POSTURAL RESPONSES OF PATIENTS WITH BILATERAL VESTIBULAR LOSS AND HEALTHY SUBJECTS TO SINUSOIDAL TILTS

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submitted by **MEHMET İMİR** in partial fulfillment of the requirements for the degree of **Master of Science in Engineering Sciences Department, Middle East Technical University** by,

Prof. Dr. Gülbin Dural Ünver
Dean, Graduate School of **Natural and Applied Sciences** ____________

Prof. Dr. Murat Dicleli
Head of Department, **Engineering Sciences** ____________

Assoc. Prof. Dr. Senih Gürses
Supervisor, **Engineering Sciences Department, METU** ____________

Examining Committee Members:

Prof. Dr. Hakan I. Tarman
Mechanical Engineering Department, METU ____________

Assoc. Prof. Dr. Senih Gürses
Engineering Sciences Department, METU ____________

Prof. Dr. Bülent Satar
Otorhinolaryngology Department, University of Health Sciences ____________

Assist. Prof. Dr. Ali Emre Turgut
Mechanical Engineering Department, METU ____________

Assist. Prof. Dr. Kutluk Bilge Arıkan
Mechatronics Engineering Department, Atılım University ____________

Date: ____________

I hereby declare that all information in this document has been obtained and presented in accordance with academic rules and ethical conduct. I also declare that, as required by these rules and conduct, I have fully cited and referenced all material and results that are not original to this work.

Name, Last Name: MEHMET İMİR

Signature :

ABSTRACT

POSTURAL RESPONSES OF PATIENTS WITH BILATERAL VESTIBULAR LOSS AND HEALTHY SUBJECTS TO SINUSOIDAL TILTS

İMİR, MEHMET
M.S., Department of Engineering Sciences
Supervisor : Assoc. Prof. Dr. Senih Gürses

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Posture control to maintain the stability of upright posture is a very complex task. It requires sensorimotor integration of all sense organs. If one of these organs loses its functionality, the person may have difficulties in maintaining postural balance. This study examines the difference in postural responses of patients with bilateral vestibular loss and healthy subjects to sinusoidal tilts. It has shown that center of mass(CoM) motions of control and patient groups were similar respect to space coordinates but different respect to platform coordinates at low frequency (f=0.05 Hz). In contrast, their both CoM motions became more distinct at high frequency (f=0.17 Hz). It is argued that vestibular loss can be compensated by other available sensory information at low frequency. However, this compensation started to inadequate for maintaining postural balance at high frequency especially in the absence of visual information. In addition, heterogeneous response characteristics of patients in this study suggest that ability of patients to compensate their vestibular sensory loss differ across patients.

Keywords: Postural Control, Balance, Vestibular loss, Vision, Center of Mass, Movable Support Surface

ÖZ

BİLATERAL VESTİBÜLER KAYBI OLAN HASTALARIN VE SAĞLIKLI KİŞİLERİN SİNÜZOİDAL SALLANMALARA VERDİKLERİ CEVAPLAR

İMİR, MEHMET
Yüksek Lisans, Mühendislik Bilimleri Bölümü
Tez Yöneticisi : Doç. Dr. Senih Gürses

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İnsanın dik duruşunu kontrol etmesi karışık bir eylemdir. Bir çok duyu organının birlikte çalışması insanın dik duruşunu dengede tutması için gereklidir. Bu duyulardan herhangi birinin yokluğu, insanın dik duruşunu dengede tutamamasına neden olabilir. Bu çalışmada vestibüler duyusunu kaybetmiş hastaların ve sağlıklı kişilerin sinüzoidal sallanmalara verdikleri cevaplar arasındaki farklar incelenmektedir. Hastalar ve sağlıklı kişiler düşük frekanstaki (f=0.05 Hz) sinüzoidal sallanmalara benzer cevaplar vermektedir. Buna karşılık, bu grupların yüksek frekanstaki (f=0.17 Hz) sinüzoidal sallanmalara verdikleri cevaplar arasında farklılıklar bulunmaktadır. Vestibüler duyunun yokluğu düşük frekansta mevcut olan diğer duyularla telafi edilebilmektedir. Fakat diğer duyular yüksek frekansta vestibüler duyunun yokluğunu telafi etmekte yetersiz kalmaktadırlar. Ayrıca, hasta grubunun bu çalışmada sinüzoidal sallanmalara verdikleri cevapların birbirinden ayrışık olması, vestibüler duyunun yokluğunu telafi etme kabiliyetlerinin farklı olduğunu göstermektedir.

Anahtar Kelimeler: İnsan Dik Duruşu, Denge, Kütle Merkezi, Vestibüler Duyu Kaybı, Hareketli Yüzey, Görme Duyusu

To My Parents

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LIST OF ABBREVIATIONS

CoM	Center of Mass
CNS	Central Nervous System
EO	Eyes Open
EC	Eyes Closed
HAT	Head, Arms, and Trunk
ϕ	Segment Orientation of Platform
θ_1	Segment Orientation of Lower Leg
θ_2	Segment Orientation of Upper Leg
θ_3	Segment Orientation of HAT
θ_{CoM}	Absolute Angular Displacement of Center of Mass
$\theta_{CoM,rel}$	Relative Angular Displacement of Center of Mass
FFT	Fast Fourier Transform
RMS	Root Mean Square

CHAPTER 1

INTRODUCTION

It is an important task for humans to control postural balance during several postural activities and positions in their daily life such as standing, sitting, and walking. Postural control is defined as the skill to maintain, achieve or restore a state of balance during any postural position or activity. The postural control system has two principal objectives: postural orientation and postural stability. Postural orientation can be defined as fixing the orientation and position of the body segments that serve as a reference frame for perception and action in space. Postural stability is the ability to maintain postural balance against gravity and keep posture orientation in optimum [14, 22].

The postural orientation and stability of human posture is not a simple skill rather it requires very sophisticated skills based on the interaction of dynamic sensory-motor processes. Because of this complexity, researchers tried to divide postural control system into many sub-components. Horak et al. (2006) described these sub-components: biomechanical constraints, movement strategies, sensory strategies, orientation in space, control of dynamics and cognitive processing [15]. On the other hand, some researchers have divided postural control four primary functions. These are an internal representation of the body orientation and its stability limits, integration of multisensory inputs for orientation and stabilization of body segments and flexible postural reactions or anticipations for balance recovery after disturbance, and postural stabilization during voluntary movements [22]. In addition to these descriptive levels of postural control, some researchers also analyze postural control through ecologically-based perspective [36]. The complexity of human postural control causes no clear agreement on about what is needed to stabilize human postural control. The essential requirements for postural control of human body could be described as below.

First, biomechanical constraints or stability margins of the body center of mass (CoM) are one of the most important parameters for postural control. The stability requirements for body CoM is not a single point but rather a region with limits according to biomechanical constraints of individuals. A person should keep to body CoM position within the stability limits with respect to its base of support to maintain body stability against to gravity [20, 25, 32, 34]. However, the position of CoM is not the only parameter to guarantee stability of the standing posture will be sustained. It has been argued that horizontal velocity of the CoM should also be considered in the stability of posture [33]. Although, it is a requirement to maintain postural balance while standing, ability of getting out of range of stability (creating an unstable equilibrium point in standing) is the subject of one of the most famous theories of voluntary movement; i.e., equilibrium point theory: to perform a movement requires shifting the initial stability margin of CoM to another position [6, 7]. Besides, the central nervous system (CNS) constructs an internal map of stability boundaries of the body, which has a major role in maintaining postural balance [25]. Several factors affect the size of stability limits and its internal representation. It has been shown that aging causes reduced stability margin and it has an effect on the postural instability of elderly people [37]. Also, central processing abnormalities within the basal ganglia like Parkinson's disease lead to the inaccurate internal representation of stability limits and postural instability [41].

Second, the different types of movement strategies can be used to maintain to postural balance. To maintain postural balance to small perturbations, people sway as a flexible single degree of inverted pendulum about their ankles with very low hip or knee motion [9]. This movement strategy is called the ankle strategy. However, when people are exposed to large perturbation, they behave like a two degrees of inverted pendulum, keeping ankle rotation small and primarily swaying about hip joints, which is called the hip strategy [21]. In addition to these strategies, people use the stepping strategy for transition from standing to walking [14].

Third, cognitive skills could have a major role in postural control. The stabilization of upright posture has usually been treated as an automatic task which requires minimal attention. This statement may be true for a well trained healthy individual performing a relatively easy task. However, there may very well be significant focus requirements for postural control, depending on the postural task, the age and the balance abilities of the individual [46]. Besides, anticipatory postural adjustments, the voluntary adjustments of the body to return its optimum posture orientation, can be treated as a cognitive skill in postural control [3].

Fourth, Stoffregen and Riccio emphasized that postural control relates not only to information about the motion of the human body relative to the local force environment but also on information about the environment such as surface of the support and about the corrective actions of the person [39]. The person's relation to the surface of the support and its characteristics play a significant role in determining the actions to maintain postural stability. The interaction of person with the surface of support have also effect on limits of stability and exploration of stability limits is essential for postural balance. Traditionally, the movement variability has been considered as postural instability. However, the movement variability can play an important role in detection and exploration of postural stability limits [40].

Fifth, the representation of body orientation in space involves an integration of different sensory information [26]. The past studies have shown that information from visual[2], somatosensory[29], and vestibular[5] system contribute to postural control. Also, proprioception, the sensation of joint motion and acceleration, is another important sensory feedback mechanisms for motor control and posture [18]. However, it is important to distinguish proprioception as an interoceptive sense from other exteroceptive senses (visual, somatosensory and vestibular). The integration of different sensory information is one of the most valuable resources for CNS to maintain postural balance. The absence of any sensory information refers that the nervous system has less information to precisely estimate CoM position or velocity. Therefore, estimation of body orientation in space become less accurate and can cause postural balance problems [17]. However, the absence of some sensory information may be compensated by re-weighting other available sensory information based on a closed loop control of postural stability [35].

All skills mentioned above are in complex interaction and absence of any skill or skills may cause instability in postural activities. Since it is the aim of this study to examine the effect of loss of vestibular sensory information on postural control, it will be described in detail. The sensory information from the vestibular system allows healthy subjects to reference and estimate the movement of their own bodies in space with a higher accuracy than can patients with bilateral vestibular loss [12]. However, one of the most dramatic effects of vestibular loss on posture is that, when patients with vestibular loss have been exposed to the variable support surface or visual environments, they are unable to suppress the influence of visual and proprioceptive inputs correctly [31]. Therefore, vestibular loss not only causes lack of information about the movement of the self (body), but also affects the reliability of other sensory systems. However, vestibular system originated information loses its importance relatively when movement of the support surface is small. The past studies have shown that vision, and vestibular information have little effect on balancing of human body posture in space at low frequency tilt of support surface, and patients with bilateral vestibular loss can successfully maintain their postural stability even the absence of visual information. However, at the higher frequencies patients tended to actively align their bodies respect to the tilting platform unlike to healthy subjects who align their bodies respect to gravity vertical when their eyes open [24]. The absence of vision at higher frequencies lead to that patients show abnormal large body sways [28].

The study of postural control requires a metric to identify the response of subjects. The center of mass (CoM) is a passive variable which represents total mass of the body as a single point in space. A lot of literature suggest that CoM is the primary stabilized variable for posture and movement coordination during standing and voluntary movements [13, 23, 34, 38, 43]. Therefore, CoM can be used as a metric to identify the postural balance of subjects.

An appropriate mathematical model is necessary to describe dynamics of the human body. The properties of the support surface, postural strategies, and biomechanical constraints of the human body should be taken into consideration for more accurate modeling. In quiet standing, the muscles at the ankle joint play a major role in controlling body sway in the sagittal plane. Therefore, most of the studies in postural control were based on one degree inverted pendulum model in quiet standing(fixed support) [17, 35, 45]. However, for moving support surfaces like tilting experiments, one degree of freedom inverted pendulum model is not sufficient to fully describe dynamics of the human body. The muscles at the hip and knee joints also contribute to postural balance in moving support surface [1, 32]. Thus, it is important to construct a mathematical model of the human body as a three degrees of inverted pendulum (ankle, knee, and hip) in the sinusoidal tilt experiments.

1.1 Vestibular System and Bilateral Vestibular Loss

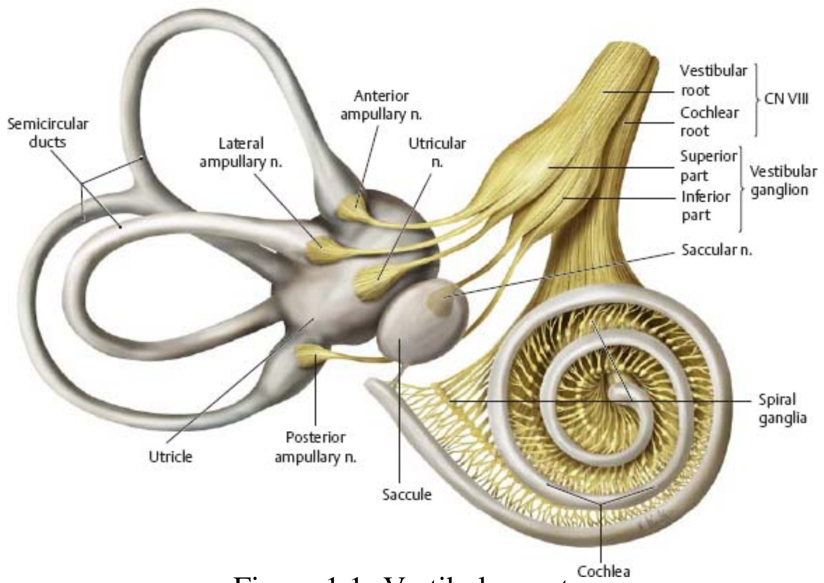


Figure 1.1: Vestibular system

The vestibular system is made up of two inner ears that have nerve connections to the brain and the eyes. The vestibular system contains three semicircular canals, the utricle, and the saccule (see Fig.1.1). Each canal contains hair cells, which are specialized epithelial cells. Angular or linear acceleration of the head leads to deflections of the hair cells in these channels. The inner hair cells transform these deflections to sound vibrations in the fluids of the cochlea and then into electrical signals. Then, these electrical signals are transmitted to CNS by nerve connections. The utricle and saccule sense linear acceleration or head tilt whereas three semicircular canals sense angular acceleration of head [10].

The vestibular system is in interaction with vision and the somatosensory to give information about postural orientation and help to maintain postural stability. If both inner ears/vestibular nerves are damaged, as determined through vestibular testing, the brain's sensory information available for guiding movement has decreased. Without any information from the inner ears, the brain becomes more dependent on sensation from other sources, such as vision and somatosensory. This loss of inner ear input can cause imbalance while walking and performing everyday tasks.

Bilateral vestibular loss implies damage to the inner ears on both sides. The possible causes for vestibular loss include ototoxic medication in childhood, Paget's disease, meningitis, bilateral Meniere's disease, congenital abnormalities, bilateral acoustic neuroma, syphilis or autoimmune inner ear disease [8]. The symptoms of bilateral vestibular loss are postural instability, difficulty in walking, visual illusion because of head movement and unsteadiness in the dark or with eyes closed.

1.2 Hypotheses

The postural control of the human body is a very complex task, and it requires the complex interaction of many skills as mentioned above. One way to analyse this complex structure of postural control is the separation of some important resources for postural control. Therefore, to analyse the effect of loss of vestibular resources on postural stability, we aim to compare postural responses of healthy subjects and patients with bilateral vestibular loss to sinusoidal tilts. The vestibular system gives information about angular and/or linear acceleration of the body and its sensitivity is poor at low frequency body perturbations [42]. Based on that, experiments have been done at low and high frequency sinusoidal tilts. The lack of any sensory input leads to sensory re-weighting of available sensory information so to examine this phenomena absence of vision has also been considered in this study. Therefore, experiments have been done at two different frequencies (f=0.05 and f=0.17 Hz), and eyes conditions (eyes open and eyes closed).

We hypothesized that the responses of healthy subjects and patients should be similar at low frequency sinusoidal tilts whether their eyes are open or closed, since vestibular system and vision has a small effect on postural control at low frequency. Furthermore, we claim that the response of these two groups should become distinct at high frequency sinusoidal tilts and this distinction should increase in the absence of vision. Finally, we try to obtain clues about whether movement variability of patients which are irrelevant to sinusoidal tilts is an incidence of instability or an exploratory movement.

CHAPTER 2

EXPERIMENTAL EQUIPMENTS

2.1 Experimental Equipments

The experimental equipments: custom made tilt platform and motion capture system will be introduced in this section. Note that, force plate sensor has also been used as a data acquisition device in this experiment, but its data have not been presented in this study.

2.1.1 Custom Made Tilt Platform

Tilt experiments are done by using a 2-dof tilting human balance testing machine (DETES), which can perform sinusoidal (antero-posterior (A-P) and medio-lateral (ML)) tilts. The tilt platform can follow sine wave in the range of 0.05 – 2 Hz of frequency and 1 - 10 degrees of the peak amplitude. The moment arm of the tilting platform is 34.5 cm below the surface of tilting platform. In other words, there is a 34.5 cm distance between the rotation axis of the tilt platform and rotation axis of the ankle. Thus, there are not only pitch and roll rotations of tilting platform but also the trajectory motion of platform which is induced by the moment arm. The human balance testing system also has a 1-dof tilting cabinet whose axis of rotation passes through the ankle rotation axis. This tilting cabinet has the same frequency response characteristics with the tilting platform.

There are three AC Servo-motors, two of them drive the tilting platform, and one of them drive the tilting cabinet. AC Servo-motors were Allen-Bradley © OEMax

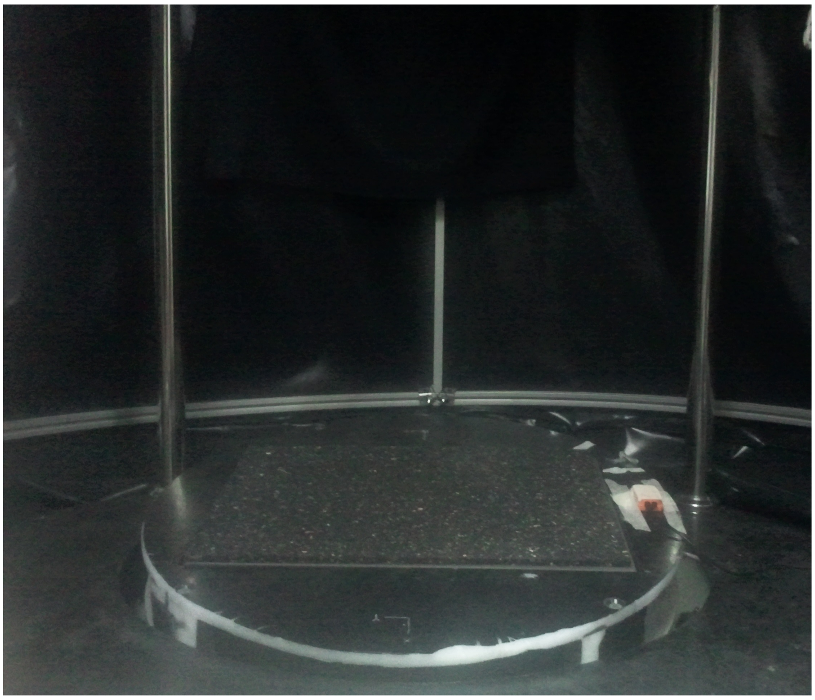


Figure 2.1: A photo of 2-dof tilting human balance testing machine (DETES)

(RD15-A) and the driver CSD3 with the specifications 1.5 kW maxpower, 4.77 Nm max torque capacity, 3000 revolutions/minute and the quadrature encoder inside with the characteristics of 2500 pulse/revolution. A reducer of x80 (reducing the angular velocity by increasing the torque) has been used in-front-of the actuators. An A/D (NI cRIO© 9073) card is implemented to the system for controlling the perturbation platform and collecting data. For further information about tilting human balance testing machine (DETES) see [11].

2.1.2 Motion Capture System

The Xsens MVN inertial motion capture system is a wearable system for tracking 3D motion of full-body human movement. The Xsens MVN inertial motion capture system is shown in Fig.2.2. The MVN consists of Xsens' state of the art miniature inertial sensors which are called motion trackers. There are three different types of motion trackers; 3D linear accelerometers, 3D rate gyroscopes and 3D magnetometers. These motion trackers measure local acceleration, angular velocity and the magnetic field vector with respect to the global coordinate system. Sensor position and orientation are obtained by integrating the angular rate data and double integrating

Figure 2.2: Motion capture system

the local acceleration data in time. Global coordinate frame is defined as follows;

Global coordinate frame:

- X positive when pointing to the local magnetic North.
- Y according to right handed coordinates (West).
- Z positive when pointing up.

Local coordinate frame of each segment is defined on the proximal end of the related segment :

- Origin: center of rotation (proximal)
- X forward.
- Y up, from distal end to proximal end.
- Z pointing right.

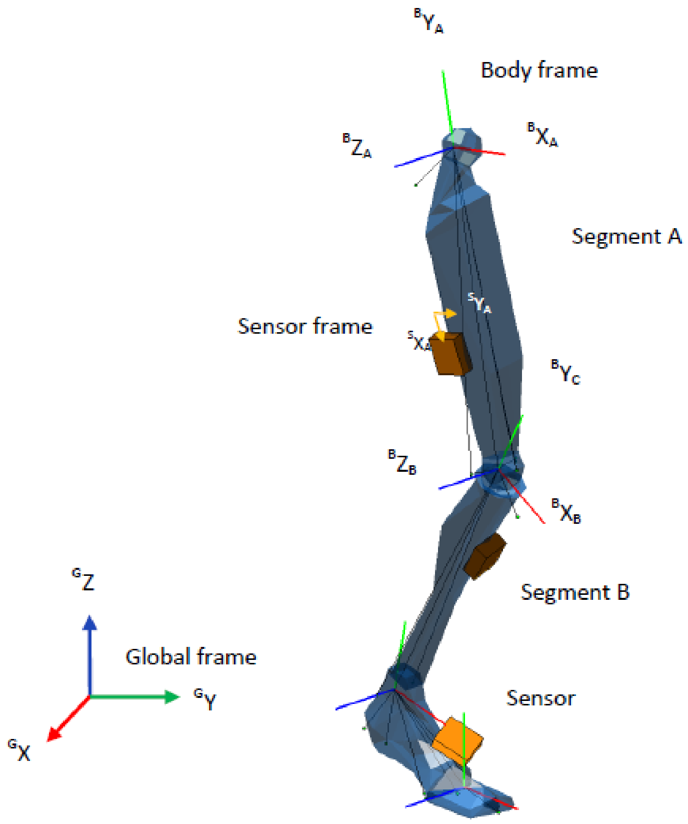


Figure 2.3: Global and local coordinate frames

The motion trackers data are converted to motion of human body movement by a well-defined 3D biomechanical model of human body. This biomechanical model is based on the assumption that two segments are, on average, coupled by a joint. If two segments are said to share a joint, there exists a point on each of the two segments that has zero average displacement with respect to each other and the location of this point is the joint position. It is necessary to input body height and foot length (shoe length at the time of the measurement) to calculate other segment lengths (based on an anthropometric model). The location of motion trackers on the human body are defined by MVN's bio-mechanical model. To obtain accurate results, motion trackers should be placed at these certain locations properly and needs a proper calibration to be performed. After the proper calibration, local coordinate axes of segments and outer casing of the related sensors are aligned. When sensor data are updated in over time, MVN uses this biomechanical model to convert motion trackers data to segment orientation data by using quaternions.

The rotation from sensor to the body segment orientation is given by quaternion ${}^{BS}q$, matching the orientation of the sensor in the global frame ${}^{GS}q$, body segment orientation with respect to global coordinate system ${}^{GB}q$ can be calculated as follows,

$$ {}^{GB}q = {}^{GS}q \otimes {}^{BS}q^* \tag{2.1} $$

where $\otimes$ refer to quaternion multiplication. Also, joint rotations are defined as the orientation of distal segment with respect to proximal segment. The joint rotation between two segments as seen from Fig.2.3 can be calculated using equation,

$$ {}^{AB}q = {}^{GA}q^* \otimes {}^{GB}q \tag{2.2} $$

where ${}^{GA}q$ is the orientation of proximal segment and ${}^{GB}q$ is the orientation of distal segment with respect to the global coordinate frame and $*$ refers to the complex conjugate of the quaternion. For detailed information about quaternions see Appendix A.

The Xsens MVN system can collect real-time data with a maximum sample rate of 120 Hz. In the experiments, a sample rate of 100 Hz has been chosen. There are several options to export data from MVN to other programs. The most common one is MVNX file format and interested data can be transferred to Microsoft Excel by using this file format.

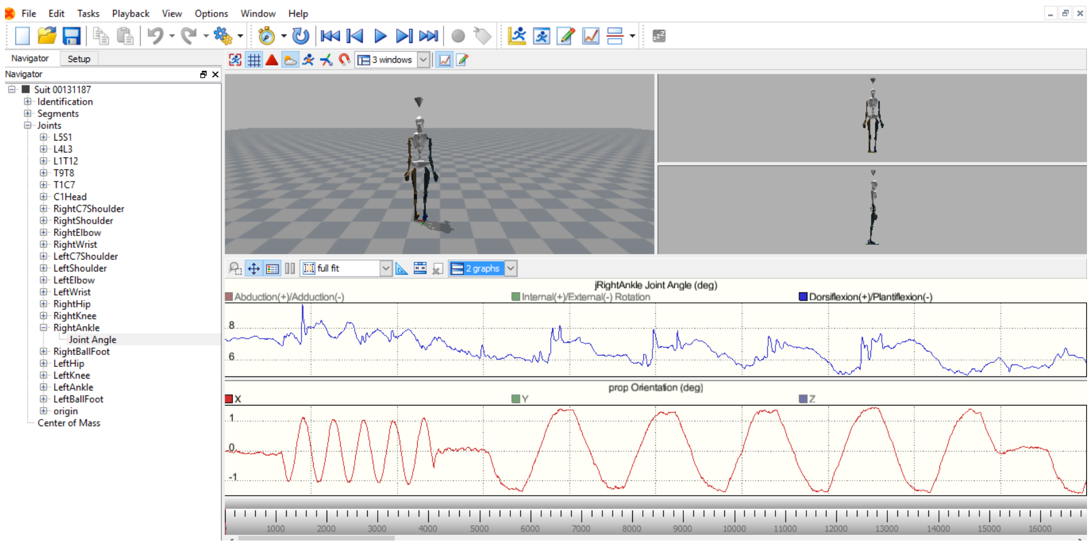


Figure 2.4: Sample data from Xsens MVN motion capture system; dorsiflexion and plantar flexion of right ankle (blue) and angular displacement of tilt platform (red)

CHAPTER 3

EXPERIMENTS AND METHODS

3.1 Experiments

The subjects have been chosen from two different groups; 6 healthy people (control group) and 6 patients with bilateral vestibular loss (patient group). Anthropometric data such as; age, height, and weight of subjects are recorded before the experiment. Table 3.1 shows the mean and standard deviation of these data for control and patient groups. Motion capture system (MTX) have been dressed to all subjects before the experiment. Subjects were told to stand without moving intentionally on the force plate sensor mounted on the custom made tilt platform. Subjects were warned about not to move their foot throughout to experiments. Data were collected from subjects when tilt platform was stationary and moved sinusoidally in the sagittal plane at two different constant frequencies: 0.05 and 0.17 Hz with 1.2 and 1-degree amplitude respectively.

Table 3.1: The mean and standard deviation of age, height and weight of control and patient groups

	Age	Height (cm)	Weight (kg)
Control group	29.3 ± 5.8	169.3 ± 8.5	72.7 ± 13.6
Patients	46.7 ± 13.8	163.3 ± 7.3	74.1 ± 19.9

There are total 12 trials for tilted experiments; 6 trials at low frequency lasting 100 seconds and 6 trials at high frequency lasting 30 seconds. Also, there is a 10 seconds long resting period between two successive trials. To answer the question: What will be the postural strategies and/or performance of patients with bilateral vestibular loss, if they lose information from another sensory system like vision, experiments have also been performed under eyes open and eyes closed conditions with 3 repetitions at two frequencies in random order. During the experiment, the only interaction with subjects is the verbal instruction about opening or closing their eyes. At the end of the experiments, a conversation with each subject about their daily postural activities, habits, life style, and impressions during the tests was made. The primary purpose of this conversation is to gain a better understanding of their postural responses during the experiments.

Figure 3.1: A photo from experiment

3.2 Data Analysis Methods

Time and frequency domain analyses will be introduced in this section.

3.2.1 Decomposition Method in Time Domain

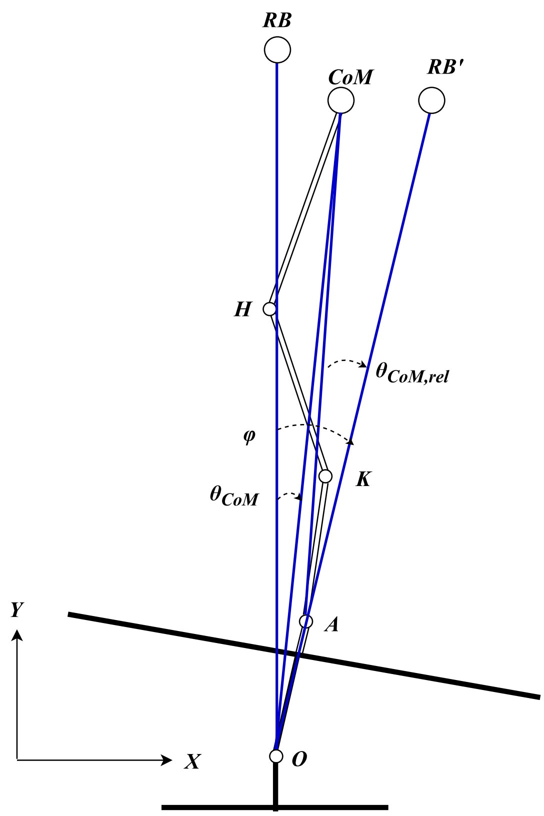


Figure 3.2: Static decomposition of subject's motion and platform's motion

Suppose a rigid body on the platform as shown in Fig.3.2. Its center of mass is located at point RB. If the angular displacement of platform respect to space is ϕ, the new location of CoM of the rigid body then becomes RB', and its angular displacement respect to space is the same as the angular displacement of platform ϕ. This displacement of the rigid body is imposed by platform's motion. Similarly, consider a subject is standing on the platform, and the location of CoM of the subject is at point RB initially. When platform moves about an angle ϕ, the new location of CoM of the subject should be at point RB', if the subject is stationary on the platform. However, in addition to platform motion, there may be also the motion of the subject with respect to the platform. Therefore, the new location of CoM of the subject is then an arbitrary point in the range of biomechanical constraints of the subject. It is important to discuss not only physiological meaning of the subject motion respect to space but also the subject motion respect to the platform.

Since rotation axis of subjects is at their ankle, it is not convenient to calculate angular displacement of CoM of subjects with respect to the rotation axis of tilt platform. $\theta_{CoM,rel}$ is angular displacement of CoM respect to tilt platform and θ_{CoM} is absolute angular displacement of CoM in space coordinate. As seen in Fig.3.3, lines |OCoM| and |ACoM| cover the same arc. Therefore, the relation between θ_{CoM} and $\theta_{CoM,rel}$ is,

$$(\theta_{CoM} - \phi)|OCoM| = \theta_{CoM,rel}|ACoM| \tag{3.1}$$

and

$$\theta_{CoM,rel} = (\theta_{CoM} - \phi)\frac{|OCoM|}{|ACoM|} \tag{3.2}$$

Note that, subjects have not been allowed to move their foot with respect to platform throughout the experiments. Therefore, it is assumed that ankle of the subject (point A) always lies on line |ORB'|.

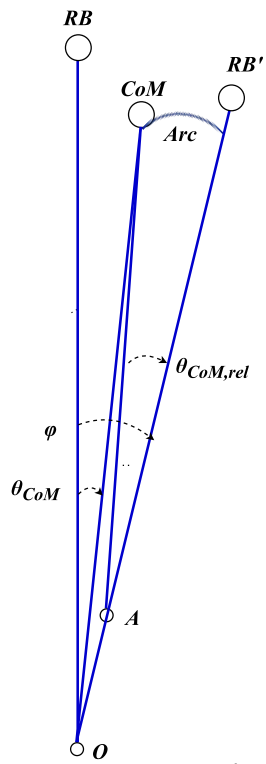


Figure 3.3: Relation between θ_{CoM} and $\theta_{CoM,rel}$

3.2.2 Time Domain Analysis

The average mean square value describes the general power of any random data.The average mean square value of a time varying function $x(t)$, over the time T can be calculated by the integral,

$$\psi_x^2 = lim_{T\to\infty}\frac{1}{T}\int_0^T x^2(t)dt \tag{3.3}$$

The positive square root of mean squared values is called the root mean square or RMS value ψ_x.

3.2.3 Frequency Domain Analysis

It is useful to describe periodic data in frequency domain since frequency domain analysis shows how much power of the signal lies within each frequency. The Fourier Transform is a mathematical method that transforms a time dependent function to a frequency dependent function. A time-dependent function $x(t)$ can be converted into frequency domain by using relation,

$$X(f) = \int_{-\infty}^{\infty} x(t)e^{-j2\pi ft}dt \tag{3.4}$$

3.2.4 Cross-Correlation Density Function

The cross-correlation between two signals; $x(t)$ and $y(t)$ describes the general dependence of the values of one signal to the other. The cross-correlation function $R_{xy}(\tau)$ of signals $x(t)$ and $y(t+\tau)$ over the time T can be calculated by taking average product of two signals as T goes infinity.

$$R_{xy}(\tau) = lim_{T\to\infty}\frac{1}{T}\int_0^T x(t)y(t+\tau)dt \tag{3.5}$$

The Fourier transform of the cross-correlation function of two signals $R_{xy}(\tau)$ is cross-spectral density function $G_{xy}(f)$.

$$G_{xy}(f) = 2\int_{-\infty}^{\infty} R_{xy}(\tau)e^{-j2\pi f\tau}d\tau \tag{3.6}$$

The cross-spectral density function has a real part $C_{xy}(f)$ and an imaginary part $Q_{xy}(f)$, called coincident spectral density function and quadrature spectral density function respectively.

$$G_{xy}(f) = C_{xy}(f) - jQ_{xy}(f) \tag{3.7}$$

One can also represent cross-spectral density function in polar form

$$G_{xy}(f) = |G_{xy}(f)|e^{-j\theta_{xy}(f)} \tag{3.8}$$

where

$$|G_{xy}(f)| = \sqrt{C_{xy}(f)^2 + Q_{xy}(f)^2} \tag{3.9}$$

and

$$\theta_{xy}(f) = tan^{-1}(\frac{Q_{xy}(f)}{C_{xy}(f)}) \tag{3.10}$$

$|G_{xy}(f)|$ and $\theta_{xy}(f)$ are both useful relation for signal analysis in frequency domain. $|G_{xy}(f)|$ is the magnitude of cross-spectral density function of signals $y(t)$ and $x(t)$. Whereas $\theta_{xy}(f)$ is the phase difference between signals $x(t)$ and $y(t)$.

Another useful relation for linear signal analysis in the frequency domain is coherence function. Consider a linear system with a single input x(t) and single output y(t) with an impulse response h(t) such that,

$$y(t) = h(t) * x(t) \tag{3.11}$$

Fouriér transformation of the auto-correlation of signals $x(t)$ and $y(t)$ are $G_{xx}(f)$ and $G_{yy}(f)$ respectively. Then Eq. 3.11 become,

$$G_{yy}(f) = |H(f)|^2 G_{xx}(f) \tag{3.12}$$

The cross-spectral density function of these signals is $G_{xy}(f)$.

$$G_{xy}(f) = H(f)G_{xx}(f) \tag{3.13}$$

The ratio of cross-spectral density function $G_{xy}(f)$ to product of auto-correlation function of signals $x(t)$ and $y(t)$ is

$$\frac{|G_{xy}(f)|^2}{G_{xx}(f)G_{yy}(f)} = \frac{|H(f)G_{xx}(f)|^2}{G_{xx}(f)^2|H(f)|^2} = \frac{|G_{xx}(f)|^2}{G_{xx}(f)^2} = 1 \tag{3.14}$$

For real systems this equation is never equal to 1, but smaller than 1. Then, define coherence function $\gamma_{xy}(f)^2$

$$\gamma_{xy}(f)^2 = \frac{|G_{xy}(f)|^2}{G_{xx}(f)G_{yy}(f)} \leq 1 \tag{3.15}$$

Coherence function tells about how the two signals are related to each-other in frequency domain. If $\gamma_{xy}(f)^2$ is close to 1, it shows that two signals are coherent. If it is close to 0, signals are incoherent.

3.2.5 Analysis of Variance (ANOVA)

Analysis of variance (ANOVA) is widely used to compare the means between the interested groups and determines any of those means are statistically significantly different from each other or not. In this study, 4-way ANOVA is used to determine the statistically significant difference between interested independent variables. These variables are subjects (control group versus patients), angular displacement of CoM (absolute vs relative), frequencies (f=0.17 Hz vs 0.05 Hz), and eyes conditions (eyes open vs eyes closed). In addition, to understand interactions between these variables, multiple-comparisons (post hoc analysis) have been performed. The significance level of 0.05 has been chosen for ANOVA test.

CHAPTER 4

MATHEMATICAL MODEL

Although center of mass (CoM) data of a subject is available in Motion Capture System (MVN), it is not well suited for our static decomposition method because the details of the biomechanical model of MVN is not known. In addition to that, the motion of the platform is a trajectory motion where its axis of rotation is 34.5 cm below the platform (see Chapter 2). Motion capture system neglects rotation of the platform for calculation of CoM motion of the subject for some good reasons. For example, it may not be important (depending on the aims) to calculate the absolute motion of CoM (with respect to the ground), when walking or running. However, during the conversation with the subjects at the end of the experiment, many of the subjects stated that they felt a motion in antero-posterior direction. Also, Motion Capture System calculates CoM with respect to subjects foot sole and ignore moment arm of the tilt platform. Thus, a mathematical model of the human body which includes ankle, knee, and hip joints is developed to calculate CoM motion of subjects.

Fig.4.1 shows a mechanical model of a subject standing on the tilt platform. The model consists of three segments which are lower leg (leg), upper leg (thigh) and HAT (head, arms, and trunk). Segments are defined as; the lower leg is from medial malleolus to femoral condyles, the upper leg is from femoral condyles to greater trochanter, and the HAT is from greater trochanter to glenohumeral joint. The model is based on the assumption that two segments are coupled by a joint. The joint position is defined at the location where each of the two segments has zero average displacements with respect to each other (ankle, knee, and hip). Subjects were not allowed to move their foot with respect to the platform during the experiments. Therefore,

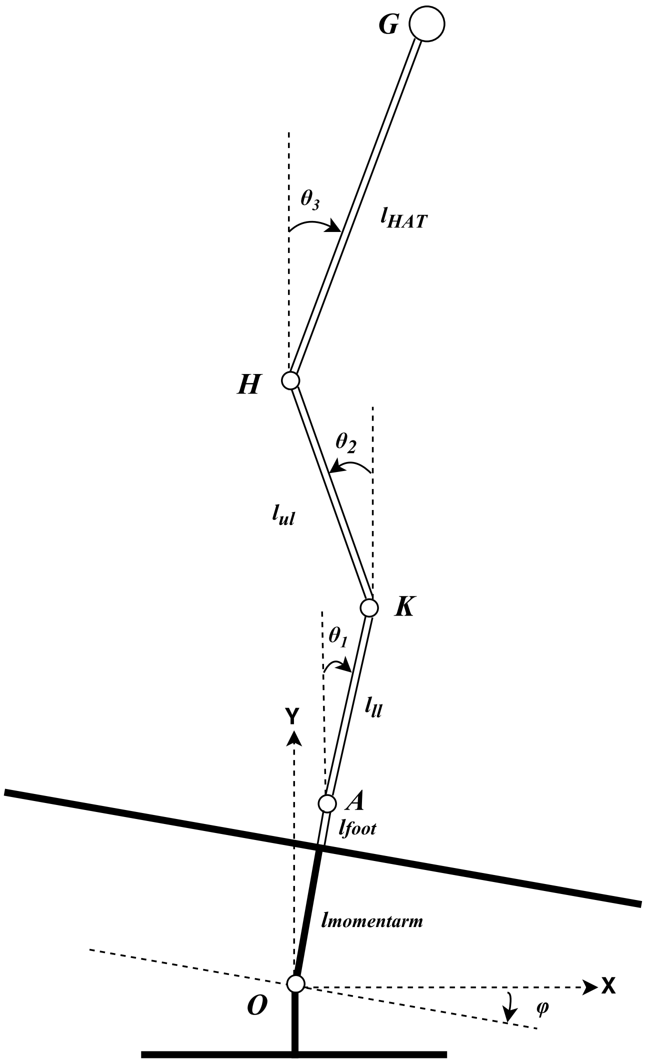


Figure 4.1: Mechanical model of a subject standing on the tilt platform

the relative motion between the foot of the subject and the platform was assumed to be equal to zero. In other words, the distance between an arbitrary point on foot and another arbitrary point on the platform remains constant over time. Thus, the foot of subject and platform are considered as a single rigid body in the mathematical model.

In Fig.4.1, distance from ankle (point A) to knee (K) represents length of lower leg l_{ll}. Similarly, distances from knee to hip (H) and hip to glenohumeral joint (G) represents lengths of upper leg l_{ul} and HAT l_{hat} respectively. The orientation of platform with respect to the global coordinate system is represented by angle ϕ. θ_1 is the segment orientation of lower leg, θ_2 is the segment orientation of upper leg, and θ_3 is the segment orientation of HAT with respect to the global coordinate system respectively.

The anthropometric data such as length, position of center of mass, and weight of body segments are calculated based on [44]. The body segment lengths are expressed as a fraction of body height H, and weight of body segments are given as a fraction of body segment weight to total body weight M. The location of the center of mass of each segment is also given as a fraction of the segment length from either the distal or the proximal end.

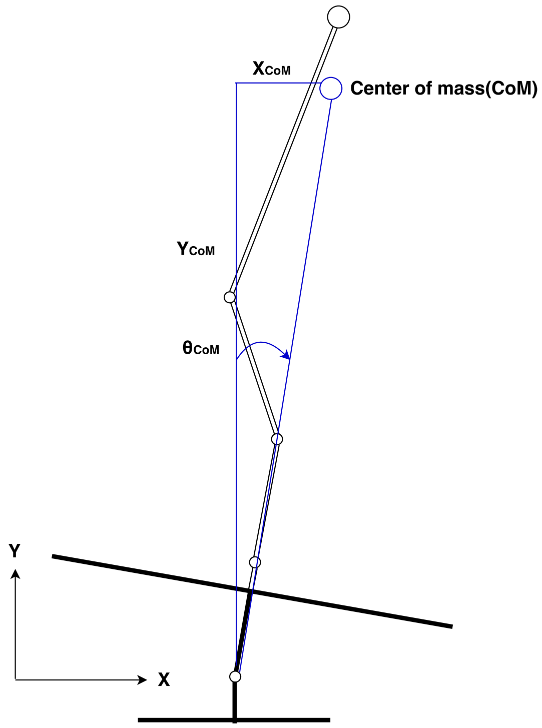


Figure 4.2: Absolute angle of center of mass (CoM)

The distance from rotation axis of platform O to ankle joint A can be expressed as,

$$l_{OA} = l_{foot} + l_{momentarm} \tag{4.1}$$

The absolute displacement of CoM can be calculated respect to stationary point O as below. First, the coordinates of center of mass of lower leg can be calculated as,

$$x_{ll} = l_{OA}sin(\phi) + c_{ll}sin(\theta_1) \tag{4.2}$$

$$y_{ll} = l_{OA}cos(\phi) + c_{ll}cos(\theta_1) \tag{4.3}$$

where c_{ll} is the center of mass of lower leg from the distal end of the segment. The coordinates of center of mass of upper leg can be calculated by using relation,

$$x_{ul} = l_{OA} + l_{ll}sin(\theta_1) + c_{ul}sin(\theta_2) \tag{4.4}$$

$$y_{ul} = l_{OA}cos(\phi) + l_{ll}cos(\theta_1) + c_{ul}cos(\theta_2) \tag{4.5}$$

where c_{ul} is the center of mass of upper leg from the distal end of the segment. The coordinates of center of mass of HAT is;

$$x_{HAT} = l_{OA}sin(\phi) + l_{ll}sin(\theta_1) + l_{ul}sin(\theta_2) + c_{HAT}sin(\theta_3) \tag{4.6}$$

$$y_{HAT} = l_{OA}cos(\phi) + l_{ll}cos(\theta_1) + l_{ul}cos(\theta_2) + c_{HAT}cos(\theta_3) \tag{4.7}$$

where c_{HAT} is the center of mass of HAT from the distal end of the segment. Finally, coordinates of center of mass of the whole body is;

$$X_{CoM} = \frac{m_{ll}x_{ll} + m_{ul}x_{ul} + m_{HAT}x_{HAT}}{M} \tag{4.8}$$

Similarly,

$$Y_{CoM} = \frac{m_{ll}y_{ll} + m_{ul}y_{ul} + m_{HAT}y_{HAT}}{M} \tag{4.9}$$

Since CoM is calculated in centimeter and sinusoidal tilts of platform ϕ is in degree, a direct comparison is meaningless. What is needed is a shared metric that enables comparisons. Thus, CoM can be converted to degree as follows;

$$\theta_{CoM} = \arctan(\frac{X_{CoM}}{Y_{CoM}}) \tag{4.10}$$

The simulation results of developed mathematical model are illustrated in following figures. Fig.4.3 and Fig.4.4 show displacements of center of mass(CoM) in sagittal plane. In addition, the absolute angular displacement of CoM is demonstrated in Fig.4.5.

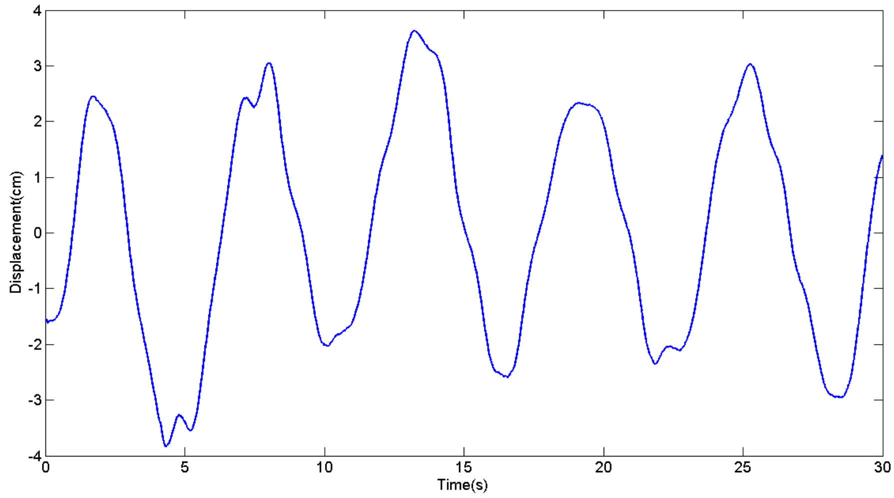


Figure 4.3: The simulation of anterior-posterior displacement of center of mass (CoM) in sagittal plane

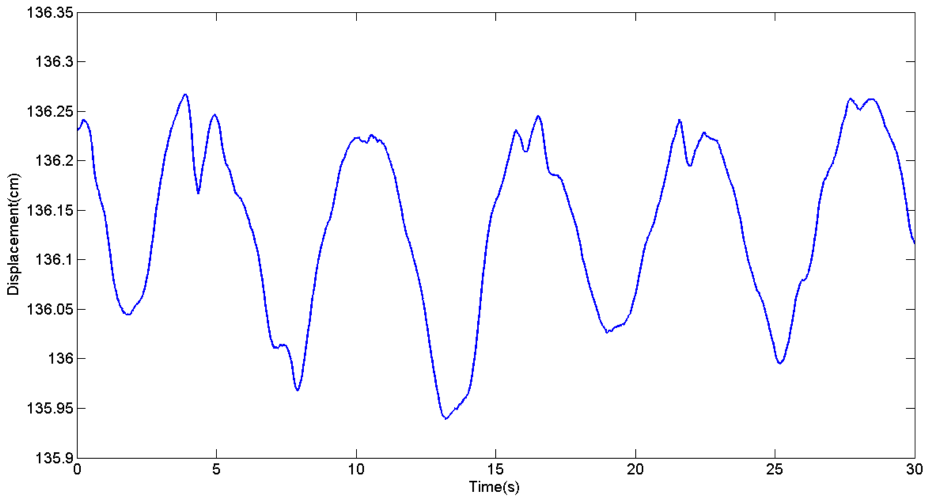


Figure 4.4: The simulation of inferior-superior displacement of center of mass (CoM) in sagittal plane

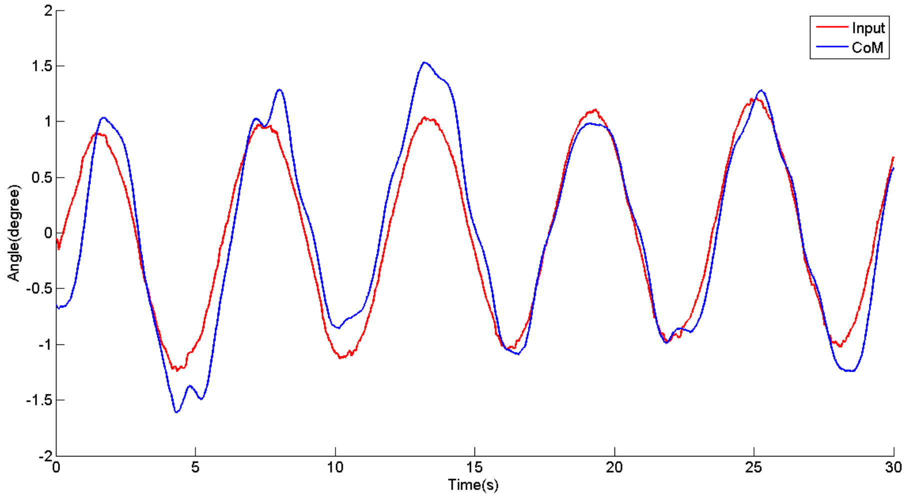


Figure 4.5: The perturbation of platform and simulation of absolute center of mass (CoM) angular displacement

CHAPTER 5

RESULTS

In this chapter statistical analysis of measured and decomposed (see Chapter 3) CoM signal are presented. These are absolute and relative CoM angular displacement of subjects to sinusoidal tilts in time and frequency domains. In addition, coherence, magnitude and phase difference between CoM angular displacement of subjects and the platform perturbation are demonstrated. Finally, ANOVA test has been applied to estimated variables to check whether the results are statistically significant or not. The first part of results includes the inter-group comparisons; i.e., between the control group and patients with bilateral vestibular loss, while the second part involves intra-group variations.

Fig.5.1 shows a control subject's (top figure) and patient's (bottom figure) absolute and relative angular displacements of CoM as a response to 1 degrees 0.17 Hz sinusoidal perturbation of tilt platform. The responses of the patient and control subject to sinusoidal tilts are similar at eyes open condition. As seen from figures, their absolute and relative CoM motion have very similar amplitude and phase characteristics. Fig.5.2 shows the same subjects' responses to 1.2 degrees 0.05 Hz sinusoidal perturbation of tilt platform at eyes open condition. The absolute CoM angular displacement of the subjects have very close amplitude and phase characteristics. However, their relative CoM angular displacements are different. Patient has much higher relative CoM angular displacement then control subject. These results suggest that patient and control subjects have different postural control strategies at low frequency. The investigation on RMS values of CoM motion of patient and control group in the time domain may give more detailed information about responses of these two groups.

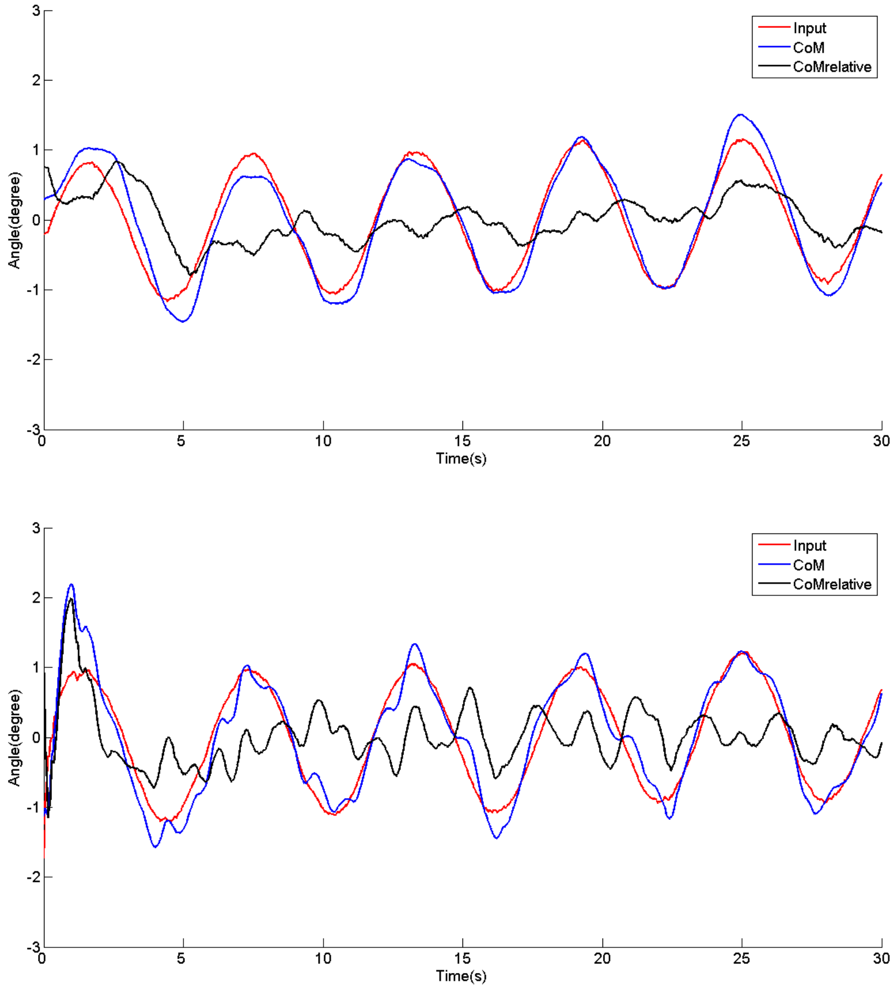


Figure 5.1: 1 degrees 0.17 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of healthy subject 4 (top figure) and patient 4 (bottom figure) at eyes open condition

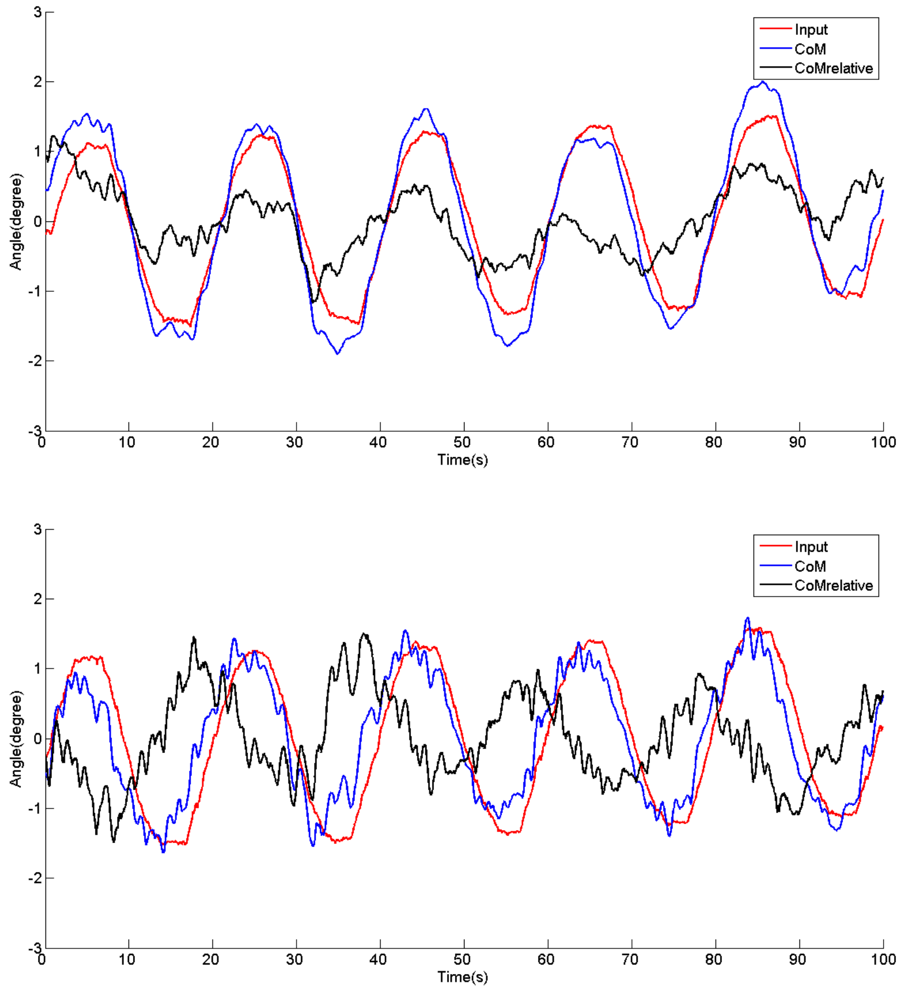


Figure 5.2: 1.2 degrees 0.05 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of healthy subject 4 (top figure) and patient 4 (bottom figure) at eyes open condition

Tables 5.1 and 5.2 show the mean and standard deviation of RMS values of CoM angular displacement for both control group and patients at different experimental conditions. The mean RMS values of patients are higher than the control group for angular displacement (both absolute and relative) of CoM ($p<0.0023$). However, more detailed analysis of this result yields different conclusions. Although the mean RMS values for absolute CoM angular displacement of patients are slightly higher than the control group at f=0.17 Hz, it is not significantly different. Also, it is very close to each other at f=0.05 Hz (see table 5.1). However, the examination of RMS values of relative CoM motion has different outcomes. It can be observed that patients have much higher mean RMS values than the control group at both frequencies (see table 5.2). However, this result is only statistically significant at f=0.05 Hz ($p<0.0992$). It is not significantly important at f=0.17 Hz because of high standard deviation at this frequency. These results support that, control group and patients have different strategies at f=0.05 Hz which is also observed at low frequency response of the two subjects in Fig.5.2. Control group remains relatively stationary compared to patients, whereas patients are seen to be active during the experiment. More detailed examination of experimental data such as coherence between the input and response of subjects may give some clues about whether the action of the patients is a preference or a result of instability.

Table 5.1: The mean and standard deviation of RMS values of control group and patients for absolute angular displacement of CoM

	f=0.17 Hz		f=0.05 Hz	
	EO	EC	EO	EC
Control group	0.646 ± 0.139	0.776 ± 0.112	0.946 ± 0.153	1.066 ± 0.0893
Patients	0.786 ± 0.516	1.141 ± 0.641	0.974 ± 0.456	1.036 ± 0.350

Table 5.2: The mean and standard deviation of RMS values of control group and patients for relative angular displacement of CoM

	f=0.17 Hz		f=0.05 Hz	
	EO	EC	EO	EC
Control group	0.445 ± 0.119	0.408 ± 0.0959	0.486 ± 0.164	0.597 ± 0.178
Patients	0.736 ± 0.519	0.940 ± 0.875	0.974 ± 0.365	0.918 ± 0.294

The tilting frequency seems to have no significant effect on control group's mean RMS values for both absolute and relative CoM angular displacements. However, there is also some gap between RMS values of the control group at f=0.17 Hz and f=0.05 Hz. It is possible that tilting frequencies greater than f=0.17 Hz may cause significant differences between RMS values of the control group across frequency. Also, note that patients mean RMS values for both absolute and relative CoM angular displacements do not differentiate across the frequency. In addition to these results, eyes conditions have no significantly important effect on RMS values of both control group and patients in our experiment.

Tables 5.1 and 5.2 also demonstrate the standard deviation of RMS values of subjects. The standard deviations of patients are much higher than the control group at all experimental conditions. It suggests that patients have very high intra-group variability. In other words, responses of patients are not homogeneous, but they are heterogeneous. Their response characteristics and control strategies differentiate from each other. On the contrary, lower standard deviations of RMS values for control group suggests that their responses are much more homogeneous than patients. It means that members of control group use similar control strategies and their response characteristics are close to each other. These results may give some clues about the effects of the loss of vestibular sensory information on postural control strategies and response characteristics of patients.

A useful method to compare responses of subjects is the estimation of the coherence between CoM angular displacement of subjects and perturbation of tilt platform. The coherence is a statistic that can be used to examine the power transfer between input and output of a linear system. If the signals are ergodic, and the system functions linearly, it can be used to estimate the causality between the input and output. Tables 5.3 and 5.4 show mean coherence values of control group and patients at different experimental conditions.

If the subject is stationary, the coherence between input and absolute motion of CoM should be high (close to 1). On the other hand, coherence between input and relative motion of CoM should be close to zero. If the subject is in anti-phase with input, coherence value between input and absolute CoM motion should be close to zero,

Table 5.3: The mean and standard deviation of coherence values of control group and patients for absolute angular displacement of CoM

	f=0.17 Hz		f=0.05 Hz	
	EO	EC	EO	EC
Control group	0.941 ± 0.0234	0.952 ± 0.0249	0.972 ± 0.0198	0.973 ± 0.0190
Patients	0.931 ± 0.0320	0.943 ± 0.0536	0.963 ± 0.0376	0.968 ± 0.0653

Table 5.4: The mean and standard deviation of coherence values of control group and patients for relative angular displacement of CoM

	f=0.17 Hz		f=0.05 Hz	
	EO	EC	EO	EC
Control group	0.554 ± 0.260	0.419 ± 0.169	0.675 ± 0.116	0.769 ± 0.109
Patients	0.674 ± 0.287	0.710 ± 0.0878	0.866 ± 0.134	0.830 ± 0.195

and coherence value between input and relative CoM motion should be close to 1. Tables 5.3 and 5.4 show that the coherence values of absolute motion of subjects are higher than coherence values of relative motion of subjects (p<0.0000). Thus, in general, subjects seem to remain stationary during to experiment.

According to ANOVA analysis, there is no difference in between coherency of absolute CoM motion of control and patient groups. This result can be observed in table 5.3. Also, the same table shows that coherence value between absolute CoM motion of subjects and platform perturbation is not varying across the frequencies. Therefore, it can be said that at both frequencies, absolute CoM angular displacement of subjects and sinusoidal tilt is very coherent (above 0.90) and frequency of stimulus have no effect on coherency for our experiment. Although above statement is true, it does not provide detailed information about effects of frequency on coherency. In other words, are strategies of subjects also same across the frequencies? Subjects having high coherence values either may remain stationary with respect to the platform or may move their CoM with some phase with respect to the platform perturbation. Table 5.4 shows mean coherence values of relative CoM motion of both groups at different experimental conditions. It can be observed that coherence values are significantly different from each other at different frequencies (p<0.0022). Subjects have relatively high coherence values at f=0.05 Hz than at f=0.17 Hz. However, this result is not observed for coherence values of patients such that their coherence values are not significantly different across the frequencies for relative angular displacement of CoM. Therefore, lower coherence values of the control group at f=0.17 Hz compared to their coherence values at f=0.05 Hz should be a reason for this result. Control group has more attention to perturbation of platform at f=0.05 Hz than 0.17 Hz. In other words, control group prefers to remain stationary at f=0.17 Hz, and they seem to be more active at f=0.05 Hz. However, patients appear to be active at both frequencies.

It is mentioned that absolute CoM motion of both control and patient groups and sinusoidal tilt is very coherent and there is no significant difference across groups. However, the comparison between relative COM motion of control and patient groups yields different results. As seen from Fig.5.3 relative CoM motions of patients are more coherent than control group. However, there is no significant difference between coherency of relative CoM motion of control group and patients at f=0.05 Hz and it is only statically significant at f=0.17 Hz (see table 5.4). These results do not imply a contradiction regarding previous results, the mean RMS values of the control group is significantly lower than the patient group at low frequency but not high frequency. The coherence implies the relation between input (perturbation of platform) and response of subjects at the interested frequency (input frequency). On the other hand, RMS value is the root mean square of CoM motions along the whole trial. It means that relative motion of patients includes not only response to perturbation of platform but also undefined CoM displacements other than input frequency. The undefined CoM displacement term refers to source and purpose of this motion. It is also important to consider the meaning of higher coherency of relative CoM motion of patients compared to control group at f=0.17 Hz before the more detailed discussion of the term, undefined CoM displacement. Since coherence shows power transfer between input and output, the response of patients is directly related to platform perturbation at high frequency contrary to at low frequency. This response can be either in phase or anti phase with the input. Both of them can cause high coherence value between output and input. However, in phase response may cause large body sways and loss of postural balance while an anti phase response can reduce the absolute CoM motion. The magnitude values of the transfer function estimation between platform perturbation and angular displacements of CoM can give a reasonable clue about this discussion. Thus, the undefined CoM displacement term can be classified either as a bad (malign) or good (benign) movement. If this movement is not voluntarily performed and causes to lose the balance of posture like large body sways, it is a bad movement. However, it is a good movement in the case that this motion enhances task performances and/or learning capabilities.

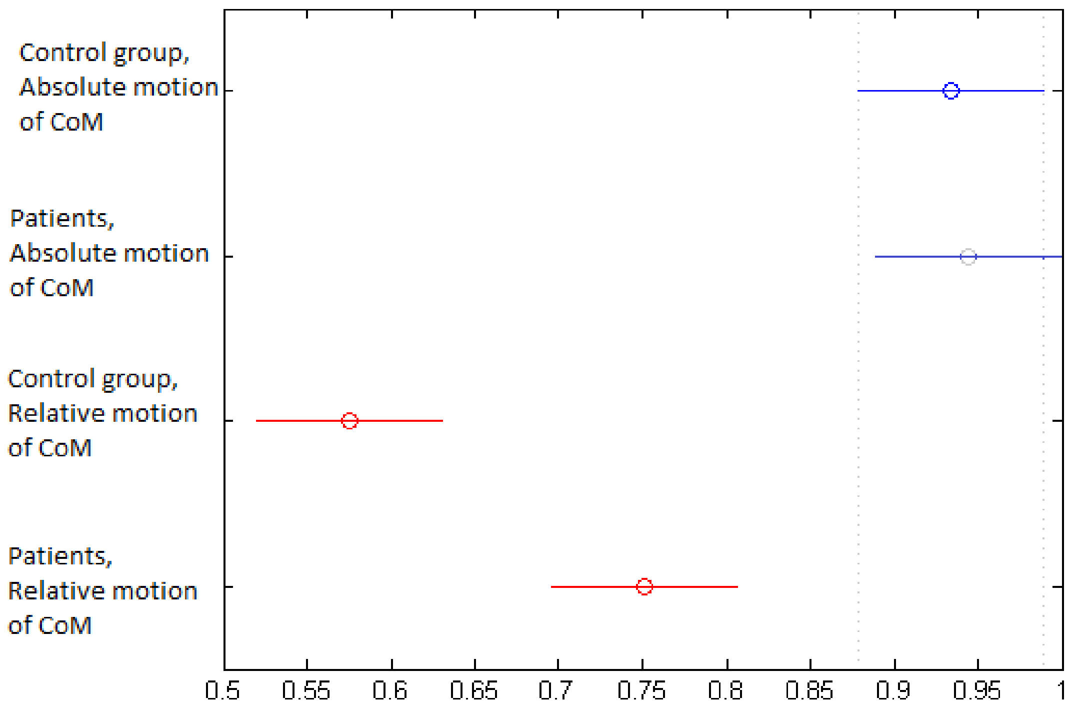


Figure 5.3: ANOVA for coherence values of absolute and relative CoM motions of subjects

Eyes conditions seem to have no effect on coherence values of both control group and patients. There is no significant difference in their coherence values across the eyes conditions (see tables 5.3 and 5.4). Also, the standard deviation of coherence values between CoM motion and platform perturbation is very low for both groups. The intragroup variability observed in RMS values of patients is not observed in coherence values. It can be argued that all subjects are well cohere with perturbation of platform.

The magnitude values of transfer function estimation between platform perturbation and absolute angular displacement of CoM are demonstrated in table 5.5. Magni-

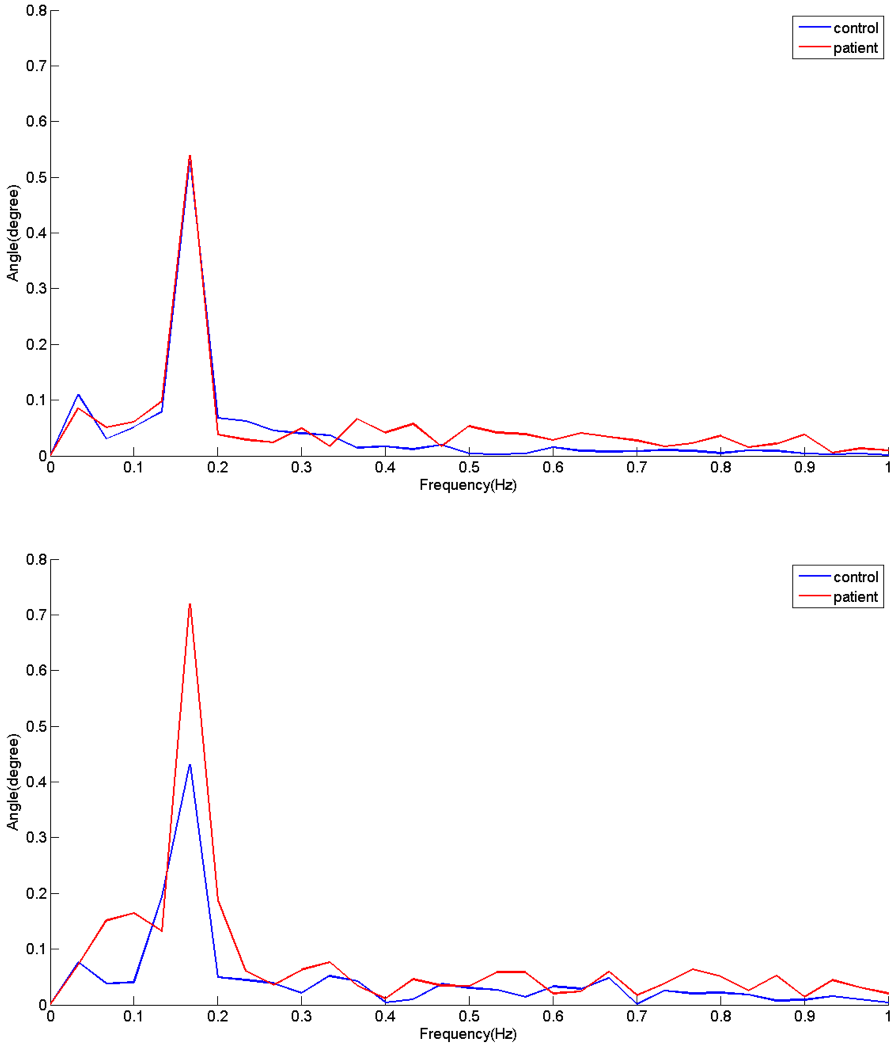


Figure 5.4: FFT of 1 degrees 0.17 Hz perturbation of tilt platform, and absolute CoM angular displacements of healthy subject 4 and patient 4 at eyes open (top figure) and closed (bottom figure) conditions

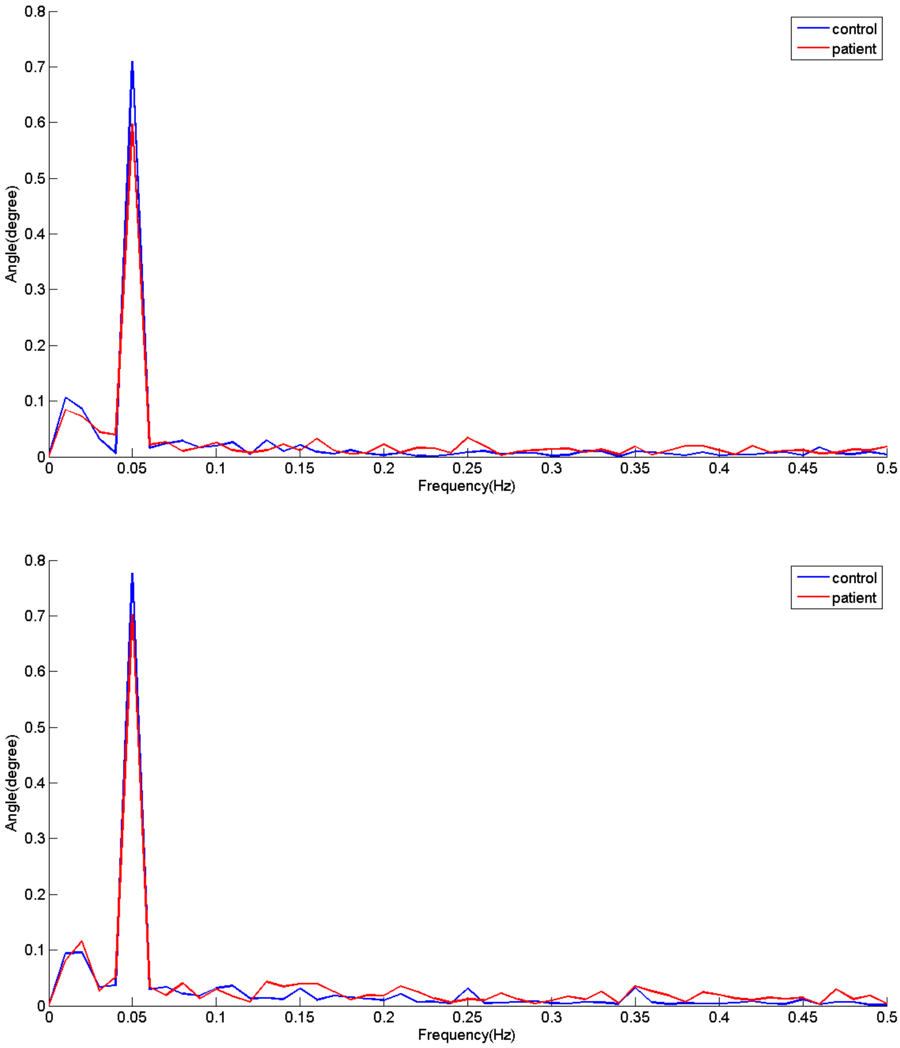


Figure 5.5: FFT of 1.2 degrees 0.05 Hz perturbation of tilt platform, and absolute CoM angular displacements of healthy subject 4 and patient 4 at eyes open (top figure) and closed (bottom figure) conditions

tude values indicate ratio of angular displacement of CoM to platform perturbation at tilting frequency. The magnitude values of control group and patients are slightly different from each other but there is no significant difference. Perturbation frequency and eyes conditions also seem to be have no significant effect on magnitude values except at f=0.17 Hz eyes closed conditions for patients. Patients have mean magnitude value 0.815 ± 0.214 when eyes open and 1.408 ± 0.366 when eyes closed condition, at f=0.17 Hz. Although magnitude values of the control group are slightly different across eyes conditions, they are not significantly different. The relatively high magnitude value of patients at eyes closed conditions indicate that the response of patients exceed perturbation amplitude of platform. Since, this overshoot is not observed at eyes open condition, it can be said that patients use their eyes as a reference for their movement. They correct their absolute CoM displacement according to feedback from eyes. This correction vanishes in the absence of visual information and the patients show large body sways. This phenomena is not observed for control groups. It suggests that control group may replace feedback from eyes with other sensory information such as vestibular feedback when their eyes closed. Fig.5.4 shows FFT of 1 degree 0.17 Hz absolute CoM displacement of a healthy subject and a patient at both eyes open and closed conditions. Patient and healthy subject have very similar response characteristics when their eyes are open (top figure). Patients have relatively high amplitude at f=0.17 Hz compared to healthy subject in the absence of vision (bottom figure). Fig.5.5 shows FFT of 1.2 degree 0.05 Hz absolute CoM displacement of same healthy subject and patient at both eyes open and eyes closed conditions. Their responses at this frequency have very similar characteristics and eyes conditions have no significant effect on their responses. We suggest that at f=0.17 Hz patients use their eyes as a reference feedback for their displacement in

Table 5.5: The mean and standard deviation of magnitude values of the transfer function estimates between platform perturbation and absolute angular displacement of CoM for control group and patients at different experimental conditions

	f=0.17 Hz		f=0.05 Hz	
	EO	EC	EO	EC
Control group	0.838 ± 0.186	1.003 ± 0.130	0.933 ± 0.157	1.041 ± 0.102
Patients	0.815 ± 0.214	1.408 ± 0.366	0.901 ± 0.368	0.994 ± 0.303

space however, this suggestion seems to be not valid at f=0.05 Hz. Absence of visual information at f=0.05 Hz does not cause large body sways for patients .

Table 5.6: The mean and standard deviation of magnitude values of the transfer function estimates between platform perturbation and relative angular displacement of CoM for control group and patients at different experimental conditions

	f=0.17 Hz		f=0.05 Hz	
	EO	EC	EO	EC
Control group	0.380 ± 0.229	0.273 ± 0.112	0.356 ± 0.159	0.457 ± 0.171
Patients	0.584 ± 0.303	0.889 ± 0.531	0.709 ± 0.214	0.682 ± 0.178

Table 5.6 demonstrates mean magnitude values of the transfer function estimates between platform perturbation and relative angular displacement of CoM for control group and patients at different experimental conditions. As seen from table patients' magnitude values of relative angular displacement of CoM are relatively higher than control group's magnitude values for all experimental conditions (p<0.0021). As mentioned above large body sways of patients at f=0.17 Hz are one of the reasons for this response. One can observe same result by comparing FFT of relative CoM angular displacement of a healthy subject and a patient. At f=0.17 Hz and eyes open condition, both patient and healthy subjects keep their angular displacement of CoM very small as seen from Fig.5.6 (top figure) and so their absolute and relative CoM motion magnitude values are similar. Bottom figure of Fig.5.6 demonstrates FFT of relative CoM's angular displacement of these subjects but with eyes closed condition. Large body sways of patient cause large gap between magnitude values of healthy subject and patient at f=0.17 Hz.

What about magnitude values of subjects at lower frequency f=0.05 Hz? The absolute angular displacement of CoM of subjects have similar characteristics across eyes conditions (see Fig.5.5). There is no large body sway of patients as observed at high frequency despite the fact that patients have higher relative magnitude values than control groups. It can be also observed in top and bottom figures of Fig.5.7. In both eyes conditions patient has much higher magnitude value than healthy subject. Healthy subject keeps her relative angular displacement of CoM close to zero. In spite of this difference, how can patients achieve the same response characteristics with control group for absolute angular displacement of CoM? It is possible that

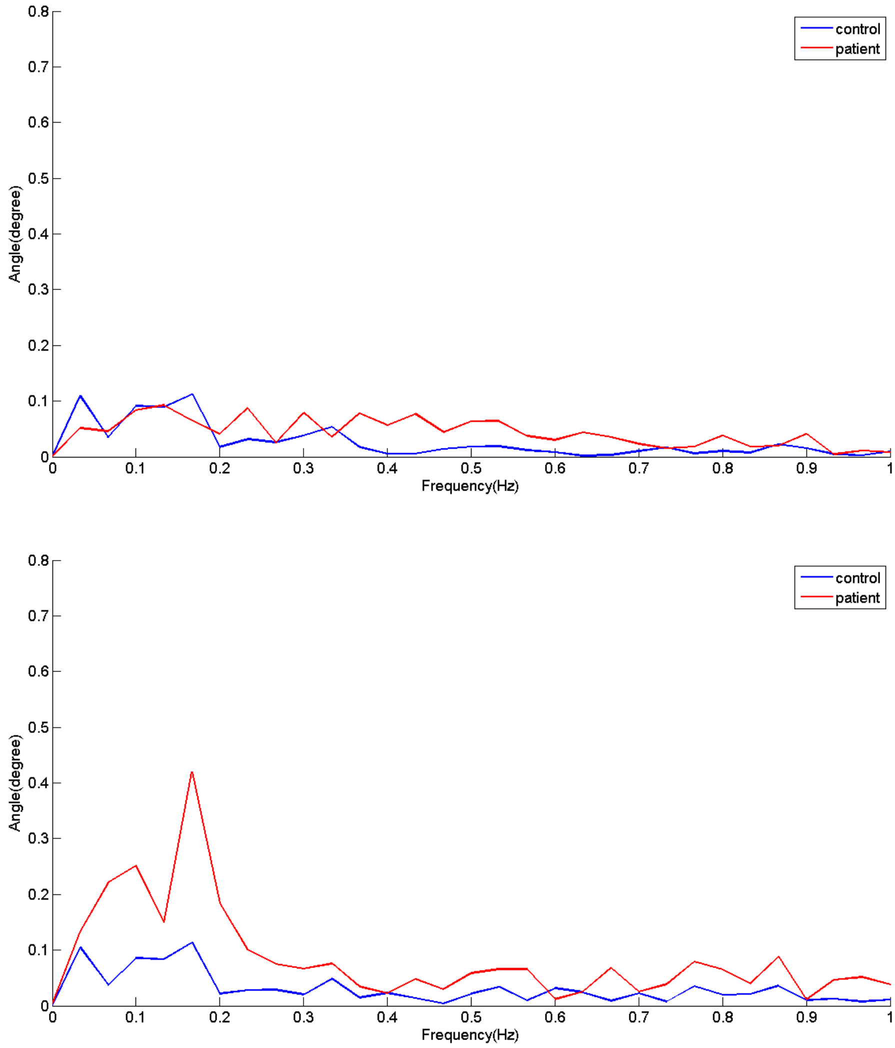


Figure 5.6: FFT of 1 degrees 0.17 Hz perturbation of tilt platform, and relative CoM angular displacements of healthy subject 4 and patient 4 at eyes open (top figure) and closed (bottom figure) conditions

response of patients to perturbation of platform is at same frequency with platform but with some phase difference. In other words, they shift their response and anticipate the motion of platform. Phase difference between angular displacement of the subjects and perturbation of platform may give more clear answer to this question.

In addition to above results, standard deviation of magnitude values of the transfer function estimates differ across the subjects. For both absolute and relative CoM angular displacements of subjects, patient group has higher standard deviation than control group. These results support the previous argument such that intragroup vari-

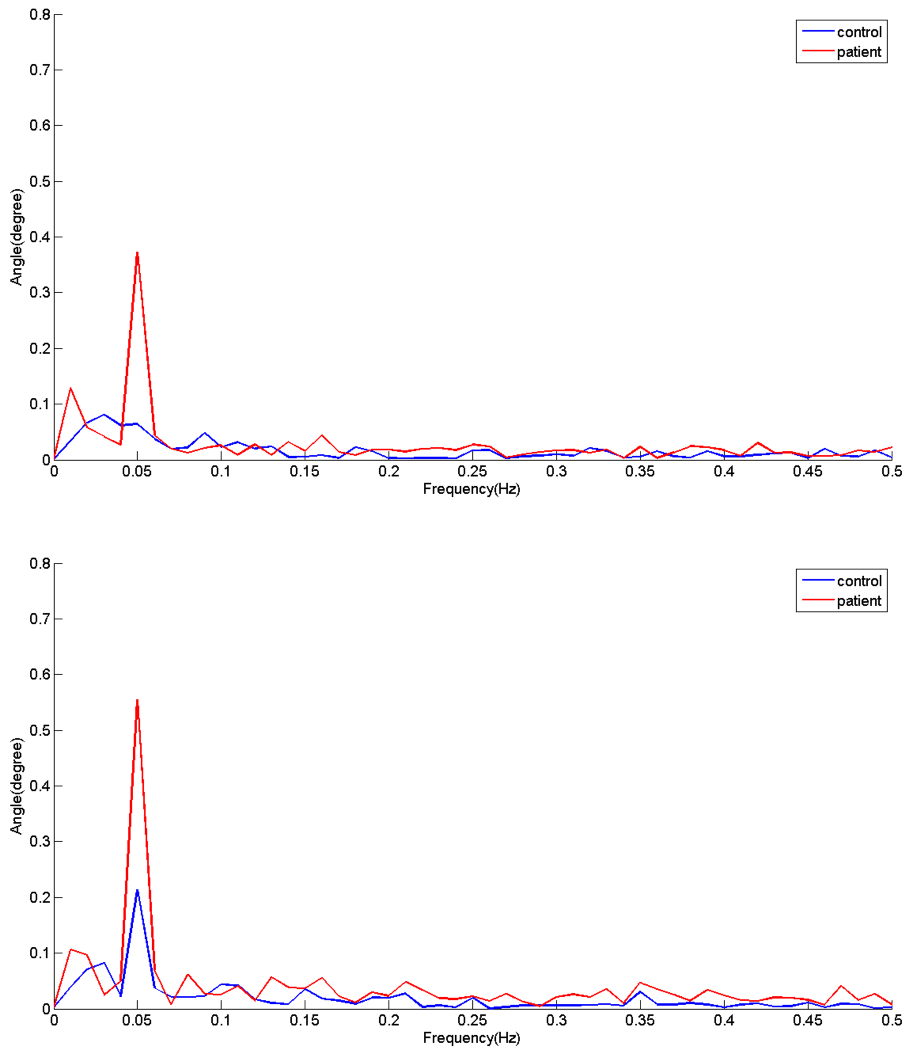


Figure 5.7: FFT of 1.2 degrees 0.05 Hz perturbation of tilt platform, and relative CoM angular displacements of healthy subject 4 and patient 4 at eyes open (top figure) and closed (bottom figure) conditions

ability of patient group is much higher than control group. It is hard to observe homogeneous response within patient group.

Table 5.7 presents phase difference between the absolute angular displacement of subjects and perturbation of platform at different experimental conditions. It is seen from table 5.7 that, the phase difference is significantly different across the frequency ($p<0.0000$). All subjects have positive phase difference at $f=0.05$ Hz, their absolute angular displacement of CoM comes before the perturbation of platform. Since performances of all subjects are successful at $f=0.05$ Hz, there is no overshoot and their responses are well cohere with perturbation of platform, which shows that they should have been anticipating the perturbation of platform. The absence of vision also has no significant effect on the phase difference between the absolute angular displacement of all subjects at this frequency. These results may give the answer the question, why does absence of vision cause large body sway for patients at $f=0.17$ Hz and why isn't it observed at $f=0.05$ Hz? Patients have successfully predicted the perturbation of platform and prevent large body sways at low frequency. However, one can argue against the same phase lead presented at $f=0.17$ Hz and eyes closed condition, and still patients presenting large body sway. It may be possible that, when their eyes are closed, patients also are trying to predict perturbation of platform to compensate the loss of feedback from vision at $f=0.17$ Hz, but can not manage to correct their movements. On the other hand, control group subjects have phase lag for both eyes conditions at 0.17 Hz, and they successfully respond to the perturbation of platform. These results suggest that control group subjects might have been compensating their movement by the use of vestibular sensory information in the absence of vision.

Table 5.8 presents phase difference between the relative angular displacement of subjects and perturbation of platform at different experimental conditions. Control and patient groups have similar phase angle at low frequency. However, patients seem to have significantly much higher phase angle at high frequency. It is hard to compare relative phase angles of these two groups because both of them have very high standard deviation. The heterogeneous response characteristics of patients once more appear in phase angles between CoM angular displacement and platform perturbation. In addition to the patient group, control group also have heterogeneous response characteristics contrary to previous results.

Table 5.7: The mean and standard deviation of phase angle between platform perturbation and absolute angular displacement of CoM for control group and patients at different experimental conditions

	f=0.17 Hz		f=0.05 Hz	
	EO	EC	EO	EC
Control group	-10.16 ± 4.82	-1.70 ± 7.13	9.52 ± 5.40	15.68 ± 6.30
Patients	-2.81 ± 10.56	4.75 ± 9.70	19.08 ± 4.56	20.52 ± 5.04

It has been argued that high standard deviations of RMS values of CoM motion of patients, and magnitude and phase estimation between platform perturbation and CoM angular displacement of patients indicate high intragroup variability for patient group. Therefore, different response characteristics within the groups should also be taken into consideration.

Fig.5.8 shows response of patient 1 (top figure) and patient 5 (bottom figure) to 1 degrees 0.17 Hz sinusoidal perturbation of tilt platform at eyes open condition. Patient 1 and patient 5 have different response characteristics. In contrast to patient 5, whose RMS value of absolute CoM angular displacement is 1.797 ± 0.105, the patient 1 has much lower RMS value 0.366 ± 0.0618 at f=0.17 Hz. Similarly, at f=0.05 Hz patient 5 and patient 1 have 1.624 ± 0.259 and 0.441 ± 0.0359 RMS values respectively (see also Fig.5.9). The relative CoM angular displacement of these patients can give enough information about the reason of this difference. As seen from both Fig.5.8 and Fig.5.9, the relative CoM motion of patient 1 is in anti-phase with platform perturbation. Therefore, the absolute CoM motion of patient 1 has much lower RMS values and amplitude than other patient. On the contrary to patient 1, the relative CoM motion of patient 5 is in in-phase with perturbation of platform and absolute

Table 5.8: The mean and standard deviation of phase angle between platform perturbation and relative angular displacement of CoM for control group and patients at different experimental conditions

	f=0.17 Hz		f=0.05 Hz	
	EO	EC	EO	EC
Control group	-125.11 ± 39.66	-3.45 ± 89.39	93.76 ± 56.10	92.24 ± 23.62
Patients	-21.16 ± 121.06	40.34 ± 69.10	120.39 ± 42.94	105.22 ± 36.59

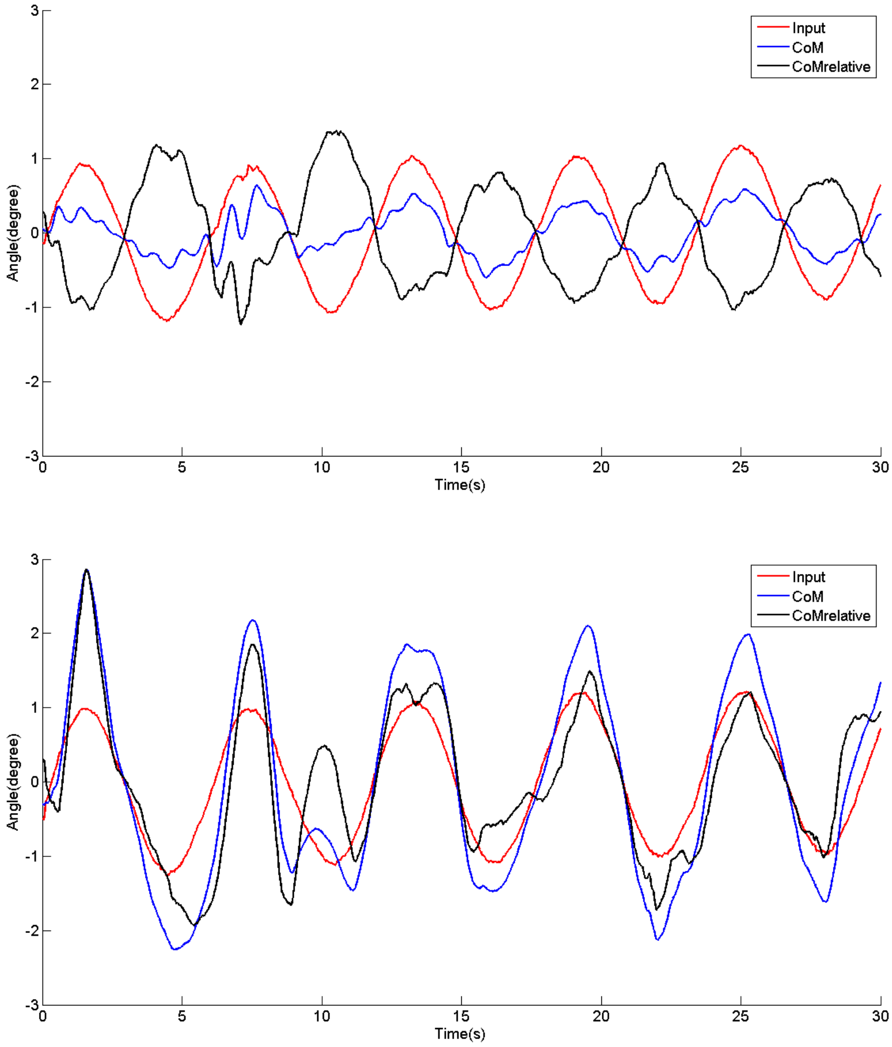
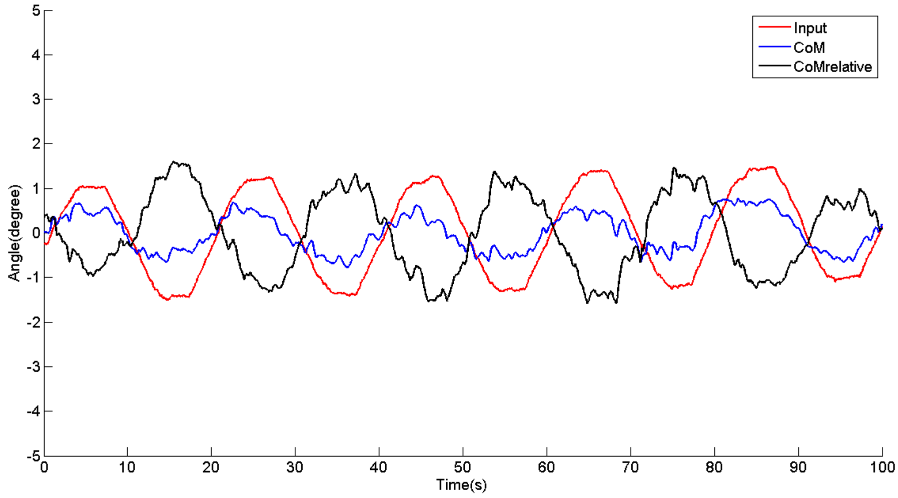


Figure 5.8: 1 degrees 0.17 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of patient 1 (top figure) and patient 5 (bottom figure) at eyes open condition



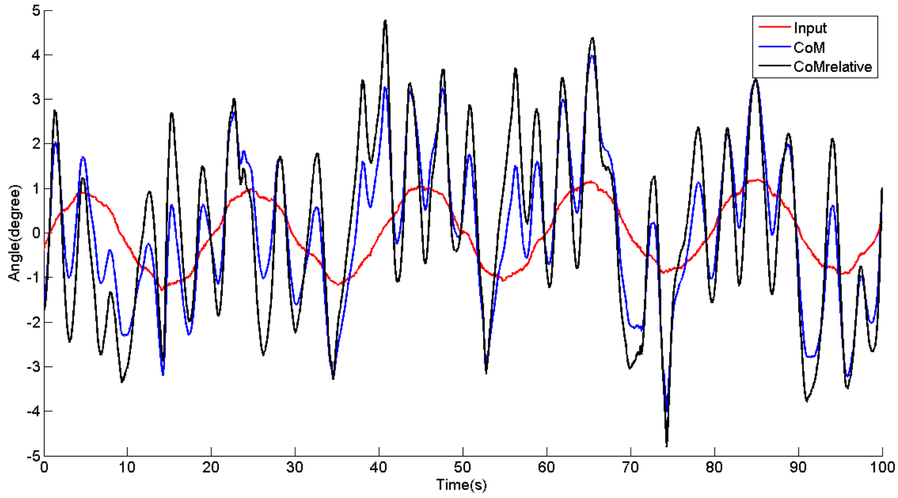
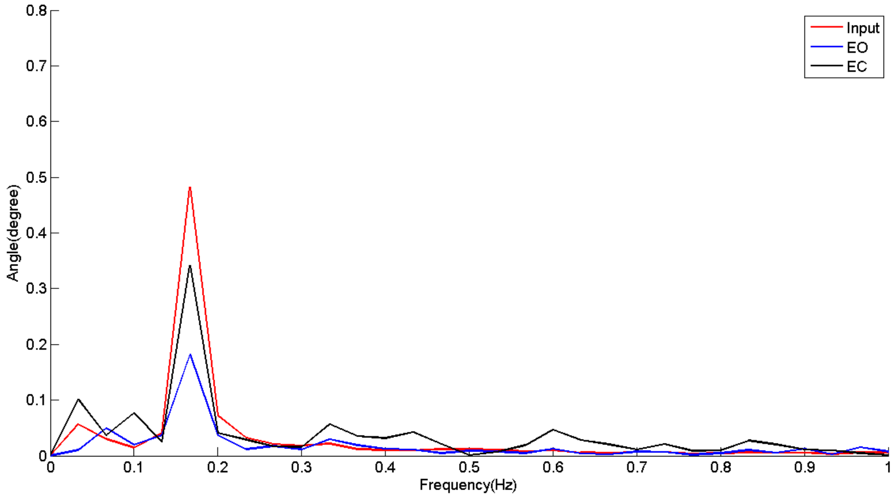


Figure 5.9: 1.2 degrees 0.05 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of patient 1 (top figure) and patient 5 (bottom figure) at eyes open condition



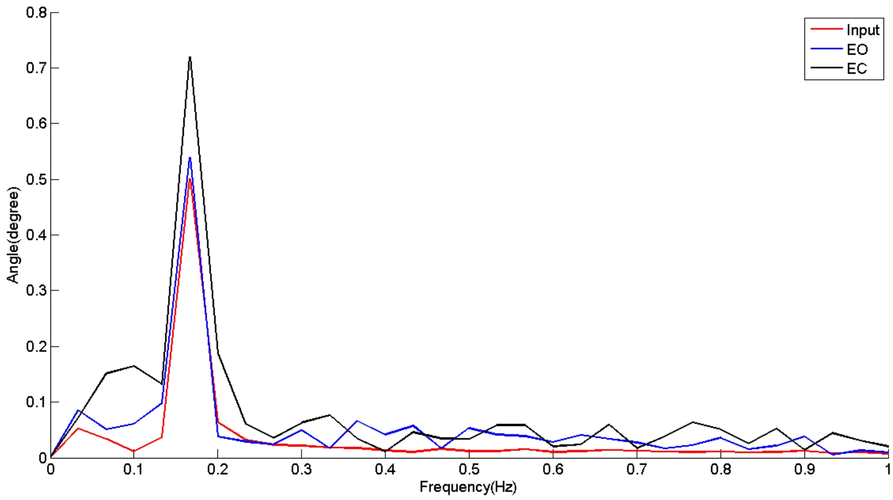


Figure 5.10: FFT of 1 degrees 0.17 Hz perturbation of tilt platform and, absolute CoM angular displacements of patient 1 (top figure) and patient 4 (bottom figure) at eyes open and closed conditions

CoM motion of patent 5 has much higher RMS values and amplitude.

At lower frequency (f=0.05 Hz), one can observe another different response characteristic of patient 1 and patient 5 (see Fig.5.9). Patient 5 not only sways at tilt frequency but also sways at different frequency higher than tilting frequency. If this sway is an undesirable movement, in other words causes postural balance problem for patient 5, it should also be observed at higher frequency (f=0.17 Hz). However, this phenomenon is not observed at the higher tilting frequency. Therefore, it is not convincing to say this a bad (malign) response or an undesirable movement. The sway that is observed at different frequencies compared to the perturbation frequency (as seen in the response of patient 5) is not observed at patient 1. However, it has been observed in responses of other patients but in smaller amplitudes.

Patient 1 also does not have large body sways at f=0.17 Hz and eyes closed condition contrary to the other patients. An example of this behavior is demonstrated in Fig.5.10. At f=0.17 Hz, patient 1 has lower amplitude than platform perturbation at both eyes conditions. However, patient 4 has lower amplitude than platform perturbation only at eyes open condition. When her eyes are closed, she shows large body sways and amplitude of CoM angular displacement, which becomes higher than the platform motion. The phase angle between body CoM and platform motion can explain difference in response of these patients. Patient 1 has mean phase differences $9,33 \pm 6.17$ and 13.06 ± 6.87 degrees, whereas patient 4 has mean phase differences -3.19 ± 2.12 and 15.23 ± 3.97 degrees at f=0.17 Hz for eyes open and eyes closed conditions respectively. Patient 1 somehow predicts the perturbation of platform accurately at f= 0.17 Hz. This successful prediction allows her to compensate for the platform movement. How can she manage to learn perturbation of the platform contrary to other patients?

Although, standard deviation of control group is very low in previous results presented in this section, there are also some different response characteristics within the control group. There are two distinct groups within the control group who have different response characteristics at f=0.17 Hz. Fig.5.11 shows FFT of 1 degrees 0.17 Hz perturbation of tilt platform and, absolute CoM angular displacement of two different healthy subjects. Subject 3 is in the first group which has relatively lower magnitude values between absolute angular displacement of CoM and platform perturbation at eyes open conditions (top figure of 5.11). In other words, they are anti-phase with sinusoidal tilt of platform at eyes open conditions and they remain stationary at eyes closed conditions. Subject 6 belong the second group and this group has no different strategies across the eyes conditions (bottom figure of 5.11). They prefer to remain stationary in both eyes conditions.

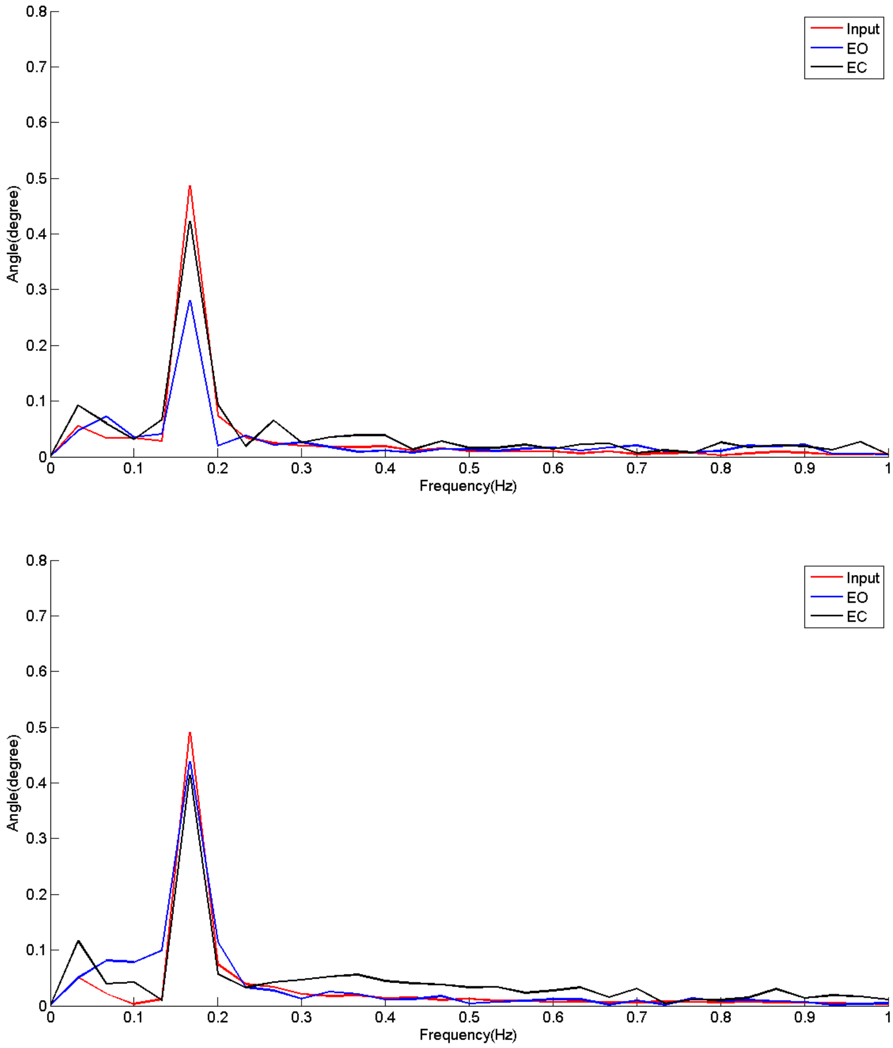


Figure 5.11: FFT of 1 degrees 0.17 Hz perturbation of tilt platform and, absolute CoM angular displacements of healthy subject 3 (top figure) and healthy subject 6 (bottom figure) at eyes open and closed conditions

CHAPTER 6

DISCUSSIONS AND CONCLUSIONS

In this thesis, postural responses of healthy subjects and patients with bilateral vestibular loss to sinusoidal tilts are analyzed. Different experimental conditions are imposed on subjects such as changing frequency and eyes conditions.

The first result of the experiment is that there is no significant difference between responses of the control group and patients, at low frequency (f=0.05 Hz), except the relative angular displacements of subjects. The RMS values of the control group and patients for absolute angular displacement of CoM are very close the each other. Also, the coherence between absolute angular displacement of CoM and perturbation of platform is very high for all subjects (above the 0.90). Besides, the absence of visual information does not cause any postural instability for both control and patient groups at f=0.05 Hz. Patients already lack sensory information from the vestibular system, and their only available sensory information are from somatosensory and proprioception at eyes closed condition. Then, how can control and patient groups have the same response characteristics? The result of a past study on patients with bilateral vestibular loss showed that the absence of visual and vestibular information have a small effect on balancing of human body posture in space at low frequencies (0-0.1 Hz) and somatosensory still provides some valuable information about the orientation of the body in space [24]. Thus, it is possible that sensory information from somatosensory and proprioception is sufficient to maintain postural balance at low frequencies. However, differences in relative CoM angular displacements of control and patient groups suggest that their control strategies may be different even if their responses are similar in space coordinates.

One of the important results of the experiments is that the RMS values of relative CoM angular displacement of patients are significantly higher than the control group. However, it is essential to interpret this result together with frequency. This result is only statically significant at f=0.05 Hz but not f=0.17 Hz. In addition, coherence between platform perturbation and relative CoM angular displacement of patients are not significantly different than the control group at low frequency. Since coherence values are estimated at the frequency of platform perturbation (f=0.05 Hz), they are directly related to platform motion. However, RMS value is the root mean square of CoM motions along the whole trial that it also includes movements at other than tilting frequency. In other words, the relative motions of patients include not only response to perturbation of platform but also other undefined CoM displacements. Therefore, there are two different reasons for higher RMS values of relative CoM angular displacements of the patients than control group. The first reason is that mean magnitude values of the transfer function estimation between platform perturbation and relative CoM angular displacement of patients are higher than control group. However, unlike the high frequency responses of patients, these higher magnitude values do not cause large body sways. In other words, their CoM angular displacements respect to space do not exceed platform motions (see tables 5.5 and 5.6). It is suggested that unlike high frequency responses of patients to platform perturbation, their low frequency responses are at the same frequency with the platform but with some phase differences (see Fig.5.2). In other words, they shift their responses and anticipate the motion of platform (see table 5.8). The second reason is that some patients have movements which are not correlated to perturbation of motions. The more extreme example of these movements can be seen in FFT of relative CoM angular displacements of patient 5 at low frequency in Fig.6.1. Patient 5 clearly has some CoM motion at different frequencies that is independent than perturbation of the platform (at f=0.05 Hz). These undefined CoM displacements can be considered as a movement variability because they are irrelevant to perturbation of platform. The source and purpose of these undefined CoM displacements may give some important cues about postural control strategies of patients. Traditionally, such movement variability have been considered as an instability. However, it is possible that such movements could have a purpose. The stability requirement for CoM angular displacement is not a single point but rather a region with limits. If CoM angular displacements of

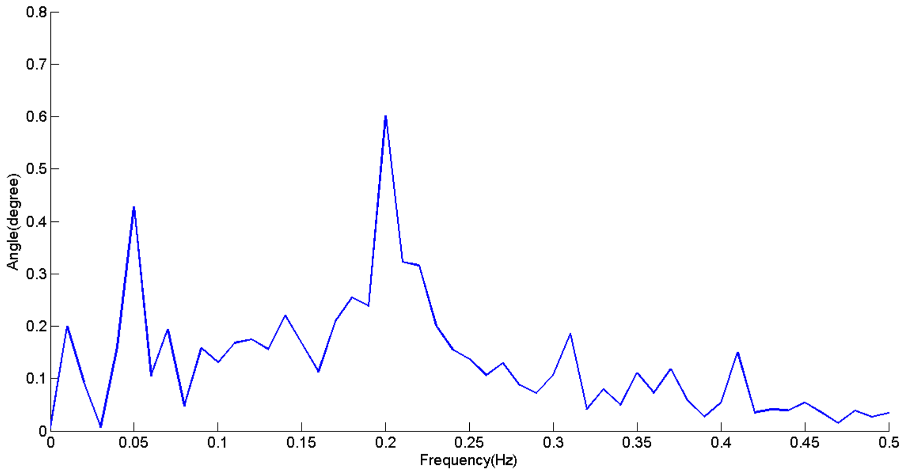


Figure 6.1: FFT of 1.2 degrees 0.05 Hz perturbation of tilt platform and, relative CoM angular displacements patient 5 at eyes open condition

patients are within their stability limits or close to its boundaries, it is unconvincing to consider these movements as bad movements or instability. Recently, some studies suggest that these movement variabilities could have been used by the subjects to detect or explore their stability boundaries [40]. In addition, Wu, in 2014 conducted an experiment such that subjects have tried to reach a specific target with their arms when their eyes closed. The authors have measured movement variability of subjects in a dimension that is irrelevant to the trajectory of the target. The authors hypothesized that if the movement variability is an exploration, then subjects who have shown higher movement variability should also have higher learning rate. If it is not an exploration but simply noise, it should have not affected the learning rate. They found that the learning rates of subjects are significantly correlated with their movement variability [47]. We suggest that movement variability of patient 5 could also be exploratory. Besides, it was argued that stability boundaries of the body have internal representation in CNS [25] and several factors like aging or basal ganglia disorders could affect the accuracy of this internal representation [41]. Although, there are no clear information and study about the effect of vestibular loss to the internal representation of stability limits of CoM, lack of vestibular information may have effects on its accuracy. Thus, the purpose of movement variability of patient 5 could be the exploration of her stability boundaries.

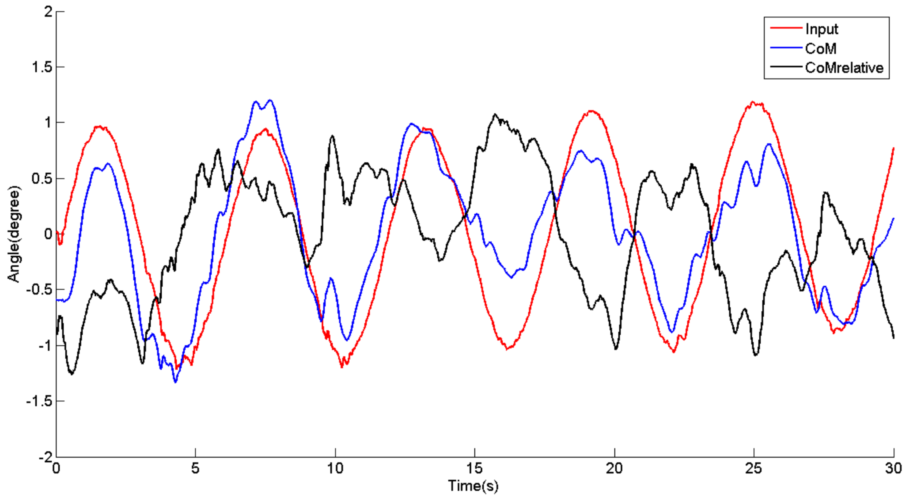


Figure 6.2: The first trial of patient 1 at eyes closed, 1 degrees 0.17 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject

Although, there are no significant differences between the response of subjects respect to space at low frequency, increase in frequency introduce some different postural characteristics between control and patient groups [28]. At f=0.17 Hz, both groups have very close gain values for absolute CoM motion when their eyes are open. Also, there are no significant differences between their RMS and coherence values at eyes open conditions. However, patients have large body sways, and so their gain values for the absolute CoM motion is higher than control group's at eyes closed condition. It can be argued that patients use their vision for stabilization of CoM motion at high frequency. If the visual information is absent, they tend to show large body sways. On the contrary to the patient group, control group continue to stabilize their CoM motion even in the absence of vision. It is possible that control group has been re-weighting their sensory information according to the available sensory information like vestibular feedback and continue to balance their CoM motion [35]. Patients also use sensory re-weighting when their eyes open, and compensate the lack of vestibular feedback to stabilize their CoM motion. However, patients failed to re-weight their available sensory information in the absence of vision. It is possible that somatosensory cues could also be inaccurate and/or insufficient at high frequency, unlike low frequency. Also, the interaction between vision and somatosensation is not well known. The somatosensation could be disrupted in the absence of vision. Therefore, proprioception is the only available sensory input for patients and is not enough to reduce postural sways at high frequency. There was one patient who continued to stabilize CoM motion even at eyes closed condition. Patient 1 had no large body sways, on the contrary, she successfully reduced the gain of absolute CoM motion in the absence of vision at f=0.17 Hz. How can she avoid large body sways at eyes closed condition, but the other patients not? The phase difference between subjects CoM motion and perturbation of platform can give an answer to this question. Patient 1 has phase lead at all 12 trials in contrast to the other patients. It can be claimed that she learned 1 degrees 0.17 Hz perturbation of platform at eyes open condition and successfully anticipated motion of platform at eyes closed condition. Besides, since perturbation of platform is very similar and straight forward (stereotyped) across frequencies, it is possible that she has learned motion of platform at the lower frequency and has used this information at the higher frequency. However, the first trial of patient 1 is 1 degrees 0.17 Hz perturbation of tilt platform at eyes closed condition as demonstrated in Fig.6.2. It is clear that she has started to anticipate perturbation of tilt platform at f=0.17 Hz and eyes closed condition. Since somatosensation and proprioception are only available sensory tools for patient 1 at this experimental conditions, she might have used feedback from these sensory systems. Nashner et al.(1976) conducted an experiment about stabilizing role of stretch reflex in postural control. He reported that the role of stretch reflex to stabilize body sway during stance could be altered: augmented if to be useful versus inhibited if is harmful or inappropriate. Some of the subjects in his experiment used long latency stretch reflex to help to reduce their postural sway [30]. These results reported by Nashner may give the answer to how patient 1 had successfully reduced her postural sway.

In addition, the patient group has a high standard deviation of RMS, magnitude, and phase angle values. It shows that patients have high intragroup variability and their response characteristics are heterogeneous. On the other hand, control group has low standard deviation for these metrics, and their responses are homogeneous. One of the reasons for intragroup variability of patients could be that the age range of patients is higher than the control group in this study, so also their standard deviation of age (see table 3.1). It was argued that aging has a very substantial effect on postural control and elderly people face postural instability because of aging [37]. However, this intragroup variability could also be an outcome of the vestibular loss. There is few hypothesis about how CNS handle the loss of sensory information to maintain postural balance. The sensory re-weighting is the most known among these hypotheses. The central principle of sensory re-weighting is based on closed loop control of posture stability. It claims that input from each sensory system is multiplied by some weight and CNS use the summation of these weighted inputs to maintain postural balance or produce a response to evoked forces. When one sensory input is absent, or accuracy of a particular sensory input is unreliable, other available or more reliable sensory inputs become more heavily weighted [4, 27, 35]. Horak, 2009 argued that the degree and accuracy of sensory re-weighting depend on the ability of a person to compensate sensory loss with other available sensory inputs. In this study, some patients with bilateral vestibular loss compensate their sensory loss better than other patients. It was shown that these patients have more complete bilateral vestibular loss than other patients with measurable vestibulo-ocular reflexes [16]. Therefore, the intragoup variability of patients in our study could be correlated to the ability of patients to compensate their vestibular sensory loss.

6.1 Future Works

In this study, it has been shown that some patients have movement variability that are unrelated to perturbation of platform which are not considered in detail. For future research, this movement variability can be studied in the frequency domain. In addition, it is very important to find a metric to measure the effect of movement variability on the response of patients such as learning rate or improvement in patient postural response to sinusoidal tilts. This metric may give us cues about whether movement variability is an instability (malign movement manifold) or an exploration (benign movement manifold).

Furthermore, some patients do not have large body sways at high frequency and eyes closed conditions contrary to other patients. We made some suggestions about this phenomenon such as modulation of stretch reflex and use of remaining sensory inputs from somatosensory and proprioception. The Electromyography (EMG) -especially monitoring and/or probing the modification of the stretch reflex- can be added to data collection to measure the electrical activity of muscles to analyse the effect of stretch reflex on this phenomena. The force plate data also can bu used to investigate the use of somatosensory inputs by patients.

The movement strategies of subjects also can be analysed in detail by calculation of subjects' joint movements and their variability. The effect of aging on postural stability is a well-known topic in postural control studies. Therefore, another consideration for future works should be that reduce to the standard deviation of the age of subjects. This also aids a more detailed discussion about the intragroup variability of the patient group.

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APPENDIX A

QUATERNIONS

In mathematics, quaternions are a non-commutative four-dimensional number system that extends two-dimensional complex numbers. They were discovered by Irish mathematician Sir William Rowan Hamilton in 1843 and applied to mechanics in three-dimensional space. Quaternions are very useful tools for describing rotations in three-dimensional space. A quaternion q can be described as,

$$q = q_0 + q_1 i + q_2 j + q_3 k$$

where q_0, q_1, q_2, and q_3 are real numbers and i, j, and k are complex part of quaternion.

A.1 Quaternion Algebra

The fundamental equation of quaternions were described by Hamilton as,

$$i^2 = j^2 = k^2 = ijk = -1 \tag{A.1}$$

and implicit in these equation is that

$$ij = i \times j = k$$

$$jk = j \times k = i$$

$$ki = k \times i = j$$

A.1.1 Addition and Multiplication

Consider quaternions q defined as above and p,

$$p = p_0 + p_1 i + p_2 j + p_3 k$$

Addition of two quaternion acts component wise,

$$q + p = q_0 + p_0 + (q_1 + p_1)i + (q_2 + p_2)j + (q_3 + p_3)k \tag{A.2}$$

The product of these quaternions should satisfies the fundamental equations of quaternions defined by Hamilton. Then quaternion multiplication $\otimes$ is that

$$\begin{aligned} q \otimes p &= (q_0 + q_1 i + q_2 j + q_3 k)(p_0 + p_1 i + p_2 j + p_3 k) \\ &= q_0 p_0 - (q_1 p_1 + q_2 p_2 + q_3 p_3) + q_0(p_1 i + p_2 j + p_3 k) + p_0(q_1 i + q_2 j + q_3 k) \\ &+ (q_3 p_2 - q_2 p_3)i + (q_1 p_3 - q_3 p_1)j + (q_2 p_1 - q_1 p_2)k \end{aligned}$$

If one insert the inner product and cross product of two vectors $\vec{q}$ and $\vec{p}$

$$\begin{aligned} \vec{q} &= q_1 i + q_2 j + q_3 k \\ \vec{p} &= p_1 i + p_2 j + p_3 k \end{aligned}$$

into the above equation, one can obtain following equation

$$q \otimes p = q_0 p_0 - \vec{q} \cdot \vec{p} + q_0 \vec{p} + p_0 \vec{q} + \vec{q} \times \vec{p} \tag{A.3}$$

Note that quaternion multiplication is not commutative unlike two-dimensional complex numbers so,

$$q \otimes p \neq p \otimes q$$

A.1.2 Complex Conjugate, Norm, and Inverse

One can rewrite the quaternion q such that

$$q = q_0 + \vec{q}$$

The complex conjugate of the quaternion q is that

$$q^* = q_0 - \vec{q} \tag{A.4}$$

and it follows that

$$q + q^* = 2q_0$$

$$qq^* = q^* q = q_0^2 + q_1^2 + q_2^2 + q_3^2$$

or in other from

$$qq^* = q^*q = q_0^2 + \|\vec{q}\|^2 \tag{A.5}$$

The norm of quaternion q is defined as

$$\|q\| = \sqrt{q^*q} = \sqrt{qq^*} \tag{A.6}$$

The definition of inverse of a quaternion q is that

$$q^{-1}q = qq^{-1} = 1$$

and it follows that

$$q^{-1} = \frac{q^*}{\|q\|^2} \tag{A.7}$$

If a norm of quaternion q is equal to 1, it is called a unit quaternion.

$$\|q\| = \|q^*\| = \|qq^*\| = 1$$

Then inverse of a unit quaternion is equal to its conjugate

$$qq^{-1} = qq^* = 1$$

$$q^{-1} = q^*$$

Lets consider to the equation A.5 for a unit quaternion

$$qq^* = q^*q = q_0^2 + \|\vec{q}\|^2 = 1$$

It implies that an unit circle exists and this unit quaternion can be rewritten in polar coordinates such that

$$q = cos(\theta) + \vec{n}sin(\theta)$$

where the unit vector $\vec{n}$ and angle θ are defined as

$$\vec{n} = \frac{q}{\|\vec{q}\|}$$

$$tan(\theta) = \frac{\|\vec{q}\|}{q_0}$$

Note that a general quaternion q can be rewritten as in the form of a unit quaternion such that

$$q = \|q\|(cos(\theta) + \vec{n}sin(\theta))$$

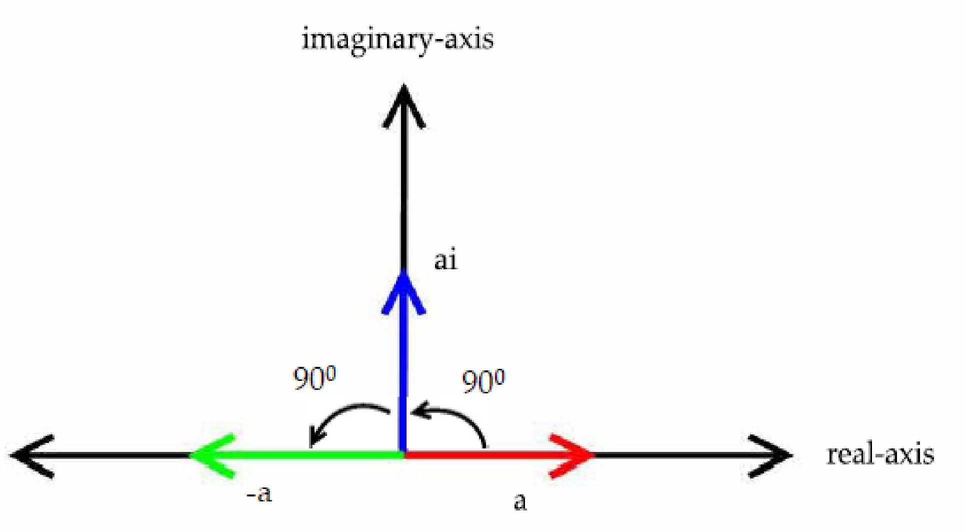


Figure A.1: Complex plane and rotations

A.2 Complex Numbers and Rotations

To better understand quaternions and how they represent rotations in three dimensions, two-dimensional complex numbers are a good example. Consider a real number a on the real-axis which initially have $0°$ rotation respect to real-axis as seen in Fig.A.1. If one multiply a by -1, it will rotate counter-clockwise by $180°$. How can one rotate number a about $90°$ respect to real-axis? Let's assume that if one multiply a by number b, a will rotate counter-clockwise by $90°$. Then, one more multiplication of a by b cause total rotation of $180°$ in counter-clockwise. Therefore,

$$b * b * a = -1 * a$$

$$b^2 = -1$$

$$b = \sqrt{-1} = i$$

If one multiply a number by complex number i, it will rotate counter-clockwise by $90°$. Thus, a rotation can be represented as a complex number.

A complex number z has a real part and an imaginary part

$$z = x + iy$$

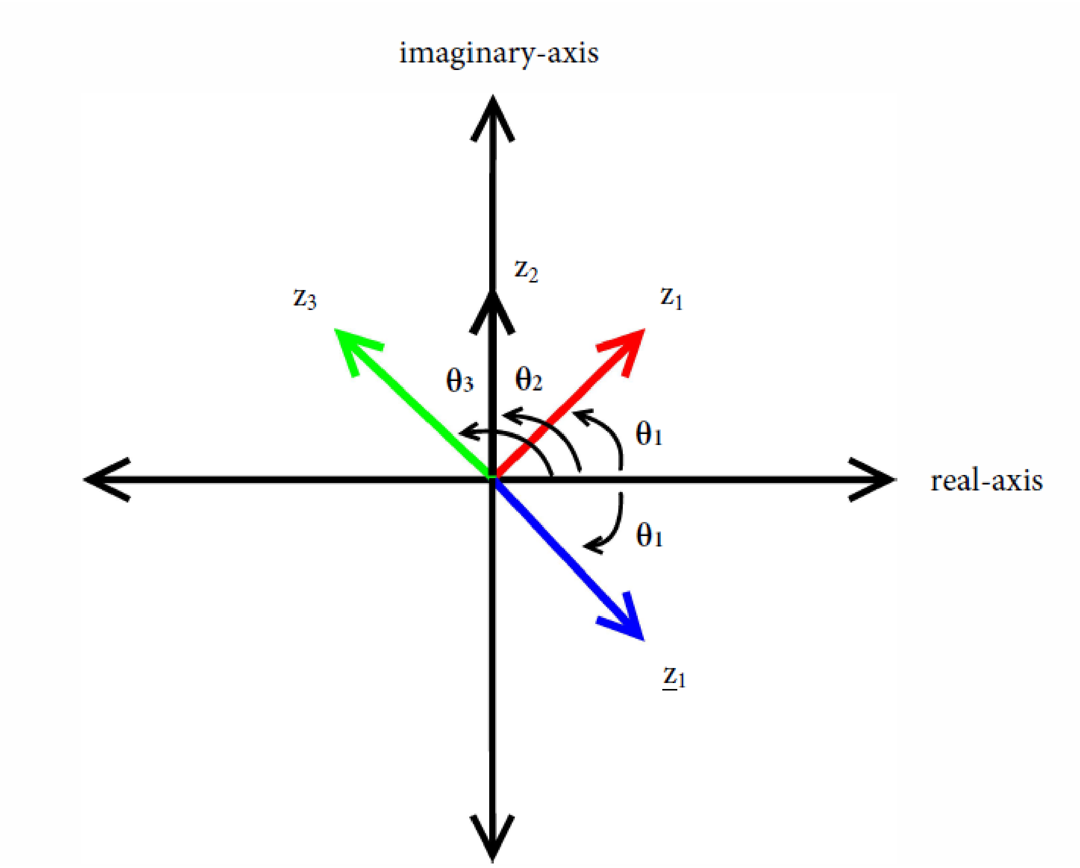


Figure A.2: An example of rotations in Complex Plane

Every complex number can be represented in polar coordinates by using Euler's famous formula

$$e^{i\theta} = cos(\theta) + isin(\theta)$$

such that

$$z = e^{i|z|\theta} = cos(|z|\theta) + isin(|z|\theta)$$

where

$$|z| = \sqrt{x^2 + y^2}$$
$$\theta = atan2(y, x)$$

A complex number z represents a rotation by θ about real-axis. If length of this complex number, $|z|$ is equal to 1, the multiplication by z does not change length of any complex number and it only rotate it by $90°$. Any real number can be rewritten as complex number with zero imaginary part.

Fig.A.2 shows some examples of rotations in complex plane. z_1, z_2, and z_3 are complex numbers and their absolute values are equal to 1. Also, $\underline{z}_1$ is complex conjugate of z_1 and it represents clockwise rotation by angle θ_1 contrary to z_1 that represents counter-clockwise rotation by angle θ_1. If one multiply complex number z_2 by z_1, it will rotate counter-clockwise by angle θ_1 and $z_2 * z_1$ represents total rotation of $\theta 1 + \theta 2 = \theta 3$. Similarly if one multiply z_3 by $\underline{z}_1$, it will rotate clockwise by angle θ_1. Therefore, $z_3 * \underline{z}_1$ represents total rotation of $\theta 3 - \theta 1 = \theta 2$ respect to real-axis.

A.2.1 Quaternion Rotation Operator

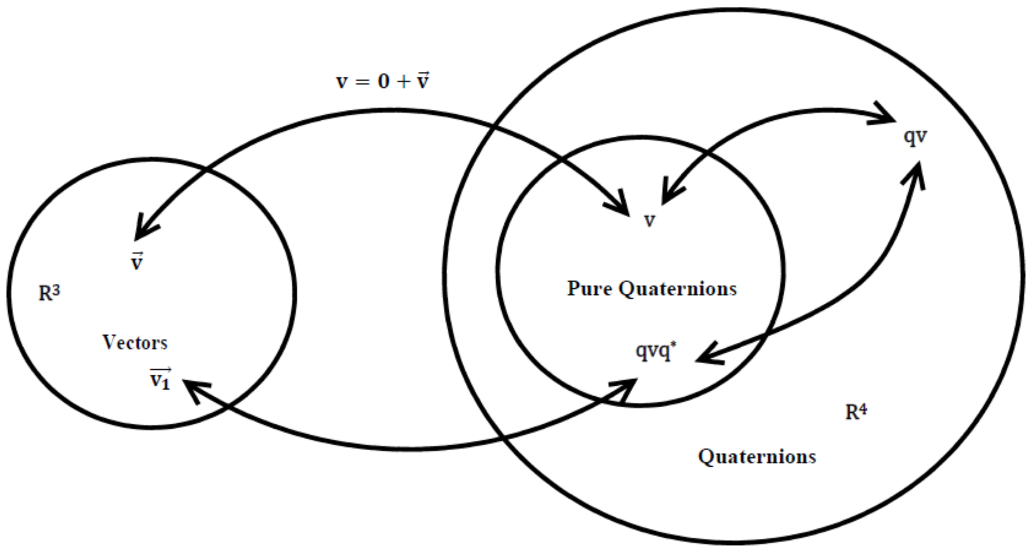


Figure A.3: Quaternion operations on vectors

A quaternion represents the rotation by angle θ about its unit axis $\vec{n}$ in three dimensions. Any vector $\vec{v}$ can be represented as a quaternion v whose real part is zero and it is called pure quaternion. Using a unit quaternion

$$q = cos(\theta) + \vec{n} sin(\theta$$

one can rotate any vector $\vec{v}$ in R^3 such that

$$L(v)_q = qvq^* = \vec{v_1}$$

where $\vec{v_1}$ is rotation of vector $\vec{v}$ through an angle 2θ about axis of the unit vector $\vec{n}$ and L_q is quaternion rotation operator. Note that, a unit quaternion does not change the length of rotated vector .

Fig.A.3 shows operations of quaternions on vectors in R^4. When a pure quaternion v multiply by unit quaternion q, it will rotate through an angle θ about the axis of the unit vector $\vec{n}$ and it will have a non-zero real part. Since it is no longer a pure quaternion, it can not return to R^3 space directly. If one multiply qv by the conjugate of unit quaternion q^*, the total rotation of $\vec{v}$ will be through an angle 2θ about the axis of the unit vector $\vec{n}$. Now, this product is equal to v_1, and it is a pure quaternion. It can return to R^3 space as a vector $\vec{v_1}$. Note that conjugate of quaternion acts like a conjugate of the two-dimensional complex number. When one multiply a pure quaternion v by vq^*, vq^* represents a rotation $-\theta$ about $\vec{n}$. However, quaternion product is not commutative, and q^*v rotate pure quaternion v by angle θ about $\vec{n}$.

In addition, quaternion rotation operator L_q may be interpreted as a frame rotation such that

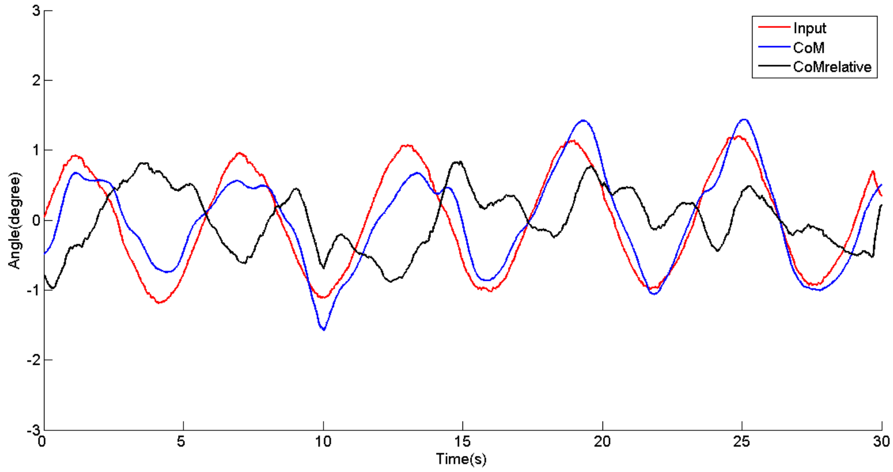
$$L(v)_q^* = q^*vq$$

L_q^* will rotate coordinate frame respect to vector $\vec{v}$ through angle 2θ about unit vector $\vec{n}$. [19]

APPENDIX B

FIGURES OF REMAINING SUBJECTS

In this appendix, time graph of absolute and relative CoM angular displacements for first trial of all subjects at different experimental conditions are presented. These experimental conditions are: 1 degrees 0.17 Hz and 1.2 degrees 0.05 Hz perturbation of tilt platform at eyes open and closed conditions.



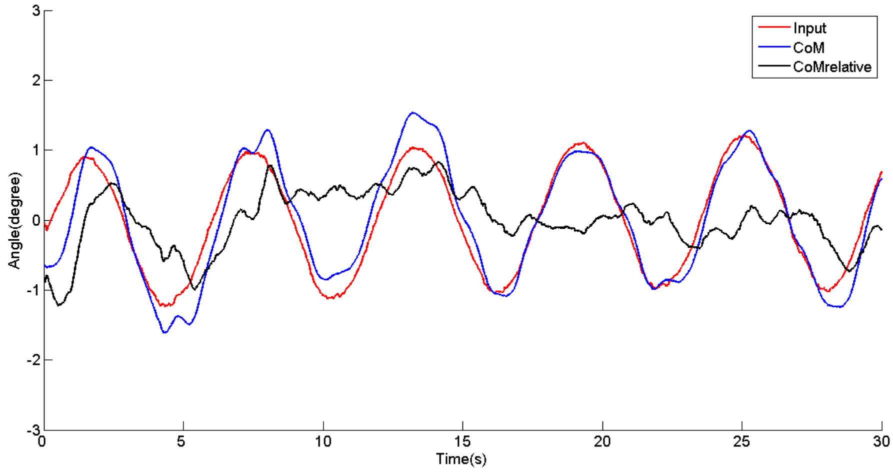


Figure B.1: Trial 1 for healthy subject 1, 1 degrees 0.17 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions

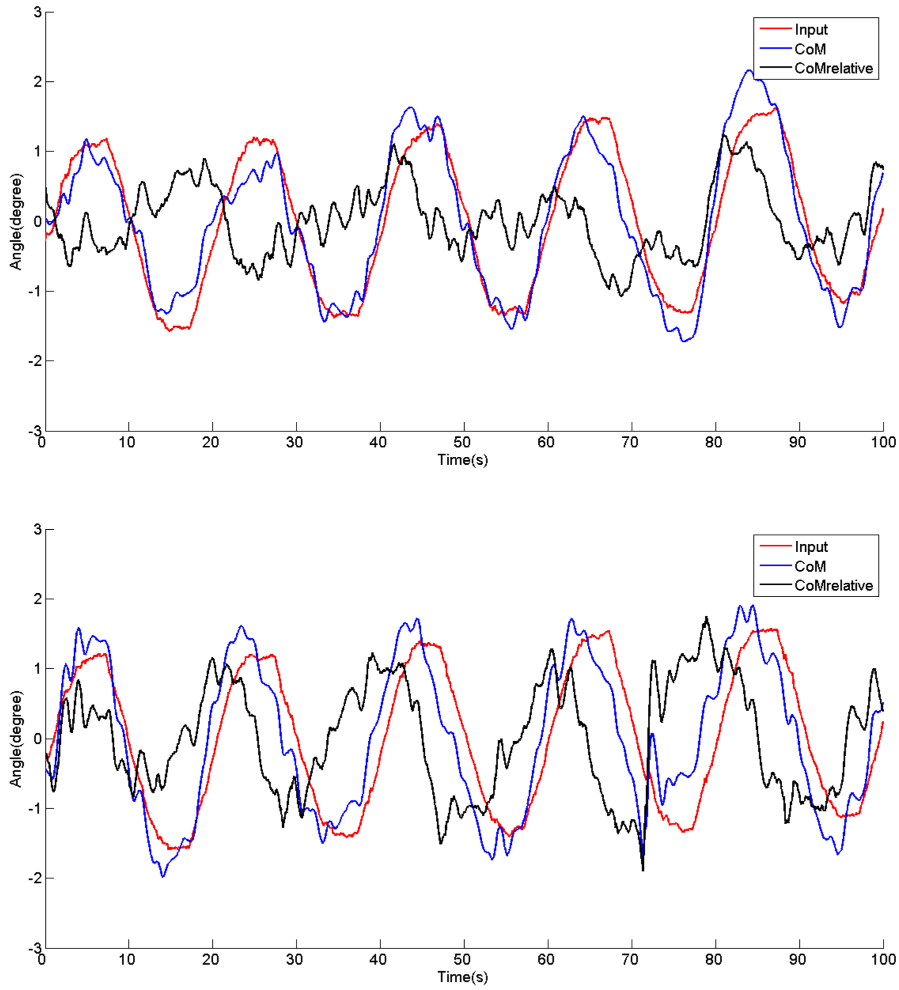


Figure B.2: Trial 1 for healthy subject 1, 1.2 degrees 0.05 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions

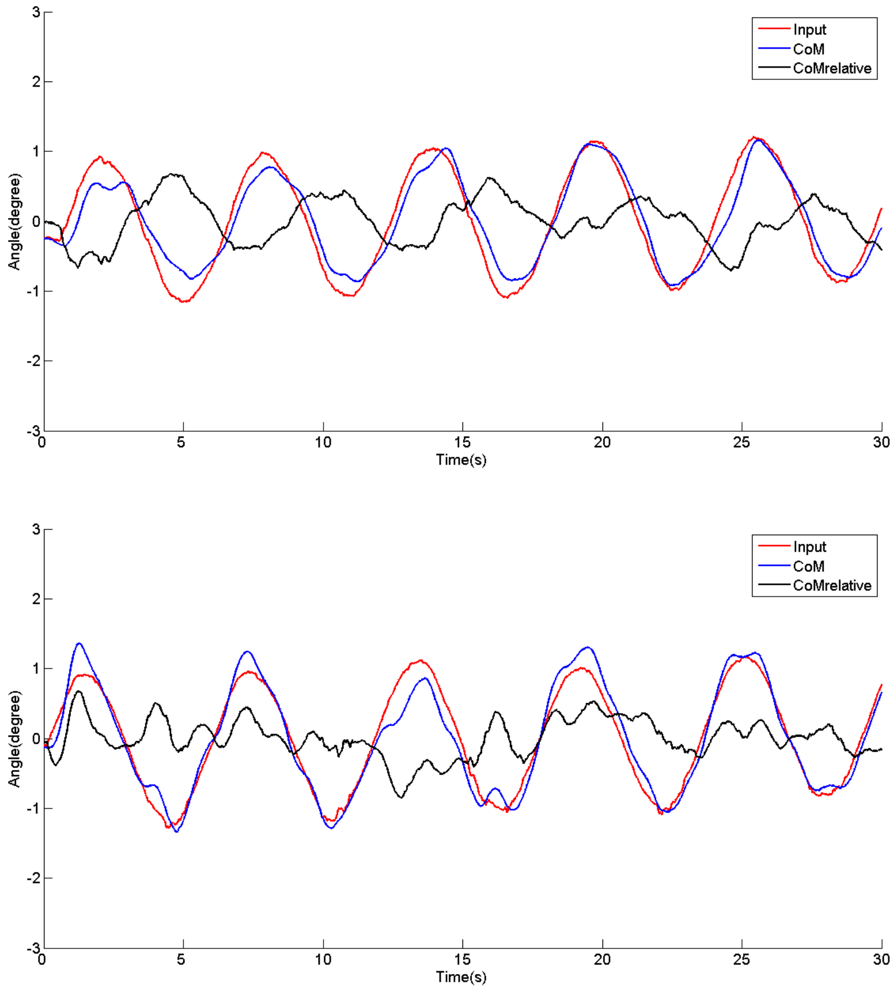


Figure B.3: Trial 1 for healthy subject 2, 1 degrees 0.17 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions

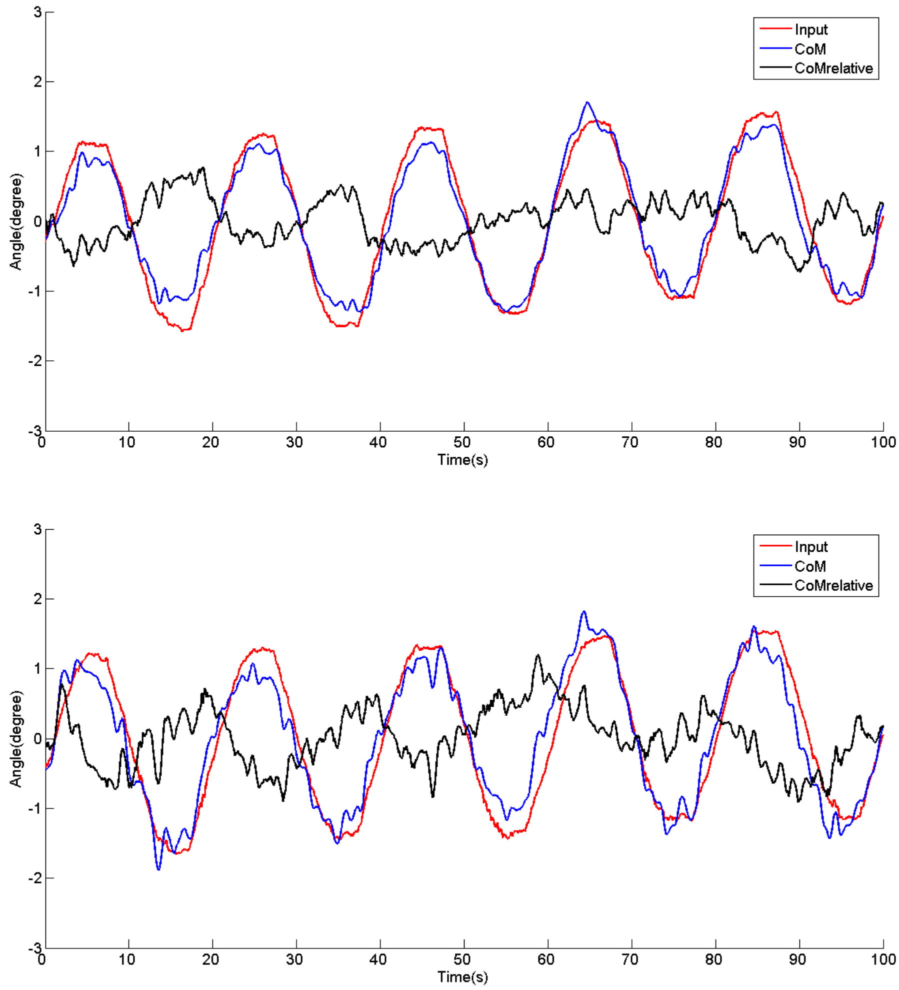
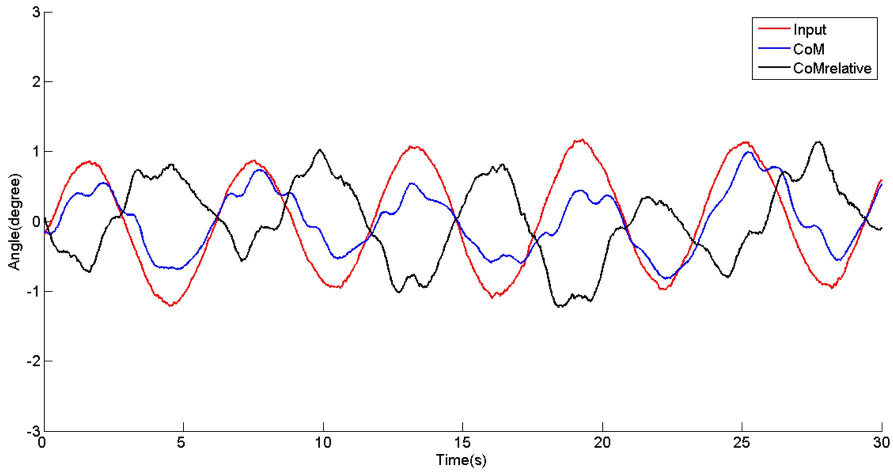


Figure B.4: Trial 1 for healthy subject 2, 1.2 degrees 0.05 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions



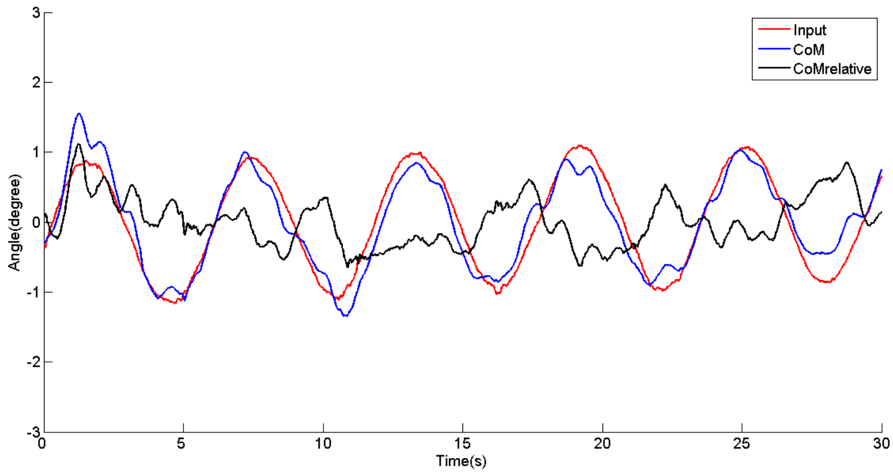


Figure B.5: Trial 1 for healthy subject 3, 1 degrees 0.17 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions

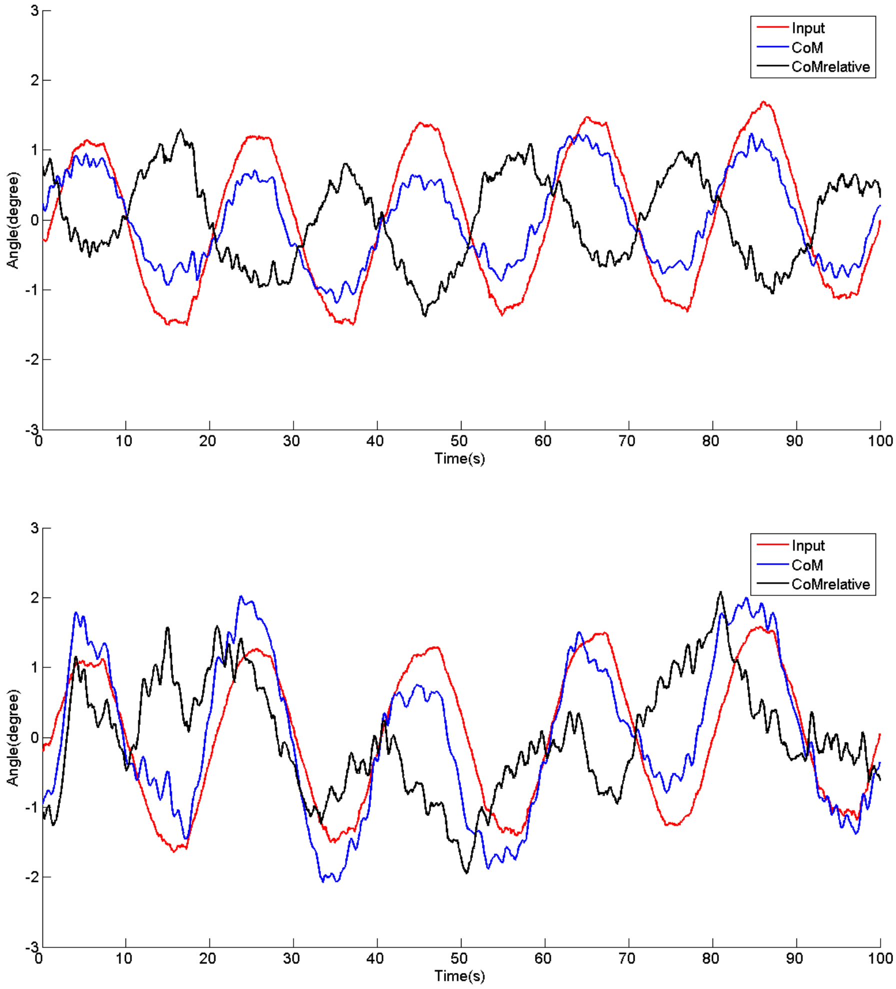
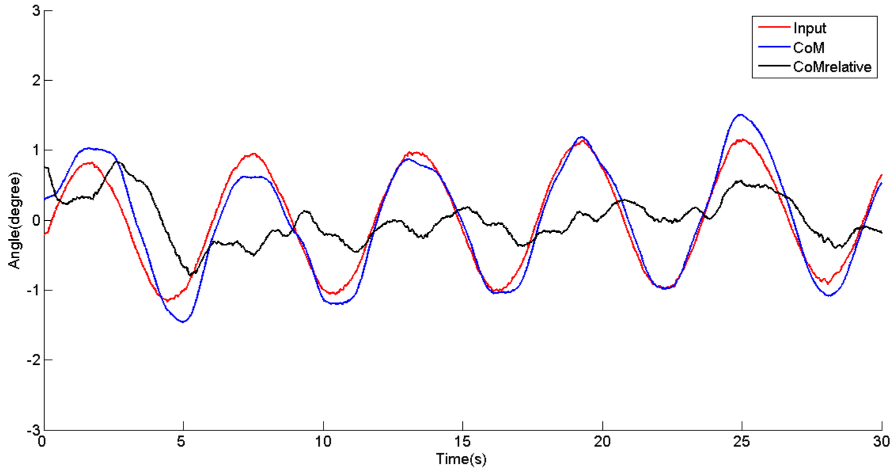


Figure B.6: Trial 1 for healthy subject 3, 1.2 degrees 0.05 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions



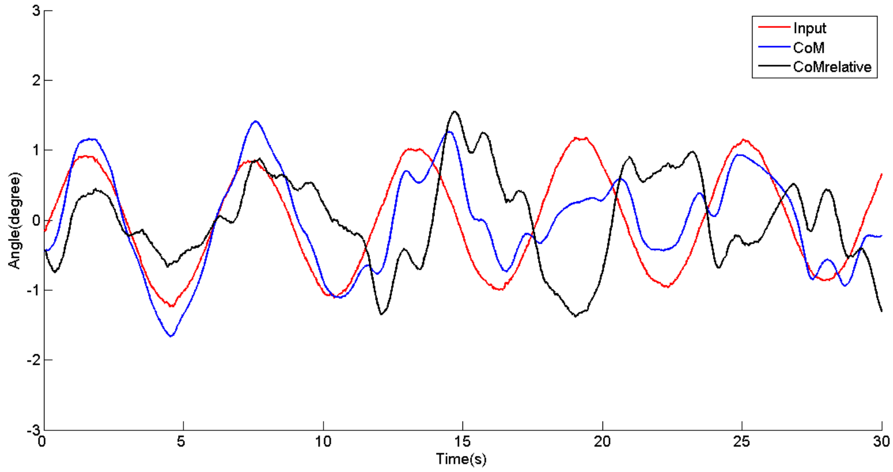


Figure B.7: Trial 1 for healthy subject 4, 1 degrees 0.17 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions

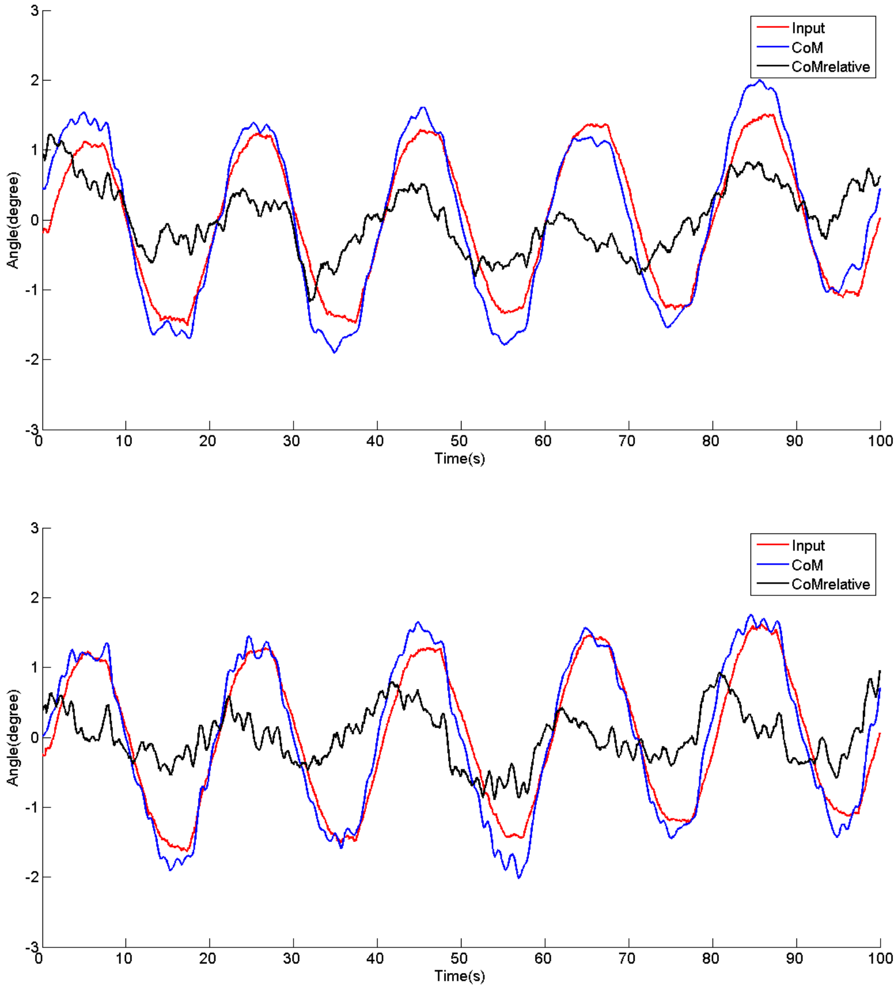
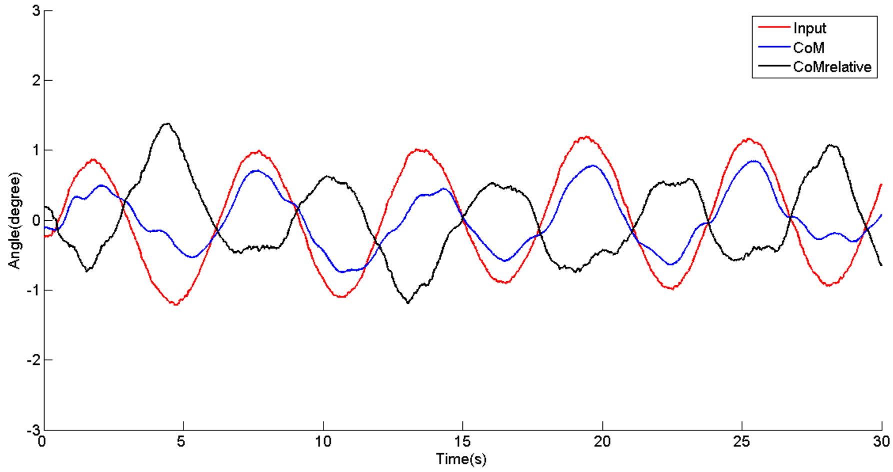


Figure B.8: Trial 1 for healthy subject 4, 1.2 degrees 0.05 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions



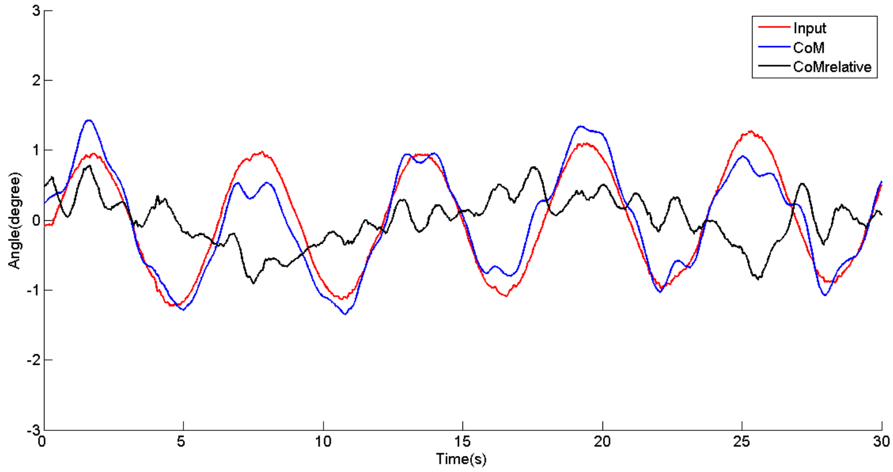


Figure B.9: Trial 1 for healthy subject 5, 1 degrees 0.17 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions

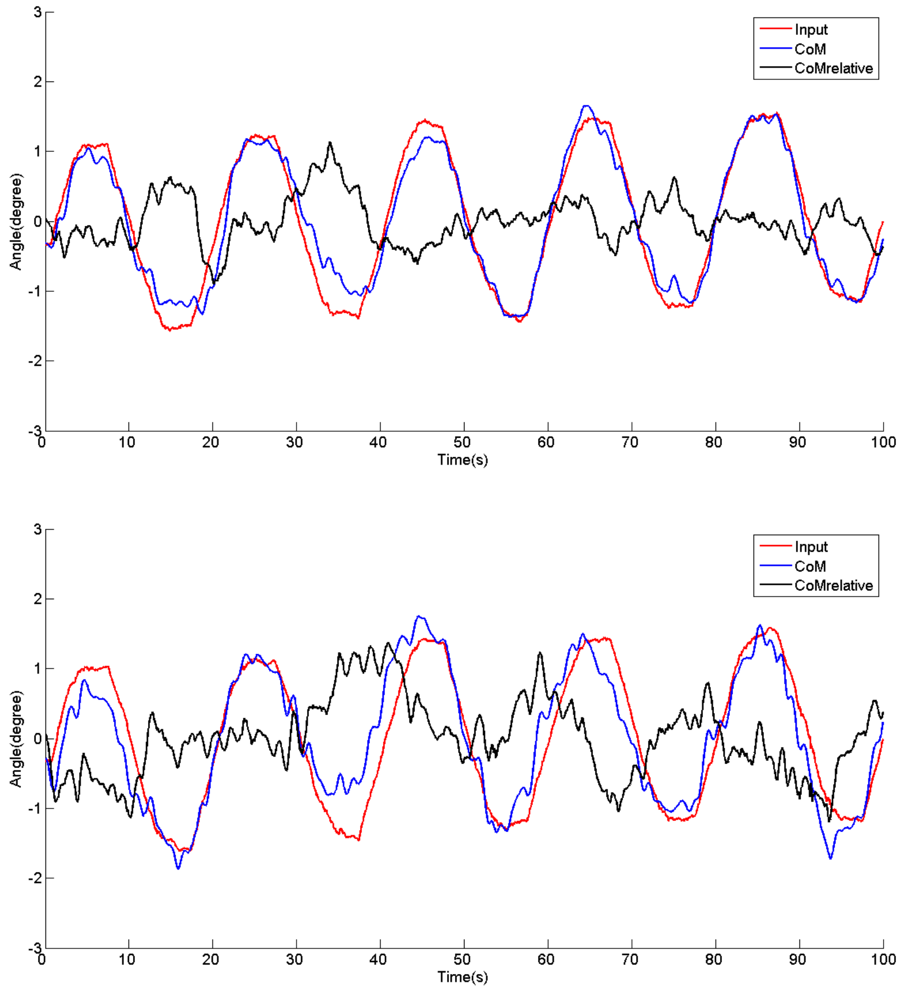


Figure B.10: Trial 1 for healthy subject 5, 1.2 degrees 0.05 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions

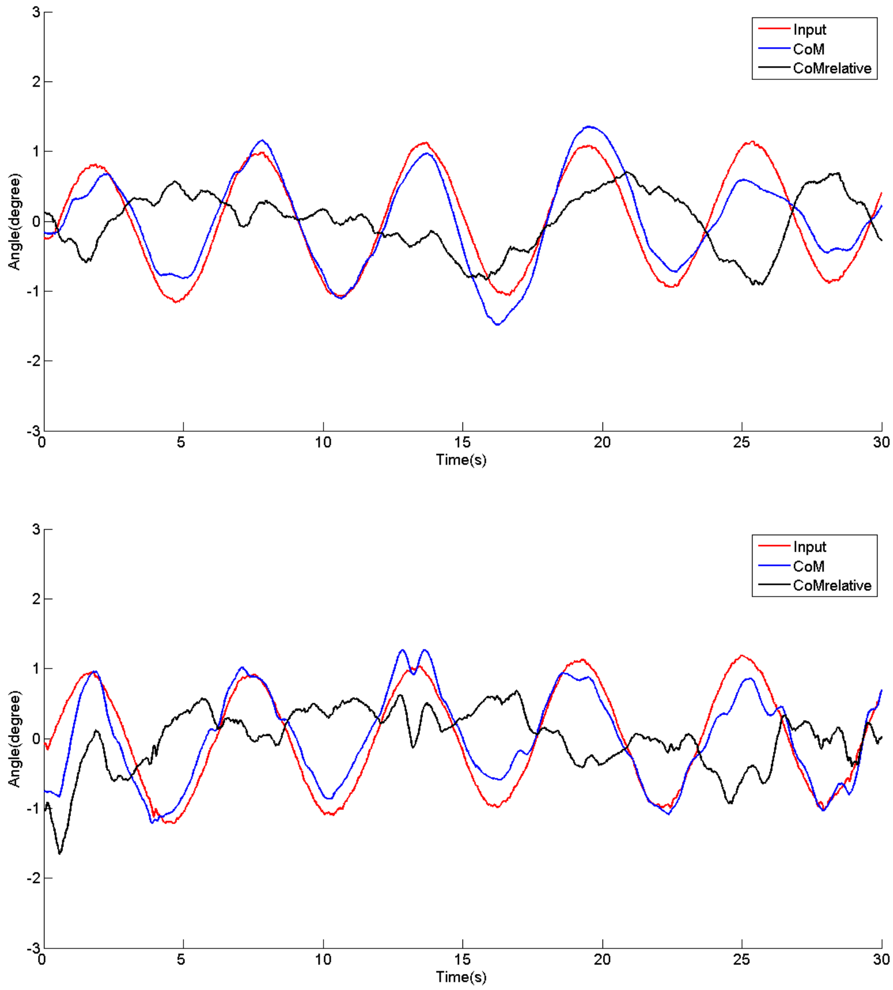


Figure B.11: Trial 1 for healthy subject 6, 1 degrees 0.17 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions

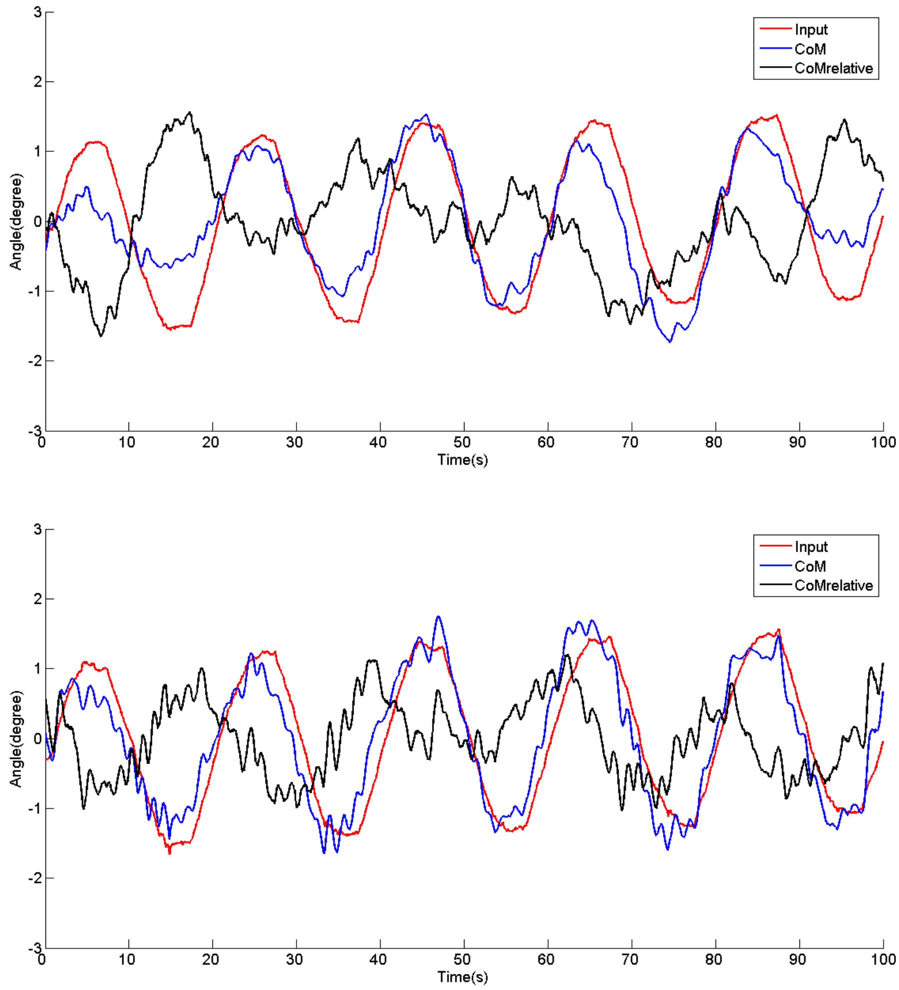


Figure B.12: Trial 1 for healthy subject 6, 1.2 degrees 0.05 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions

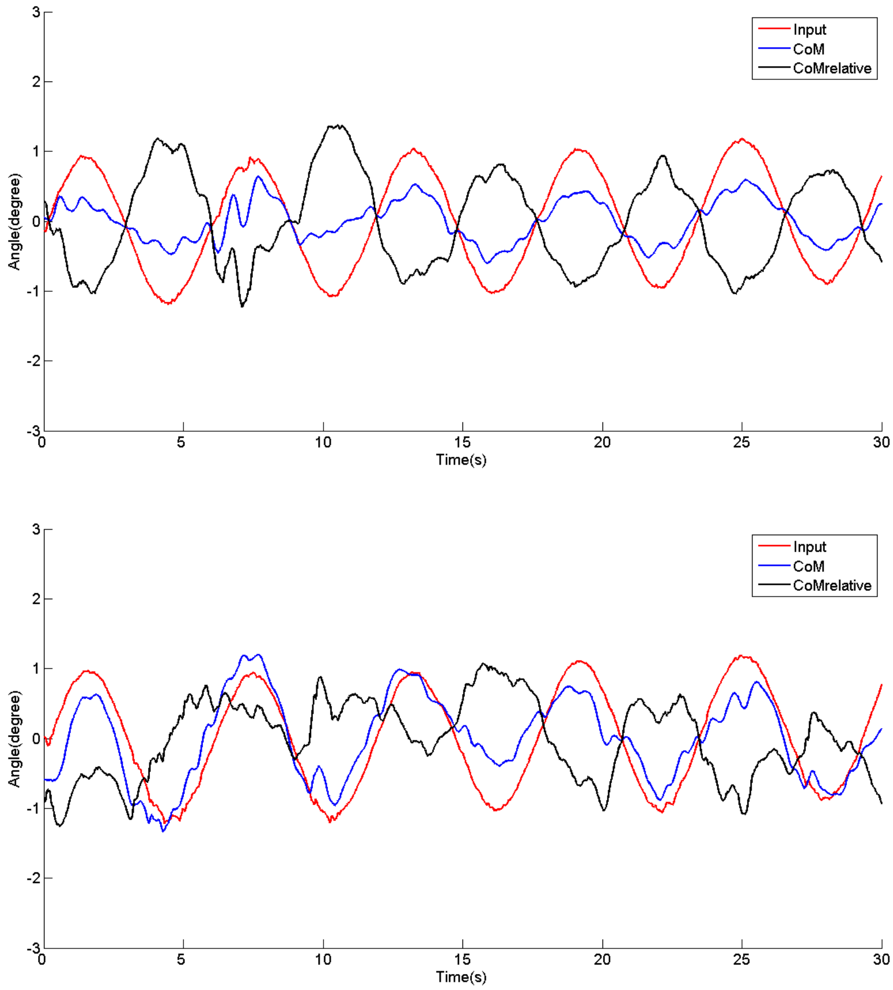
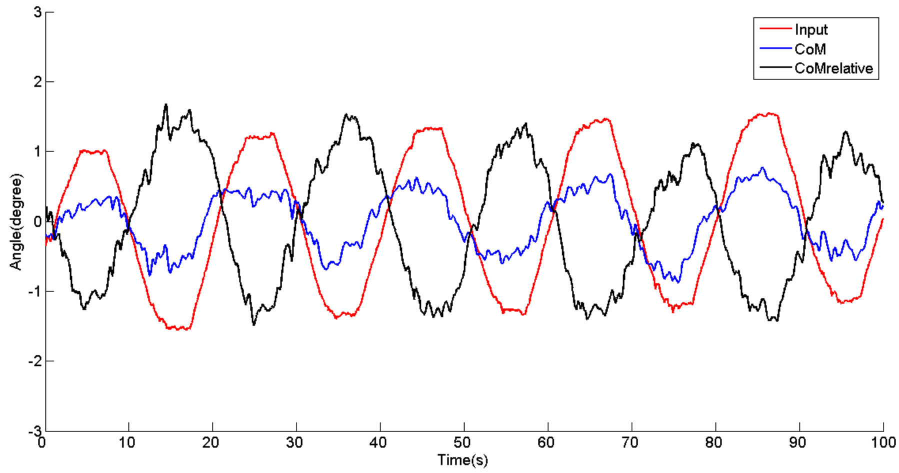


Figure B.13: Trial 1 for patient 1, 1 degrees 0.17 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions



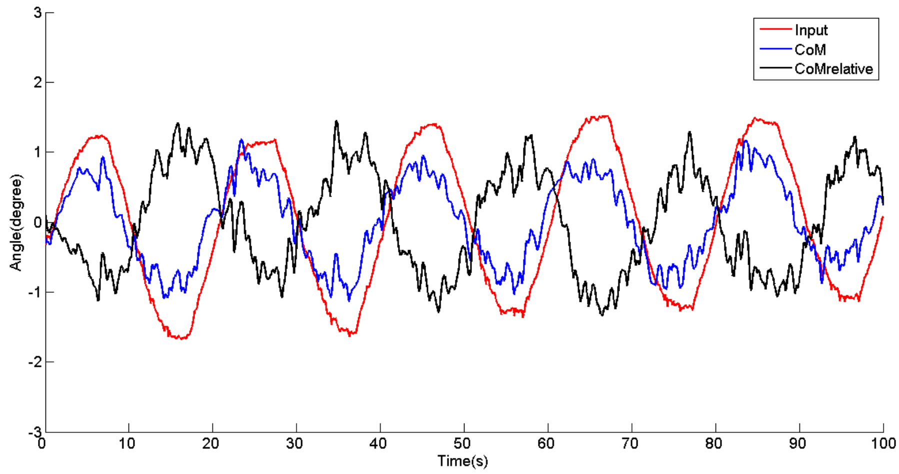


Figure B.14: Trial 1 for patient 1, 1.2 degrees 0.05 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions

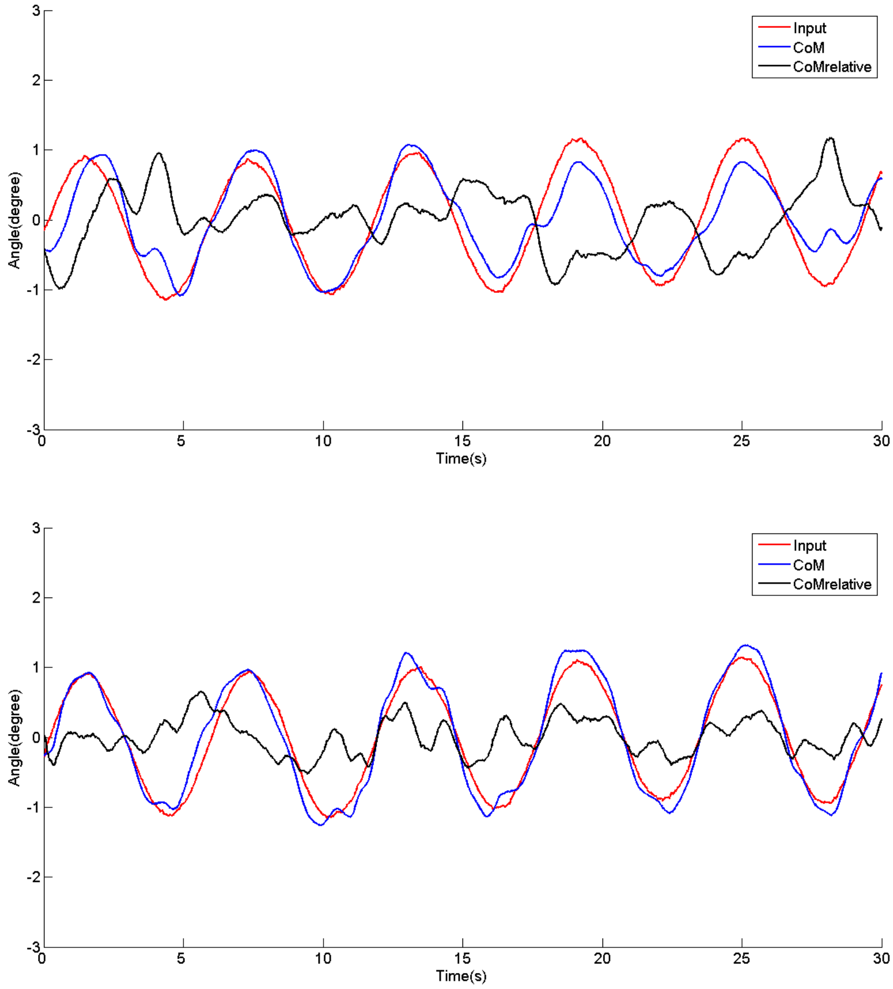


Figure B.15: Trial 1 for patient 2, 1 degrees 0.17 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions

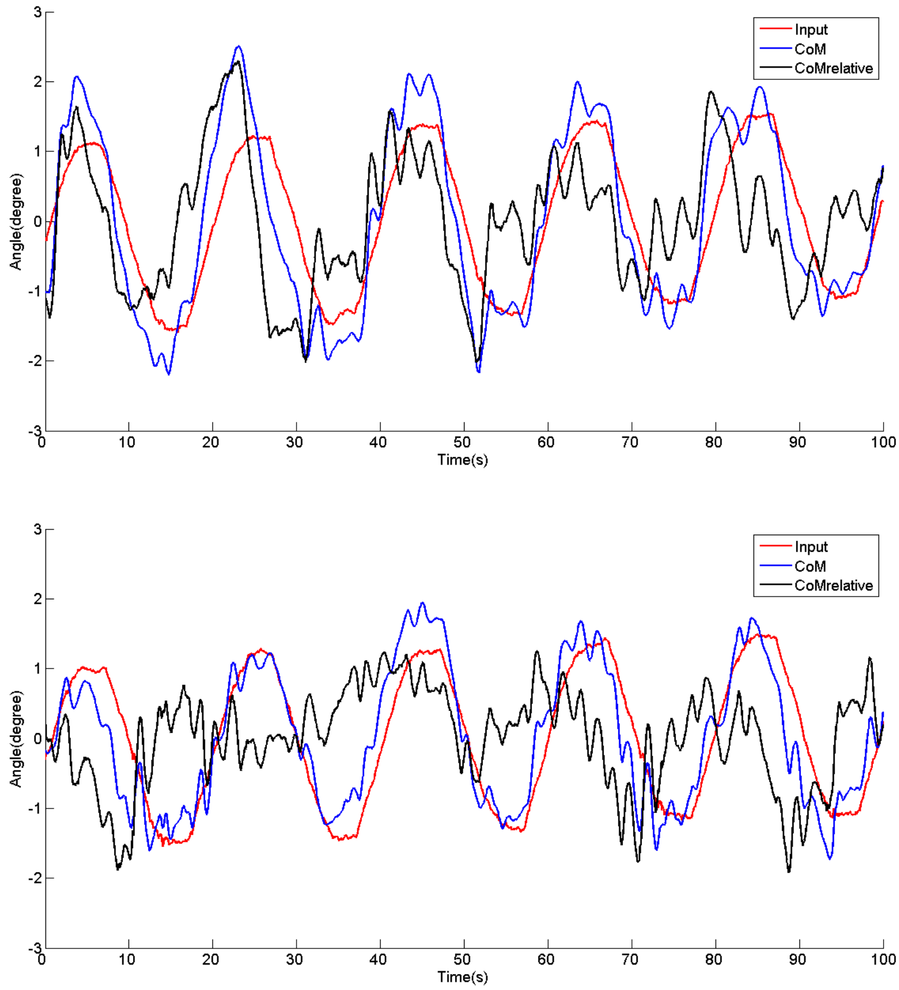
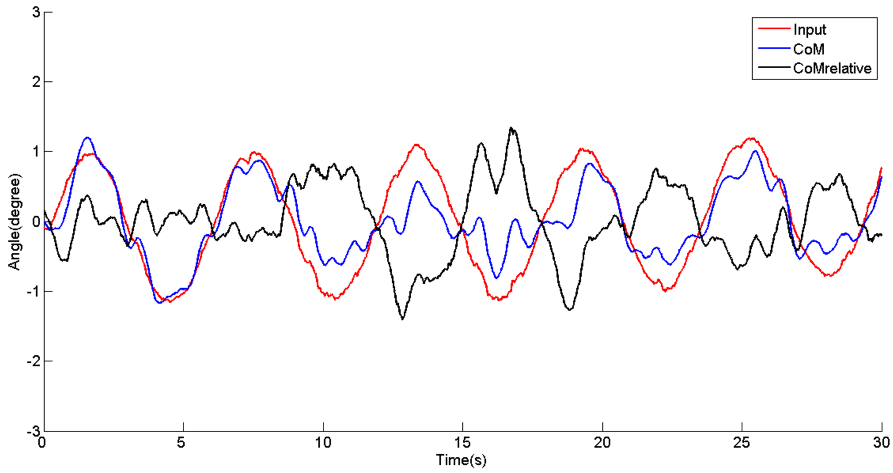


Figure B.16: Trial 1 for patient 2, 1.2 degrees 0.05 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions



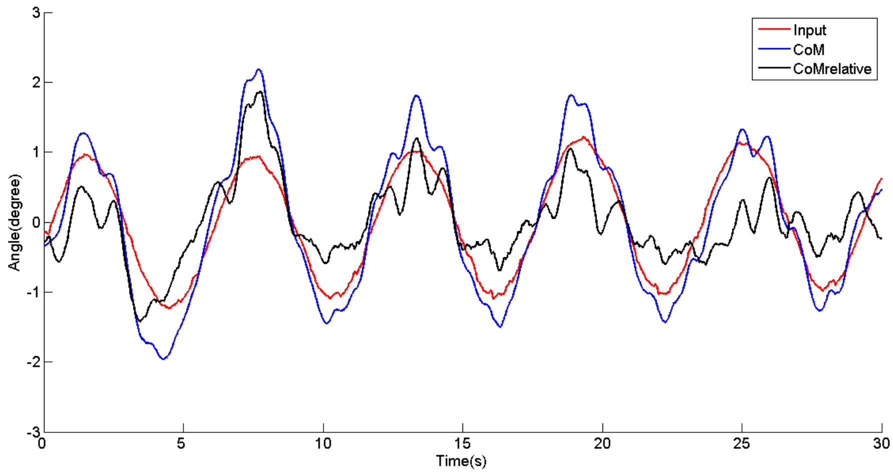
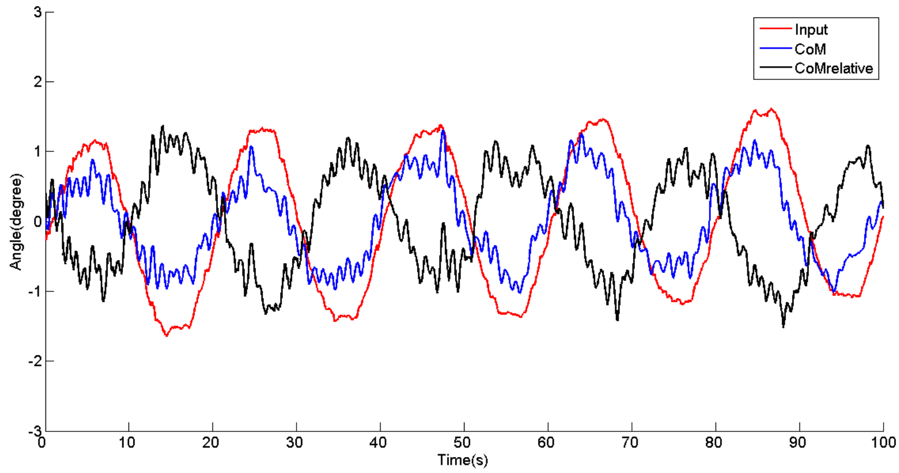


Figure B.17: Trial 1 for patient 3, 1 degrees 0.17 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions



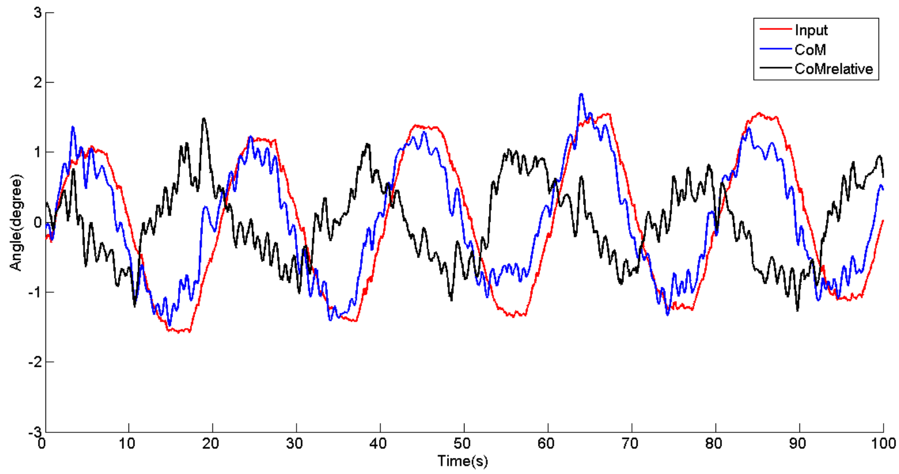


Figure B.18: Trial 1 for patient 3, 1.2 degrees 0.05 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions

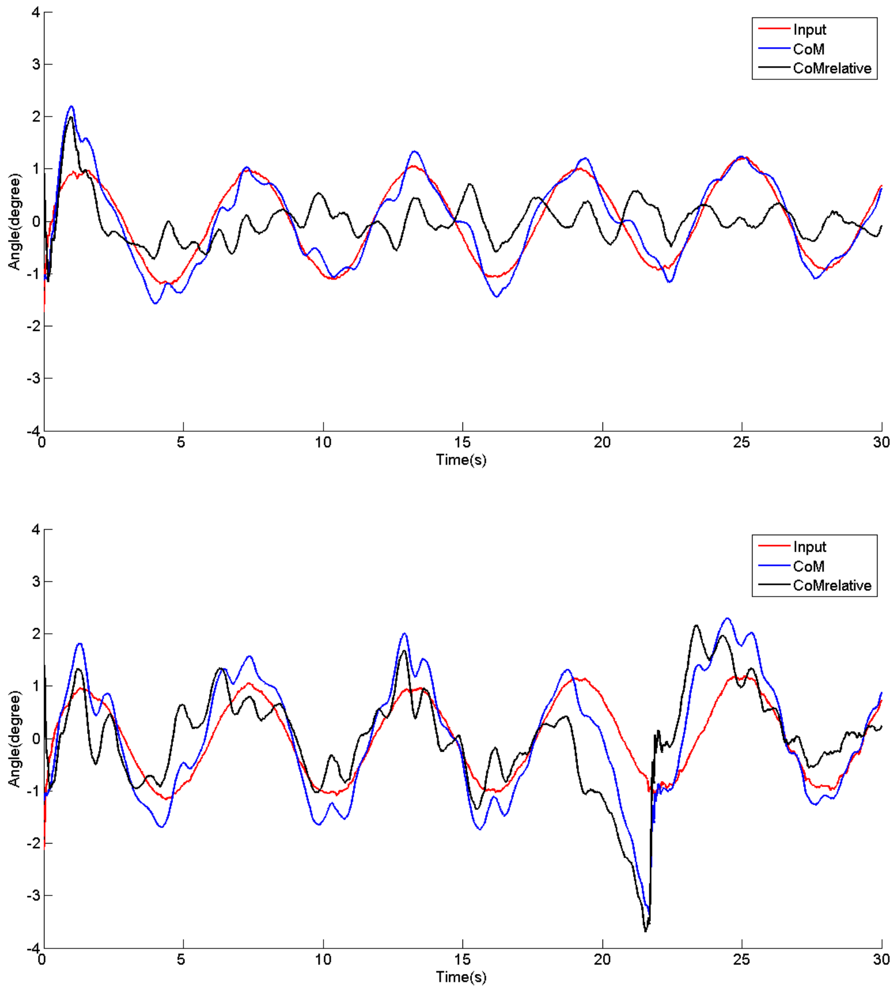
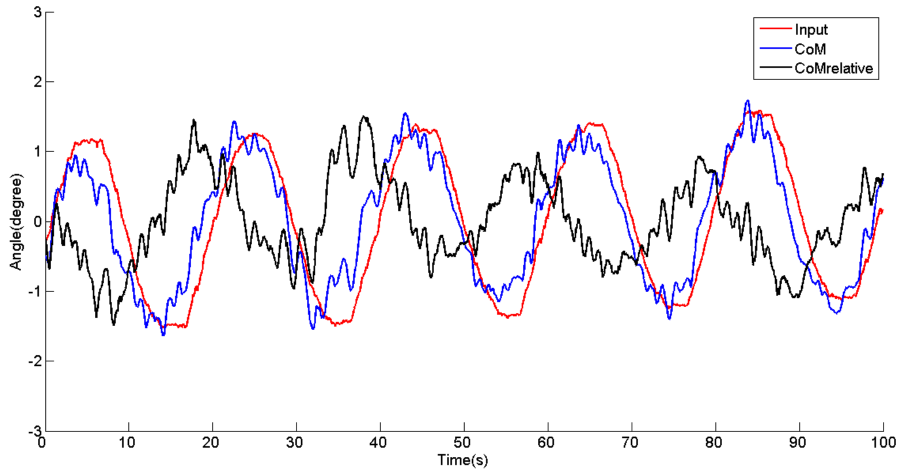


Figure B.19: Trial 1 for patient 4, 1 degrees 0.17 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions



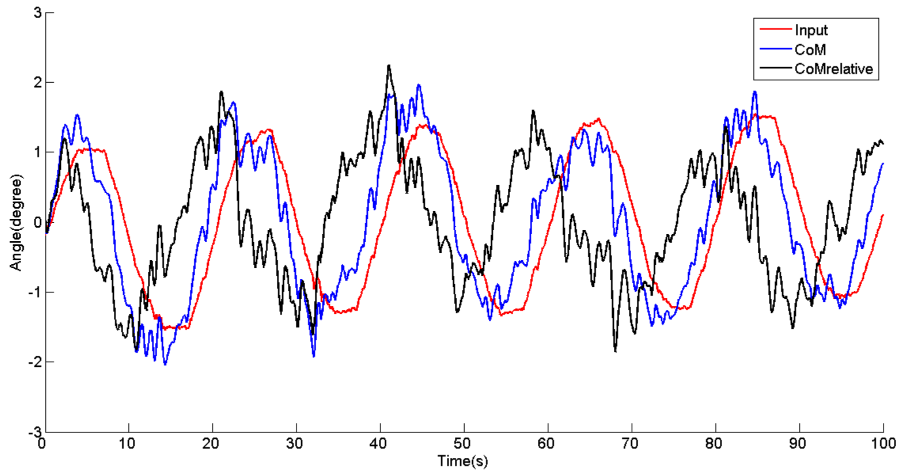


Figure B.20: Trial 1 for patient 4, 1.2 degrees 0.05 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions

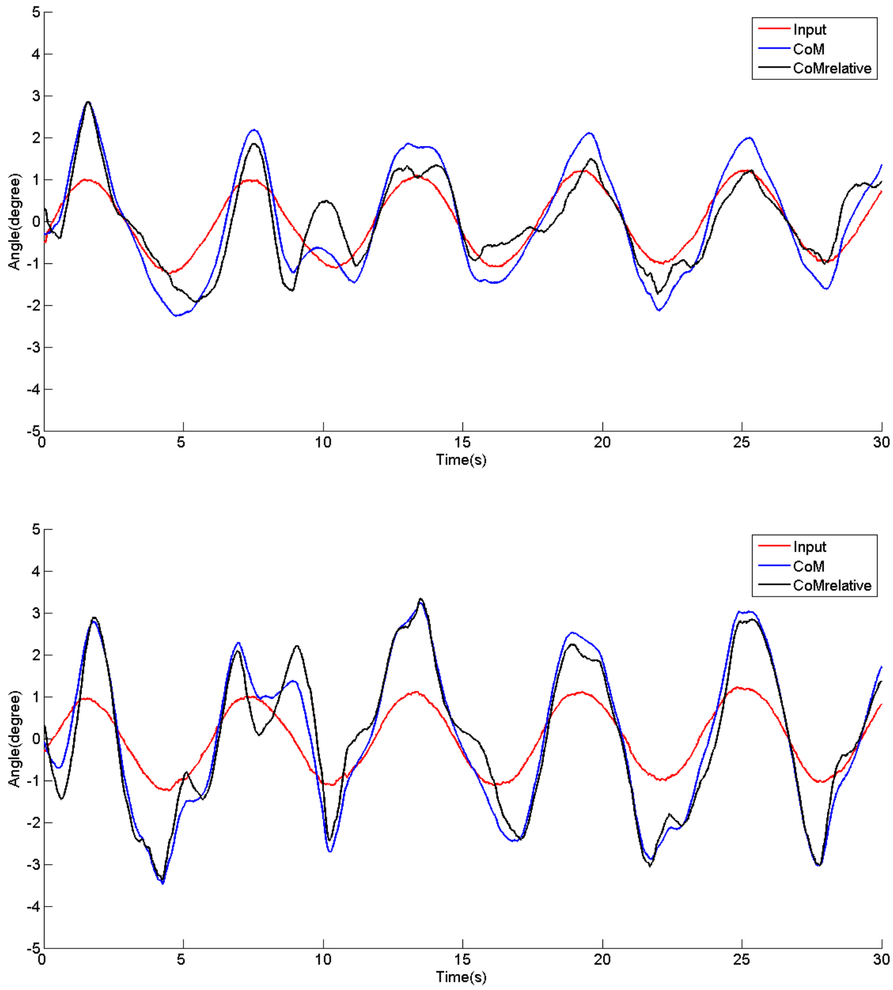
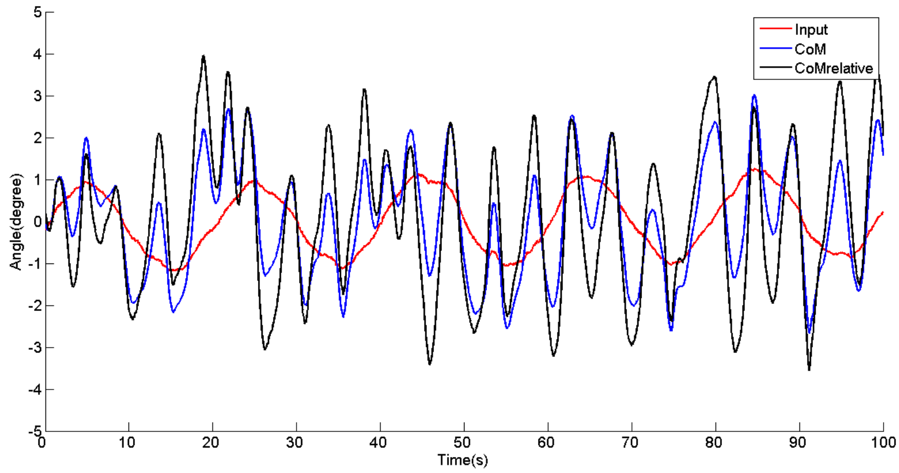


Figure B.21: Trial 1 for patient 5, 1 degrees 0.17 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions



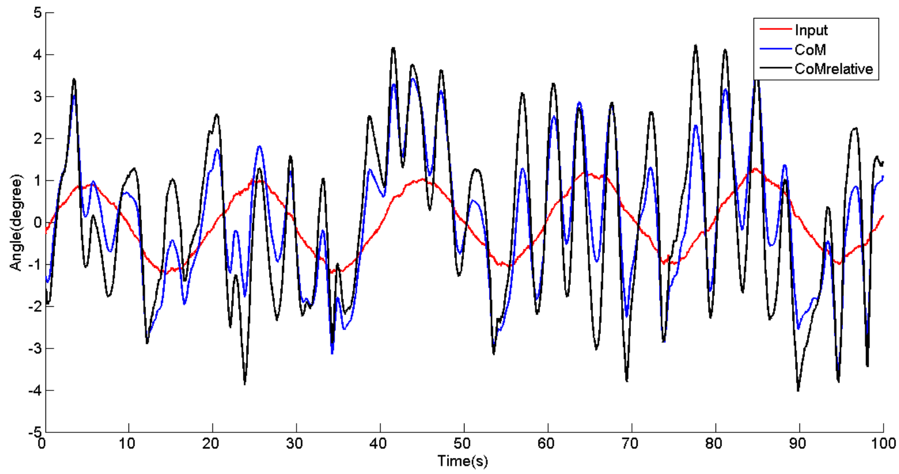


Figure B.22: Trial 1 for patient 5, 1.2 degrees 0.05 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions

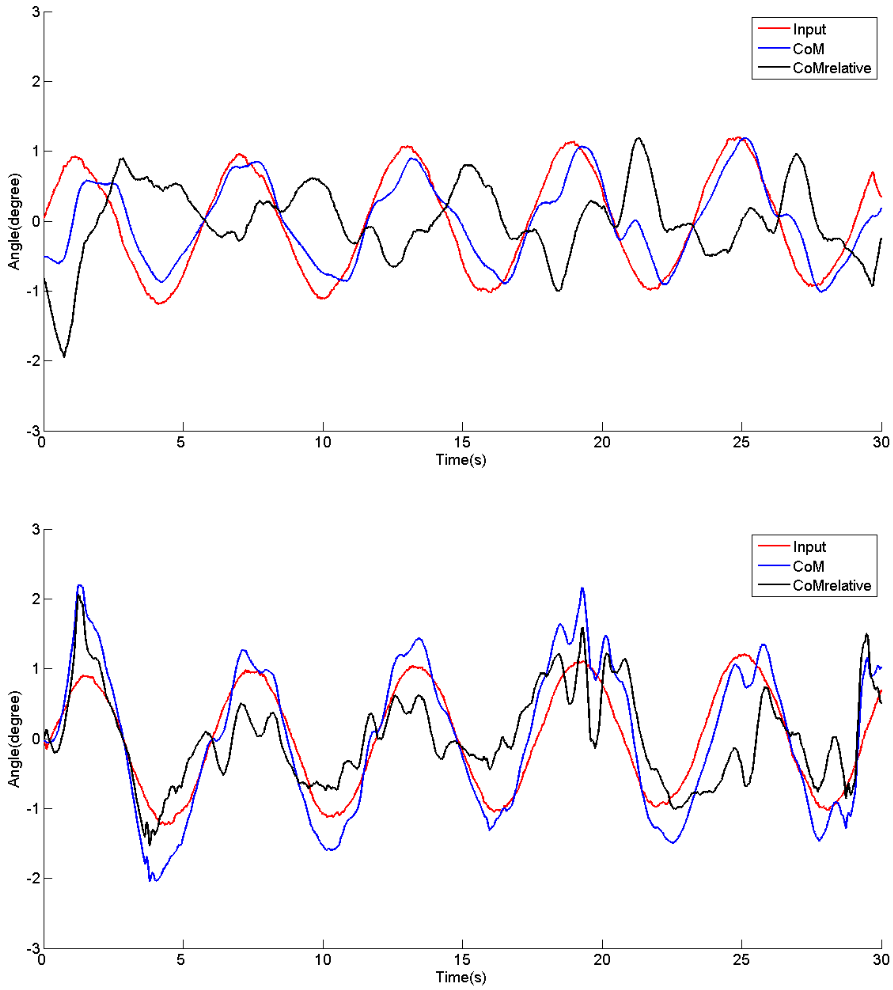
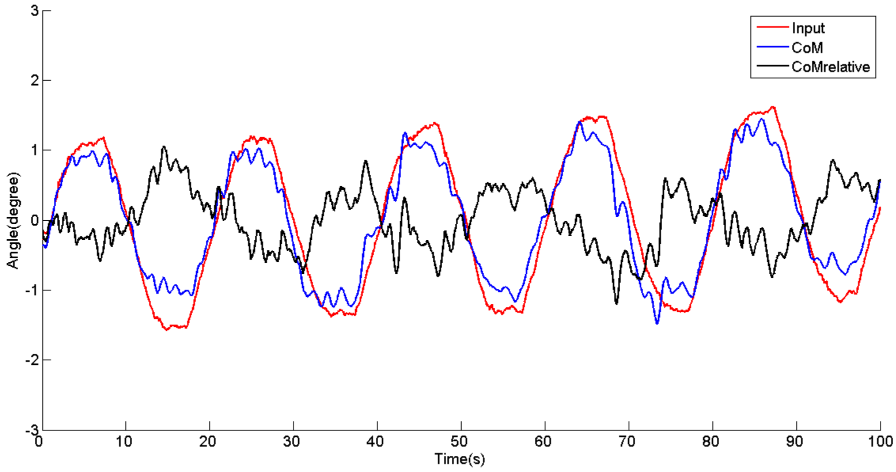


Figure B.23: Trial 1 for patient 6, 1 degrees 0.17 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions



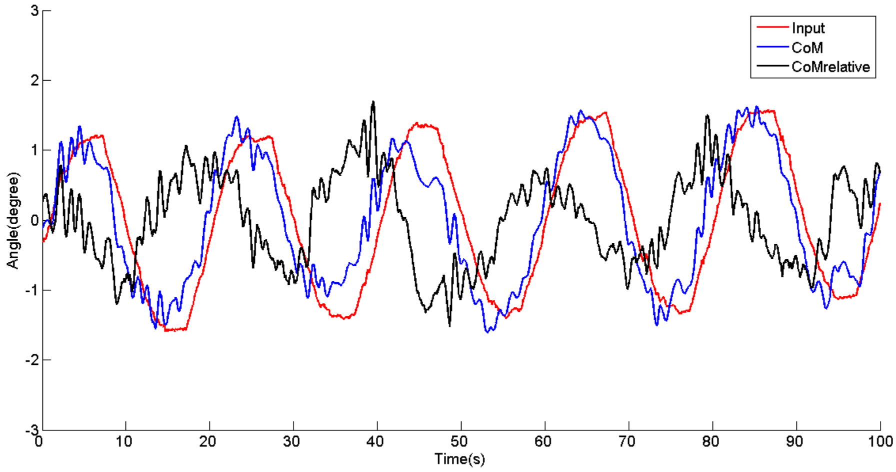


Figure B.24: Trial 1 for patient 6, 1.2 degrees 0.05 Hz perturbation of tilt platform, and absolute and relative CoM angular displacements of the subject at eyes open (top figure) and closed (bottom figure) conditions