ISSN 0280-5316
ISRN LUTFD2/TFRT--5755--SE

Experimental Study of Adaptation and Postural Control after Sleep Deprivation

Per Almbladh

Department of Automatic Control
Lund Institute of Technology
August 2005

Department of Automatic Control
Lund Institute of Technology
Box 118
SE-221 00 Lund Sweden

Document name
MASTER THESIS

Date of issue
August 2005

Document Number
ISRNLUTFD2/TFRT--5755--SE

Author(s)
Per Almbladh

Supervisor
Rolf Johansson at Automatic Control in Lund.
Per-Anders Fransson and Måns Magnusson at
Dept. of Otorhinolaryngology in Lund

Sponsoring organization

Title and subtitle
Experimental Study of Adaptation and Postural Control after Sleep Deprivation (Experimentell studie av adaption och balanskontroll vid sömndeprivering)

Abstract
This thesis investigates the influence of sleep deprivation on the human's postural control and conceptions of adaptation. Seventeen subjects were investigated with posturography after 24 h and 36 h of sleep deprivation, and a control posturography trial was performed either one week before or one week after the sleep deprivation trials. Stabilizing torques and forces were recorded by a force platform and the body sway was recorded by a 3D ultrasonic positioning system. Stimulation according to a pseudo random binary sequence (PRBS) was applied using two vibrators attached to the calf muscles of the subjects. Posturographic tests were performed with eyes open and eyes closed. Subspace based modeling was used to assess postural control after sleep deprivation. Also, quantitative variance and correlation analysis were applied to the recorded data.
The results of the investigation showed that it is possible to identify MIMO models with stimulus as input signal and torques and body sway as output signals. When body sway is characterized by the variance of the position data, statistical evidence was found for an increase of body sway after 24 hours of sleep deprivation. Also, sleep deprivation impairs postural adaptation.

Keywords
Postural control, sleep deprivation, adaptation, subspace identification, posturography

Classification system and/or index terms (if any)

Supplementary bibliographical information

ISSN and key title
0280-5316

ISBN

Language
English

Number of pages
62

Recipient's notes

Security classification

The report may be ordered from the Department of Automatic Control or borrowed through:University Library, Box 3, SE-221 00 Lund, Sweden Fax +46 46 22 42 43

Contents

Acknowledgements

I wish to thank my supervisors Per-Anders Fransson and Måns Magnusson at the Department of Otorhinolaryngology, Lund University Hospital and Rolf Johansson at the Department of Automatic Control, Lund Institute of Technology for their help and support with this thesis. Also, many thanks to the people at ÖNH for their help to carry out the experiments. Finally, I am very greatful to the people who were willing to be subjects in this study.

Chapter 1

Introduction

The human balance system is a complex control system using multiple sensory inputs together with the nervous system and skeletal muscles to maintain upright stance. One can look at the balance control system as a dynamic feedback and feedforward control system, where the input signals originate from the sensory input from visual, vestibular and somatosensory receptors, which sense changes in position and velocity of the body [11, 15]. This information is integrated by the central nervous system resulting in response actions, body motions, to maintain stability. If a repeated disturbance of postural control is introduced, the body respond with an adaptation process striving to reduce body movements [9, 14, 18].

There are many ways of introducing a postural perturbation. Usually, one of the sensory input is stimulated, i.e. proprioceptive stimulation by muscles or muscle tendons [7], vestibular stimulation by galvanic transmastoidal currents or visual stimulation by altering or moving the visual surroundings. The vision supplies important feedback information and some feedforward ability to respond with anticipatory control actions to movement. By the removal of vision, a postural disturbance is created which requires increased postural response to maintain body stability. If one sensory input is stimulated, the resulting body movements are detected by the other sensory receptors and the new information are processed and reweighted by the central nervous system to base the control actions on the more reliable sensory inputs.

In most cases, this process is highly effective and the postural control can be managed without conscious control. However, a certain mental alertness may be needed when the balance control have to adapt to new balance perturbations that perhaps require a new movement pattern behavior. Schlesinger *et al* (1998) give evidence for that attention plays a role in postural control when higher levels of sensory integration or motor coordination are required, such when disturbing one of the sensory inputs by stimulations. [19]. A reduced mental alertness could be caused by medication or alcohol. Previous studies suggest that the effects of sleep deprivation influence the attention negatively, consequently resulting in reduced cognitive performance [19]. The subjects in these studies have been subjected to different levels of sleep deprivation. However, persons with 24 hours or less of sleep deprivation did not in these studies show deteriorated postural control [22, 19]. We make the assumption that attention deficits may disturb the postural control system, especially the adaptation processes, during periods where higher levels of postural control are needed. In this thesis, the decreased attention is the result of up to 36 hours of sleep deprivation.

The postural control problem can be analysed using control theory. System identification methods make it possible to describe properties of the postural control system with mathematical models. These models can describe input-output relations, for instance between the induced balance disturbance and the response actions exercised by the subject to maintain balance [12]. The use of control theory to analyse postural control is however a difficult task since the biomechanical complexity of the human body may change in response to the balance task. Small balance perturbations and quiet stance may result in a response pattern where it is sufficient to move around the ankle joint like a segmented inverted pendulum to maintain balance [12]. However, when high intensity balance perturbations are applied, it is possible that the subject also may use movements of knees and the hip resulting in a multi-segmented inverted pendulum dynamics behavior [12, 17]. Consequently, the choice of model order is highly dependent of the postural control strategy.

The aim of this thesis is to study the effect on postural control and the ability to adapt new movement patterns when mental alertness is reduced by sleep deprivation. In this study, postural control of seventeen subjects were investigated with posturography after 24 hours and 36 hours of sleep deprivation. The subjects were exposed to calf vibration stimulus in order to induce balance perturbations. Forces and torques exerted by the feet are recorded by a force platform and the body sway was measured by a 3D ultrasonic positioning system. A control test after normal sleep was performed either one week before or one week after the sleep deprivation test.

Methods of system identification are used to assess whether changes in sway patterns and ability to adapt to a balance perturbation are changed due to sleep deprivation. Two subspace identification algorithms are used, N4SID and MOESP, to identify models between input data (stimulus) and output data (torques and body sway). Also, a statistical analysis of the variance of recorded body sway is performed as another method to assess possible changes of adaptation and postural control due to sleep deprivation. A correlation analysis of the movements of different body sections is used to characterize the movement pattern during time. This analysis also investigates the likelihood to use a more complex movement pattern when being sleep deprived.

The remainder of the thesis is organized as follows. In chapter two, the biological background of postural control is presented. Chapter three deals with the experimental setup and how the test trials were performed. Theory of analysis methods applied in the thesis including the mathematical system modeling and identification are covered in chapter four. Chapter five describes the implementation and modeling procedure and strategies used when analysing the data from the experiments. The results of the study are presented in chapter six and the conclusions are stated in the subsequent chapter.

Chapter 2

Background

2.1 Postural control

A human's balance control masks a complex process that involve many subsystem like sensory receptors and the central nervous system together with skeletal muscles to maintain postural stability. The sensory information is used to detect changes in position and velocity of the body that violate the equilibrium. The information is then processed by the central nervous system so that appropriate counteract response of antigravity muscles and postural reflexes can be initiated. To prevent falling, high sensitivity of the sensory system is needed to be able to detect changes of the body position at early stage and rapidly make suitable correction movements. Therefore, postural control uses both feedback and feedforward (predictive) control involving the visual, vestibular and somatosensory receptors. The possibility of the human body to determine its own orientation and velocities of different body parts is called kinaesthesia and relies on the highly effective sensory system.

The central nervous system plays an integral part in postural control when processing the sensory information. The visual, vestibular and somatosensory receptors deliver a certain level of complementary data which is used for increasing the likelihood of an accurate response. Another advantage of using complementary data is that reductive ability is added. This is important in cases where damages to one or more of the receptors have occurred, maybe as a result of diseases or injuries. In a situation like this, the information from other receptors are given higher weights in the processing procedure to compensate for the information loss. Over time new methods and strategies can be learned in order to maintain body stability even though the sensory system has functional deficits.

2.2 The somatosensory receptor system

The somatosensory receptors can be divided into two subgroups. Position and tension in muscles, tendons and joints of different body sections are sensed by the *proprioceptive* receptors. The *exteroceptive* receptors register the pressure and forces acting on the skin.

The proprioceptive receptors are neuromuscular spindles located in the muscles, tendon receptors (Golgi tendon) and joint receptors. The muscle spindles are being activated both in case of muscle stretch and in the case of muscle contraction. In this manner, the length of a muscle can be sensed at any time and this information is sent to the central nervous system for further processing. The tendon receptors operate in an antagonist manner to the muscle spindles. In case of muscle contraction, the Golgi receptors sense the corresponding stretch of the tendon. Even though muscle contraction is the main stimulus, ability to sense passive tendon stretch is possible. The joint receptors are located near the joint capsule. Activation are done by joint movement and stretching the muscle attached to the joint capsule.

The exteroceptive receptor system consists many mechanoreceptors located in the skin. They have the ability to sense and detect force, velocity and acceleration of indentations of the skin [23, 13].

2.3 The vestibular system

The vestibular system located in the internal ear plays an important role in postural control. In addition, the information from the vestibular system contribute to maintain fixation of the eyes when a rapid movement of the head occur [1]. Therefore, the quality of the visual information is enhanced by the vestibular system resulting in fully sufficient visual perception despite movements of the head. However, studies have shown that unilateral loss of vestibular function only initially causes falling tendencies, nystagmus and vertigo of the patient [4]. After some days of compensations, adequate postural control can be achieved through visual and proprioceptive reception alone. In this case, postural function disturbances can only be detected when the patient is being exposed to conflicting sensory stimulation [4].

The vestibular system consists of the labyrinthes, which are located on both sides of the head. The labyrinthe contains three semicircular canals placed in different planes forming the kinetic labyrinth. Thus, the angular acceleration of the head can be measured in three dimensions. The labyrinthe is filled with endolymph fluid which in addition to its wave propagation property used for the conscious perception of sound also has an important role when detecting movements of the head. A movement of the head causes angular acceleration which induce a flow of the endolymph fluid of the semicircular canals. At the ends of each semicircular canal there is an expansion, the ampullae, which contain the sensory epithelium (crista ampullaris). Together with hair cells (cilia) the cupula is formed. When a movement of the head is initiated near a plane of one semicircular canal, it causes a lag of the endolymph fluid due to inertia resulting in a swing of the cupula in the opposite direction to that of the movement of the head [1]. When the movement ceases, the moment of inertia of the endolymph fluid induce a movement of the cupula in the other direction. The fine hair cell are bent by this movement of the cupula. Depending on direction, the polarization are either increased or decreased of the endings of the nerve cells, evoking an action potential used for neuromechanical transmission.

The static labyrinth consists two vesicles, the utricle and the saccule, filled with endolymph fluid. In the macula located in the utricle and in the saccula contains special hair cells associated with calcium crystals called otoliths. When the head is being exposed to linear acceleration, e.g. gravitational pull, or transient movement, moment of inertia of the otoliths are induced. The cilias are bent by the movements of the otoliths and, like in the semicircular canals, this cause either an increase or decrease of the polarization of the vestibular afferent neurons. Thus, the position of the head with respect to gravity can be determined.

The vestibular system has an estimated frequency range of > 0.1 Hz for the semicircular canals and < 0.1 Hz for the otolith organs [5, 25].

2.4 The visual system

The light sensitive photoreceptors of the retina collect the visual information and send it through the optic nerve for further processing. By the use of the visual information, both feedback and some feedforward ability is provided. However, the visual system is not necessary for posture control [5], but it provides complementary information to the other sensory systems which increases the likelihood of proper postural control in situations of vestibular or proprioceptive function loss. Furthermore, it is commonly known that performing balance tasks with eyes closed is more difficult than in the case of open eyes due to less information to base the stabilizing movements upon. However, if the balance task is of simple nature, the absence of vision input does not immediately effect the postural control system as long as other reliable sources of balance information, e.g. somatosensory and vestibular inputs, are available [8].

2.5 The central nervous system

All the information registered by somatosensory, vestibular and visual systems are integrated by the central nervous system. This information processing is done using three stages of between the stimulus and the resulting motor command.

2.6 Body mechanics

The biomechanical body design of the human yields restrictions to postural control by the means of possible movement patterns. Limitations are caused by the geometrical configurations of body parts and the maximum muscle force that can be generated by stabilizing muscle used in postural control [11]. These facts make it difficult to analyze the body mechanics, both theoretically and experimentally. Although, the biomechanical structure obviously has the ability to generate appropriate stabilizing motions which are needed since the physical structure of the body is clearly unstable at upright stance.

Depending on the characteristics of the body motions due to various disturbances, the body motions have been classified into groups, e.g. the 'ankle' strategy and the 'hip' strategy [9]. Using the ankle strategy, for instance, the body is stiff and move around the ankle joint. One can resemble the body dynamics in this case with an inverted pendulum which is unstable in the upright position and need some kind of controller action for position control [12]. When the control task become harder, e.g. when the induced disturbances are more intense, a more complex body dynamics is adopted and the inverted pendulum model is afflicted with larger model errors.

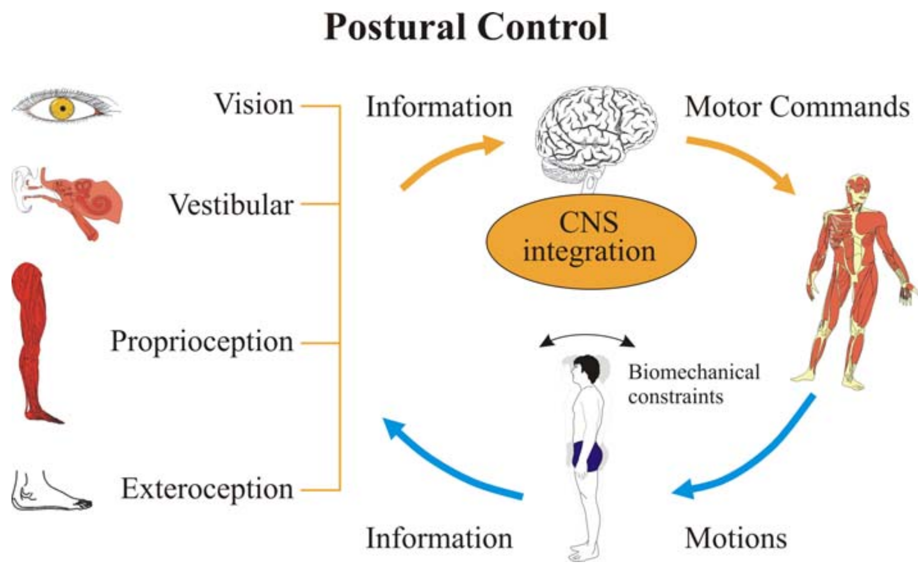


Figure 2.1: Schematic illustration of the human postural control system. The sensory inputs are integrated with motor commands to maintain a stable stance.

Chapter 3

Experimental setup

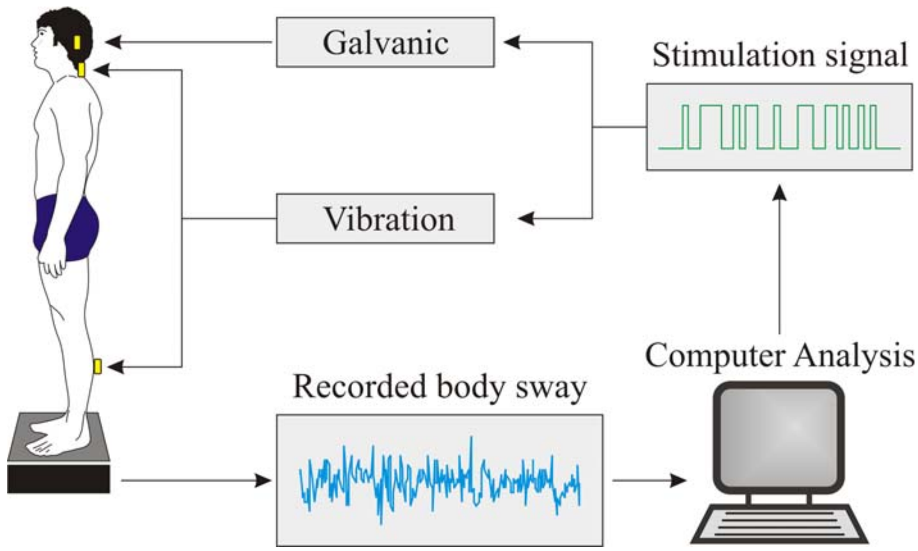


Figure 3.1: Schematic illustration of the posturographic measurement system. In this thesis, only stimulus to the calf muscles is applied.

3.1 Subjects

The 17 subjects, nine male and eight female aged between 16 and 38 years, that participated in the study were healthy and without any history of diseases of the central nervous system or injury of the lower extremities. Mean age was 23.4 years, $\sigma = 5.08$; mean weight 74.3 kg, $\sigma = 17.29$ and mean height 1.77 m, $\sigma = 0.09$. Furthermore, no one had at the time of investigation taken any medication or other drugs. Neither did they consume any alcohol for at least 48 hours prior the posturographic tests. The subjects were instructed not to consume any beverages containing caffeine during the awake period. Activities like intense computer gaming were not allowed due to the fact that this may affect the oculomotor response test. The experiments were performed in accordance with the Helsinki declaration of 1975 and approved by the local ethics committee.

3.2 Equipment

The posturography was done using a custom made force platform developed at the Department of Solid Mechanics, Lund Institute of Technology. The force platform was able to record shear forces and torques in three dimensions, making a total of six degrees of freedom. Recorded data were sampled with the software Postcon running on a PC connected to an AD converter.

The vibratory proprioceptive stimulation was done using vibrators attached to the belly of the gastrocnemius muscles of both legs. The vibrations were generated by a revolving DC-motor having a 3.5 g weight placed 1.0 mm eccentric at one end. The motor was embedded in a cylindrical plastic shell with diameter 1 cm and length 6 cm. The motor was fed with an pseudo random binary sequence (PRBS) in order to reduce effects of anticipatory behavior of the subjects response to the stimulation. By the use of a PRBS signal, the subjects have no possibility to predict when future stimulation sequences will occur. Hence, the subjects have no feedforward ability to base their postural control strategy on anticipations of the stimulation signal.

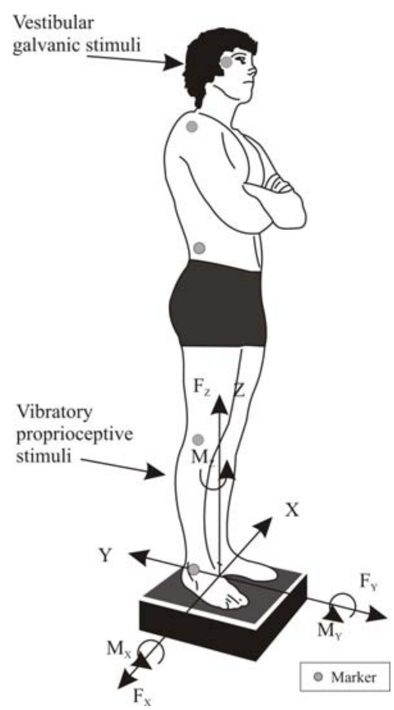


Figure 3.2: Schematic illustration of the Zebris system for measurement of body movements. The placement of the position markers on a subject standing on the force platform are shown.

Body movements were measured by Zebris CMS-HS measuring System for

3D Motion Analysis. The Zebris system uses six position markers placed on the subjects body on ankle, knee, hip, shoulder, head and neck. By the use of three ultrasonic transmitters arranged in a triangle which continuously are sending pulses at 40 kHz, the positions of the markers can be determined by triangulation. The markers receive the pulses by small microphones and due the different running times of the pulses and phase characteristics the distance between transmitter and marker can be determined. Hence, the exact positions of the markers were determined in three dimensions by its x- y- z- coordinates.

To make sure that the subjects have stayed awake during the test period, EEG equipment was used to record the brain activity. If a subject has fallen asleep, this can be seen in the EEG-data. The EEG-data was collected over the frontal pole using three electrodes, EEG signal, ground and reference signal. The ground channel was attached to the forehead, the EEG channel to the upper temple and a reference electrode placed on the mastoid bone. Data was recorded by a device, Embletta XactTrace EEG X50, using a sample frequency of 2000 Hz. The data were then down sampled to 100 Hz to reduce the size of data. EEG brain activity has much lower frequency, about 10 Hz. Hence, it is possible to down sample data without any loss of information or unwanted aliasing distortion. The EEG device was equipped with 16 Mb of memory making 12 hours of maximum recording time. The subjects were wearing the EEG recorder during the night before the initial measurement. At this time, the 12-hour data was downloaded to a computer and the EEG recorder was prepared for the forthcoming 12-hour recording during the day.

Also, the EEG equipment had ability to record movements of the device, which can be used as a measure of body movements during the awake period. If movement of the equipment are being recorded, the person will almost certainly not be asleep at this time.

3.3 Test procedure

To measure effects on postural control and adaptation after sleep deprivation, posturographic test using a force platform and oculomotor response test was made. These measurements were performed after 24 h and 36 h of sleep deprivation. As comparison, a control test including health and sensitivity tests were done after normal sleep.

The following randomization strategy was used during the test to avoid biased results. The group of subjects were split into two smaller groups where the first group of subjects performed the control test without any sleep deprivation prior the real test. The other group did the sleep deprivation posturographic test first and the control test afterwards. In this procedure, unwanted effects like adaptation to the posturographic test itself caused by repetition was minimized.

Also, each one of the two groups were subdivided into two smaller groups performing the tests with open and closed eyes in different order as another randomization strategy. Otherwise, the results may be effected by the subjects' habitation to the posturographic test with increased performance during the second test as a consequence.

3.3.1 Sleep deprivation test

The subjects were instructed to stand on the force platform without shoes in an erect and relaxed posture with the arms crossed over the chest. The angle between the feet was about 30° and the heels placed approximately 3 cm apart. The posturographic test used vibratory proprioceptive stimulation of the calf muscles in order to introduce balance perturbations. Each subject was submitted to this stimulation two times, one with eyes open and one with eyes closed. In the open eyes test, the subjects were told to focus on a photograph placed 1.5 meters away on the wall. Between the two posturographic tests, the subjects stepped down from the force platform and relaxed for about three minutes. During the tests, the subjects were wearing headphones with classical music in order to suppress auditory feedback from the stimulation equipment and surroundings.

The oculomotor response was tested using five electrodes attached to the eye surroundings measuring muscle activities of the oculomotor control. One electrode was placed right below the eye, and another one above, measuring the eye movements in vertical directions. In a similar way, two other electrodes were placed in horizontal direction measuring the horizontal movements. A ground channel was attached to the forehead. Two tests were performed. The saccade test measured how rapid the subject could move the line of vision between two points. Three angles of eye movement were examined in the saccade test, 20°, 40° and 60°. The other test method, slow pursuit, tested how well the subject could follow a moving point with angular velocity of 10, 20, 30 and 40 degree/s.

When the subjects arrived to lab they were told to make a subjective estimation of their sleepiness using a scale from zero to ten where zero is extremely alert and ten is extremely sleepy. To avoid problems of guidance from the grade numbers, a slider without grading was used. In this way, the subjects could put the marker somewhere on the non graded scale. The number between zero and ten was then read off from the other side of the stick.

During the sleepless period 12-36h, EEG (electroencephalogram) data from the subjects were recorded which made it possible to analyse the brain wave activity. The main purpose of recording the EEG data was to use it as measure of the awakeness of the subjects during the test period. If the subject accidentally had fallen asleep, this would be seen in the EEG data as changes in brain wave characteristics.

3.3.2 Control test

To be able to analyse the effect of sleep deprivation on postural control, a test under normal circumstances has to be made. The control test was performed after normal sleep and include posturography, oculomotor response test, sensitivity test and finally an examination of a doctor assuring that the subject does not have or have had any diseases of the central nervous system, the vestibular system or other parts of the body that might effect postural control.

The exteroceptive receptors of the feet senses pressure and forces acting on the skin. These receptors are part of the somatosensory system which collect information that are processed in the central nervous system for maintaining postural control. Therefore, it is important to examine the subjects' sensitivity of the exteroceptive receptors of the feet prior the posturographic test. The test examined the sensitivity to vibration and pressure on the larger and smaller digits of the feet together with the heel of both feet and was performed during the control trials.

3.3.3 Some notations

For convenience, the following notations of trials are used in the remainder of the thesis.

- CC - Control trial, Closed eyes
- CO - Control trial. Open eyes
- SD24C - 24 hour Sleep Deprivation trial, Closed eyes
- SD24O - 24 hour Sleep Deprivation trial, Open eyes
- SD36C - 36 hour Sleep Deprivation trial, Closed eyes
- SD36O - 36 hour Sleep Deprivation trial, Open eyes

Chapter 4

Analysis methods

This thesis uses two main methods to analyse adaptation and postural control; variance analysis and system identification methods. First, a characterization of recorded data is performed using spectral analysis method, e.g. coherence and power spectrum. The subsequent sections cover the variance and correlation analysis together with the mathematical modeling using system identification methods.

4.1 Spectral analysis

When the data properties should be characterized, the use of spectral analysis often shows many interesting things. The power distribution can be studied with a power spectrum which correlates the power to the frequency. Moreover, a coherence spectrum is a good test for linearity of input-output relations [10].

4.1.1 Power spectrum

The autospectrum and the power cross spectrum of the signals u and y are obtained by taking the fourier transform of the autocovariance function and the cross-covariance function. Hence,

$$S_{uu}(i\omega) = \mathcal{F}\{C_{uu}(\tau)\} \tag{4.1}$$

$$S_{uy}(i\omega) = \mathcal{F}\{C_{uy}(\tau)\}, \tag{4.2}$$

where the autocovariance function C_{uu} and cross-covariance function C_{uy} are given by

$$C_{uu}(\tau) = \lim_{T\to\infty} \frac{1}{2T} \int_{-T}^{T} u(t)u^{\star}(t-\tau) \tag{4.3}$$

$$C_{uy}(\tau) = \lim_{T\to\infty} \frac{1}{2T} \int_{-T}^{T} u(t)y^{\star}(t-\tau) \tag{4.4}$$

4.1.2 Coherence spectrum

The coherence spectrum between two signals is defined as the ratio

$$\gamma_{xy}^2(\omega) = \frac{|S_{xy}(i\omega)|^2}{S_{xx}(i\omega)S_{yy}(i\omega)}. \tag{4.5}$$

The quadratic coherence, γ_{xy}^2, always has a value between 0 and 1. If the noise level is low, the value of the quadratic coherence will be close to 1. In this case, there is a linear response between input and output. The coherence function can be described as a correlation function between two signals in the frequency domain.

In this particularly case of posture control, coherence spectrum can be calculated to express how well the input (stimulus) and the outputs are correlated in the frequency domain. However, it should be pointed out that there is three things that will effect the coherence. First, a person has normally a large part of spontaneous body sway when standing upright. For control of an inverted pendulum, it is necessary to move around the unstable equilibrium point and respond with appropriate control actions. Hence, this fact will decrease the coherence between stimulus input and outputs. Second, the coherence is expected to be higher in the anteroposterior plane than in the lateral plane since calf vibration stimulus are being used. Third, the body movements have low frequency characteristics. Most of the power has frequencies below 0.5 Hz. Above this frequency, the coherence spectrum is expected to be close to zero.

4.2 Variance and correlation analysis

One way to study the body sway is to perform an analysis of the recorded variance. In the trials, data from force platform i.e. forces and moments together with the positions of the body are measured. A natural measure of body sway is the variance of the measured data. For instance, the position itself of the body is not interesting, it is the variance of the movements that characterizes the body sway. This quantitative analysis of calculated variance values of measured torque correspond to the energy of the response movements toward the force platform expressed by the subject to maintain postural stability [16]. Furthermore, the correlation analysis between the movements of knee, hip, shoulder and head is another interesting quantitative method that reflects the movement pattern used. High correlation between different body segments suggest a movement pattern that can be modeled like an inverted pendulum.

Using the same notation as previously, the data set was divided into five periods where the first period corresponds to the 30 a of quiet stance before onset of vibration. The latter four periods (I, II, III and IV) are of length 50 s making a total stimulation time of 200 s. If there is statistical significant differences between the variance of these periods, conclusions can be drawn about adaptation of posture control. For example, if the variance of the last period is less than the first stimulus period, there is some kind of adaptation process initiated by the stimulus that reduces the amount of sway caused by the stimulation.

Previous findings have shown that the torque variance values have a dependence of the test subjects' squared height and squared weight. For normalization, the torque variance data were divided with the squared height and squared weight. The variances of the body segment movements were normalized only with respect to the subjects' squared length.

The statistical analysis was done using SPSS software to obtain significance values. The Wilcoxon non-parametric test was used to find statistical differences between the six test trials since the values were not normally distributed according to the Kolmograv-Smirnov test and the Shapiro Wilks non parametric test. Also, the Wilcoxon non parametric test was used to obtain statistically significance for differences of the variances between the four stimulus periods and for the correlation analysis. The same methodology was applied to the position data recorded by the Zebris equipment. P-values less than 0.05 are considered significant.

4.3 Mathematical modeling

The second methodology applied in this thesis to analyse the properties of postural control after sleep deprivation is mathematical modeling using system identification methods. Most of the identification is performed with subspace algorithms (MIMO) which make it possible to estimate the system directly from input-output data.

The mathematical description of the deviations from the equilibrium point during perturbed stance can be written using a state space approach,

$$\begin{aligned} x_{k+1} &= Ax_k + Bu_k + v_k \\ y_k &= Cx_k + Du_k + e_k, \end{aligned} \tag{4.6}$$

where

- k denotes the sample time index;
- A, B, C and D denotes coefficient matrices which is to be estimated by fitting the model to data;
- v_k and e_k is stochastic disturbances acting on the process, i.e. the body mechanics, and on the measurement, respectively;
- $\{u_k\}$ is the stimulus vibration to the calf muscles;
- $\{y_k\}$ is the response sequence to the stimulation, i.e. positions of body segments and force-data from the platform;
- x_k denotes state variables, like position, velocity and other 'storage' variables.

4.3.1 Inverted pendulum dynamics

When the body is exposed to perturbations in the sagittal plane, e.g. proprioceptive stimulation of the calf muscles, the subject respond with body movements to regain stability. If the perturbations are small, it is sufficient to use the 'ankle' strategy, i.e. the muscles rotate the body around the ankle joint, to maintain upright stance at equilibrium [10] p. 476. When the control task become harder, it is more likely for the subject to incorporate movements around the hip joint. Hence, the body dynamics can be modelled as an single linked or multi-linked inverted pendulum [12]. The physical model of a one-segment inverted pendulum can be described as follows.

The following assumptions are to be made [12]:

- **Assumption 1:** The body is stiff and has the mass m (kg).
- **Assumption 2:** The body center of mass is located l (m) from the force platform.
- **Assumption 3:** There is an equilibrium between the forces acting on the body and the torque accentuated by the feet.

If a person does not respond to the momentum caused by gravitational forces, the body is modeled by the force equilibrium of an inverted pendulum. Let the body moment of inertia around the ankle joint be given by J, then a person standing upright and subjected to a gravitational force g fulfills the equilibrium

$$J\frac{d^2\theta}{dt^2} = mgl\sin\theta(t), \quad J = ml^2. \tag{4.7}$$

It is clear that this equilibrium is unstable. Hence, a person who does not counteract the gravitational force g with a stabilizing torque will certainly fall. Including a stabilizing torque, $T_{bal}(t)$, and a disturbance torque, T_d, originating in the proprioceptive stimulation, the following two assumptions are necessary.

- **Assumption 4:** There is a stabilizing torque $T_{bal}(t)$ around the ankle.
- **Assumption 5:** There is a disturbance torque $T_d(t)$ from the environment (i.e. induced by the vibrating stimulation of the calf muscles).

This now yield the following equilibrium equation

$$J\frac{d^2\theta}{dt^2} = mgl\sin\theta(t) + T_{bal}(t) + T_d(t), \quad J = ml^2. \tag{4.8}$$

Now, introduce the state vector

$$x = \begin{pmatrix} x_1 \\ x_2 \end{pmatrix} = \begin{pmatrix} \theta(t) \\ \dot{\theta}(t) \end{pmatrix}. \tag{4.9}$$

Under the assumption of small movements around the equilibrium point, linearization around $\theta = 0$ can be made.

$$\begin{aligned} \frac{dx}{dt} = A_c x(t) + B_c q(t) = \begin{pmatrix} 0 & 1 \\ \frac{g}{l} & 0 \end{pmatrix} \begin{pmatrix} x_1 \\ x_2 \end{pmatrix} + \begin{pmatrix} 0 \\ \frac{1}{ml^2} \end{pmatrix} (T_{bal} + T_d) \\ y(t) = C_c x(t) = \begin{pmatrix} l & 0 \end{pmatrix} \begin{pmatrix} x_1 \\ x_2 \end{pmatrix} \end{aligned} \tag{4.10}$$

Since system 4.10 is a continuous system, it has to be transformed to a discrete system to fit the form of system 4.6. Note that $T_{bal} + T_d$ is replaced with input signal u, the vibration PRBS signal. This is not the same as the two torques T_{bal} and T_d, but used as input for all estimated models. Having this, the main problem is to estimate the matrices A, B, C and D using system identification.

4.3.2 Subspace based identification

In recent years, the use of subspace identification methods have grown considerably [20]. The basic idea is to estimate the state variables, i.e. the matrices A, B, C and D, directly from the input output data.

Again, consider the discrete-time time-invariant linear system of equation 4.6 with the process and measurement noise given by the zero mean sequences $\{v_k\}$ and $\{e_k\}$ with the covariance matrix

$$\Sigma = E[\begin{pmatrix} v_k \\ e_k \end{pmatrix} \begin{pmatrix} v_k & e_k \end{pmatrix}] = \begin{pmatrix} \Sigma_v & \Sigma_{ve} \\ \Sigma_{ve}^T & \Sigma_e \end{pmatrix} > 0 \tag{4.11}$$

The orthogonality condition

$$E[\begin{pmatrix} x_k \\ u_k \end{pmatrix} \begin{pmatrix} v_k^T & e_k^T \end{pmatrix}] > 0 \tag{4.12}$$

is believed to be true. For a more thorough description of subspace identification, see [3, 10].

In this thesis, two algorithms of subspace identification are used for identification of the state space system given in equation 4.6.

- Numerical algorithms for Subspace State Space System IDentification (N4SID).
- Multivariable Output-Error State Space Algorithm (MOESP).

Multivariable Output-Error State Space Algorithm (MOESP)

The MOESP algorithm is a very effective approach for state-space identification by Verhaegen and Dewilde [24].

1. Organize the data in Hankel structure and define the following matrices:

$$Y_{rs}^{(h)} = \begin{pmatrix} y_k & y_{k+1} & \cdots & y_{k+s-1} \\ y_{k+1} & y_{k+2} & \cdots & y_{k+s} \\ \vdots & \vdots & \ddots & \vdots \\ y_{k+r-1} & y_{k+r} & \cdots & y_{k+s+r-2} \end{pmatrix} \tag{4.13}$$

$$U_{rs}^{(h)} = \begin{pmatrix} u_k & u_{k+1} & \cdots & u_{k+s-1} \\ u_{k+1} & u_{k+2} & \cdots & u_{k+s} \\ \vdots & \vdots & \ddots & \vdots \\ u_{k+r-1} & u_{k+r} & \cdots & u_{k+s+r-2} \end{pmatrix} \tag{4.14}$$

and

$$X = \begin{pmatrix} x_k & x_{k+1} & \ldots & x_{k+s-1} \end{pmatrix} \tag{4.15}$$

2. Make the QR-factorization

$$\begin{pmatrix} U_{rs}^{(1)} \\ Y_{rs}^{(1)} \end{pmatrix} = \begin{pmatrix} R_{11} & 0 \\ R_{21} & R_{22} \end{pmatrix} \begin{pmatrix} Q_1 \\ Q_2 \end{pmatrix} \tag{4.16}$$

3. Calculate the column space of the extended observability matrix using the singular value decomposition, $R = USV^T$

$$R_{22} = \begin{pmatrix} U_n \\ U_n^{\perp} \end{pmatrix} \begin{pmatrix} S_n & 0 \\ 0 & S_n^{\perp} \end{pmatrix} \begin{pmatrix} V_n^{\perp} \\ (V_n^{\perp})^T \end{pmatrix} \tag{4.17}$$

   Now, by inspection of the magnitude of the singular values of S, the order can be determined. Up to a non-singular similarity transformation matrix $T \in R^{n\times n}$, U_n is an estimate of the extended controllability matrix

$$U_n T^{-1} \tag{4.18}$$

   which can be used to calculate the estimates $\hat{A}, \hat{C}$ of A and C.

4. Apply least-square estimation using the estimates $\hat{A}$ and $\hat{C}$ to the state-space model

$$\begin{aligned} x_{k+1} &= \hat{A}x_k + Bu_k + Kwk \\ y_k &= \hat{C}x_k + w_k, \end{aligned} \tag{4.19}$$

   to find estimates of the B matrix.

N4SID algorithm

The N4SID algorithm is similar to the MOESP except in the step where the QR-factorization is made. N4SID makes no QR-factorization in step 2. Instead, an oblique projection O_i of the row space of Y along the row space of U in the row space of block Hankel matrix $\mathbf{W}$ consisting of inputs and outputs is made. The name oblique refers to non orthogonal projections and can be interpreted as follows. Make the orthogonal projection of the row space of Y into the joint row space of U and $\mathbf{W}$ and decompose the result along the row space of W. The oblique projection O_i is denoted $O_i = Y/_U\mathbf{W}$. Steps two and three in the MOESP algorithm are now replaced with the following:

- Make the oblique projection $O_i = Y/_U\mathbf{W}$
- Calculate the SVD composition of the weighted O_i:

$$W_1 O_i W_2 = \begin{pmatrix} U_1 & U_2 \end{pmatrix} \begin{pmatrix} S_1 & 0 \\ 0 & 0 \end{pmatrix} \begin{pmatrix} V_1^T \\ V_2^T \end{pmatrix} = U_1 S_1 V_1^T, \tag{4.20}$$

  where W_1 and W_2 denotes weighting matrices. For more information, see Overschee and De Moor (1999) [3].

4.3.3 Model reduction

The model order of a system can be reduced using the balanced realization of the original system. For a more detailed description of balanced model reduction, see Johansson (2004) [10].

The observability and reachability of the states answers the two question:

- *What states can be reached with a given input energy assuming zero initial state?*

- *What state energy is necessary for $u_k = 0$ in order to obtain specified output energy?*

The reachability and observability of a system can be analysed using the reachability and observability Gramians, P and Q. Furthermore, there exists a transformation $z_k T x_k$ such that the reachability Gramian Pz equals the observability Gramian Q_z. This property is obtained by choosing a state space representation z where

$$P_z = Q_z = \Sigma = diag(\sigma_1, \sigma_2, ...), \quad \sigma_i = \sqrt{\lambda_i(PQ)}. \tag{4.21}$$

The ith eigenvalues of matrix PQ is denoted $\lambda_i(PQ)$ and Σ is a diagonal matrix with elements σ_i. Now, the magnitude of the diagonal elements σ_i of the Gramian Σ expresses the relative importance of each state. For small values of σ, the corresponding state have less importance and can be eliminated. This procedure is known as balanced model reduction and can be performed in MatLab with `balreal` and `modred`.

4.3.4 A PID approach

Considering a model of reduced order, i.e. rigid body structure, the postural control can be analysed from a PID (proportional, integrating, derivative) point of view. Having the force equilibrium in eq. (4.8), we make the assumption that PID control via the ankle torque T_{bal} is sufficient to maintain upright stance. The P, I, D components can be written as

$$\text{P} \quad : \quad -mgl \sin\theta(t) - kJ\theta(t) \tag{4.22}$$

$$\text{I} \quad : \quad -\rho J \int_{t_0}^{t} \theta(t)dt \tag{4.23}$$

$$\text{D} \quad : \quad -\eta J\dot{\theta}(t) \tag{4.24}$$

and is characterized by the parameters k, η and ρ. These parameters have the following interpretation. The parameter k can be seen as a spring constant, the parameter η describe the damping of the system and the parameter ρ is related to the importance given to the integral component of the control system. It can be pointed out that the integral component is not initially necessary for stability. However, it is used to eliminate any *bias* of the controlled system.

When a subject is exposed to proprioceptive stimulus, the vibration v introduces disturbances of the sensory inputs, leading to misperception of the position θ and the angular velocity $\dot{\theta}$ yielding the following modification of the P and D actions.

$$\text{P} \quad : \quad -mgl \sin\theta(t) - kJ\theta(t) + b_1 v(t) \tag{4.25}$$

$$\text{I} \quad : \quad -\rho J \int_{t_0}^{t} \theta(t)dt + b_2 v(t), \tag{4.26}$$

$$\tag{4.27}$$

where b_1 and b_2 denotes the relative influence of the disturbance v to the perceptions of θ and $\dot{\theta}$.

Using PID-control together with the force equilibrium in eq. (4.8) yield

$$J\frac{d^2\theta}{dt^2} = mgl \sin\theta(t) + T_{bal}(t) + T_d(t), \tag{4.28}$$

with the control law

$$T_{bal}(t) = -mgl\sin\theta(t) - kJ\theta(t) - \eta J\dot{\theta}(t) - \rho J\int_{t_0}^{t}\theta(t)dt + (b_1 + b_2)v(t). \quad (4.29)$$

Elimination of T_{bal} from eq. (4.29) gives

$$J\frac{d^2\theta}{dt^2} = -\eta J\frac{d\theta}{dt}(t) - kJ\theta(t) - \rho J\int_{t_0}^{t}\theta(t)dt + T_d(t) + (b_1 + b_2)v(t). \quad (4.30)$$

In eg. (4.30), there are three states that affect motion: the angular velocity $d\theta/dt$, angular position θ and the *bias* compensation achieved by integral action. Following Johansson (1988), a Laplace transformation and algebraic simplification gives the transfer function

$$\theta(s) = \frac{\frac{1}{J}(b_1 + b_2)s}{s^3 + \eta s^2 + ks + \rho}V(s) + \frac{\frac{1}{J}s}{s^3 + \eta s^2 + ks + \rho}T_d(s). \quad (4.31)$$

For small movements around the equilibrium $\theta = 0$, $\sin\theta$ is approximately equal to θ. Linearization around the equilibrium gives a transfer functions between stimulus V and disturbance T_d:

$$T_{bal}(s) \approx (Js^2 - mgl)\theta(s) - T_d(s) \quad (4.32)$$

$$= \frac{(b_1 + b_2)(s^3 - \frac{g}{l}s^2)}{s^3 + \eta s^2 + ks + \rho}V(s) - \frac{\eta s^2 + (k + \frac{g}{l}) + \rho}{s^3 + \eta s^2 + ks + \rho}T_d(s). \quad (4.33)$$

Normalization with respect to frequency yields the stimulus dependence

$$T_{bal}(s) = \frac{(b_1 + b_2)\left((\frac{s}{\omega_0})^3 - \frac{g}{l\omega_0}(\frac{s}{\omega_0})^2\right)}{\frac{s}{\omega_0}^3 + \frac{\eta}{\omega_0}(\frac{s}{\omega_0})^2 + \frac{k}{\omega_0^2}(\frac{s}{\omega_0}) + 1}V(s). \quad (4.34)$$

Introducing the parameters *swiftness*, *stiffness* and *damping*, a more functional representation of the transfer function properties is obtained.

- Swiftness: $\omega_0 = \rho^{1/3}$
- Stiffness: k/ω_0^2
- Damping: η/ω_0

The swiftness parameter is a bandwidth (rad/s) and provides information about the highest angular frequency of the disturbance for which the postural control system can make appropriate correction. The stiffness and damping are dimensionless stability parameters. If the swiftness parameter is large, the subject has rapid response to disturbance, i.e. rapid compensation for small deviations from the equilibrium point. The damping parameter characterize the damping of sway velocity and a high value means a large damping.

These three parameters of postural control can be fully determined from the denominator of the transfer function. At this point, we have two possible methods to identify the parameters *swiftness*, *stiffness* and *damping*.

- Estimate a higher order MIMO or SISO model using the previously described subspace approach. Perform balanced model reduction such that a third order model is obtained. From the third order state space system, find a transfer function from u to y using spectral factorization [10]. The last step involves solving the Riccati equation

$$P = APA^T - APC^T(CPC^T + \Sigma_e)^{-1}CPA^T + \Sigma_v, \tag{4.35}$$

where $\{e_k\}$ and $\{v_k\}$ are noise sequences acting on kinematics and measurements. The transfer function is then given by

$$H(z) = C(zI - A)^{-1}K + I, \tag{4.36}$$

where $K = APC^T(CPC^T + \Sigma_v)^{-1}$. For further details, see Johansson (2004) [10]. Finally, calculate the parameters *swiftness*, *stiffness* and *damping* from the denominator of the transfer function $H(z)$.

- Estimate a higher order ARMAX model of the discrete transfer function between input $u(t_k)$ and output $y(t_k)$ and perform model reduction until a third order model is obtained. The ARMAX model is given by

$$y(t_k) = \frac{B(q)}{A(q)}u(t_k) + \frac{C(q)}{A(q)}e(t_k), \tag{4.37}$$

where $\{e(t)\}$ denotes a noise factor. The three parameters *swiftness*, *stiffness* and *damping* can be determined from the autoregressive polynomial $A(q) = q^3 + a_1q^2 + a_2q + a_3$. When using the ARMAX approach, only SISO systems can be considered.

Chapter 5

Identification procedure

All identification was done using the system identification toolbox in MATLAB. Before any models could be estimated, it was necessary to pre-process the data recorded by the force platform and by the Zebris position system.

5.1 Pre-processing of data

The raw force data is afflicted with the characteristics from the force platform itself and need to be normalized in order to get the actual forces and torques. This was done by a MATLAB routine written by Per-Anders Fransson.

Furthermore, the position data recorded by the Zebris system had occasionally spikes due to signal loss from the microphones. The simplest way to reduce the amount of spikes is to remove the spike and interpolate with a straight line between the gap. Like the normalisation procedure, this was performed by a MATLAB routine written by Per-Anders Fransson.

The force data and the Zebris data was synchronized using the input PRBS signal which make it possible to put all data into a single matrix. For identification purposes, the data was arranged in `iddata` objects with the desired number of outputs and the PRBS stimulus sequence as input.

The sampling frequency used was 50 Hz. However, the human postural dynamics is a system with much less frequencies of body movements, typically below 0.5 Hz. When the sample time is near zero, all system dynamics are described by integrators [21]. Hence, almost all poles and zeros are located close [Re Im]=[1, 0] in the unit circle. In this case, problems may arise in the identification process due to bad conditioned matrices. To avoid this, the data was re-sampled with 1/10 of the original sample frequency after low pass filtering resulting in data series with a sample frequency of 5 Hz.

Finally, constants and linear trend were removed before any identification as this are considered to be disturbances in system identification. All channel were normalized with respect to the amplitude of the signal. If one channel has a much larger signal range than the other outputs, this may cause problems in the subspace identification process.

5.2 Model order

It can be argued what model order that is appropriate. If the stimulation disturbance is small, the subject has little problem to maintain postural stability. In this case a suitable model of body dynamics would be to resemble the body as an inverted pendulum. Necessary states for the inverted pendulum are position and velocity of mass centrum in space. Since the variances of the movement along the z-axis is very small compared to the body sway in the $x-y$-plane, the degrees of freedom can be reduced by neglecting movements along the z-axis. Hence, a model of order four can model movements of a rigid body structure.

However, as we are using high intensity stimulus, more complex movement patterns may be adopted by the subject. One strategy is to move around both the ankle joint and the hip joint, like an double linked inverted pendulum. A double linked pendulum can be modeled with 8 states which imply a model of 8th order. Therefore, the estimated MIMO models are of eighth order.

5.3 Evaluation

The estimated models need to be evaluated in some way to assess their simulation or predictive ability. The following methods are used:

- Simulation. The estimated model is driven by an identical input sequence as the PRBS stimulus sequence used in the experiments and the model response is compared to the measured output of the true system. Also, the k-step ahead predictor can be studied to evaluate the predictive ability of the model. If k equals infinity, pure simulation is achieved.

- Percentage Variance Accounted For (VAF) between simulated and true output. VAF is defined as:

$$\left(1 - \frac{Var(y-\hat{y})}{Var(y)}\right) \cdot 100\% \tag{5.1}$$

  VAF should be interpreted as a measure how well the model explain the variation of the output variables. If the simulated output is identical to the true output, VAF equals 100 %. Low values indicate that the model only explain a small fraction of the total variation of the output variables. However, it should be pointed out that a subject always has a large part of spontaneous sway that cannot be explained from the stimulus to the calf muscles. Furthermore, it is possible to achieve negative values of VAF in cases of larger variance of the error signal than the variance of the signal itself.

5.4 Model estimation

As described previously, the data consist of recordings from six trials for each subject. The maximal number of outputs are 24, where 18 are (x, y, z)-coordinates of the six position markers and 6 are forces and torques recorded by the force platform. The approach was to estimate one model for movements in the anteroposterior direction and one model for the lateral movement. In the anteroposterior model, the y-coordinates of the knee, hip, shoulder and head together with the force F_y and torque M_x were used as outputs. The neck marker was considered redundant due to its high correlation with the head marker. Furthermore, the ankle marker was excluded since this point naturally move with very small amplitude. In this way, the reduced data set consists of one input (stimulus) and seven outputs. The lateral models were estimated from data sets with the same structure, but with the use of the x -coordinates and the force F_x and the torque M_y instead.

The data sets were divided into five periods with quiet stance being the first 30 s period. The four stimulus periods, are denoted period I-IV, have all the length 50 s. All data series used were detrended to remove offsets and linear trends in the data. The delay time in the system was set to 0.2 s which is known from previous experiments to be a good approximation. The initial state vector was estimated for highest accuracy in the simulations.

The discussion concerning the dynamics of a double segment inverted pendulum suggests models of 8th order. However, models of less or of higher order could be of interest. Therefore, model orders from 4 to 12 were tested and Akaikes Final Prediction Error (FPE) was used to decide appropriate model order. However, in many cases the FPE value suggested a model order higher than 8. As the model order increases, the risk for unstable estimated systems also increases resulting in simulations that "explodes". Hence, the best thing is to use a model order that capture most of the major dynamics of the system without being unnecessary high. Considering the above, models of 8th order were chosen.

One model on each of the four stimulus periods was estimated using the N4SID algorithm. A fifth model was estimated on the whole stimulus period of 200 s. Two evaluation strategies were used. First, the estimated model of stimulus period I was validated against itself and the subsequent three periods. The purpose of this was to see variations of VAF when the first period is validated against all four stimulus periods. For instance, if there is some kind of time varying phenomenon due to any adaptation process, a model estimated on period I may not be a good model on later time segments since the system may have changed. If this is the case, it will be seen in the VAF values when comparing simulated and true output of the system.

As another validation strategy, the four models estimated periods I, II, III and IV were validated against the same time segment. Here, the hypothesis was that the subject may adapt to the stimulus resulting in reduced body sway. In other words, the stimulus will cause smaller part of the recorded body sway due to this adaptation. Consequently, the calculated VAF values will decrease for models of forthcoming time segments if this hypothesis is true.

The mean values of the VAF values were calculated and plotted for all simulation strategies.

The parameters *swiftness*, *stiffness* and *damping* were calculated for periods I-IV and the complete stimulus period for every trial. This was done both with the anteroposterior torques Mx and the shoulder position as outputs. Statistical analysis of differences between periods I-IV, and between the test trials CC, CO, SD24C, SD24CO, SD36C and SD36O was performed using the Wilcoxon nonparametric test since the values were not normally distributed.

Chapter 6

Results

The subjective scoring of sleepiness showed that 15 of the 17 subjects considered themselves more sleepy after 36 hours of sleep deprivation. The eye test (oculomotor response) indicated a reduced alertness due to sleep deprivation. This was most evident in eye movement velocity which was statistically lower during sleep deprived conditions, especially in cases of large amplitudes in movements.

The recorded EEG data suggested that all subjects had stayed awake during 36 hour test period. Due to measurement failure, data was lost from the posturographic examination of one of the subjects reducing the number of subjects to 16.

The amplitudes of postural body sway were always largest in the anteroposterior direction with a distinct peak at the time of onset of the vibration stimulus. This peak can be resembled as some kind of initial step response.

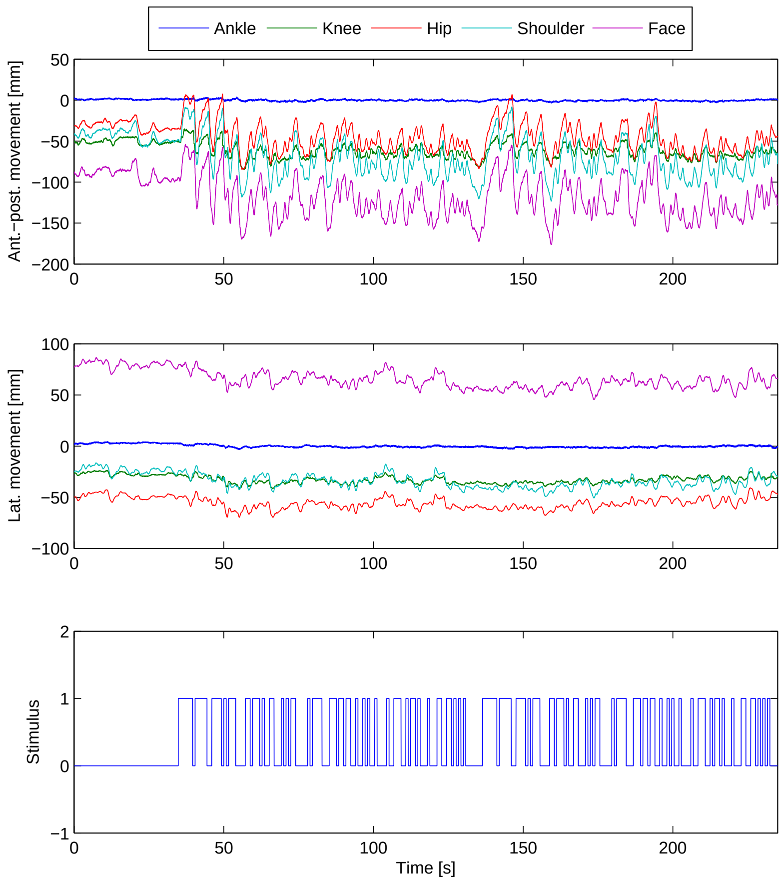


Figure 6.1: Recorded body sway of one subject. Body sway is clearly greater in magnitude in the anteroposterior direction than it is in lateral direction. Body movements along the z-axis is negligible compared to anteroposterior and lateral movement and are therefore not plotted.

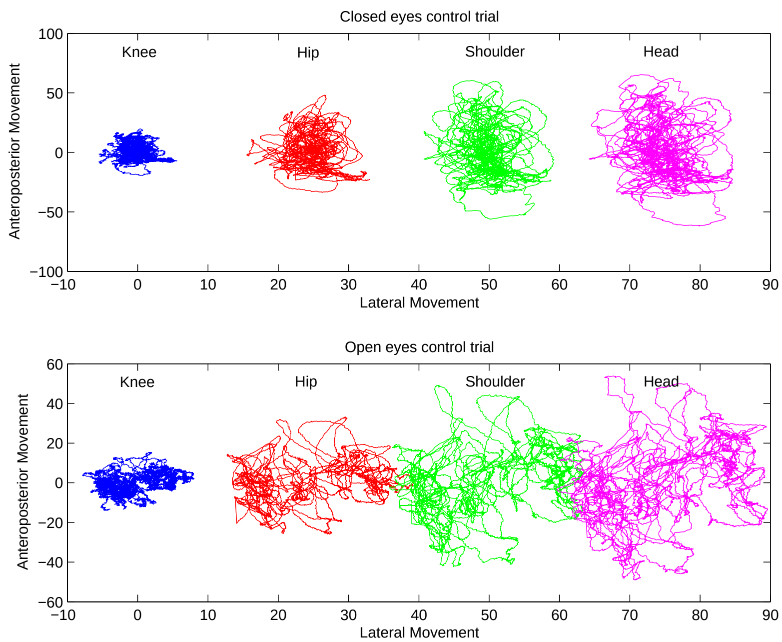


Figure 6.2: XY-plot of recorded body movements from one subject during the complete test sequence including quiet stance and stimulus period. Both with closed eyes and open eyes, the body sway is greater in magnitude at the head and progressively less in magnitude down towards the feet. Constants have been added to the lateral data series for better visualisation.

6.1 Variance and correlation analysis

6.1.1 Variance of torque data

The variance analysis of measured torque in closed and open eyes tests in the anteroposterior direction can be seen in figure 6.3. In the closed eyes trials, the body sway was significantly larger for three of the four stimulus period compared with closed eyes control trials. For the 36 hour closed eyes trials, the body sway was significantly lower in the first stimulus period when comparing with control trials. This was not the case in subsequent stimulus periods. However, the most interesting fact is that the body sway measured in the 24 hour trial seemed to be significantly larger than the 36 hour recording (figure 6.3, period I and period IV).

The open eyes anteroposterior torque variance results do not suggest that the postural control is effected by sleep deprivation. The only significant difference was obtained for the third stimulus period where both the 24 hour and the 36 hour recording clearly proved a larger body sway than the open eyes control trials. Since the statistical difference was more evident in the closed eyes trials where statistical difference from the control trial was determined in period II, III and IV, it can be concluded that postural control is effected by 24 hour sleep deprivation with closed eyes.

Figure 6.4 show the results of the lateral torque variance. In the closed eyes trials, where there is statistical difference suggesting larger body sway in 24 hour recording in period II, III and IV comparing with closed eyes control tests. Furthermore, periods III and IV have higher significance level with $P < 0.01$. Also, the 36 hour sleep deprivation recording produced significant difference ($P < 0.05$) in period III and IV suggesting weaker postural control in the second half of the stimulus period.

The open eyes lateral torque variance analysis produced significant difference in the 24 hour trial in period II, III and IV. In the 36 hour recording, there was only a significant difference in period IV.

From the results, effects on postural control due to sleep deprivation is more distinct in lateral data than it is in anteroposterior recordings. In the lateral results, larger body sway can be concluded in later stimulus periods in both closed and open eyes trials. The anteroposterior results do only suggest larger body sway in closed eyes trials.

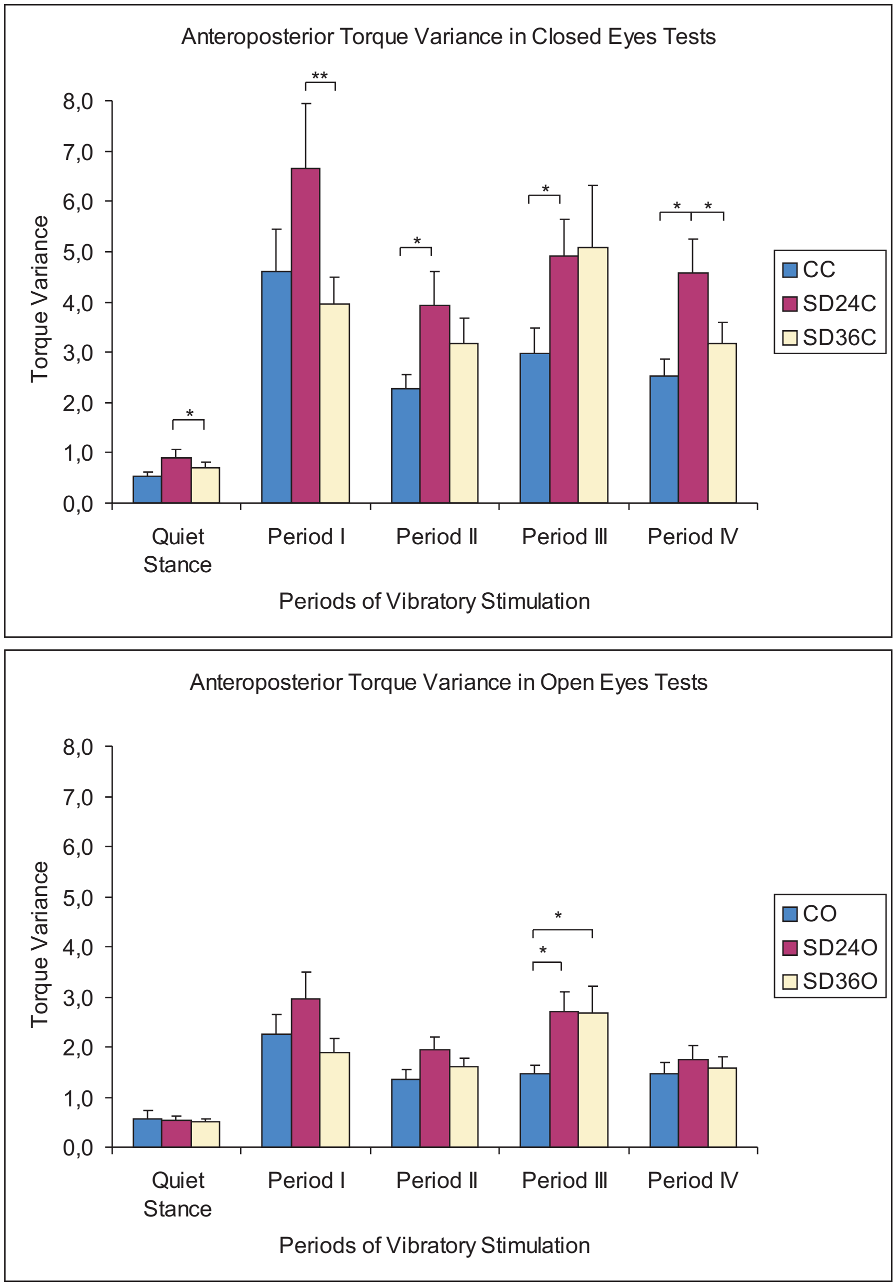


Figure 6.3: Anteroposterior results of torque variance recorded by force platform. * denotes a significance level of $P < 0.05$ and ** denotes significance level with $P < 0.01$.

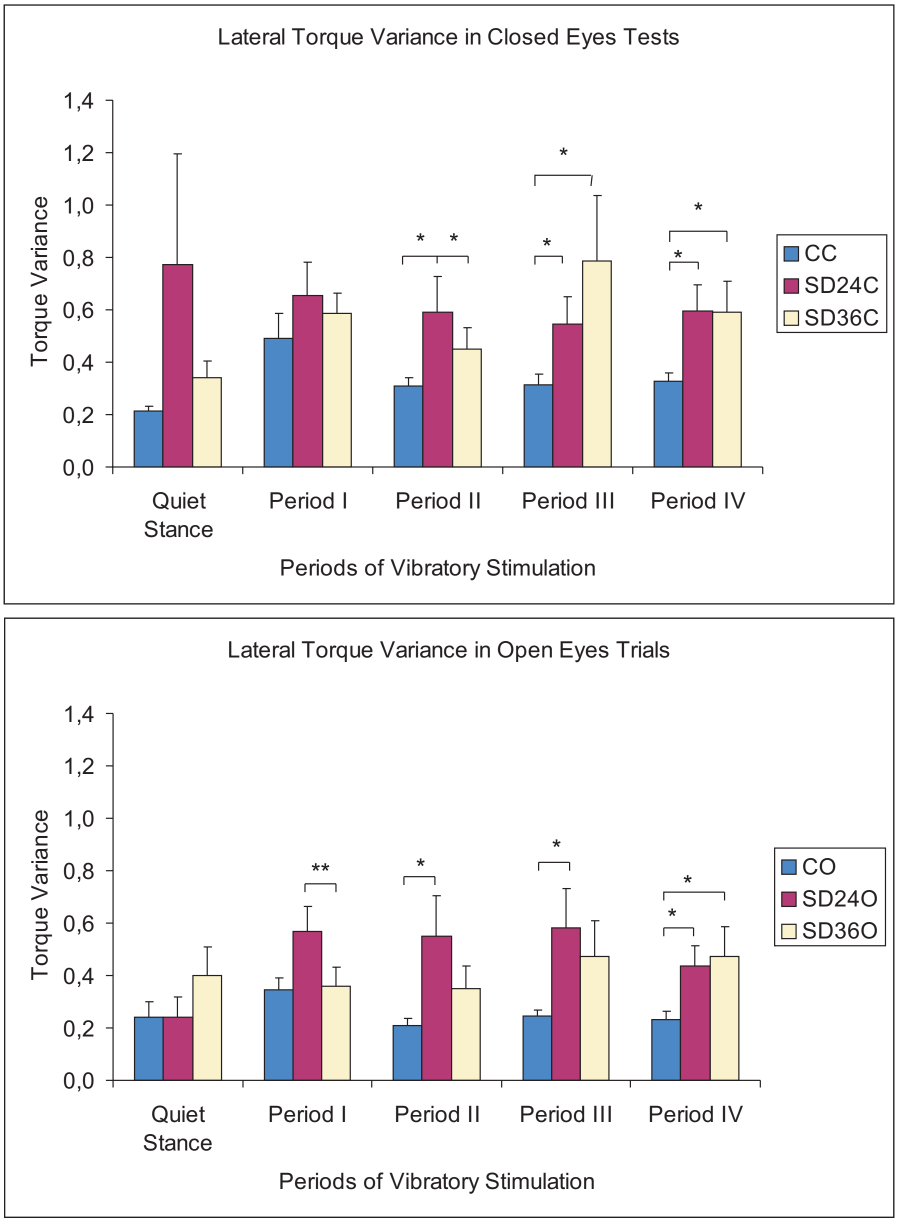


Figure 6.4: Lateral results of torque variance recorded by force platform. * denotes a significance level of $P < 0.05$ and ** denotes significance level with $P < 0.01$.

The study of adaptive behavior to the stimulus sequence, i.e. comparisons of torque variance between period I and period IV, some indications for decreased adaptive ability were found. In the anteroposterior torque, no statistical differences could be concluded between period I and period IV in the trials SD24C, SD24O and SD36C. Consequently, poor anteroposterior adaptation to the stimulus is suggested in three of four sleep deprivation tests. Laterally, all trials except the open eyes control test (CO) were found to have non significant differences when comparing period I with period IV. The statistical results of the comparisons period I-IV and QS-period I are presented in table 6.1.

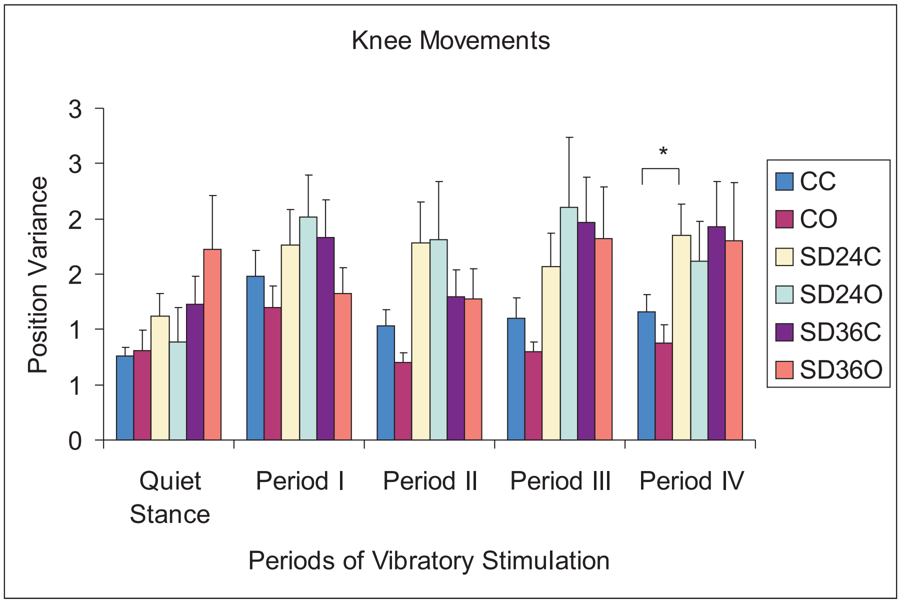
Table 6.1: Summary of the statistical result found between the torque variance values during quiet stance (QS), period I and period IV. P-values less than 0.05 are considered significant and ns denotes non significant ($p > 0.05$).

	Anteroposterior		Lateral	
	QS-I	I-IV	QS-I	I-IV
CC	$p < 0.00$	$p < 0.01$	$p < 0.01$	ns
CO	$p < 0.01$	$p < 0.05$	$p < 0.05$	$p < 0.05$
SD24C	$p < 0.00$	ns	ns	ns
SD24O	$p < 0.00$	ns	$p < 0.01$	ns
SD36C	$p < 0.00$	$p < 0.05$	$p < 0.05$	ns
SD36O	$p < 0.00$	ns	ns	ns

6.1.2 Variance of position data

Considering anteroposterior movements, the analysis of recorded body motions show almost no significant difference between the control trials and both the 24 hour and the 36 hour sleep deprivation trials.

Laterally, no statistical differences were found between the control trials and the sleep deprivation trials in period I. However, in period IV, the lateral data show statistical differences between the control test and 24 hours of sleep deprivation at the knee ($p < 0.05$), the hip ($p < 0.05$), shoulder ($p < 0.05$) and the head ($p < 0.05$). Also, statistical differences between control test and 36 hour of sleep deprivation in period IV was found at the hip ($p < 0.05$). This result can be seen in figures 6.5 and 6.6.



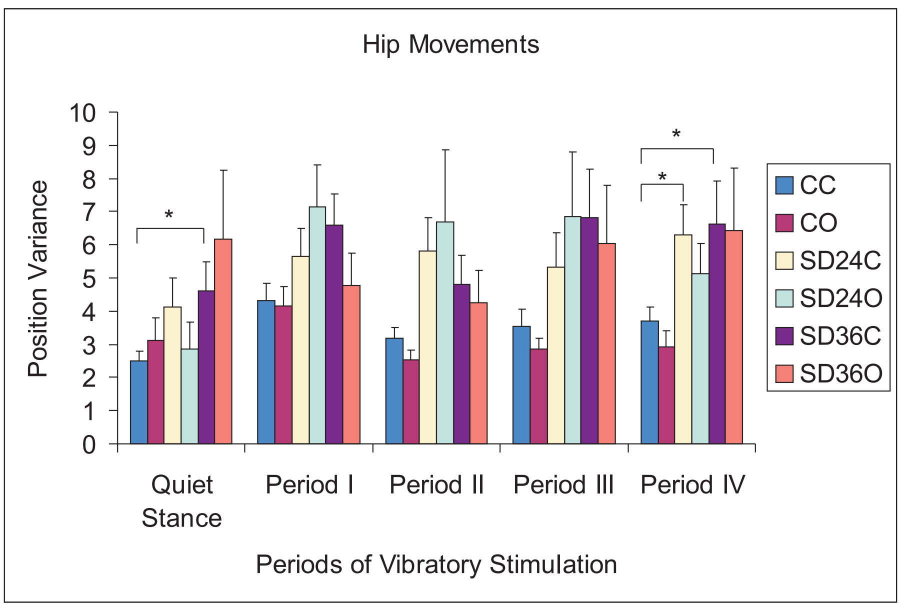
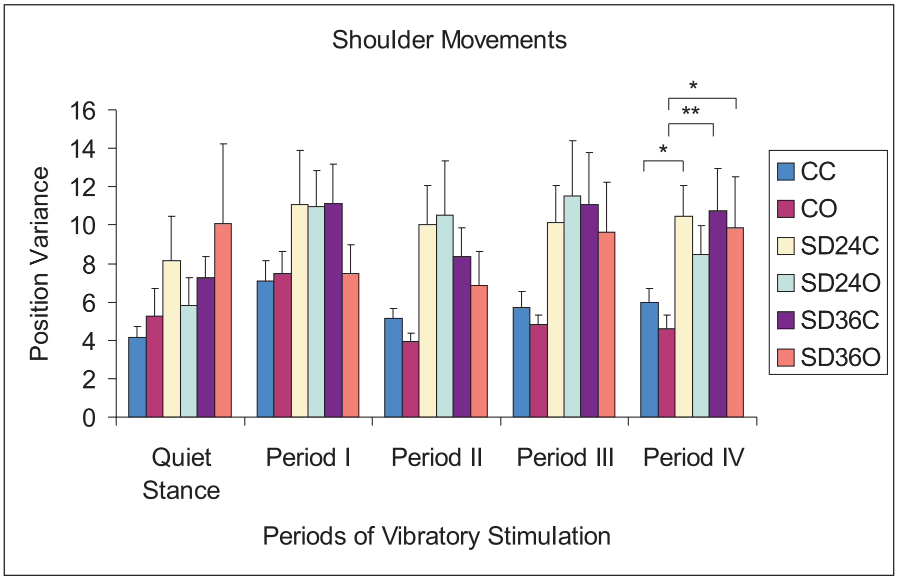


Figure 6.5: Lateral variance of the knee and the hip. Statistical differences are found in period IV between CC and SD24C at the knee, between CC and SD36C and SD24C at the hip.



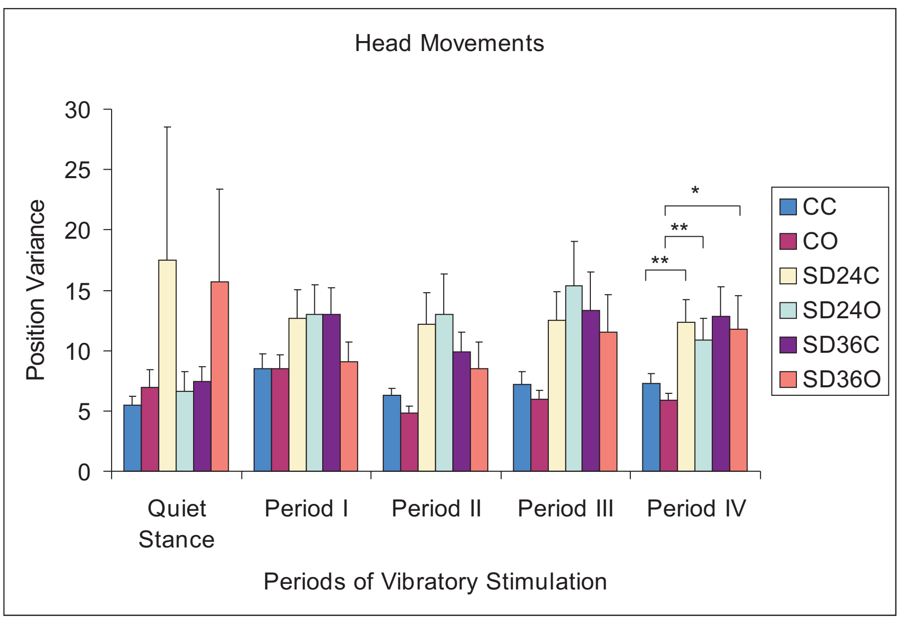


Figure 6.6: Lateral variance of shoulder and head position. The statistical differences in period IV between sleep deprivation trials and control trials are more distinct than for knee and hip movements.

The initial response to the onset of vibratory stimulation is an increase of body sway. This can be seen in the statistical analysis performed on the position recordings presented in table 6.2 where variance of period I is compared with variance of period IV. In every trial, CC, CO, SD24C, SD24O, SD36C and SD36O, there are significant differences of anteroposterior body sway between quiet stance (QS) and the first stimulus period (period I), see table 6.2. A comparison between stimulus period I and period IV show statistically differences of body sway (position variance) in both closed eyes and open eyes control trials and in 24 hour sleep deprived closed eyes trials (CC, CO and SD24C). However, in the 24 hour open eyes and the two 36 hour sleep deprivation tests, there is statistical evidence suggesting no difference between the recorded body sway of period I and period IV. Thus, the decrease of body sway found in the control trials is not present in the sleep deprived tests, especially in case of 36 hour sleep deprivation.

The lateral body sway was found to decrease in the control test with eyes open (CO) and in the two trials with 24 hours of sleep deprivation (table 6.2). In all the other trials, no significant differences of variance were found between period I and period IV.

Table 6.2: Statistical analysis of calculated variances of position data. For p-values larger than 0.05, non significance (ns) are concluded.

		Anteroposterior		Lateral	
		QS-I	I-IV	QS-I	I-IV
CC	Head	$p < 0.00$	$p < 0.01$	$p < 0.01$	ns
	Shoulder	$p < 0.00$	$p < 0.01$	$p < 0.01$	ns
	Hip	$p < 0.00$	$p < 0.01$	$p < 0.05$	ns
	Knee	$p < 0.00$	$p < 0.05$	$p < 0.05$	ns
CO	Head	$p < 0.01$	$p < 0.05$	ns	ns
	Shoulder	$p < 0.01$	$p < 0.05$	ns	ns
	Hip	$p < 0.00$	$p < 0.01$	ns	$p < 0.05$
	Knee	$p < 0.01$	$p < 0.05$	ns	$p < 0.05$
SD24C	Head	$p < 0.05$	$p < 0.01$	$p < 0.05$	ns
	Shoulder	$p < 0.01$	$p < 0.05$	ns	ns
	Hip	$p < 0.01$	$p < 0.05$	ns	ns
	Knee	$p < 0.01$	$p < 0.05$	$p < 0.05$	ns
SD24O	Head	$p < 0.00$	$p < 0.01$	$p < 0.01$	ns
	Shoulder	$p < 0.01$	$p < 0.05$	$p < 0.01$	$p < 0.05$
	Hip	$p < 0.00$	ns	$p < 0.01$	ns
	Knee	$p < 0.00$	ns	$p < 0.01$	ns
SD36C	Head	$p < 0.00$	$p < 0.01$	ns	ns
	Shoulder	$p < 0.00$	ns	ns	ns
	Hip	$p < 0.00$	ns	ns	ns
	Knee	$p < 0.00$	ns	$p > 0.05$	ns
SD36O	Head	$p < 0.05$	$p < 0.05$	ns	ns
	Shoulder	$p < 0.01$	ns	ns	ns
	Hip	$p < 0.01$	ns	ns	ns
	Knee	$p < 0.00$	ns	ns	ns

6.1.3 Correlation

Figures 6.7 and 6.8 show calculated correlation coefficients between movements of different body section in anteroposterior and lateral plane, respectively. In every period including quiet stance, correlation coefficients are calculated between head-shoulder, head-hip, head-knee, shoulder-hip, shoulder-knee and hip-knee. Evidently, there is a distinct increase of correlation of anteroposterior movement at the onset of stimulus in period I. This suggest that a more rigid body posture is adopted, and hence, the assumption of inverted pendulum dynamics in the mathematical modeling section is supported.

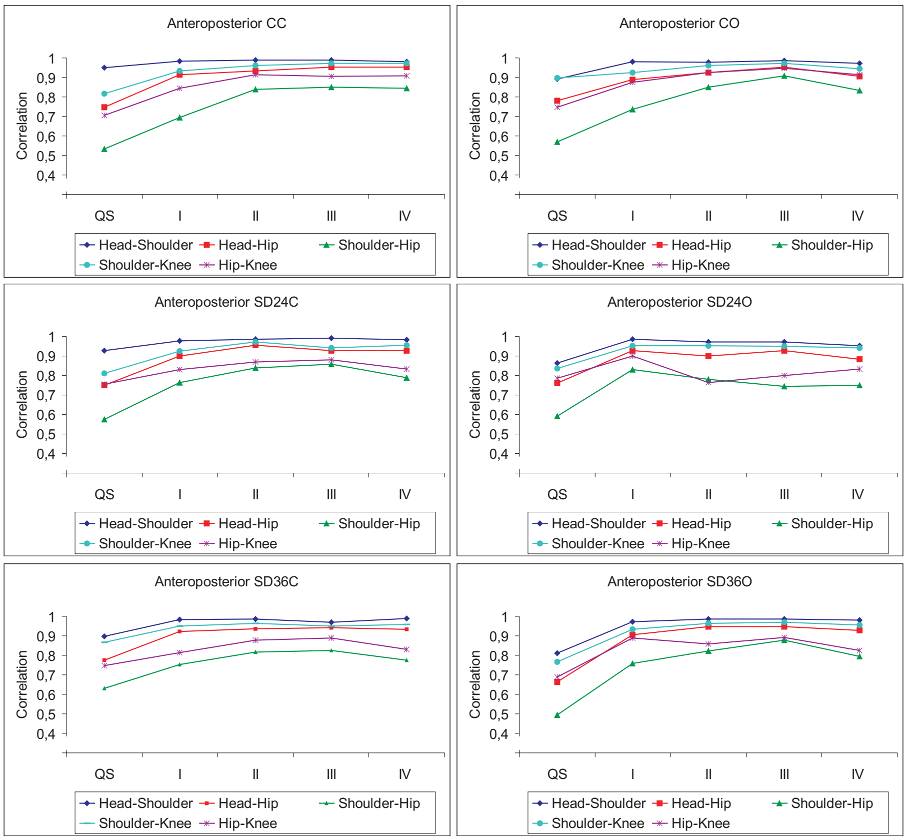


Figure 6.7: Anteroposterior correlation between different body sections over time. The correlation values suggest that the subjects have adopted a movement pattern with more coordinated movements of the body sections due to the vibration stimulus.

The correlation between lateral moving body segments do not produce the same result as in the anteroposterior case. Here, there is obviously a high correlation in all periods. However, indications can be seen in the lateral open eyes trial suggesting that some adaptation of a more rigid posture occur in the first stimulus period (period I).

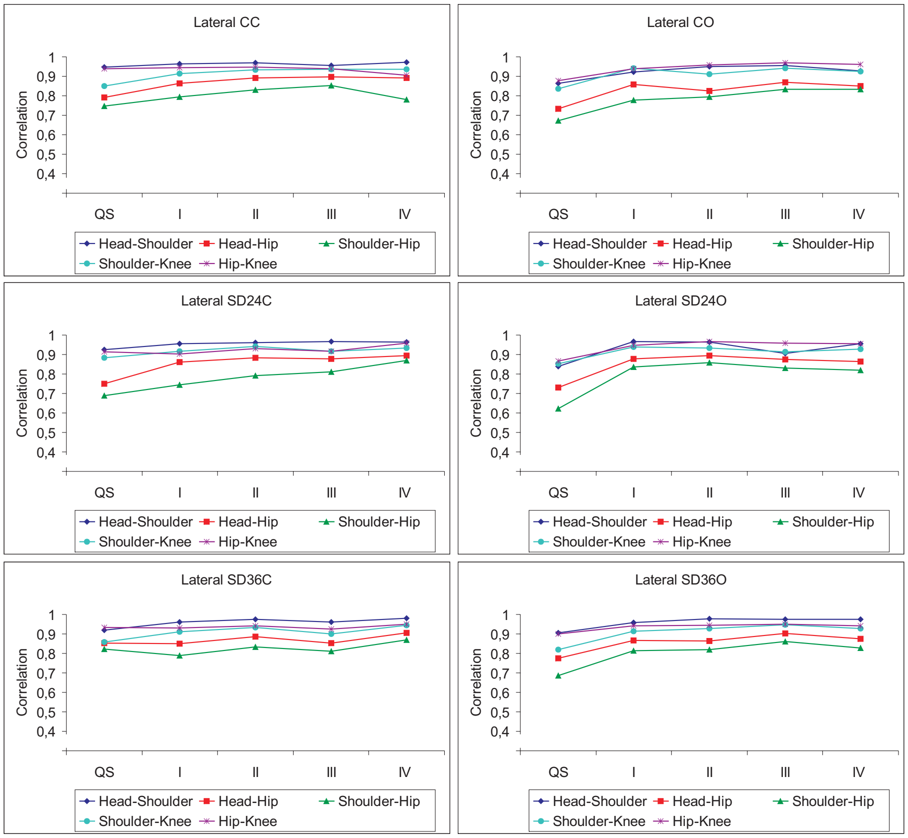


Figure 6.8: Lateral correlation between different body sections over time. There is a small increase of correlation in the open eyes trial, especially in the 24 hours of sleep deprivation with (SD24O).

6.2 System identification analysis

In this section, the results of the system identification analysis are presented. First, coherence spectrum and power spectrum for a typical subject are shown.

6.2.1 Coherence spectrum

The quadratic coherence spectrum was high between the input (stimulus) and outputs of the anteroposterior plane for frequencies below 0.1 Hz. Above this frequency, the coherence values were close to zero. In the lateral plane, the coherence spectrum show low values for all frequencies suggesting that the input-output relation has large noise levels or large amounts of spontaneous sway. Interestingly, the coherence for the open eyes trials were systematically lower than in the closed eyes trials. Figure 6.9 shows the coherence between stimulus and anteroposterior and lateral torques, respectively, of a subject in a closed eyes control trial.

It can be concluded that the data from the posturographic trials is not the easiest data to model. As input signal we have a vibration signal of a certain frequency which is turned on and off according to a PRBS schedule. This input signal differs a lot from the outputs (forces, moments and positions) and the coherence between input and output are therefore not necessarily optimal, i.e. close to 1.

Furthermore, it should again be pointed out that the recorded body sway consists of partly stimulus induced sway and partly spontaneous body sway from the normal balance control. Depending on the relation between these two types of body sway, the coherence is effected. Subjects with large amounts of spontaneous sway are more difficult to model since the coherence between stimulus and sway will be low.

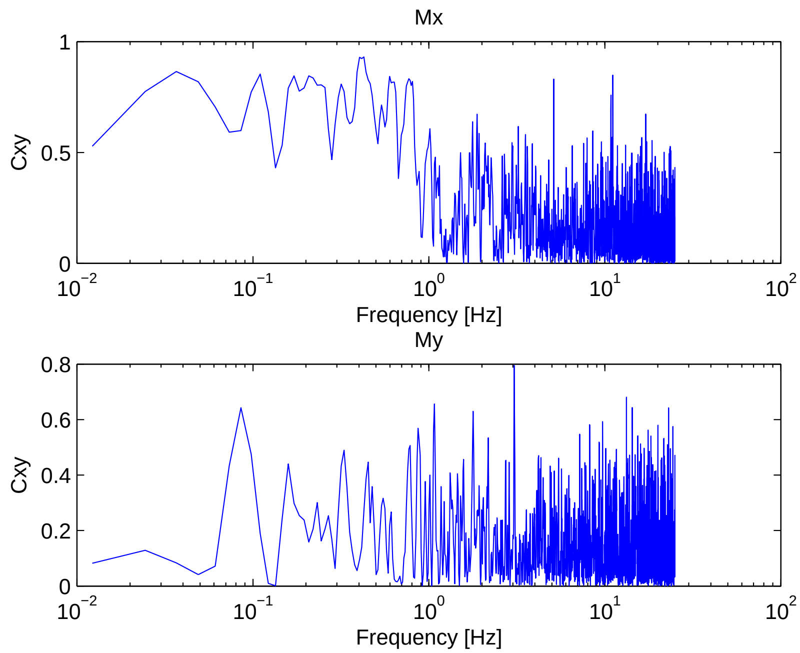


Figure 6.9: Coherence spectrum between stimulus and anteroposterior and lateral torques (Mx and My) for closed eyes control test (CC). The coherence between stimulus anteroposterior torque (Mx) is higher than in the lateral case.

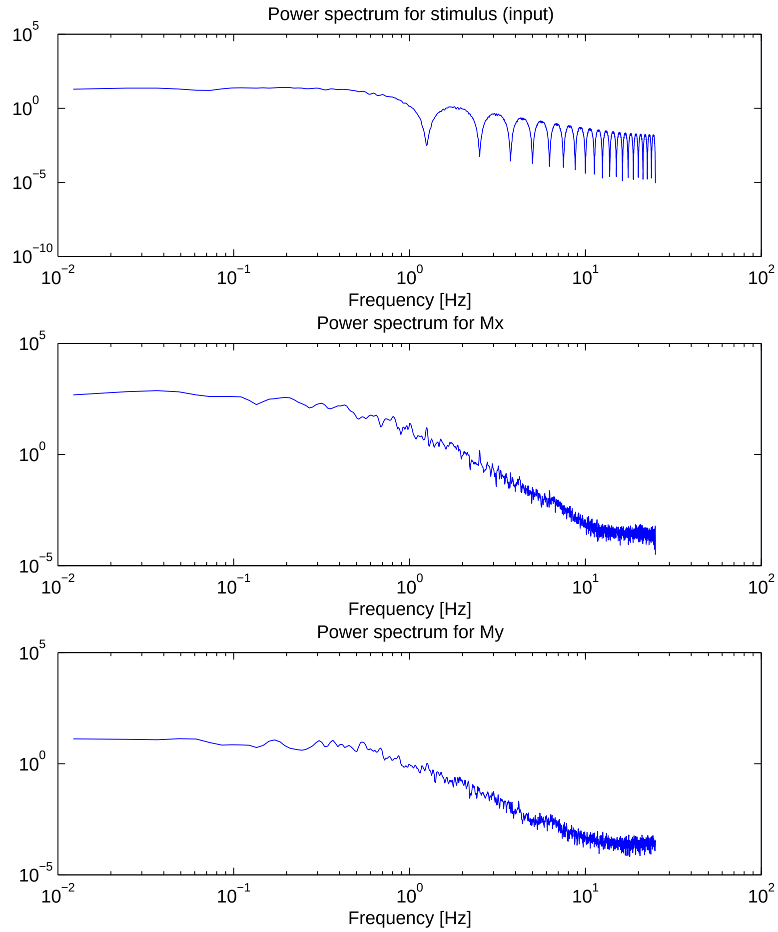


Figure 6.10: Power spectrum for stimulus (input), anteroposterior torque (Mx) and lateral torque (My) for closed eyes control test (CC). The plots show that the stimulus signal used as input has appropriate bandwidth according the power spectrum of the outputs.

6.2.2 Anteroposterior modeling

The following figures show the mean VAF values for the six trials, CC, CO SD24C, SD24O, SD36C and SD36O, when the model of stimulus period I is validated against all four periods. The most noticeable fact is that the VAF value decreases when the validation is performed on subsequent data segments. However, it is obvious that output error, the difference between simulated and true output, will be larger when a model is estimated on one time segment and simulated on a later time segment. Therefore, the calculated VAF value is expected to decrease. Interestingly, the VAF values seem to increase in when the third stimulus period is used as validation set. The explanation of this would probably be that the stimulus sequence of the third period contains longer intervals of vibration which results in a larger response of body sway. The fraction of spontaneous sway that can not be explained from the model will be lower and hence, the model will have greater ability to explain the measured body sway.

In some cases, the error signal has a larger variance than the variance of the true output which result in negative VAF values. Negative VAF values can be a result of an unstable model with eigenvalues of the system matrix A outside the unit circle.

However, the variance of the error will also be large if the characteristics of the system have changed during time. If the validations on subsequent time segments result in many negative VAF values compared to the validation on the same data set and the estimated system have all poles inside the unit circle, i.e. the system is stable, changes of the system dynamics are suggested. In the anteroposterior case, the model estimated on the first stimulus period give some negative VAF values when the validation is performed on period three and four. This suggest that some kind of time varying phenomenon is present. No clear difference between the six trials can be noticed in these anteroposterior models.

When looking on the four models estimated on each single stimulus period and validated against the same data set, the calculated VAF values are higher, around 50-60 % for all periods. Interestingly, there is an increase in VAF in period three which can be explained by the longer stimulation intervals of this period. A comparison between the six trials suggests that the two trials after 36 hours of sleep deprivation have higher values of VAF in period three and four. This can not be seen in the control trials. Evidently, a larger part of the body sway measured in the two last periods in the SD36C and SD36O can be explained from the stimulus to the calf muscles. This may be a result of decreased attention which negatively effect the adaptation to the stimulus. It should be mentioned that all of the estimated models have non-negative VAF values.

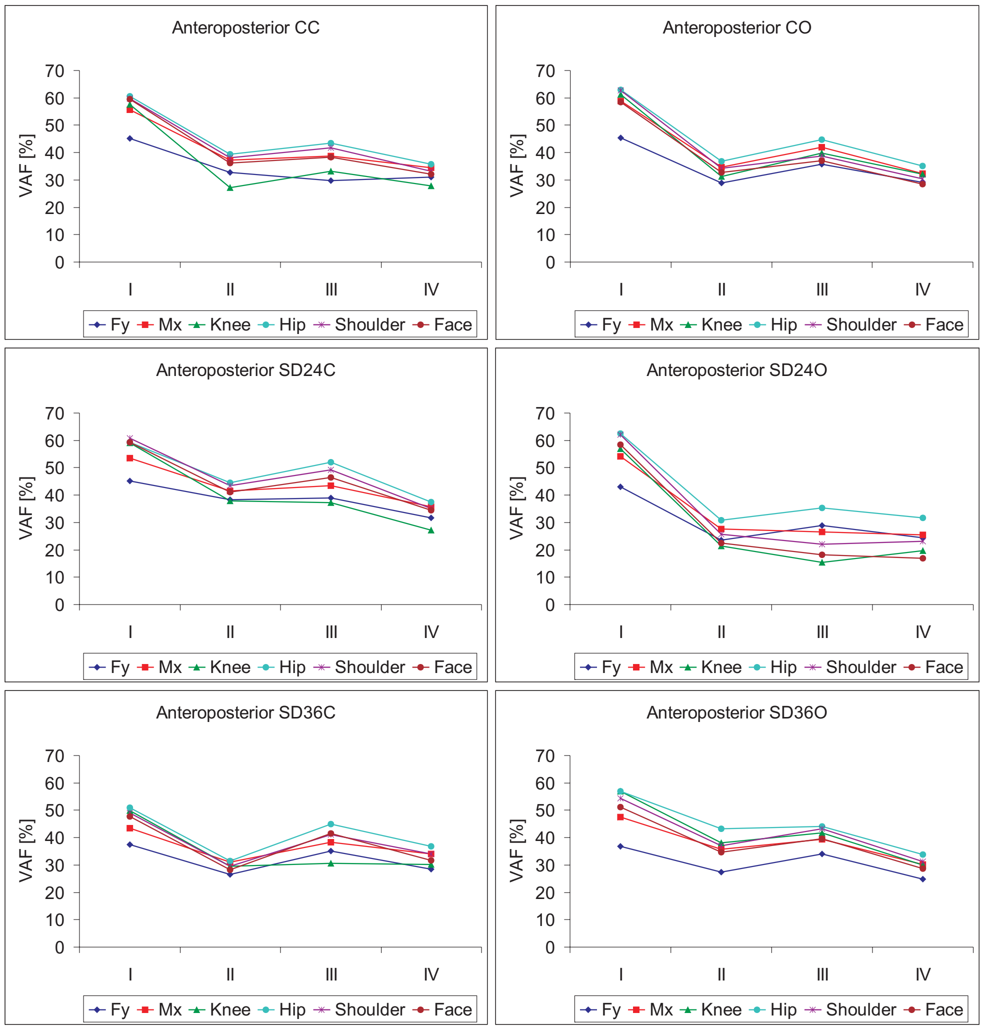


Figure 6.11: Mean VAF of anteroposterior model 1 estimated on the first stimulus period. Simulation and calculation of VAF values are done on all four stimulus periods.

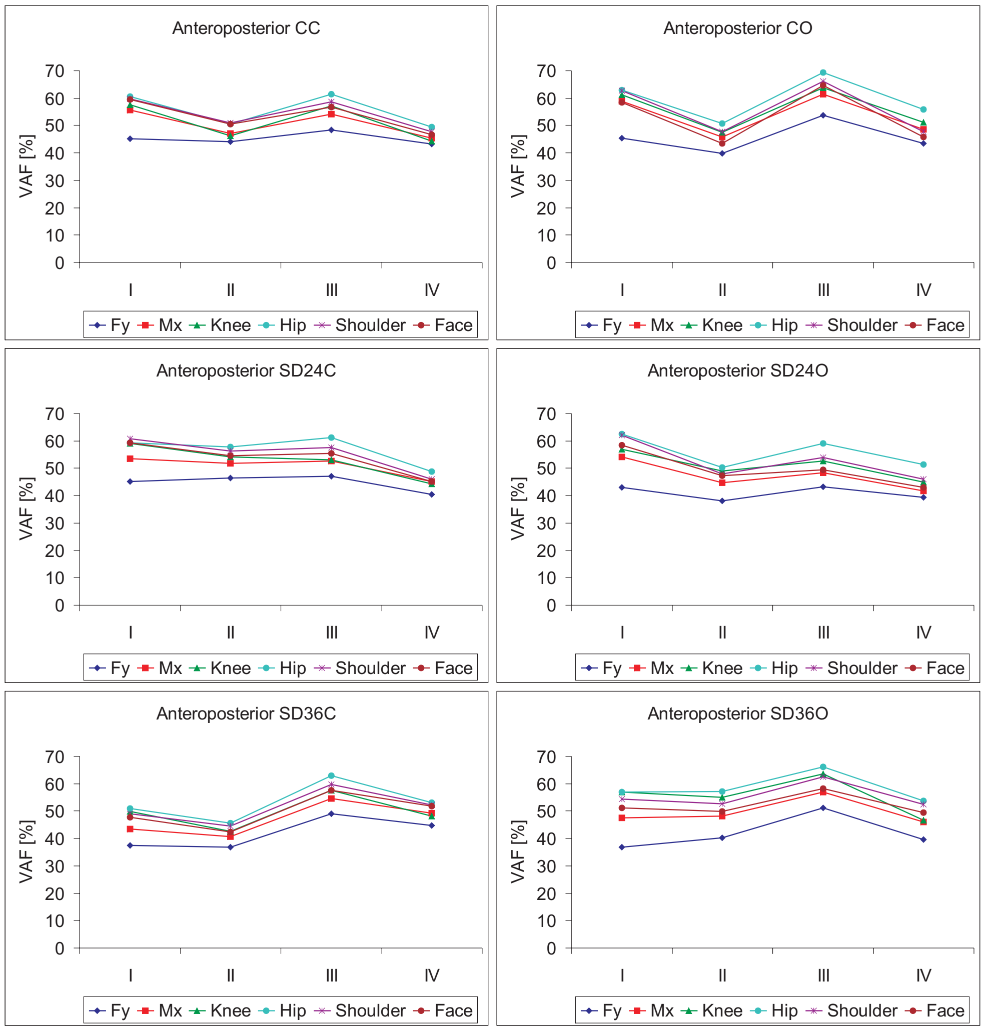


Figure 6.12: Mean VAF of the four anteroposterior models estimated on period I to period IV. Simulation and calculation are done using the same data set as for the estimation process.

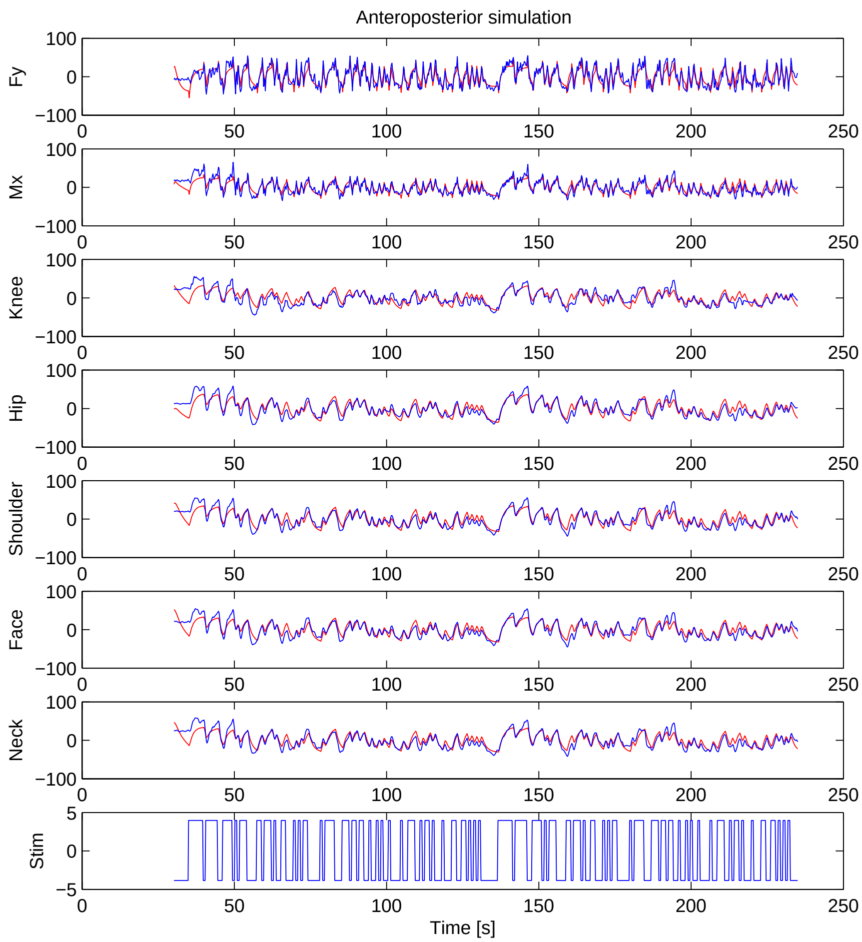


Figure 6.13: Eight order model estimated on the whole stimulation period. Outputs are position of knee, hip, shoulder, head and neck and the force and torque in anteroposterior direction (Fy, Mx).

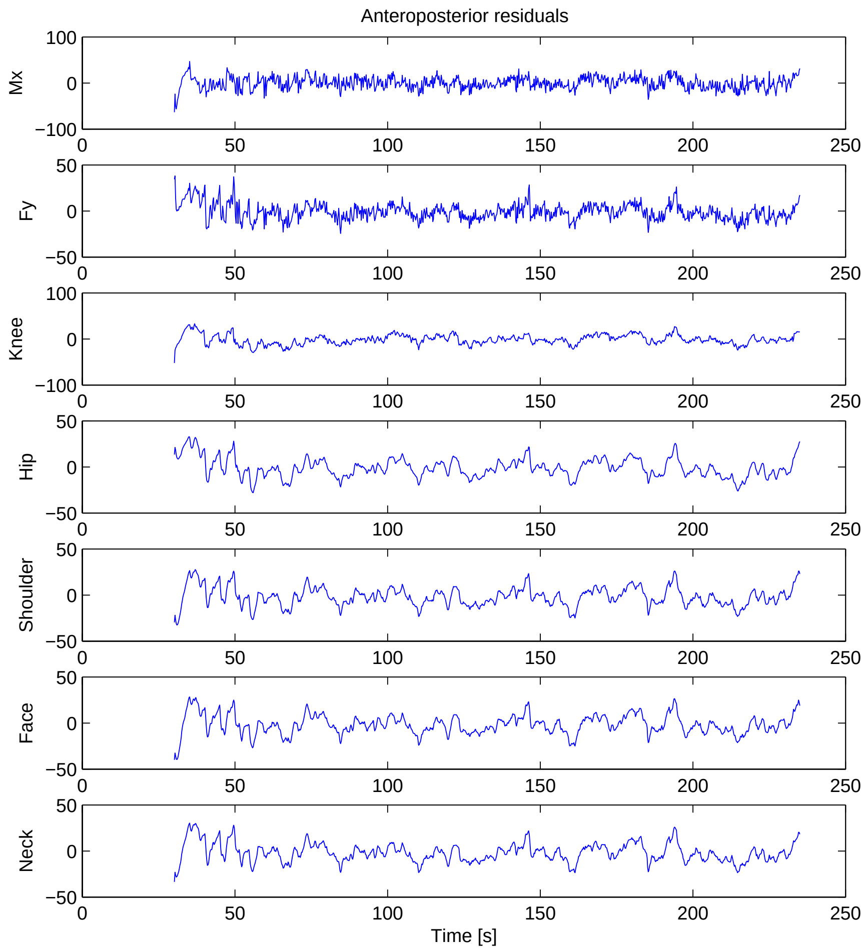


Figure 6.14: Residual series of the model simulation in figure 6.13.

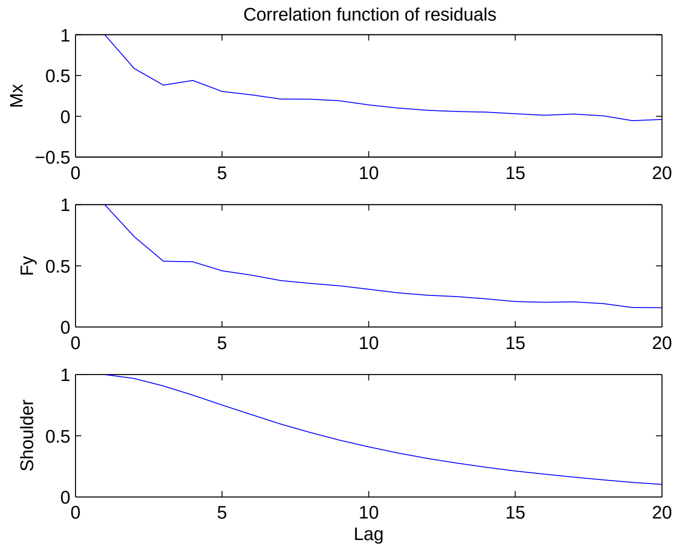


Figure 6.15: Correlation function of residuals from the anteroposterior model in figure 6.13. The correlation function for the residuals of Mx, Fy and shoulder are plotted for lags up to 20 samples.

6.2.3 Lateral modeling

The same analysis on models with lateral movements as outputs indicate some differences compared to the anteroposterior case. First, the response to the stimulus is not as "direct" as when anteroposterior movements are studied. Remembering the coherence analysis, this is naturally since the coherence between stimulus and the response variables are significantly lower in the lateral case than in the anteroposterior case. It is therefore expected that the VAF values calculated for simulated output are lower than for the corresponding anteroposterior model. Second, the validation of the model on the first stimulus period on subsequent periods result in a large part of negative VAF values.

In cases of negative VAF values, i.e. the variance of the error signal is greater than the variance of the measured signal, the model fails to explain any of the variation of the output variables. In figure 6.16, cases of negative VAF values are counted as zero which explain the large decrease of mean VAF values when the first lateral model is simulated and compared with data from subsequent periods. However, in five of the six trials, there are indications of a further decrease of VAF during period II-IV. This suggest, that the model estimated on the first stimulus period performes worse when simulated and compared against data from period IV than when compared with data from period II. Hence, the subject is using some kind of adaptation strategy suppressing the influence of the stimulus disturbance.

The lateral models estimated on the full stimulus period showed poor performance with negative VAF values or close to zero. Evidently, the lateral data is more non-stationary than the anteroposterior data which causes bad modeling properties. The lateral results should be interpreted with these restrictions in mind.

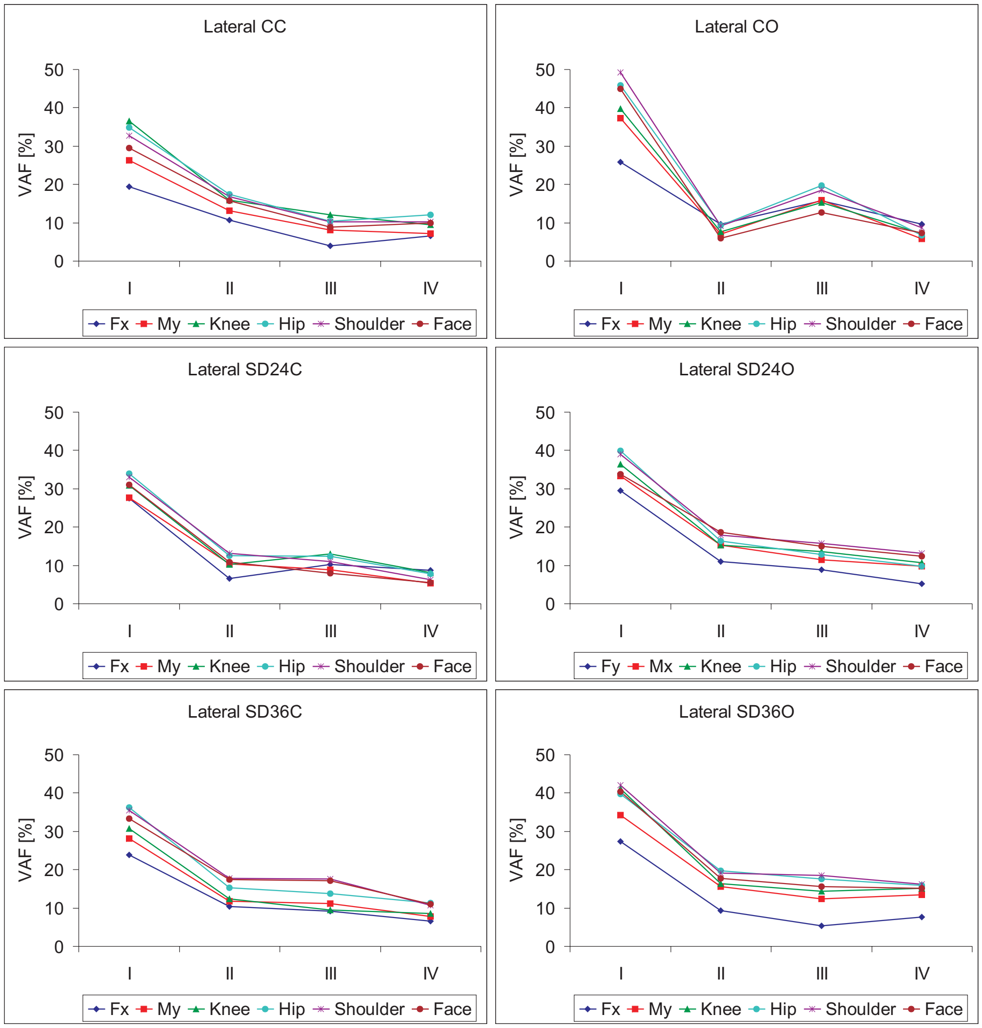


Figure 6.16: Mean VAF of lateral model 1 estimated on the first stimulus period. Simulation and calculation of VAF are done on all four stimulus periods.

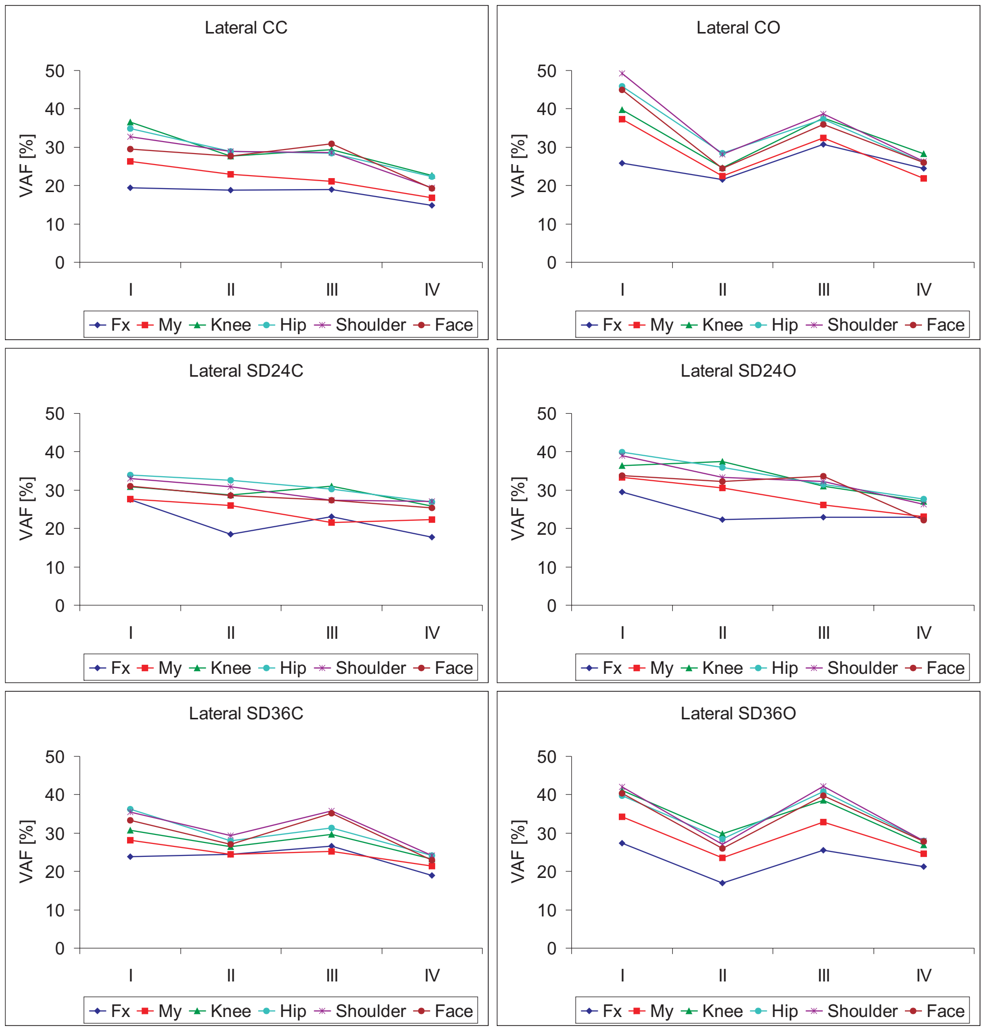


Figure 6.17: Mean VAF of the four lateral models estimated on period I to period IV. Simulation and calculation are done using the same data set as for the estimation process.

6.2.4 Parameter extraction

In this section, the main results of the analysis of the parameters swiftness, stiffness and damping are presented. Statistical analysis using the non-parametric Wilcoxon test showed little significance for differences, both when comparing different trials and when comparing different stimulus periods. The most clear differences were found in the stiffness parameter when considering the complete stimulus period and the sub-periods I-IV, see figure 6.21. In figures 6.18 to 6.22, the star notation is used for significance levels. A significance level with P-values less than 0.05 is denoted with one star.

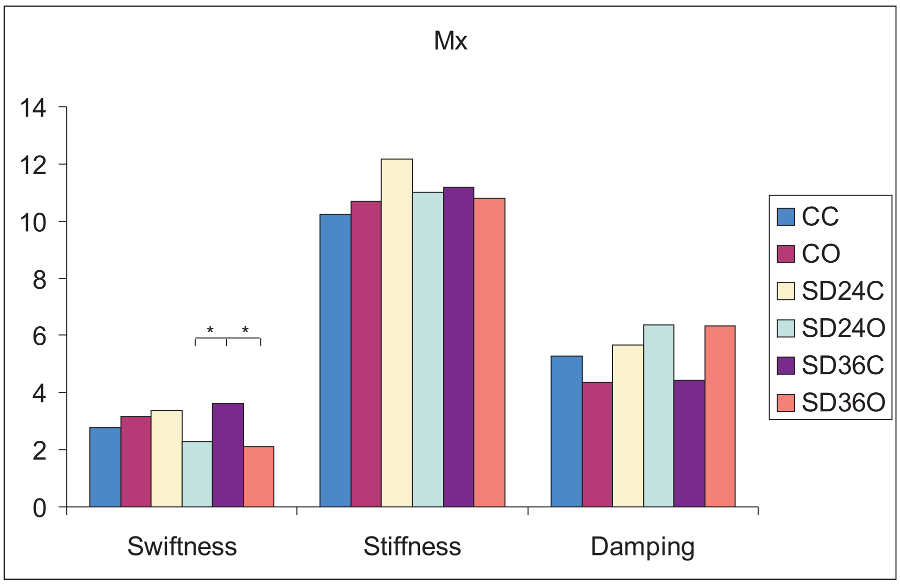


Figure 6.18: The parameters swiftness, stiffness and damping extracted from an ARMAX model with the anteroposterior torque Mx as output signal. Little evidence is found for difference between the trials. * denotes significance level with $P < 0.05$.

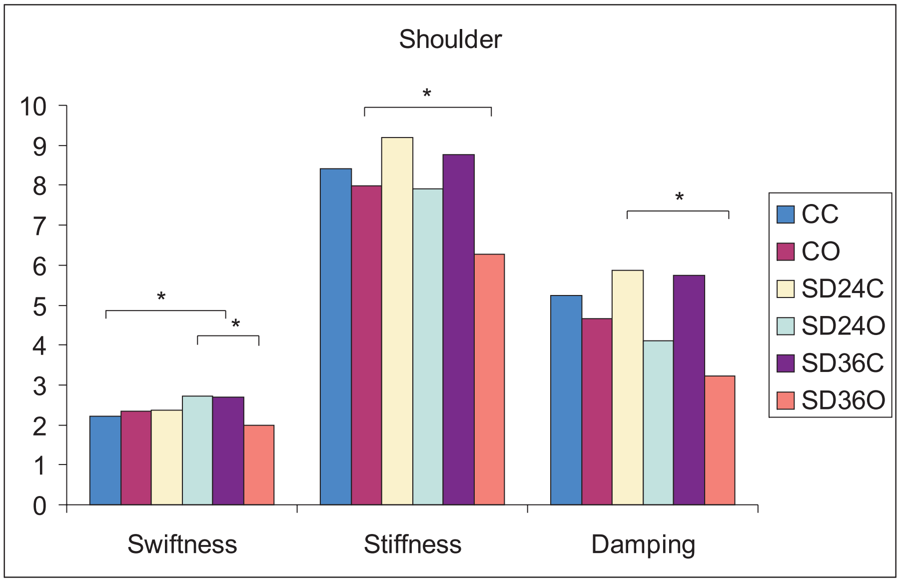


Figure 6.19: The parameters swiftness, stiffness and damping extracted from an ARMAX model with the shoulder position as output signal. Only a few trials have statistical evidence for differences in the parameters. Again, * denotes a significance level of $P < 0.05$.

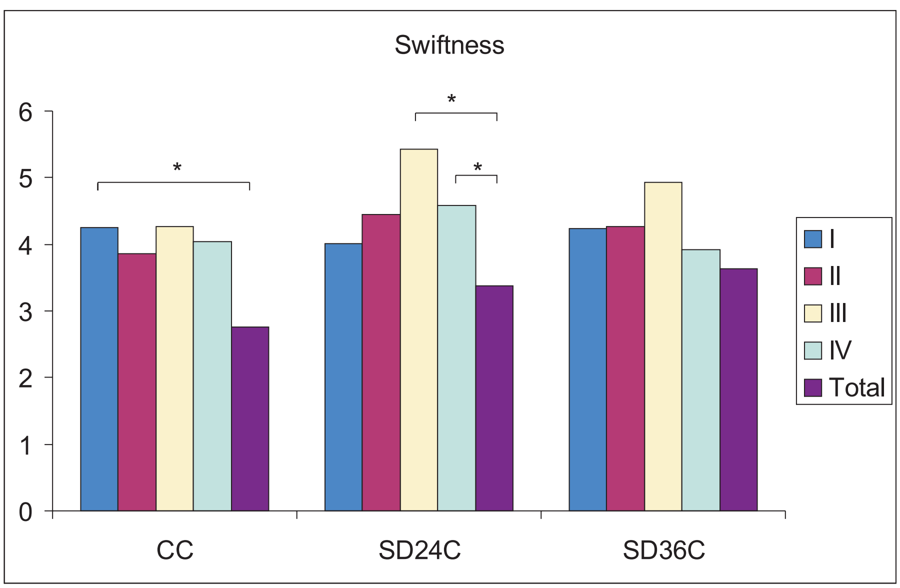


Figure 6.20: The swiftness parameter calculated from third order ARMAX model with anteroposterior torque as output signal. Only closed eyes trials are considered.

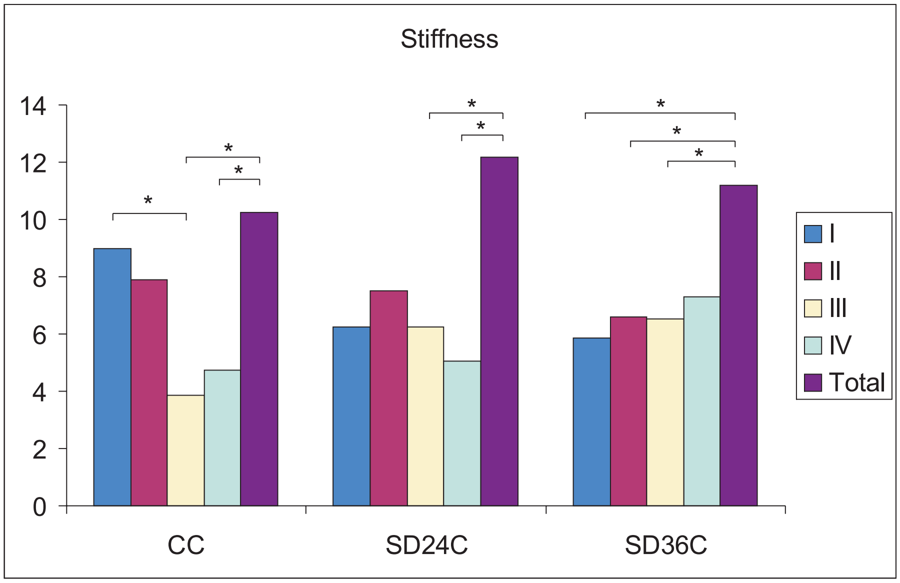


Figure 6.21: The stiffness parameter calculated from third order ARMAX model with anteroposterior torque as output signal. Only closed eyes trials are considered.

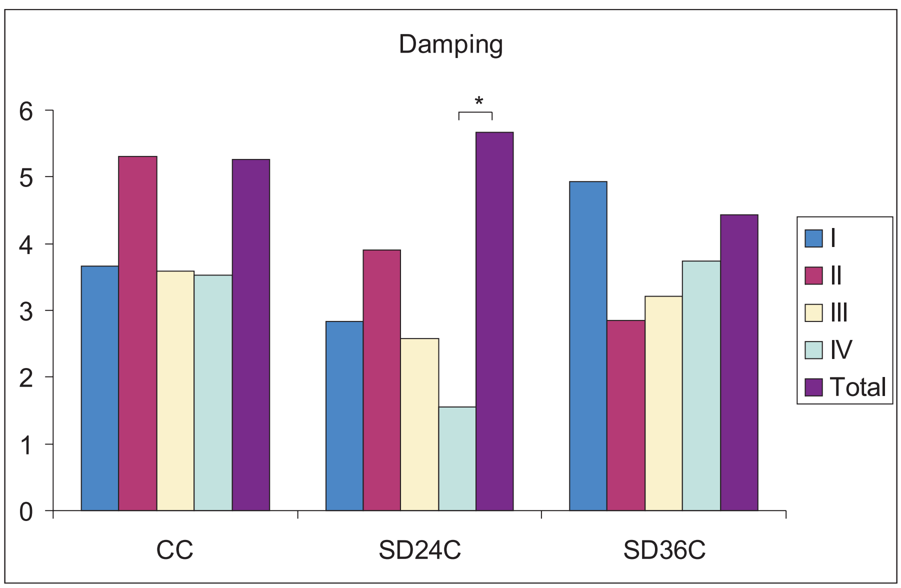


Figure 6.22: The damping parameter calculated from third order ARMAX model with anteroposterior torque as output signal. Only closed eyes trials are considered.

Chapter 7

Discussion

In this thesis, properties of postural control and adaptation of 17 subjects has been investigated after sleep deprivation up to 36 hours. Both the assessment of mathematical models describing input-output relations and statistical analysis of variance have been applied to the recorded data.

7.1 Variance and correlation

The characterization of recorded body sway using variance analysis produces significant results where it can be concluded that attention influences the human postural stability. It is suggested that anteroposterior stability decreases in closed eyes tests with 24 hours of sleep deprivation. Also, the lateral stability is affected by sleep deprivation of 24 hours, both in trials with open and closed eyes. The Zebris recordings of position data support these findings from the force platform data. The variance analysis of position data show a worsened stability in trials with 24 hours of sleep deprivation, but only in the lateral direction. Hence, the conclusion that 24 hours of sleep deprivation decreases postural stability more than 36 hour of sleep deprivation can be drawn.

Strongest evidence for decreased postural stability was found in periods after the first stimulus period, i.e. period II-IV. Previous research have shown that a task is more sensitive to sleepiness if it is long, monotonous and demand continuous attention [2]. Consequently, it may be the case that the increased level of attention at the onset of vibratory stimulation during period I maintains the postural activity. With time, the postural task to preserve balance under the perturbations caused by the vibratory stimulation become more monotonous and some attention is lost. This reasoning support the findings of worsened stability in periods II-IV.

The attention may play another important role in this study of postural control and sleep deprivation. The results from torque data and position data showed that reduced stability were more often recorded in the lateral direction than in the anteroposterior direction. One hypothesis involving attention, is that the subject uses high attention to respond to the anteroposterior disturbances caused by the vibratory stimulus of the calf muscles. The lateral perturbations are not as strong as in the anteroposterior direction, and hence, little attention is given to lateral balance control.

The interesting fact that the 36 hour tests and calculated significance values indicate a 'recovery' where the body sway again decreases. Intuitively, the body sway is expected to increase with the level of sleep deprivation resulting in larger sway after 36 hours. The explanation of this finding can be habituation the posturographic test procedure itself. Every subject perform the 36 hour sleep deprivation trial after they have performed the 24 hour trial. Thus, they have greater experience and ability of how to respond to the balance perturbations and consequently, the recorded body sway is reduced in the 36 hour trial.

The correlation analysis shows that the immediate response to the stimulation onset was that a more rigid body posture with higher correlation between the movement of different body sections. This is also in the line of the work of Fransson (2005) [6].

7.2 System identification

The subspace identification using N4SID and MOESP algorithms have proven to be very effective methods to model the body sway and torques exerted by the feet with the stimulus signal as input. Since the data is non-stationary, the main approach have been to segment the data series in order to estimate sub models on shorter time intervals. The comparison and the use of different validation and simulation strategies suggest that the characteristics of data have changed during time in almost every posturographic trial. When dividing the stimulus period into four periods and estimate models on each sub intervals, higher VAF values were obtained compared to the model estimated on the complete data set. This indicates time dependent properties, for example a change of system dynamics during time. However, it was difficult to find analysis methods to characterize the time dependence of the system dynamics.

One idea was to apply balanced model reduction to the higher order anteroposterior models for stimulus periods I-IV and the total stimulus period. Reducing the model order to three, it is possible to consider the postural control problem from a PID perspective. The postural control is in this case performed via the ankle torque by the use of proportional, integrative and derivative control actions. The advantage of this approach is that the three parameters of motion, swiftness, stiffness and damping easily can be extracted from the model. These parameters characterize the human dynamics and have been used to assess changes in the modelled system. However, the statistical analysis of calculated parameter values show little evidence for parameter changes. Only in a few cases, statistical evidence is found for the hypothesis that the subject may change control strategy due to increased difficulties to maintain balance as an effect of being sleep deprived. Consequently, it is suggested that the subjects have a rigid body posture in all trials which resembles a single-linked inverted pendulum. This conclusion does not violate the findings from the variance analysis, where increased body sway was found in 24 hour sleep deprivation trials with closed eyes. An increase of body sway may appear although the system dynamics is described by a rigid body structure.

The lateral models were found to be less reliable with bad simulation properties. Probably, this is a consequence of the poor coherence between stimulus and the lateral movement. The lateral data do not have such a direct response as the anteroposterior movement, which can be concluded by inspection of recorded data and comparing with the stimulus sequence. There are two possible conclusions that can be drawn from the lateral modeling. The first hypothesis, is that the lateral data is afflicted with larger time varying characteristics than in the anteroposterior data. This implies, that a model estimated on one of periods I-IV, performs very poorly when the simulated output is compared with data from another period. Furthermore, the fact that models estimated on the complete lateral data series had bad performance and negative VAF values or close to zero support this conclusion of larger time varying characteristics in lateral data.

The other possibility is that the lateral movement simply cannot be described with a model with the stimulus sequence only as input. Hence, one cannot reject the risk of making conclusions on incorrect basis. Therefore, the conclusions concerning the lateral modeling should be taken with some caution.

If any conclusions should be drawn from the system identification analysis, it would be that sleep deprivation does not degrade postural control and the ability to use adaptation pattern to handle the induced balance perturbations. At least, this can not be supported by the analysis methods used in this study. According to the findings of Schlesinger *et al.*, sleep deprivation only effect postural control in cases where higher levels of sensory integration or motor coordination is required [19]. The balance task in this study is fairly simple with the most difficult test condition during closed eyes trials where the removal of visual input implies that the postural control action have to be based entirely on somatosensory and vestibular inputs. However, on healthy subjects the integration by the central nervous system is highly effective and despite the reduced mental alertness due to sleep deprivation, balance can be maintained without problems when perturbations to the proprioceptive sensory system is introduced.

7.3 Difficulties

In every experimental study there are many sources of error that will affect the outcome. First, we have the experimental methodology involving the subjects. Even though hard efforts were made to establish equal test conditions, there is always a risk of results influenced by difference in test conditions. One major problem in a study like this, is the habituation to the test method itself and especially to the posturography test. Worse postural stability is highly possible when a person performs a posturography for the first time. Since the 'first time posturography' either is control trial or 24 hours of sleep deprivation, these effects will not have influence on the 36 hour sleep deprivation test. Unfortunately, it is almost impossible to eliminate this phenomenon in the results.

In cases of large amounts of spontaneous movements in the normal balance control used to maintain upright stance, it is difficult to apply system identification methods. Subjects with this property has poor coherence between stimulus and response in terms of movements. Methods of data filtering can be used to reduce the amounts of spontaneous sway. However, one should be careful when applying filter techniques due to the difficulty to determine and characterize which part of the response data that can be derived from the stimulus and the part that originates from spontaneous motion.

The extraction of the dynamical parameters swiftness, stiffness and damping from MIMO-models resulted in strange parameter values. Consequently, the ARMAX approach was used instead. However, it is desirable to be able to perform the same parameter extraction from the higher order MIMO-models and compare the results with the ARMAX method.

The quantitative results and the results from the system identification methodology showed some discrepancy. The most interesting results, the deteriorated adaptation after sleep deprivation, is assessed using quatitative variance analysis. Naturally, it is of great interest to support these evidence using system identification methods. This matter needs further investigation.

Chapter 8

Conclusions

From this study of postural control and the effects of sleep deprivation, the following conclusions can be drawn.

- Analysis of variance suggest a weaker postural stability after 24 hours of sleep deprivation, especially in lateral direction.
- Analysis of variance suggest a deterioration of adaptation, both after 24 hours and 36 hours of sleep deprivation. Apparently, sleep deprivation impairs postural adaptation.
- Correlation of body segments indicate adaptive behavior in movement patterns. However, no distinct differences due to sleep deprivation can be concluded.
- Subspace based identification can successfully be used for the identification of MIMO models between stimulus perturbation and movement response of the subject.
- Evaluation of subspace based modeling using VAF values give evidence for time dependant system characteristics which can be connected to adaptation.
- System identification methodology used in this thesis does not give any clear support of deterioration of postural stability due to sleep deprivation, nor can this hypothesis be excluded.

Bibliography

[1] Murray L. Barr and John A. Kiernan. *The human nervous system - An anatomical Viewpoint*. J.B. Lippincott Company, fifth edition, 1988.

[2] L. De Gennaro et al. Visual search performance across 40 h of continuous wakefulness: Measures of speed and accuracy and relation with oculomotor performance. *Physiol Behav*, 74(1-2):p. 197–204, 2001.

[3] B. De Moor, P. Van Overschee, and W. Favoreel. Numerical algorithms for subspace state-space system identication - an overview, 1999.

[4] J. Dichgans and H.-C. Diener. The contribution of vestibulo-spinal mechanisms to the maintenance of human upright posture. *Acta Otolaryngol (Stockh)*, 107:338–345, 1989. Pärm 2 K.

[5] H. C. Diener, J. Dichgans, B. Guschlbauer, and M. Bacher. Role of visual and static vestibular influences on dynamic posture control. *Hum neurobiol*, 5:105–113, 1986.

[6] P. A. Fransson. *Analysis of Adaptation in Human Postural Control*. Phd thesis, Department of Otorhinolaryngology, Lund University, 2005.

[7] GM. Goodwin, DI. McCloskey, and PB. Matthews. Proprioceptive illusion induced by muscle vibration-contribution by muscle spindles to perception. *Science*, 175:1382–1384, 1972.

[8] A. Hafström, P. A. Fransson, M. Karlberg, T. Ledin, and M. Magnusson. Visual influence on postural control, with and without visual motion feedback. *Acta Otolaryngol (Stockh)*, 122(4):392–397, 2002.

[9] F.B. Horak and L.M. Nashner. Central programming of postural movements: Adaptation to altered support-surface configurations. *J Neurophysiology*, 55:1369–81, 1986.

[10] R. Johansson. *System Modeling and Identification*. Prentice Hall, Englewood Cliffs, N.J. USA., 2004.

[11] R. Johansson and M. Magnusson. Human postural dynamics. *CRC Crit Rev Biomed Eng*, 18(6):413–437, 1991.

[12] R. Johansson, M. Magnusson, and M. Åkesson. Identification of human postural dynamics. *IEEE Trans Biomed Eng*, 35(10):858–869, 1988.

[13] RS. Johansson and ÅB. Vallbo. Spatial properties of the population of mechanoreceptive units in the glabrous skin of the human hand. *Brain Res*, 184(2):353–366, 1980.

[14] E.A. Keshner, J.H.J. Allum, and C.R. Pfaltz. Postural coactivation and adaptation in sway stabilizing responses of normals and patients with bilateral vestibular deficits. *Exp Brain Res*, 69:77–92, 1987.

[15] M. Kleiber, GA. Horstmann, and V. Dietz. Body sway stabilization in human posture. *Acta Otolaryngol (Stockh)*, 110:168–174, 1990.

[16] M. Magnusson, R. Johansson, and J. Wiklund. Galvanically induced body sway in the anterior-posterior plane. *Acta Otolaryngol.*, 110(1-2):11–17, 1990.

[17] L. Nashner and G. McCollum. The organization of human postural movements: A formal basis and experimental synthesis. *Behav and Brain Sci*, 8:135–172, 1985.

[18] L.M. Nashner. Adapting reflexes controlling the human posture. *Exp Brain Res*, 26:59–72, 1976.

[19] A. Schlesinger, MS. Redfern, RE. Dahl, and JR. Jennings. Postural control, attention and sleep deprivation. *Neuroreport*, 9(1):49–52, 1998.

[20] R. Shi and J. F. MacGregor. A framework for subspace identification methods. *Proceedings of the American Control Conferance*, pages 3678–3683, 2001.

[21] Karl J. Åström and Björn Wittenmark. *Computer Controlled Systems - Theory and Design.* Prentice Hall, third edition, 1997.

[22] S. Uimonen, K. Laitakari, R. Bloigu, and M. Sorri. The repeatability of posturographic measurements and the effects of sleep deprivation. *J Vest Res*, 4(1):29–36, 1994.

[23] JP. Vedel and JP. Roll. Response to pressure and vibration of slowly adapting cutaneous mechanoreceptors in the human foot. *Neurosci Lett*, 34(3):289–294, 1982.

[24] M. Verhaegen and P. Dewilde. Subspace model identification. part 1: The output-error state-space model identification class of algorithms. *Int. J. Control*, 56(5):1187–1210, 1992.

[25] VJ. Wilson and JG: Melville. *Mammalian vestibular physiology.* Plenium Press, New York, 1979.