significance, α = 0.05)

Perturbation	Non- Randomized					
	Regular		Blocks		Pivots	
	Normality	Probability	Normality	Probability	Normality	Probability
1	Yes	0.126	Yes	0.166	Yes	0.404
2	Yes	0.060	Yes	0.525	No	0.028 *
3	Yes	0.688	No	0.600	Yes	0.961
4	No	0.138	Yes	0.681	No	0.249
5	No	0.893	Yes	0.382	Yes	0.707
6	Yes	0.174	No	0.345	Yes	0.933
7	Yes	0.110	No	0.917		

Figures 4.58-4.60 show the error signal peaks with standard error for all perturbations of eyes opened and eyes closed trials of the first, second, and third experiments. It is noticeable that despite that there is some difference between EO and EC trials, there is no pattern.

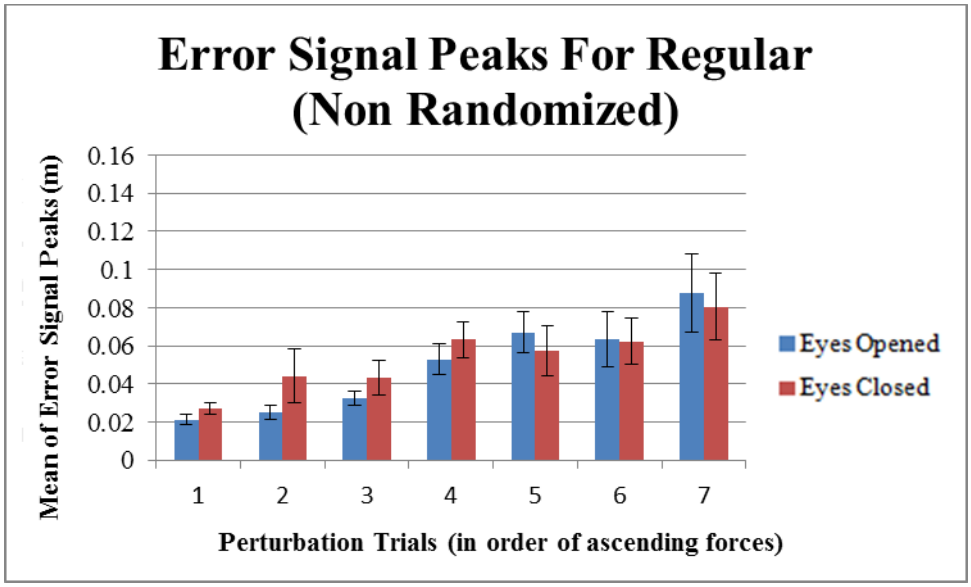

Figure 4.58 The means of the error signal peaks and standard error bars for eyes opened and eyes closed trials of the first experiment (Regular, Non-Randomized).

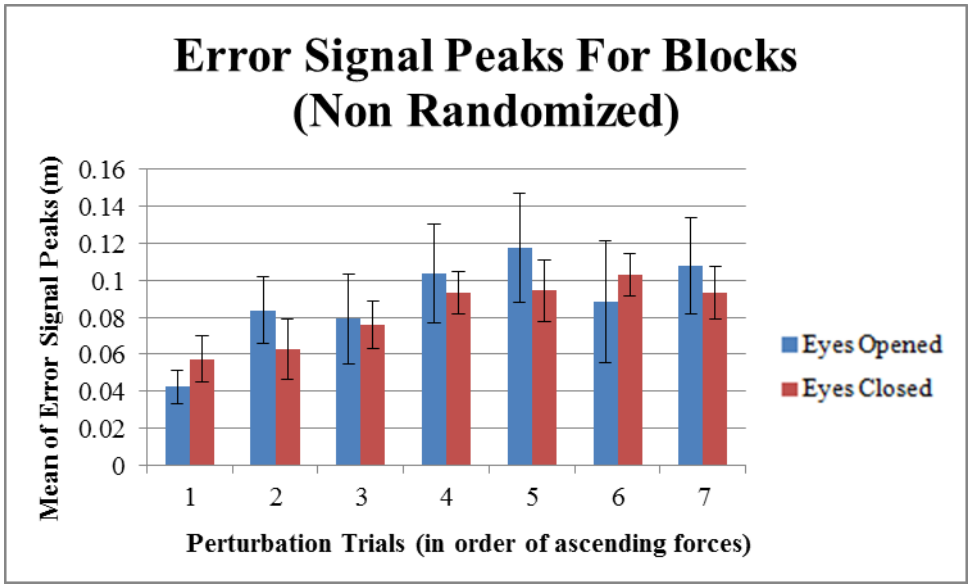

Figure 4.59 The means of the error signal peaks and standard error bars for eyes opened and eyes closed trials of the second experiment (Blocks, Non-Randomized).

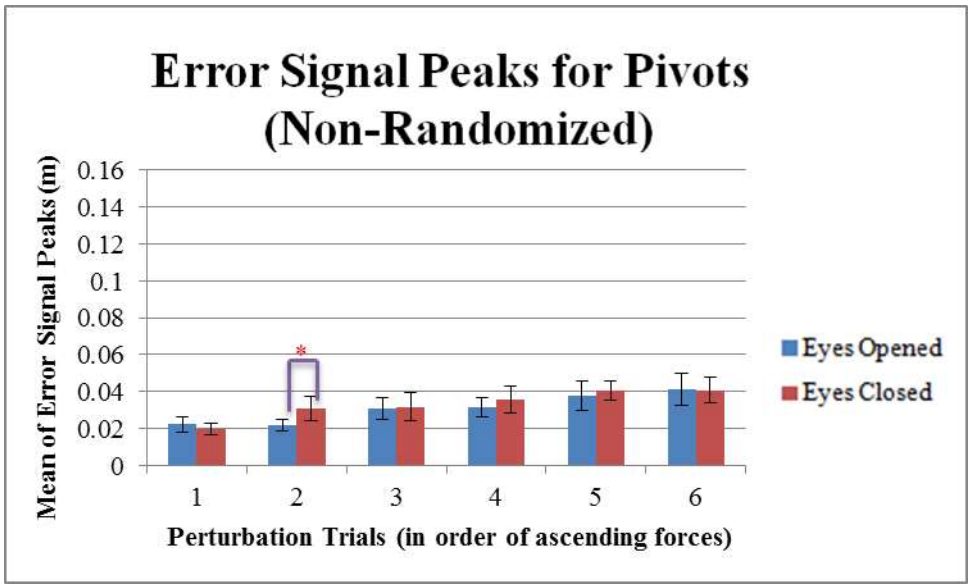

Figure 4.60 The means of the error signal peaks and standard error bars for eyes opened and eyes closed trials of the third experiment (Pivots, Non-Randomized), (* indicates significance, $\alpha = 0.05$).

6) Error signal peaks (Randomized, EO vs EC, Regular, Blocks, and Pivots).

Table 4.24 shows the statistical analysis results for the comparison of the error signal peaks between EO and EC trials of Regular, Blocks and Pivots experiments of the Randomized group. Figures 4.61-4.63 shows the error signal peaks for all perturbations of eyes opened and eyes closed trials of the first, second, and third experiments. It shows that there is no significant difference in error signal peaks between EO and EC of the three experiments.

Table 4.24 Eyes opened and Eyes Closed Comparison Statistical Analysis Summary for Error Signal Peaks of Eyes opened and Eyes Closed Trials of the First Experiment (Regular, Blocks and Pivots, Randomized)

Perturbation	Randomized					
	Regular		Blocks		Pivots	
	Normality	Probability	Normality	Probability	Normality	Probability
1	Yes	0.126	Yes	0.299	Yes	0.183
2	Yes	0.060	Yes	0.257	No	0.500
3	Yes	0.688	Yes	0.487	Yes	0.425
4	No	0.138	Yes	0.107	No	0.225
5	No	0.893	Yes	0.747	Yes	0.526
6	Yes	0.174	Yes	0.336	Yes	0.640
7	Yes	0.110	Yes	0.647		

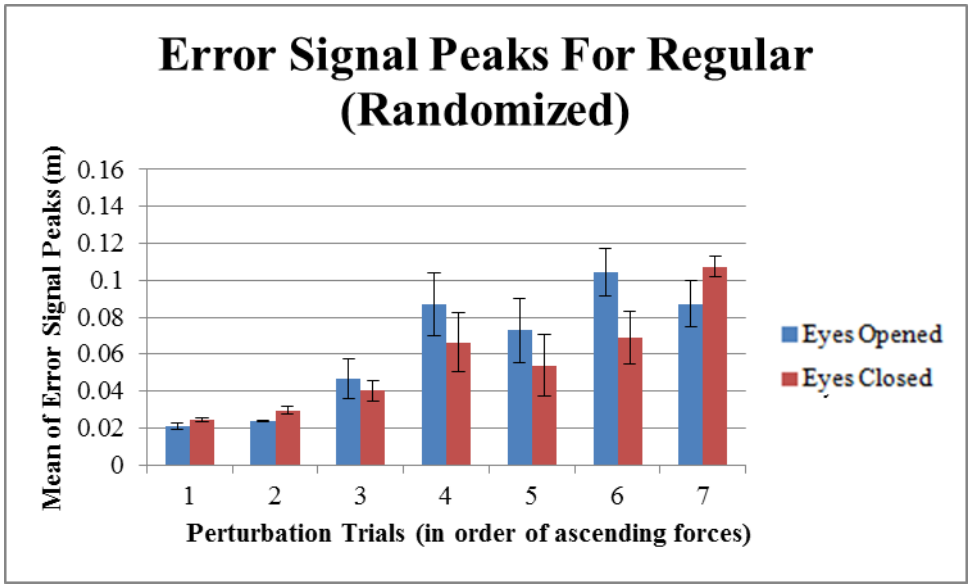

Figure 4.61 The means of the error signal peaks and standard error bars for eyes opened and eyes closed trials of the first experiment (Regular, Randomized).

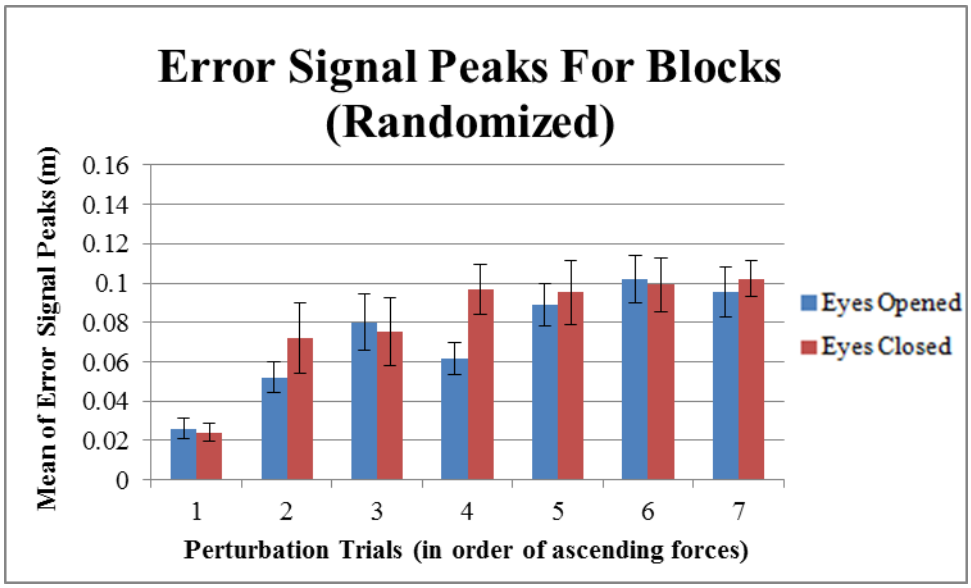

Figure 4.62 The means of the error signal peaks and standard error bars for eyes opened and eyes closed trials of the second experiment (Blocks, Randomized).

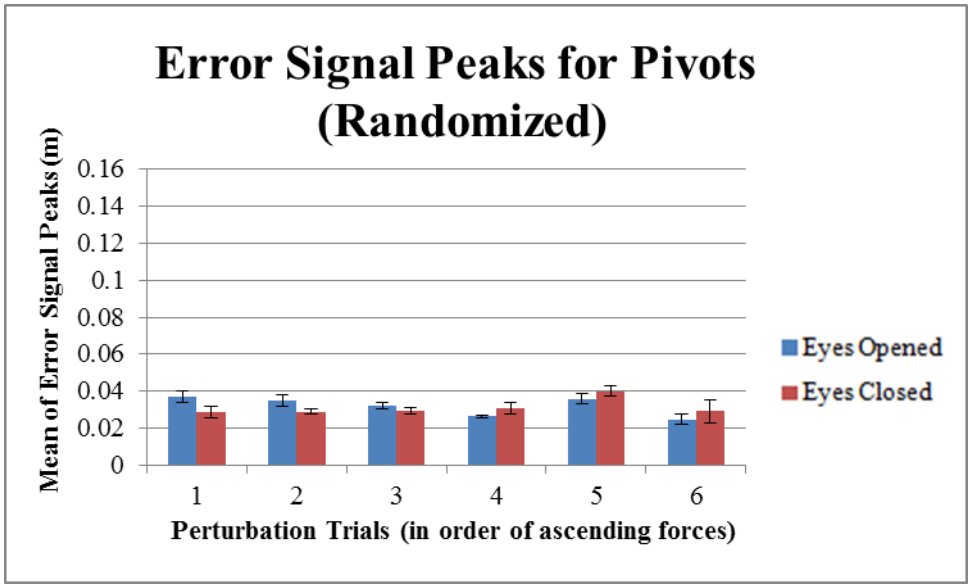

Figure 4.63 The means of the error signal peaks and standard error bars for eyes opened and eyes closed trials of the third experiment (Pivots, Randomized).

7) Correlation between COM and COP (Non-Randomized, EO vs EC, Regular, Blocks, and Pivots).

The last comparison is for the correlation between COM and COP between EO and EC trials. Table 4.25 below shows a summary of the statistical results which show that there is no significant difference in the correlation coefficients between EO and EC trials for the three experiments of the Non-Randomized group.

Table 4.25 Eyes opened and Eyes Closed Comparison Statistical Analysis Summary for the Correlation between COM And COP of Eyes opened and Eyes Closed Trials of the Three Experiments (Regular, Blocks, and Pivots, Non-Randomized)

Perturbation	Non- Randomized					
	Regular		Blocks		Pivots	
	Normality	Probability	Normality	Probability	Normality	Probability
1	Yes	0.520	No	0.1150	No	0.345
2	Yes	0.912	No	0.4620	Yes	0.135
3	No	0.917	No	0.5990	No	0.600
4	Yes	0.688	No	0.3440	No	0.463
5	Yes	0.628	No	0.7520	Yes	0.497
6	Yes	0.869	No	0.9160	Yes	0.847
7	Yes	0.793	No	0.3440		

Figures 4.64 to 4.66 show correlation coefficients for all perturbations of eyes opened and eyes closed trials of the first, second and third experiments: Regular, Blocks, and Pivots. Despite that there is no significant difference between them, the correlations coefficients for EC trials are smaller than EO trials, and this can be explained since subjects are less stable while there is no visual feedback.

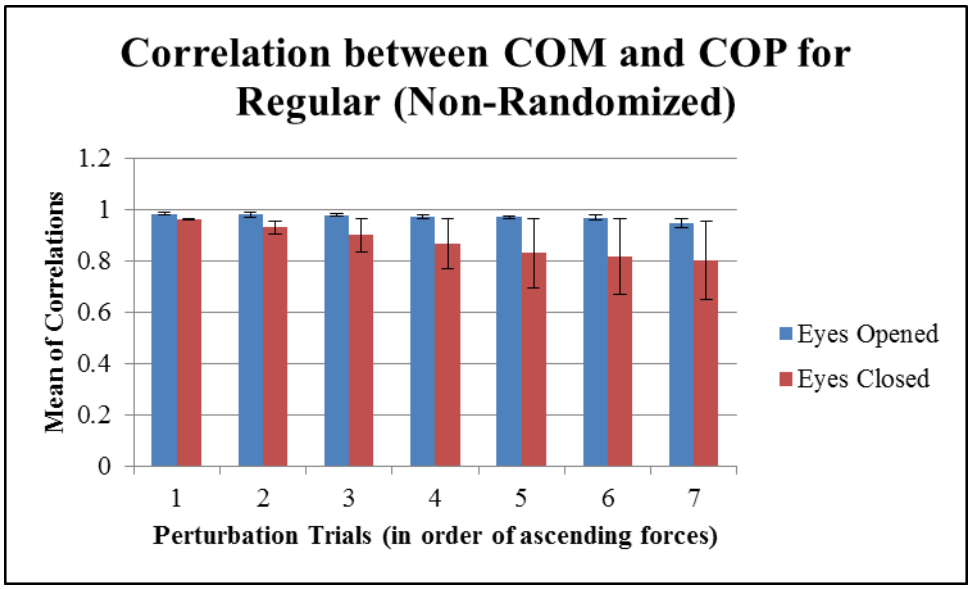

Figure 4.64 The means of the correlation between COM and COP in addition to the standard error bars for eyes opened and eyes closed trials of the first experiment (Regular, Non-Randomized).

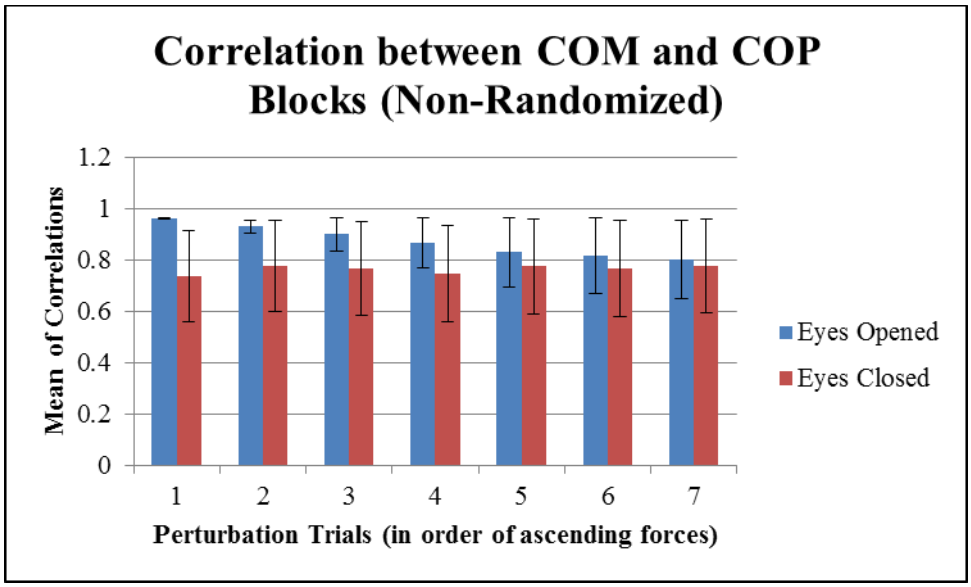

Figure 4.65 The means of the correlation between COM and COP in addition to the standard error bars for eyes opened and eyes closed trials of the second experiment (Blocks, Non-Randomized).

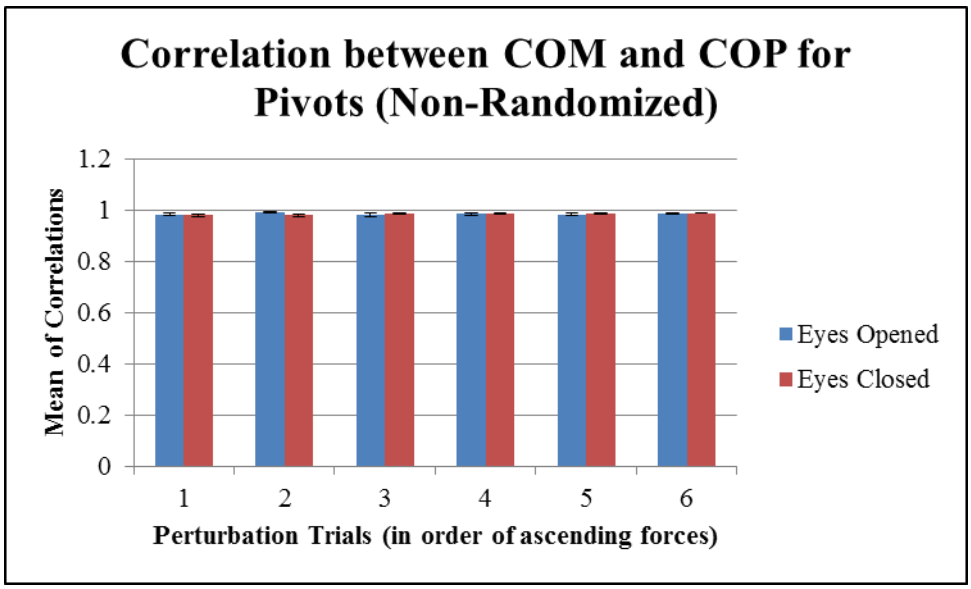

Figure 4.66 The means of the correlation between COM and COP in addition to the standard error bars for eyes opened and eyes closed trials of the third experiment (Pivots, Non-Randomized).

8) Correlation between COM and COP for each perturbation (Randomized, EO vs EC, Regular, Blocks, and Pivots).

Table 4.26 below shows a summary of the statistical results. There is no significant difference between them for any perturbations except the fourth perturbation of EO and EC trials, Randomized, Regular.

Table 4.26 Eyes opened and Eyes Closed Comparison Statistical Analysis Summary for the Correlation between COM and COP Of Eyes opened and Eyes Closed Trials of the Three Experiments (Regular, Blocks, and Pivots, Randomized), (* indicates significance, α = 0.05)

Perturbation	Randomized					
	Regular		Blocks		Pivots	
	Normality	Probability	Normality	Probability	Normality	Probability
1	Yes	0.462	No	0.522	Yes	0.748
2	No	0.463	No	0.491	Yes	0.106
3	Yes	0.144	No	0.138	Yes	0.511
4	No	0.028 *	No	0.512	Yes	0.336
5	Yes	0.235	No	0.257	Yes	0.658
6	No	0.600	No	0.225	Yes	0.219
7	Yes	0.345	No	0.721		

Figures 4.67 to 4.69 show the correlation coefficients of the correlation between COM and COP for all perturbations between eyes opened and eyes closed trials of the three experiments (Regular, Blocks, and Pivots). It is obvious that the correlation coefficients look similar to each other in the Figures.

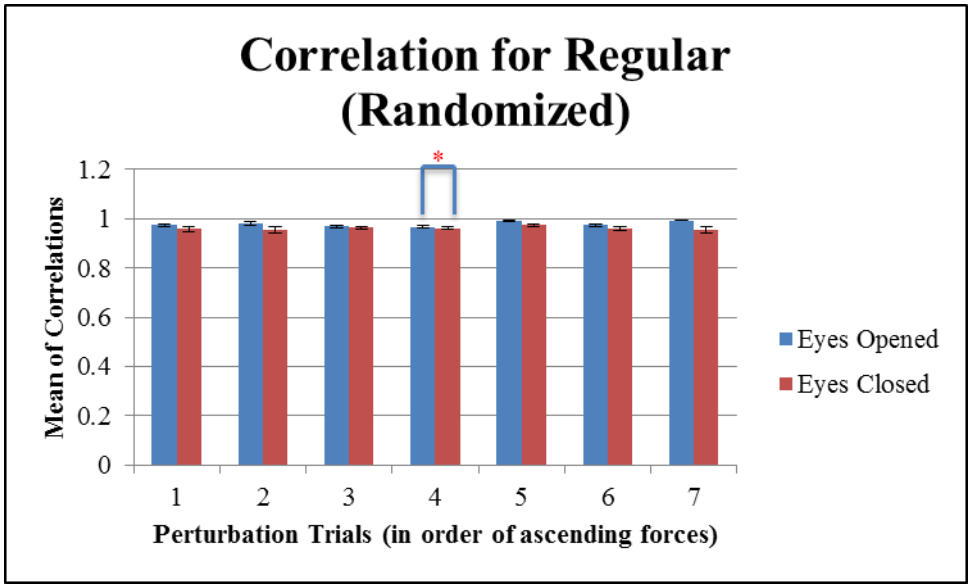

Figure 4.67 The means of the correlation between COM and COP in addition to the standard error bars for eyes opened and eyes closed trials of the first experiment (Regular, Randomized) , (* indicates significance, $\alpha = 0.05$).

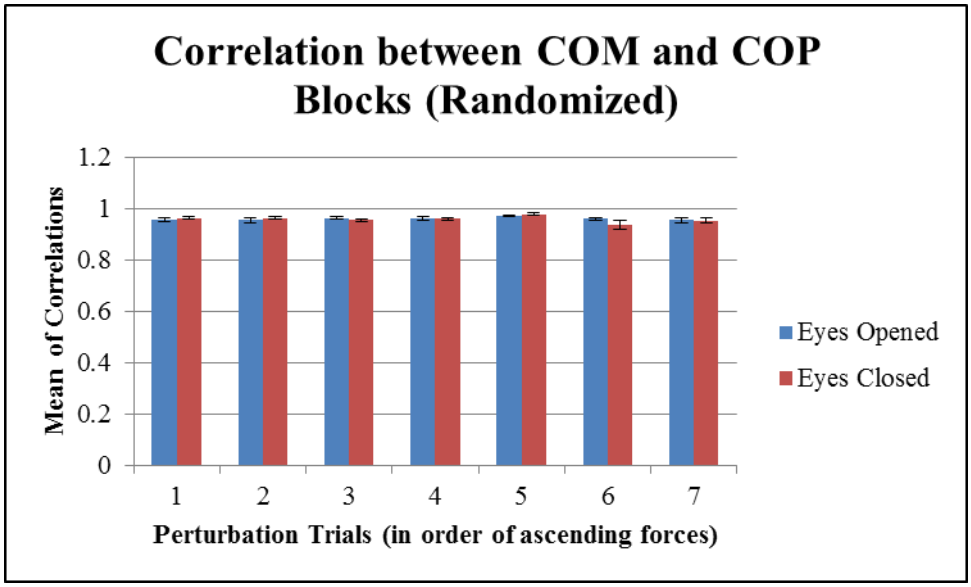

Figure 4.68 The means of the correlation between COM and COP in addition to the standard error bars for eyes opened and eyes closed trials of the second experiment (Blocks, Randomized).

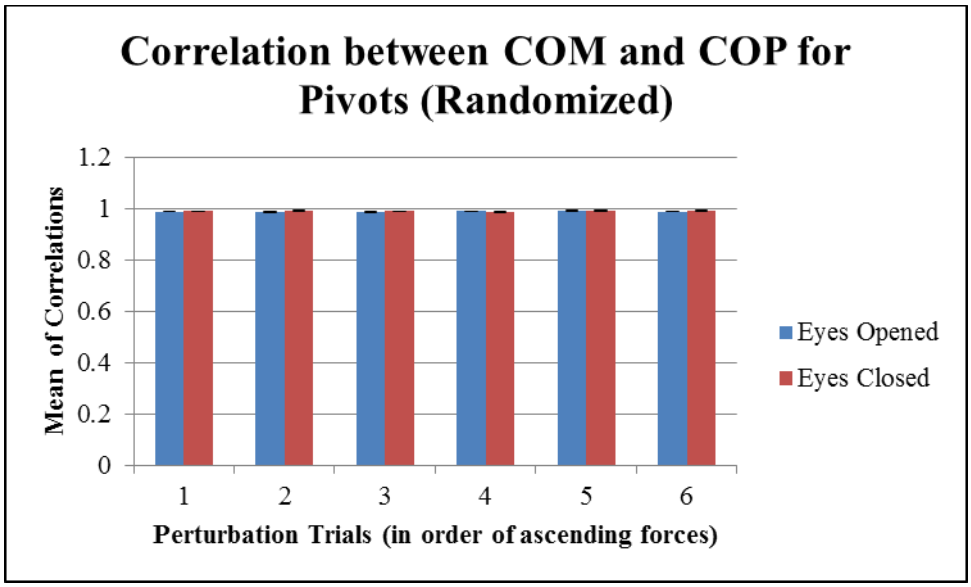

Figure 4.69 The means of the correlation between COM and COP in addition to the standard error bars for eyes opened and eyes closed trials of the third experiment (Pivots, Non-Randomized).

CHAPTER 5

Conclusions and Future Directions

4.1 Conclusions

The main conclusion of this study is that sensor-motor substitution using hand controlled trekking poles offers the potential to maintain balance in quiet standing and in response to small destabilizing perturbations. The statistical analysis of stability index, error signal peaks, and correlations comparing the Pivots experiment to Regular experiment support this conclusion. In addition, the high correlation coefficients between COM and COP in quiet standing on Pivots, and in the Pivots experiment with perturbations, and the high correlation coefficients of the correlation between COP and the trekking poles trajectories indicates that the trekking poles are working in a good way to control COP. We can say that using the trekking poles, though the response to perturbations does not match the biological response, it is proved to be effective in maintaining balance in quiet standing and perturbed quiet standing specially for small perturbations.

Conclusions for each Aim

Aim 1 Assembling the apparatus for a Perturbation/Motion capture system

The apparatus functioned as designed. The actuator moved forward in the sagittal plane with seven force-controllable perturbations, and the motion capture system was synchronized with the perturbation system. Several limitations were identified such as: while eight EMG channels were required to record EMG activity for Rectus Femoris and Biceps Femoris muscles of the thigh, the DAQ board had only four channels available for EMG. EMG for lateral Gastrocnemius and Tibialis Anterior were recorded. In addition, it was not possible to

record COP under each foot separately since only two force plates are available. One of them was used for the subject to stand on and the other was used to find the COP under the foot when the subjects stepped forward.

Aim 2 Studying normal human body response to perturbations of different forces (Regular), in A/P plane.

- The apparatus and custom developed COM computations performance is excellent since they generate COM and COP trajectories that matches expectations and are consistent with the literature.

- An important drawback of this experiment is the noise produced by the actuator while moving, which meant that subjects are expecting a perturbation, this makes the reaction to them less unexpected.

Aim 3 Investigation of the human capability to adjust to reduction in COP range with and without visual feedback, while perturbed with different forces in A/P plane, using the shoes with small Blocks.

- Total forward displacement and number of steps are larger for Blocks vs Regular.

- Error signal peaks and SI of Blocks are larger compared to Regular.

- The correlation coefficients of the correlation between COM and COP for Blocks are the lowest.

Keeping in mind that the stability index is defined as the destabilizing torques over the stabilizing torques, we can say that larger stability index indicates less stability. The error signal is the difference between COM and COP, and since the stability depends on how close and fast COP follows COM, we can say that the larger error signal indicates less stability. It can be concluded that standing on the Blocks, which reduces the boundary of the BOS leads to reduction in stability. However, an important finding is that humans can quickly

accommodate to a reduction of the BOS and adapt their ankle/hip and stepping strategies to maintain balance (while less stable) with little to no training.

Aim 4 Investigation of the human capability to use hands with trekking poles to accommodate for the confined COP range with and without visual feedback, while perturbed with different forces in A/P plane.

- Perhaps the most important observation in this study is that subjects were able to maintain balance using the pivot shoes and trekking poles. When using the Regular and block shoes, subjects employed their long-practiced ankle, hip and stepping strategies. Since the ankle and hip strategies were rendered useless, since the pivot prevented the ankle torque from altering the COP, subjects were able to maintain quite standing balance substituting their hands and arms as actuators of COP movement. It is impressive that with essentially no practice, subjects could make a motor substitution that was very effective.

- Error signal peaks and SI of Pivots are very close to Regular especially for the two first perturbations.

- The correlation coefficients of the correlation between COM and COP for Pivots are larger than for both Regular and Blocks. This is a good indication that COP follows COM smoothly. A possible explanation for this is that the spring and damping coefficients of the Pivots and the trekking poles are different from ankle joints.

All of this indicates that using trekking poles to maintain stability is a good approach.

- A drawback for Pivots experiment is that subjects did not use stepping strategy at all for any perturbation. This means that although using trekking poles is a good approach to replace ankle and possibly hip strategies, especially for small perturbing forces, it may not be good for replacing the stepping strategy. There are many possible reasons to explain this:

    - Legs muscles are already performing a motor task (to balance on Pivots), it was recognized that gastrocnemius muscle was active all of the time, and so they cannot perform another motor task (stepping strategy) at the same time.

    - The subject did not have enough practice to use pivot shoes and trekking poles to learn stepping. Further training could give better results.

- The weight of the platforms with Pivots, trekking poles, and shoes could impede stepping.

- There is no sensory feedback from the proprioception system in the foot since the only contact point between the foot and the base on the floor except under Pivots, thus the proprioception system can sense the ground reaction forces and the COP only at that point, and so there is less than required or even wrong incoming feedback to the balance system to turn on stepping strategy.

- Although it was demonstrated that trekking poles are a good approach to control balance, since all the objective parameters using hands/arms strategy (trekking poles) are very close to those of ankle/hip strategies (Regular experiment) Figures 4.1 and 4.2 show that processes are not entirely equivalent. In Figure 4.1, the movements of the COM and COP increase in the anterior direction in response to increasing perturbation force. In Figure 4.2, the excursion of the COM and COP approach the anterior boundary of the BOS in nearly equally in all following perturbations, regardless of the force magnitude. One can conclude that at least in these unpracticed experiments, the hand arm method affects the COM and COP movements differently. This could be a result of the fact that the spring and damping factors of Pivots and trekking poles are different than ankle joints, or since the use of the trekking poles that are attached to the outer frontal edge of each force plate make it a quadrupedal process which affect the reaction and probably resists the stepping strategy.

  Another possibility is that the hand/arm interaction plays a more complex role beyond controlling the COP. From the literature, the ankle strategy is thought to be primarily related to the COP control, while the hip strategy allows the individual to exercise some control over the anterior movement to the COP. The pivot shoes remove not only the ability to generate ankle torques, but also severely limit the subjects' ability to control hip torques. It may be that the arm/hand control of the trekking poles allows the user to provide torques that replaces the ankle strategy and to move the COM in the posterior direction. This needs to be studied further.

- To my knowledge the findings corresponding to the second (Blocks) and third (Pivots) experiments are entirely new and have not been represented in the literature.

Non-Randomized vs Randomized Perturbations

There is no significant difference in total forward displacement, number of steps, SI, error signal peaks between Non-Randomized and Randomized Regular experiments. However, the stability index is larger for non-randomized than randomized, and since randomized group

subjects do not have any idea about the magnitude of the coming perturbation force, they are less stable.

Eyes opened vs eyes closed

There is no statistically significant difference in all parameters for most perturbations between eyes opened and eyes closed trials in all the three experiments (Regular, Blocks, Pivots) This can be explained depending on the role of the visual system in balance which is locomotion planning and avoiding obstacles, and since all of the experiments include perturbation for subjects in quiet standing, the response does not need planning and so does not need a visual feedback to take place. This indicates that the visual feedback is not critical in quiet standing balance. However, it was observed that many subjects opened their eyes as they were perturbed, or did not wait till their response is complete, so they used their visual feedback to step not only to get back to the original position as they were instructed.

General Conclusions

There is a wide range of differences in the way human body retains balance and responds to perturbations. These differences are between subjects and within the same subject. Many factors can affect the balance system response including differences in human body's height and weight, anthropometrics, balance system efficiency, neuron's elasticity and plasticity, Muscle and ligament properties (including stiffness and damping properties) that is based on a person being relaxed or tensed (alert, anticipating, happy, nervous, tiered, sleepy,…).

An example of between subjects differences are subjects that responded in a very different way in the second experiment (Blocks) from the rest of the subjects and from each other. The first one found out that she can tilt her feet forward, and tip on her toes to extend the BOS to balance instead of using stepping strategy despite that the Blocks were high enough for the

rest of the subjects to make them step. The second subject compensated in a very different way to perturbations while wearing the shoes with Blocks, he tilted his upper body (trunk, head, and arms) to the back in a way to adjust COM to keep it within the BOS instead of letting it to go forward and making the COP follow it by stepping.

5.2 Future Directions

- An extended research for the efficiency of using trekking poles to balance by means of stepping strategy with large perturbations. Using larger perturbation forces drive the COM in A/P plane to exceed platforms front edges (BOS) to force subjects to step, this is to be done after dealing with possible reasons that limit the use of stepping strategy:

  - ❖ Finding another way different from using Pivots to make ankle strategy useless to maintain balance. A suggestion is using a very firm ankle brace. This way legs muscles will not be busy with another motor task (balancing over the Pivots) and see if this can enhance using stepping strategy while using the trekking poles to balance.

  - ❖ Give subjects more practice standing and walking with Pivots and trekking poles, further training could give better results.

  - ❖ Rebuilding Pivots-trekking poles system with a lighter material making it easier to step.

  - ❖ Performing training sessions before running the experiment and checking if this improves performance.

- Studying static balance using trekking poles in the M/L plane.

- Studying dynamic balance using trekking poles in both A/P and M/L planes. Keeping in mind that during the stance (single or double) the trekking pole can be used to adjust the COP under the stance foot for balance.

- Since there is no sensory feedback from the proprioception system in the foot since the only contact point between the foot and the base on the floor except under Pivots, and since paraplegic people do not have a proprioceptive feedback too, an alternative is needed. An alternative feedback approach is under research in our laboratory that is a vibrotactile hand actuators that gives feedback to the hands instead of the impaired proprioceptive system of the feet to be used for TREKKER.

- Another under research project is to build 2 DOF active ankle joint to ensure a perfect ankle and stepping strategy in both A/P and M/L directions for TREKKER exoskeleton.

- Studying the effect of cognitive tasks on the efficiency of using the trekking poles to maintain balance.

- Adding force sensors on the trekking poles to study the interaction between the human body and them in an extensive way, to find torques and study the components of forces that participate in the balance control, and to understand how subjects interact with them (do they push down more or push forward more?).

- Using more force plates or pressure sensors under each foot to get COP under each foot and to be able to build a free body diagram and find the net hip, knee and ankle torques.

- Adding force sensors to the trekking poles to measure the torque applied to the poles, which results from tangential forces applied to the poles and the force applied downward on the poles. Such downward force would be opposite user force necessary to lift the foot as part of a stepping strategy.

- Study of sensory substitution of COP information normally available via the mechanoreceptors on the soles of the feet by substituting vibrotactile feedback to the hands.

- A further study of this would involve the use of Blocks of varying lengths, thus allowing one to examine whether or not the reduction in balance stability is inversely proportional to the distance between the block and Regular anterior BOS boundary.

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