Bachelor thesis

BALANCE AND STEP RESPONSES WITH AN IMPAIRED ANKLE TORQUE

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Abstract

Previous studies have reported an asymmetry in balance contribution for unilateral diseased patients, like Parkinson Disease or stroke patients. The aim of this study is to examine the underlying effects on balance and step responses when having such an impaired ankle torque. This study uses healthy subjects and mimics the impaired torque with the use of wooden blocks, one with a foot-size length and one half the foot-size length. The same subjects were tested in the normal situation for a comparison. Prior to the main experiment were Base of Support (BoS) trials where the subjects maximal feasible CoP was examined. The BoS for the small block was considerably smaller than on the foot-size block. The BoS of the foot-size block was about the same length as the without-block feet BoS. The main experiment used transient platform perturbation to disturb subjects balance. These trials showed that the CoP was confined by the BoS. A novel finding was that the CoP of the impaired ankle does not reach up to the BoS, even though the CoM exceeds it. These findings indicate that the impaired ankle does not contribute to its maximum capacity, but scales to the healthy ankle torque.

1 Introduction

1.1 Clinical relevance

Balance control is essential in erect stance and locomotion. In particular elderly suffer from impaired postural control. Fall injuries are the leading cause of injury and death among elderly. In de United states 80% of the deaths in 2008 were caused by falling [1]. The costs for these injuries are high, as the healthcare has to be financed by insurance companies. The epidemiology is under-recognized and more research has to be done on balance and locomotion.

Several balance tests were developed over the years in order to quantify balance. In these examinations the motor deficits of a patient are assessed. Common used tests are the Berg Balance Scale, Functional Reach test and the Dynamic Gait Index. The accuracy of these tests are examined for Parkinson Disease (PD) patients in [2], where the researchers concluded that collective interpretation of multiple tests valid for a diagnostician of a PD patients' fall risks.

Neurologic impairments such as Parkinson's Disease (PD) or CerebroVascular Accident (CVA) can manifest asymmetrically, i.e both legs contribute differently to task execution. The relation between weight-bearing and balance control in stroke patients is non-linear [3]. Similar results were gained with PD patients [4].

1.1.1 Cerebrovascular accident

A CVA[1] is the rapid loss of brain function due to disturbance of blood supply to the brain. The majority are caused by ischemia, blockage of blood flow leading to dysfunction of brain function in the affected area. The other category is hemorrhage, lack of blood flow leading to accumulation of blood elsewhere. A stroke can manifest silently, leavening the patient unaware of the occurred stroke. This can be lead to permanent neurological damage or dead.

[1]A disease commonly known as stroke

1.1.2 Parkinson's Disease

PD is the second most common degenerative disease of the nervous system [5]. Most patients with PD are diagnosed with idiopathic PD and only a small proportion of them can be attributed to known genetic factors. Common symptoms of PD are progressive postural instability, hypokinesia, rigidity and tremor. Movements of PD patients are impaired due to progressive loss of dopamine neurons in the substantia nigra. Defects the in motor system are not only associated with walking, but also the stability in quiet stance can be affected. The deficits in balance result in increased an fall risk, as well as a loss of movement control and sensory deficits balance [5]. Novel research shows promising results in managing symptoms by deep brain stimulation [6].

1.2 Related work

Only a few studies are addressed to balance with a paretic and non paretic ankle. G. Brus studied in his bachelor thesis [7] balance and stepping and mimicked the asymmetry with wooden blocks. This study comprised three experiments: a multisine experiment to evaluate weight-bearing, a static trail to analyse the body mass velocity as well as a step experiment. The findings of the step experiment were that the step time and reaction time did not differ significantly. However, the subjects with a wooden block showed a significant increased step length.

Van der Kooij at al. studied balance contribution of the paratic and non-paratic ankle was studied for PD patients [8] and van Asseldonk et al. for stroke patients [3]. These studies reveal that the linear relation between weight-bearing and balance contribution (existing in healthy subjects) is absent for PD and stroke patients.

1.3 Objective

The goal of this study is to gain more insight in the effects of a reduced balance due to an impaired ankle, and thereby contribute to the development of rehabilitation strategies. The working of the underlying human balancing system remains unclear, in particular the case of asymmetrical diseases. Since not much research was done on impaired stepping responses, this study mimics the asymmetrical disease and subsequently examines the balance responses compared to the normal and impaired situation. In this study a main focus will be on two questions. Firstly, the validity of the mimicked posture imbalance is investigated, whereafter the step response of such a posture imbalance and a healthy balance are compared. Subsequently the cause for, the expected, imbalance is tried to be examined. The causes impaired balance and increased stepping responses when having an impaired ankle torque are investigated.

There are several hypothesis. The first one is that stepping is used as a last method, only used when other balancing strategies are ineffective. An attenuated ankle torque will make it harder to maintain balance and will result in more stepping. The same principle may hold for higher perturb amplitudes, a positive regression is hypothesised between amplitudes and the amount of corrective steps. Forward steps were evoked by backward perturbations, and in the same manner were backward steps evoked by forward perturbations.

2 Theoretical background

A basic understanding of the theory is essential for analysing the results obtained from measurements. Also, justified predictions of the experiments are made based on the theoretical framework. In this study, the balance and balance control are examined for subjects standing in the upright posture.

2.1 Balance control

2.1.1 Closed feedback system

The human body perceives various inputs that contribute to orientation and balance. The afferent input signals have to be processed by the Central Nervous System (CNS) in order to generate a corrective torque with the limb muscles. The balance process could be expressed as a closed loop system with three basic elements control, plant and sensor [8]. The balance control model is presented in figure 2.1.

The CNS can be perturbed with both internal (T_{int}) as external perturbations (T_{ext}). This is sensed by the CNS, and this sensor signal has three input contributors:

· Visual input
· Proprioception input
· Vestibular input

. These inputs are then compared with a reference, which is the desired posture of the human body. If there is a difference between the desired and sensed posture, a proportional correction is carried out to get back to the desired position. To be able to study single leg contributions for this correction, a division between left and right leg is made in the model. The torque that corrects the current posture into the desired posture is subdivided in an active and a passive torque. The passive torque (as well as the internal disturbance) can not be measured individually. But, the active torque is stimulated by the CNS and therefore has a lumped neural delay. This neural delay can be measured with Electromyography (EMG).

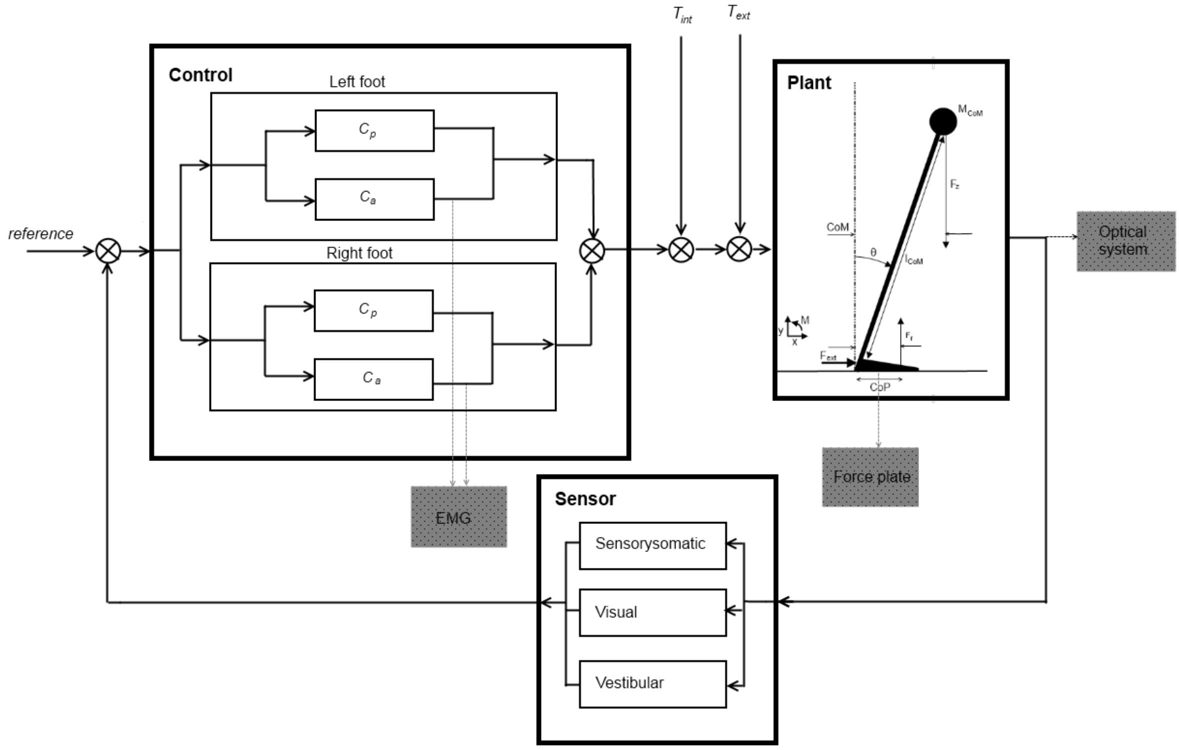

Figure 2.1: Balance control system operates like a closed loop system, with a *control*, *plant* and *sensor* block. The *control* represents the CNS, which evaluates the current stance with the reference. The *plant* the body dynamics and the *sensor* as the sensor input

2.1.2 Balance strategies

In two third of the human body mass is located in the upper body, making the balance control a difficult task. Small perturbations can be corrected by creating a corrective torque around joints, usually accompanied with a rotation of joints. Large perturbations can be corrected with one as well as grasping a table object in the environment. The balance system has a preference for the ankle strategy in order to maintain the upright stance. Thus, minor perturbations can be corrected without lifting the feet or help of the surrounding. In theory also the neck can create a corrective torque, but due to a matter of course this is very inconvenient. Fortunately, the balance system does not resort to this strategy and tries to keep the head as stable as possible, reducing head acceleration and maintaining the upright posture [9].

The hip strategy is a highly effective balancing strategy. This is due to the strong upper leg and trunk muscles which are capable of producing a high moment around the hip. Moreover the Head, Arms and Trunk (HAT) second moment of inertia is relatively small. However, it is important to note that the ankle and hip strategy are not mutually exclusive [10],[11]. Most perturbations are corrected with a combination of the ankle and hip strategy. The hip strategy can be studied separately by a subject standing on a beam,

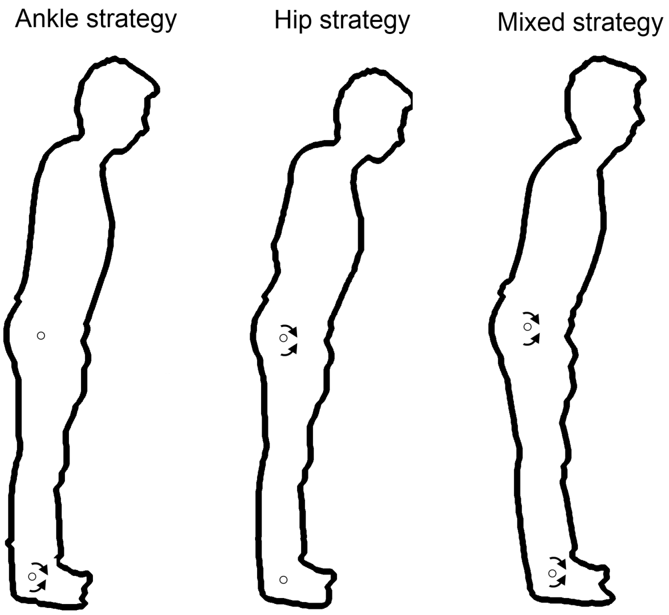

Figure 2.2: The three major used strategies to maintain balance in the saggital plane.

excluding the ankle strategy. The ankle strategy could be isolated with an orthosis, which constrains hip movement. The three most used balance strategies in the sagittal plane are presented in figure 2.2.

Furthermore, other joints contribute in maintaining the upright posture. In particular the knee and the shoulder joints could generate a corrective torque whenever a high perturbation is imposed.

2.1.3 Muscles involved

Maintaining the upright stance is impossible without the contribution of the muscle-skeleton. Posture is mainly an active process, in particular controlled by slow twitch muscle fibres. The contractile force is a function of both length and velocity.

A large portion of corrective torque by small deviation in the upright stance is corrected by passive ankle torques. Passive ankle torque is generated with the intrinsic muscle, tendon surrounding ligaments properties. The muscles fibres and tendons have elastic and damping features.

The Anterior/Posterior (A/P) [1] sway is controlled by dorsiflexors/plantarflexors using the tibilia anterior and medial gastrocnemius respectively.

Perturbations in Medial/Lateral (M/L) [2] direction are not only corrected by the ankle joints, the hip contributes too. The hip abductors and adductors load and unload the two limbs. The load/unload mechanism is marked by out of phase vertical reaction forces.

[1] Forward and backward sway; in the saggital plane

[2] sideways sway; in the frontal plane

Hip movements are necessary because the ankle, with the small foot width, is incapable to generate enough torque in M/L direction. The two mechanisms work independent of each other. In M/L direction the corrective torque is controlled by the ankle invertors/evertors. The invertors/evertors are also dorsiflexors/plantarflexers, however not used for corrective A/P torques [12].

Table 2.1: Muscles involved in generating the corrective ankle and hip torque in the saggital plane.

Joint	**Movement**	**Muscle**
Ankle	Plantarflexion	Gastrocnemius
		Soleus
	Dorsiflexion	Tibialis anterior muscle
		Extensor hallucis longus muscle
		Extensor digitorum longus muscle
		Peroneus tertius
Hip	Flexion	psoas
		iliacus
		Rectus femoris
		Sartorius
	Extension	Gluteus maximus muscle
		Semimembranosus muscle
		Semitendinosus muscle

2.1.4 Perturbation methods

Voluntary movements lead to internal disturbances. A voluntary movement can be evoked by a sensory conflict. Sensory information, described in section 2.1.2, can be disturbed in various ways. The visual input can be tricked by rotating the visual surrounding, i.e creating a 'villa volta'. Sensorysomatic input can be tricked by vibrating the ankle at high frequencies. In this way, the muscle spindles will react as if they are stretched. Another way to fool the cognitive input is to distribute the vestibular input. This could be achieved by poring warm liquid in the patients ear which disturbs the galvanic vestibular system [13].

Transient external disturbances can be imposed to measure the passive corrective properties of the muscles. Most common external perturbations are surface translations and surface tilts. BRON

2.1.5 Stepping

The human body can correct A/P perturbations with a set of strategies, described in section 2.1.2, corrected with a torque generated at the ankle, hip or both. In addition, a step can be made. Stepping is used as the last method, when other balancing strategies are ineffective.

When analysing the force sensor data, one can distinguish three different phases: [14]

· Symmetric feet in place responses
· Asymmetric feet in place responses
· Step responses

. Early automatic neural responses are present after the onset of a perturbation. Small perturbation can be corrected without a step and typical symmetric feet ground reaction force are measured. With high perturbations a corrective step has to be made and stepping responses are measured. An intermediated response can take place when the perturbation disturbs balance but no step has to be made. This intermediate response is characterised by a lateral weight shift, which could be explained as the preparation of making a step.

2.2 Kinematics

Kinematics are classical mechanics, that describe the relationship between motion, mass, forces and torques. There are several ways of measuring human movement with the use of kinematics [15]. The easiest way is with a goniometer, which measures a joint's angle. Another feasible way to measure kinematics is by use of an accelerometer. However, for a more detailed analysis with an optical system is required. A regular film camera provides data for quantitative analysis.

2.2.1 Optical imaging

A common method in analysing human kinematics is recording markers with an infra-red sensitive camera. An infra-red camera has several advantages. Firstly, when using passive markers, no wires are needed. Secondly, a high number of sensors, reflective markers, can be used. The main disadvantage of infra-red optical systems are the high costs and the risk of marker occlusion.

2.2.2 Force plates

Ground reaction forces can be measured with a rectangular force plate. Force platforms are used in studies to quantify balance, gait or other biomedical parameters. There are several types of force sensor transducers. The simplest force plates contain a single plate with only vertical force sensors. For a weight-bearing assessment two load cells are required. More advanced models can measure shear forces and thus measure in 3 degrees of freedom (DOF), or could in addition measure the torque around 3 axes, which results in 6 DOF. Force plates are essential when evaluating inverse dynamics, see section 2.3.

2.3 Kinetics

Kinetics is the study that relates motion and forces. One method of evaluating kinematics is equating Newton's laws[3]. The basic tools are Newton's second law of motion and the thereof derived torque law:

$$\begin{aligned} \sum F &= m \cdot a \\ \sum \tau &= I \cdot \ddot{\alpha} \end{aligned} \tag{2.1}$$

2.3.1 Definitions

Center of Mass

The Center of Mass (CoM) represents the location of the mean body mass (located around the waist), the weighted average of each body segment mass. . The CoM position is a passive variable expressing in combination with the CoM length the body sway. The common way the derive the CoM is through optical imaging [15].

Center of Pressure

The Center of Pressure (CoP) is the location where the average pressure is applied on the ground. The CoP is acquired with one force plate under both feet, a netto CoP, or with the use of two force plate, then representing a separate CoP for both feet individually. The CoP is an active variable that controls the position of the CoM.

Base of Support

The CoP is not a fixed location, it moves under the foot area, thereby physical limited by the foot length. The Base of Support (BoS) represents the boundary CoP location, i.e the maximal and minimal CoP that could be created by the subject.

2.3.2 Symmetric sagittal model

This study is restricted to forward and backward perturbations and therefore the theoretical model is made in the sagittal plane.

The body mass is simplified as a point mass, indicated as M. The angle between the CoM and the vertical axis is denoted as θ. A gravitation force $M \cdot g$ acts on M , denoted as $F_{\mathtt{y}}$, which has to be counteracted by a ground reaction force, $R_{\mathtt{y}}$. The CoP moves under the foot area and is defined to be zero at ankles. The inverted pendulum model is depicted in figure 2.3. The CoP has a `x` and a `y` component. As mentioned before, this study was restricted to the sagittal plane, thus the CoP and CoM discussed in the study is the CoP and CoM in `x` direction. The external disturbance is denoted by F_{ext},

[3]Other methods are elaborated in appendix A.

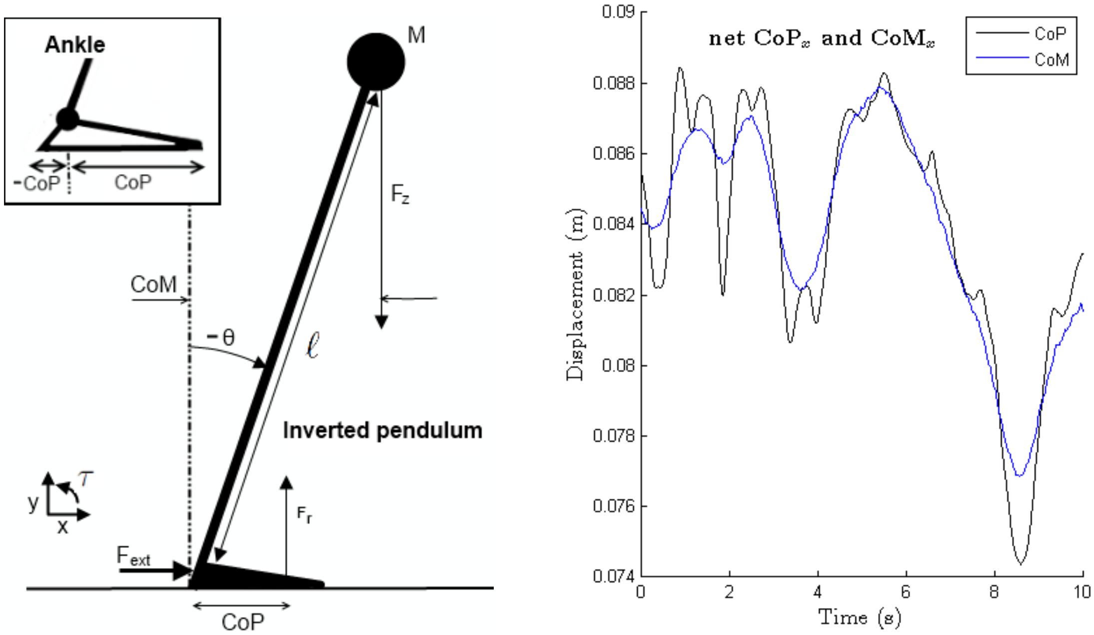

Figure 2.3: Left: the human body simplified as an inverted pendulum. The ankle is magnified to indicate the CoP sign definition. Right: *CoP* and *CoM* for subject 1 measured at erect stance. Note that the *CoP* excursions are larger than the *CoM* excursions in order to correct the *CoM* deviations.

a horizontal perturbation force generated by platform accelerations, ($\ddot{x}_{sb}$). The CoP is expressed as:

$$CoP = \frac{\tau + F_x \cdot y_0}{R_{\mathrm{y}}} \tag{2.2}$$

Where τ is the torque and R_{y} the vertical reaction force, measured by the sensors in the moveable platform. F_x is the horizontal force on ankles and y_0 is the ankle height.

Erect stance is modelled with rigid body dynamics and an inverted pendulum. While using the inverse pendulum model several assumptions are made.

- Movements are restricted to the sagittal plane
- There is no movement in the hip, or other joints than the ankle. Consequently, the length of the pendulum is constant.
- Ankle height and ankle mass are neglected
- There is no foot movement, which means that there is no stepping or toe and heel lifting and no horizontal movement (e.g caused by slipping).
- The feet are kept straight to the medial line, side by side.

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Newton laws, equations 2.1, state that the summed torques must equal the angular velocity times the moment of inertia of the body. As shown by figure 2.3, the clockwise torque is produced by gravitation and the angular acceleration. The counter-clockwise torque is produced by the muscles around the ankle, creating a reaction force. In equation form:

$$\tau_+ = CoP \cdot R_\mathrm{y} \tag{2.3}$$
$$\tau_- = CoM \cdot F_\mathrm{y} + I \cdot \ddot{\alpha} \tag{2.4}$$

Where τ_+ is the counter-clockwise torque and τ_- is the clockwise torque. Using the small angle approximation, vertical accelerations are neglected. Hence, the angular acceleration is estimated with the horizontal acceleration of the CoM and reaction force becomes equal to the gravitation force.

In erect stance, the case of $CoP > BoS$ will result a in step or a fall. A CoP larger than the foot length can only be generated by the ankle when the foot is fixated to a larger object (for instance the floor). In practice, the maximal CoP is a few centimetres before the toes.

If $CoM \cdot F_\mathrm{y} > CoP \cdot R_\mathrm{y}$, the body will experience a negative angular acceleration, resulting in a forward sway. To compensate for this, the body will increase the CoP so that $CoM \cdot F_\mathrm{y} < CoP \cdot R_\mathrm{y}$, results in backwards acceleration. In this case a backward sway will follow.

The relation between CoP and CoM are derived with:

$$CoP \cdot F_\mathrm{y} - CoM \cdot R_\mathrm{y} = I\ddot{\theta} \tag{2.5}$$
$$I\frac{\ddot{\theta}}{\ell} \approx I\frac{\ddot{x}}{\ell} \tag{2.6}$$
$$R_\mathrm{y} \approx F_y$$
$$(CoP - CoM)F_\mathrm{y} = I\frac{\ddot{x}}{\ell} \tag{2.7}$$
$$CoP - CoM = \frac{I}{F_\mathrm{y}\ell}\ddot{x} \tag{2.8}$$
$$CoP - CoM = C \cdot \ddot{x} \tag{2.9}$$

Where g is earth's gravity, I the moment of inertia of the total body and $\ddot{x}$ the horizontal linear acceleration of the CoM. The constant C combines the parameters that are invariable in time. The latter equation describes a linear relation between the left and right equations, thus predicting a strong correlation between $CoP - CoM$ and $\ddot{\theta}$, as shown in figure 2.3.

2.3.3 Asymmetric saggital model

With unilateral pathology [4] the CoP and vertical reaction force in both ankles are asymmetrical. [8],[3].

The netto CoP during double limp support can be expressed as[9]:

$$CoP_{net} = CoP_l \frac{R_{\mathrm{y},l}}{R_{\mathrm{y},l} + R_{\mathrm{y},r}} + CoP_r \frac{R_{\mathrm{y},r}}{R_{\mathrm{y},l} + R_{\mathrm{y},r}} \tag{2.10}$$

Where CoP_l, CoP_r and CoP_{net} are the CoP of the left, right foot, and netto respectively. $R_{\mathrm{y},l}$ and $R_{\mathrm{y},r}$ represent the vertical reaction force under the left and right foot.

2.3.4 Platform perturbations

The platform accelerations perturb human balance with an external torque. The torque imposed by the platform on the body is expressed as [3]:

$$\tau_{\mathrm{ext}} = -M \cdot \ell \cdot \ddot{x}_{pl} \tag{2.11}$$

Where M is the total body mass, ℓ the CoM length and $\ddot{x}_{pl}$ the support base perturbation imposed by the platform.

[4] e.g patients who suffer from stroke, PD, or underwent an amputation.

3 Materials and methods

The purpose of this study is to examine the effects of an impaired ankle torque. This study uses healthy subjects in an attempt to mimic the unilateral diseases. The experiments were conducted in a controlled environments using transient perturbation to disturb the subjects balance.

3.1 Protocol

3.1.1 Mimicking unilateral disease

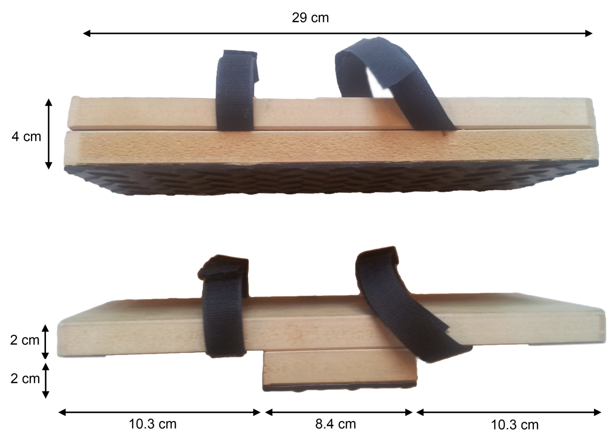

Figure 3.1: Above: the wooden block worn by the subjects under the non-preferred foot. Below: the wooden block worn under the preferred foot.

To mimic unilateral diseases, an approach that uses a wooden block was devised. The wooden blocks had to be bound under the subjects feet. The asymmetry was introduced with a difference in surface length. One block had an average foot size length (29 cm) while the other block length was roughly half of the average foot size. Therefore, one ankle had a smaller BoS, confining the maximal torque. The small block mimics the unilateral paretic leg. The blocks used in this study are similar to the blocks used in the study conducted by [7]. However, the block used in present study had larger surface contact length of 8.4 cm instead of 5 cm.

It was predicted that the subject has a preferred leg, which dominates over the non-preferred leg, i.e the preferred leg is used more frequently to step out. The small wooden block was mounted under the preferred leg. Designating the non-preferred leg as the impaired leg can influence the stepping behaviour. Therefore, the preferred leg was assessed prior to the experiment conducting a facile experiment. The subject stood in the upright position and was informed that his stepping behaviour needed to be evaluated. However, the exact intention of the experiment was not noticed, since this may increase anticipation which can influence the results. The observant pushed the subject in the back and wrote down which leg was used to step out. This experiment was executed in threefold to reduce causality.

3.1.2 Perturbation signal

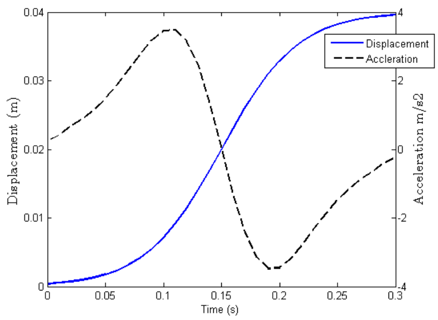

Figure 3.2: The sigmoid function used as to describe the platform during a perturbation.

The subjects balance was disturbed by transient platform translations. The platform was controlled with Simulink, which imposed an translation at a given amplitude. The platform does not behave linear with an acceleration above 8 m/s^2. As a result, the platform was incapable in producing transient changes in accelerations. For that reason a sigmoid function was used for the transient platform translation:

$$f(t) = \frac{A}{1 + \exp(-9.19(t - 0.05))} \tag{3.1}$$

With A as the translation amplitude. The input signal is shown in figure 3.2. The average perturbation signal magnitude is determined after the 'pilot trials'. The perturbation signal consists of an equal number anterior as posterior translations. One trial had 20 perturbations with five different amplitudes, every amplitude exerted twice forward and twice backward. The amplitude sequence is randomised. The duration of one perturbation was 300 ms. Subsequently, the platform remained in diverted position for five seconds in order to collect the data. The platform moved back slowly using a sinus function for the position. One trail had a duration of 168 seconds.

Table 3.1: Set amplitudes for the experiment. Second and third row are the corresponding peak velocities and accelerations.

Amplitudes (m)	-0.08	-0.07	-0.06	-0.05	-0.04	0.04	0.05	0.06	0.07	0.08
Velocities (m/s)	-0.61	-0.53	-0.46	-0.38	-0.30	0.30	0.38	0.46	0.53	0.61
Accelerations (m/s^2)	-6.95	-6.08	-5.21	-4.34	-3.47	3.47	4.34	5.21	6.08	6.95

3.1.3 Weight-bearing

Patients with a paretic ankle have a tendency to lean on their healthy ankle [3]. This bias stepping, therefore an equal weight-bearing during the trials is preferred. The weight-bearing was computed with the fraction between the two vertical force sensors. Providing the subject with a real time view of the weight-bearing adds a cognitive process and could influence the balance control [16]. Therefore, it was decided to monitor the weight-bearing by the observant. If necessary, the observant requested the subject to adjust his weight-bearing.

3.1.4 Marker placements

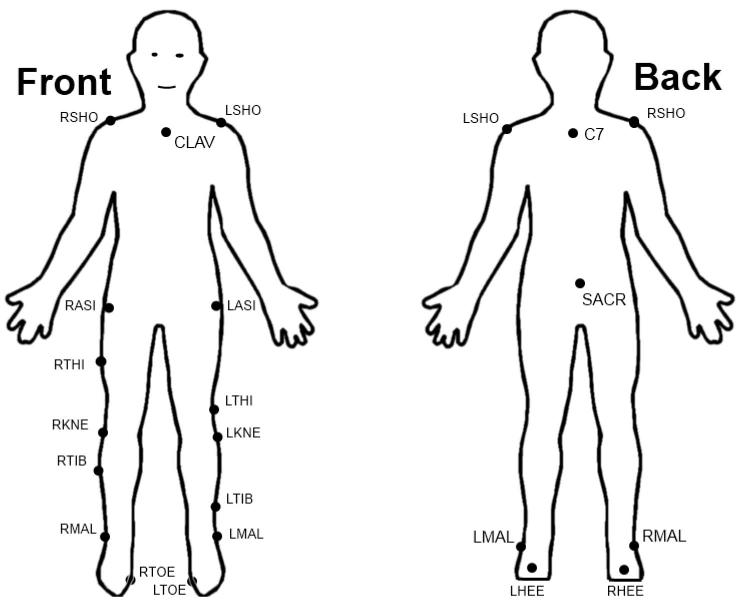

Figure 3.3: Left: anterior view of the markers. Right: posterior view of the markers.

Markers were placed at anatomic landmarks. The subject segment positions are required to derive the CoM position. The segments positions are obtained from the optical system (section 3.2.3). An overview of the marker placements is shown in figure 3.3. To anchor the markers firmly, subjects wore shorts and a sleeveless shirt and were barefoot. Three markers were attached on the feet: one on the big toe, one on the heel and on one the malleosus. The marker on the malleolus was considered as the ankle joint. The lower leg was defined by the malleolus marker, the tibia marker and a marker at the lateral epicondyle. The upper leg consisted of the knee markers, the thigh marker and the left anterior superior iliac spine (ASIS) marker. The marker on the ASIS was considered as the hip joint and the marker on the knee as the knee joint. The HAT (Head Arms and Trunk) was mapped by the ASIS markers, a sacrum marker, a marker at cervical vertebrae 7 (c7), a marker between the clavicles and a marker on both shoulders. Three additional markers were attached to the platform which served as a reference.

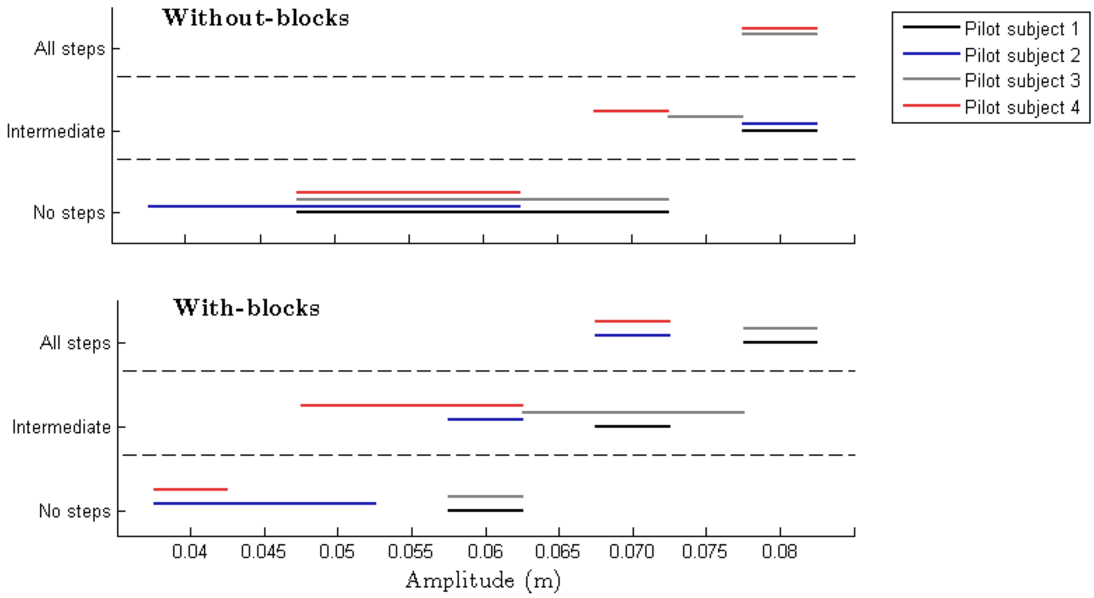

Figure 3.4: Platform amplitudes executed on the pilot subjects, together with the relative response frequency. In one pilot trail consisted of 12 platform translations. No steps refer to less than three steps, all steps refer to more than 9 steps and intermediate to four - eight steps.

3.1.5 Pilot trials

Six pilot trails were conducted prior to the main experiment. The main reason for doing pilot trials was to ascertain suitable perturb amplitudes. The minimal amplitude was defined as the minimal amplitude that was required to evoke one step in a trial containing 12 transient translations. The amplitudes were raised incrementally until the subject stepped for all perturbations. The experiment was repeated with the wooden blocks, shown in figure 3.1

Prior to the first pilot experiment, the minimum amplitudes were predicted in four different cases, based on a study by McIlroy and Maki [14]. Here the step responses were examined with the following platform velocities.

Forward	without-block	$\sim$ 0.06 m
Backward	without-block	$\sim$ 0.07 m
Forward	with-block	$\sim$ 0.04 m
Backward	with-block	$\sim$ 0.05 m

The minimum and maximum amplitudes were determined for these amplitudes. The full range of amplitudes imposed on the pilot subjects and their responses are represented in figure 5.1.

Based on these results (keeping in mind the maximum platform acceleration) the chosen perturb amplitudes for the experimental trials are `[0.04 0.05 0.06 0.07 0.08]`.

Corrections were made to the protocol as a result of the pilot sessions. A manual was written in addition to the protocol, mainly to simplify and shorten the protocol. The protocol and manual are attached in appendix I, and appendix H.

3.1.6 Experiment trials

There were eight experimental sessions, each with a different subject. The same subjects were used for the with-block and without-block experiment.

The protocol started with anthropological measurements. The subject's total height and leg length, as well as the distance between the greater trochanter and the floor were measured with a tape line. This was done while the subject was standing in the anatomical position.

Six habituation trails were preceded to the record trials. This way the subject got the possibility to get used to the perturbations, which reduced learning/carry over effects. First, the subject was exposed low perturbations, by means of translating the platform with an amplitude of 0.04m. The perturbations were raised with 0.02m in two subsequent steps. The same habitation procedure was executed while the subject was wearing the wooden blocks.

Subsequently, a set of six BoS trails were conducted. In these trials the subjects bended slowly forward and backward, until a step had to be made. The BoS trial was performed three times constraining the subject to the three different balance techniques, ankle, hip and mixed. The three strategies were executed for the with-block and without-block condition. This experimental BoS gives insights in the maximum CoP, related to the torque generated by the ankle. Theoretically, no CoP should be measured by the force plate with the hip strategy.

The experimental trials contained 12 trails. These were divided into four cycles, each containing three trials. Each trial consisted of 20 perturbations, randomised in order and direction: forward and backward. Between the perturbations was a random ample time in which reduced anticipation, four or twelve seconds. The trials in the perturbations experiment are given by:

2x3	Without-blocks	±[0.04 0.05 0.06 0.07 0.08]
2x3	With-blocks	±[0.04 0.05 0.06 0.07 0.08]

Trial sequence was determined prior to the experiment. The trail block sequence was randomised in order for all subjects.

3.2 Set-up

Measurements were conducted in the VRLab at the University of Twente. Subjects stood on a dual force plate, embedded on a 6 degrees of freedom platform. Kinematics were recorded with an optical system and ground reaction forces were recorded with a dual force plate.

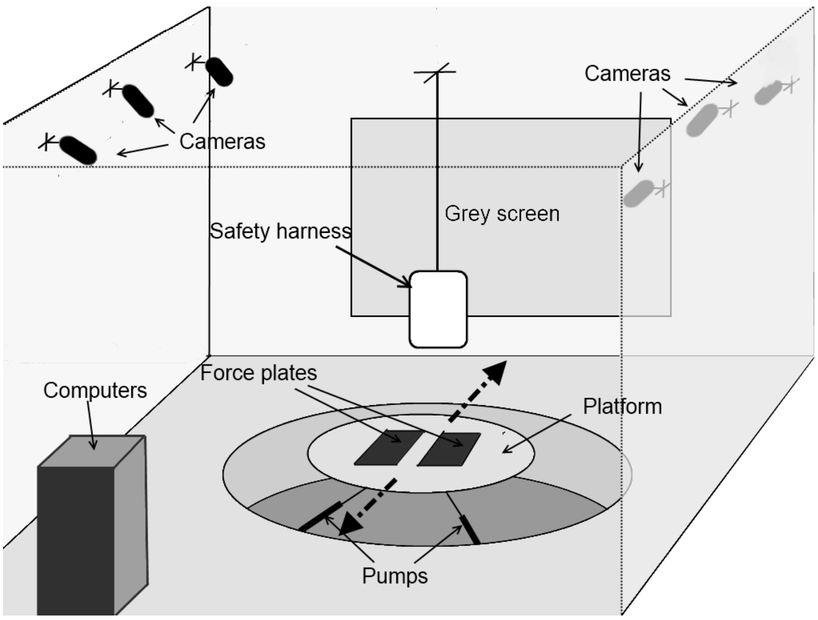

Figure 3.5: The set-up used for the experiments. Data was recorded with six optical cameras and two force plates.

3.2.1 Subjects

Eight healthy male subjects volunteered to participate in this study. The average age was 22.0 (std 1.6) years. The average height and weight were 185.3 (std 7.1) cm and 83.5 (std 15.5) kg respectively. An overview of the participants anthropometric measures are presented in table 3.2. None of the participants had a history of neurological or balance disorders. No medical ethical committee approval was required for this survey. All subjects gave their written informed consent prior to the start of the experiment (appendix H).

The subjects faced a light grey screen. The subject was obliged to wear a safety harness whenever he was standing on the platform. The safety harness was suspended from the ceiling and was worn around the trunk. The safety harness height was adjusted for each subject to prevent constraints in movements necessary for balancing or stepping.

Table 3.2: Subjects anthropometric measured prior to the experiments. W represents the subjects weight, H the body height and Leg the leg length. Pref. Foot represents the subjects preferred foot and Seq the trail sequence. The means and standard deviations are shown in the last row.

#	Sex	Age	W (kg)	H (cm)	Leg (cm)	Shoe size (EU)	Pref. Foot	Seq
1	M	22	65	182	94	43	Left	[3,1,2,4]
2	M	19	78	189	99	45	Left	[2,3,4,1]
3	M	24	105	178	90	44	Right	[4,2,3,1]
4	M	21	66	182	95	41.5	Left	[1,3,2,4]
5	M	24	85	184	100	45	Right	[3,4,1,2]
6	M	22	76	181	95	43.5	Right	[3,2,4,1]
7	M	22	88	201	110	48	Left	[1,2,3,4]
8	M	22	105	185	94	46	Right	[2,3,4,1]
		22.0±1.6	83.5±15.5	185.3±7.1	97.2±6.1			

3.2.2 Moveable platform

Participants stood on a 6 degrees of moveable platform (Hydraudyne, HSE-6-MS-8-L-2D). The platform was powered by three servo-controlled hydraulic pumps. This made it possible to translate the platform in three axes and to rotate it around three axes.

There were two 6 degrees of motion forces transducers (ATI-Mini45-SI-580-20) mounted on the platform. The force plate dimensions were 15×17.5 cm. The platform computer was controlled by Matlab and Simulink (R2010b). The input signal is described in section 3.1.2.

3.2.3 Optical system

Movements were recorded with an optical system (VICON Oxford Metrics, Oxford, UK)). The six high speed cameras recorded the positions of reflective markers. The cameras make use of infra-red lightning to record 2D images, which could be reconstructed in 3D.

3.3 Statistics

3.3.1 Regression analysis

Human behaviour is in the general case non-linear. Therefore, it was hypothesised that there would not be a linear regression between step frequency and amplitude. The balance system would be able to correct low perturbations rather good, until a threshold. At that threshold a steep increase in step frequency would be noticeable. Subsequently the gain in step frequency will attenuate until the amplitude where every perturbation results in a step.

Nonetheless, the difference between the with-block and without-bock observed frequencies could be evaluated. Plainly, the small block would impair balance, resulting in a higher step frequency. If the latter assumption would hold, the difference between with-block and without-block could be captured in a linear regression. The static model is described with the first order polynomial:

$$\hat{y}_i = \hat{\beta}_0 + \hat{\beta}_1 x_i + \epsilon_i, \quad i = 1, \cdots, n \tag{3.2}$$

Where x is the amplitude ranging from 0.04 to 0.08. $\hat{\beta}_0$ is the intersect with the vertical axis, reflecting a shift in the step frequency. The slope is captured in $\hat{\beta}_1$ and reflects a increased step frequency with every amplitude. The ϵ_i is the error, the difference between the y_i and $\hat{y}_i$. The values of $\hat{\beta}_0$ and $\hat{\beta}_1$ are derived with minimization of sum of squared residuals.

3.3.2 Step incidence analysis

The steps per amplitude were summarised in a fraction for the with-block and the without-block situation. The fraction represents the step incidence proportion, ranging from zero to one, where zero mean that no steps were made at the specific amplitudes and one that all perturbations resulted in a step. Step incidence is a rank and therefore not normal distributed. The Wilcoxon signed-rank test is an alternative for the paired t-test, designed for non-parametric populations. The Assumptions for a Wilcoxon signed-rank test are:

· Data is measured on an interval scale.
· Data comes from the same population and is paired.
· Each pair is chosen randomly and independent.

. The steps incidence proportions means were compared for four different cases:

Forward with-block	Forward without-block
backward with-block	backward without-block
Forward without-block	backward without-block
Forward with-block	backward with block

3.3.3 With-block versus without-block comparison

Various parameters could be compared when analysing the with the without-block conditions. this study restricts to the mean peak velocity for describing balance, and the peak

CoP for describing balance responses. It is assumed that these parameter are normally distributed and therefore the paired t-test used to compare without-block and with-block means. The assumptions for a paired t-test are:

- Data is normally distributed
- Data comes from the same population and is paired.
- Each pair is chosen randomly and independent.

.

4 Data processing

4.1 Raw data

Optical data

Positions of these markers are measured by the optical system and expressed in an 3D Cartesian coordinate system. The marker positions were used to describe body segment positions and motions. The data was digital filtered using a second order low pass recursive Butterworth filter with a 10 Hz cut off frequency.

Marker data that was partially obstructed during the trail was reconstructed using interpolation. Markers that were invisible during the whole trail, or falsely interpolated data was attempted to reconstructed with the help of surrounding markers. An overview of this method is presented in appendix B

Force plate data

The raw force plate data, an example attached in appendix C, were resampled from 600 Hz to 120 Hz and subsequently digital filtered using a second order low pass recursive Butterworth filter with a 5 Hz cut off frequency.

The dual force plate had a slight offset. To correct for the offset, the data was multiplied by a calibration matrix. The force plate was mounted on the platform and had to be corrected for the moment of inertia and the regarding force plate top layer mass. This is shown in the following equation:

$$R = F_{fp} - m_{fp} \cdot a_{fp} \tag{4.1}$$

R represents the corrected force, corresponding to reaction forces produced by the subject. F_{fp} is the force measured by the force plate, m_{fp} the force plate mass and a_{fp} the force plate acceleration. The forces and torques where subsequently multiplied by a calibration matrix and multiplied by a transformation matrix. The accelerations in y and z were neglected. The forces and torques during an experimental trail are presented in appendix C.

Table 4.1: Segments used for computing of the CoM, together with the corresponding segment's mass and segment mass relative location. [9]

	Segment	Markers	Mass	Proxal-distal
	HAT	(RSHO & LSHO) - (LASI & RASI & CLAV)	0.678	0.626
2x	Upper leg	ASIS - KNEE	0.100	0.433
2x	Lower leg	KNEE - MAL	0.0465	0.433

4.2 Processed data

4.2.1 Variables

Center of Mass

Position of the segments markers that represent the feet, under-, upper legs and the HAT are monitored by the optical system. The anthropometric table **??** describes the positions and mass fractions of the different segments. The CoM can estimated using:

$$CoM = \frac{1}{M} \sum_{i=0}^{n} m_i x_i \tag{4.2}$$

Where M is the the total body mass, m_i the segment mass and x_i the proxal-distal position of the CoM. The segments used in this study are shown in table 4.1. There are several methods of defining segments masses and CoM's, the used values are obtained from cadaver studies.

As discussed, inverted pendulum boundary conditions are violated during a step. The segment mass computation does no longer provide the correct inverted pendulum CoM and therefore the CoM was not estimated whenever a foot was lifted.

Center of Pressure

The CoP becomes prone to errors when F_y was very low (i.e. when the foot gets lifted). A small error in F_y corresponds to a large error in CoP. Equation 2.9 states that CoM and netto CoP should equal over the entire trail, provided that start and final position of the CoM are identical. Observations of the CoP and CoM in trajectories revealed that this relation ship was violated. It was assumed that the CoM was correct, thus the CoP required a correction. This was done by adding the average CoM difference to the netto CoP.

Weight distribution

The weight distribution represents the weight-bearing which is measured by the vertical force sensors. The vertical axis is normalised to the body weight. Both force plates have four vertical force sensors, laying in the rectangle (0,0),(0,Z),(Z,X),(X,0). The fraction between both vertical forces represents the weight fraction. An example of a lateral weight shift of subject 1 during a forward perturbation is shown in figure 5.2.

4.2.2 Identification of characteristic points in time

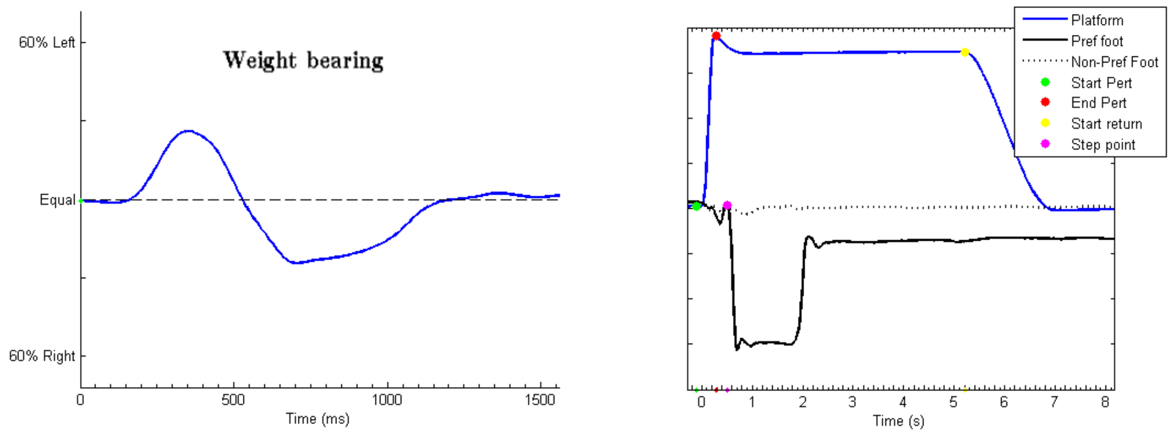

Figure 4.1: Left: an example of lateral weight shift, evoked by a perturbation. Right: the characteristic points in time required for data analysis. The platform's horizontal position indicates the corresponding perturbation and the preferred foot the evoked step response.

The observant can easily check the number of steps made by a subject during the experiment. However, a computerised calculation is required to make sure that every excursion above the same threshold is identified as a step. All points were determined with the marker position data.

The three markers attached to the platform indicate the platform position. One platform translation has three characteristic points:

· Start of the perturbation

· End of the perturbation

· Start of the return motion

. Functional data were collected during the start of the perturbation and the start of the returning motion. The time between start and end perturbation served as a verifying method for the translation duration.

The position of the foot relative to the platform had to be derived in order to address markers displacement as a step. Subsequently the foot positions were computed by averaging the three foot markers and by marking the begin position as zero. A step was recognised as a foot displacement above a fixed threshold, namely 0.05m. An overview of the discussed characteristic time points are given in figure 5.2.

4.2.3 Platform amplitudes

The original protocol was designed with five forward and five backward translations, $\pm$[0.04 , 0.05 0.06 0.07 0.08]. Inspection showed that the devised amplitudes did not correspond with the actual measured amplitudes produced by the platform. Not only was one amplitude missing in both forward and backward translations, the values did not correspond with the associated input values. Moreover, the amplitudes were dispersed for subject 7 (appendix C). Therefore, the mean output amplitudes were investigated amplitude which had more than a 0.002m deviation from the mean amplitudes were excluded. A more detailed discussion about the platform is given in section (6.4).

Table 4.2: the protocol had ten different amplitudes. The mean output amplitudes had only eight amplitudes and a slight offset.

#	A1	A2	A3	A4	A5	A6	A7	A8	A9	A10
Input (m)	-0.08	-0.07	-0.06	0.05	-0.04	0.04	0.05	0.06	0.07	0.08
measured a (m)	-0.074	-0.065	-0.055	-0.043	0.044	0.055	0.066	0.077		

4.2.4 Wiggling

A undevised response with regards to a perturbation was encountered when subjects wore blocks. During the experiments it was observed that subjects wiggled on the small block repeatedly. This flaw of the blocks biased stepping, (i.e. the amount of steps during a trail was reduced due to the wiggling). The wiggling is in principle the same as the heel and toe raise, which was a not allowed strategy. Nevertheless, the wiggling appeared to be a common reflex, not unlearned with repeated instructions. It was decided to included a wiggle as a step. A wiggle was defined as the toe marker minus the heel marker exceeding a 1.5 cm threshold.

5 Results

5.1 Base of support

There was a clear difference in the BoS between the without-block and with-block condition, the BoS of the small block was considerably smaller. This corresponds to the theory, where contact surface length should relate to the BoS.

The small block's BoS exceeded in most cases the theoretical limit, the block length. The BoS for a typical subject is depicted in figure 5.1. The subject was instructed to make use the ankle strategy exclusively. All the BoS are positively shifted in the x direction compared to the theoretical limit.

The complete set of BoS plots are shown to appendix E. Theoretically, the isolated hip strategy would change the CoP on the force plate. This was not the case, the hip BoS had generally the same length as the ankle BoS. This was presumably caused by not adequately following the instructions and nonetheless using the ankle strategy.

The tight coupling between the CoM and CoP indicates a low angular acceleration, which was instructed by observant by 'bending slowly'.

5.2 Step incidence

Wiggling (described section 4.2.4) occurred for all subjects. However, some subjects were more prone to wiggling on the small block. Wiggling above the 1.5 cm threshold was assigned as a step. The wiggle proportion for every subject per amplitude are presented in appendix F. Forward steps were always evoked by backward platform translations and the same analogy holds for backward steps and forward platform translations. There were significant results between the forward/backward perturbations and between the with-block and without-block condition.

Forward with-block	Forward without-block	P<0.000
backward with-block	backward without-block	P<0.000
Forward without-block	backward without-block	P<0.000
Forward with-block	backward with block	P<0.002

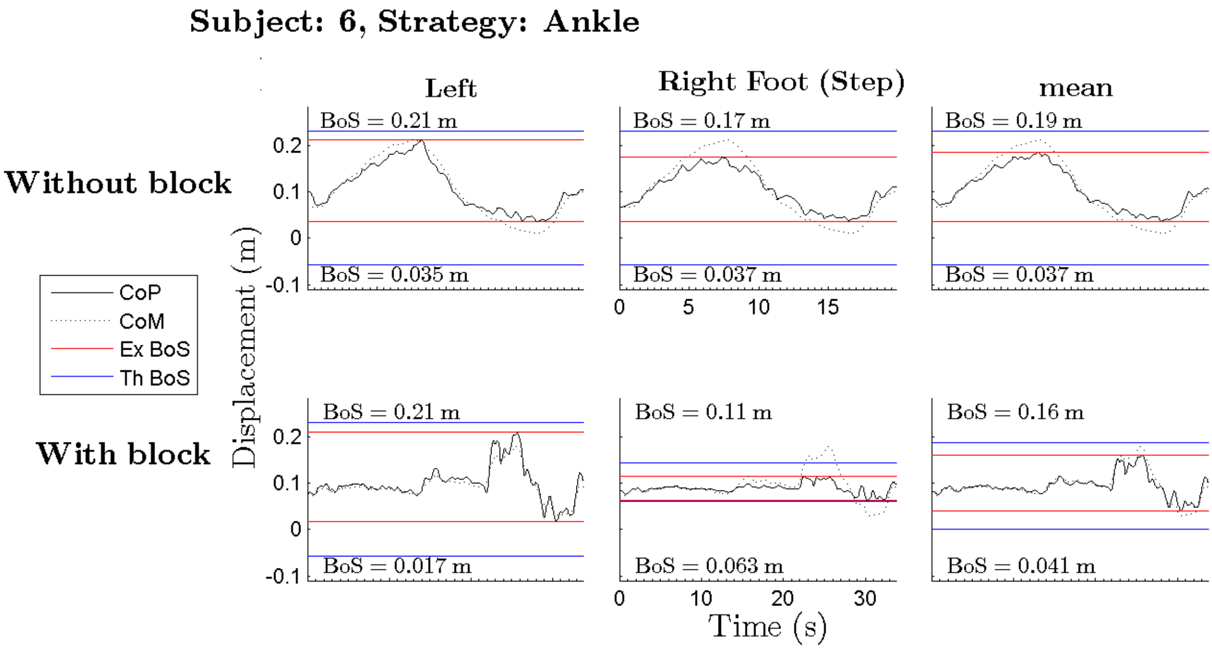

Figure 5.1: BoS measurement results for subject 6 for both with-block and without-block. The block was bound under the right foot. The Exp BoS represents the experimental BoS, the maximal and minimal CoP measured during the trial. The BoS with-block is considerably smaller. Note that the subject was instructed to use only the ankle strategy.

5.3 Weight-bearing

Unequal weight-bearing was not noticed by the observant in without-block trials. On the other hand, unequal weight-bearing did occur occasionally in the ample period during with-block trials and was noticed by the observant. The weight-bearing feedback, by means of instructions by the observant, were followed adequately by the subjects.

The expected lateral weight-bearing shift did occur adjacent to amplitudes required to evoke a step. The shift in weight-bearing occurred for both with-block as without-block subjects. Bar charts of the incidences of lateral weight shift are presented in appendix G. Those bar charts indicate a large variant in weight-shift. In example is shown in figure 5.3. In this figure an subject is perturbed with a forward platform translation. The weight-bearing graph shows a shift in weight-bearing after the onset of the perturbation. This shape of weight-bearing proportion line was characteristic for all weight-bearing shifts. The figure also shows a wiggle on the small block. This has an effect on the CoP (right foot), which makes an oscillation. The CoM exceeds the small block BoS but due to the compensation torque of the left foot the CoP of the left foot remains between the BoS boundaries. Four more examples are included in the appendix (D).

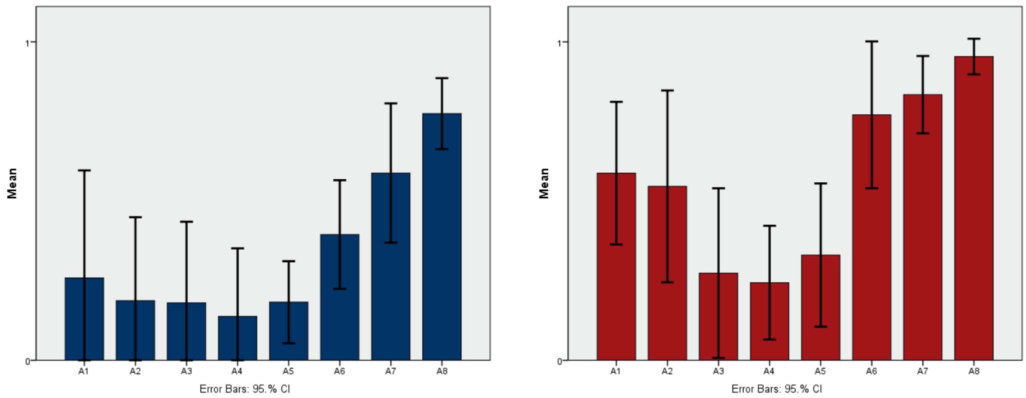

Figure 5.2: Step incidence proportions at the eight different amplitudes. The bar heights indicate the average steps proportions over the eight subjects, the error bars show at a 95 % confidence interval. Left: without-block. Right: with-block. The bars show an indication for an increased step proportion for with-block and for backward platform translations.

5.4 Center of pressure

One assumption for the sagittal inverted pendulum model was the neglectable M/L forces and torques. CoP in the z direction were indeed neglectable, as shown 5.4. In this graph the CoP and CoM trajectories are plotted after a backward perturbation. It is also shown that the both CoM as CoP return to the initial position after the perturbation. Similar result were seen for forward and backward, with-block and without-block trajectory plots, the trajectory plots for the four conditions are presented in appendix C.

Forward perturbations resulted in a backward CoP excursion and a backward perturbation in a forward CoP excursion. The CoP excursion were transient (typical durations of 1 second) and the netto CoP exceed the CoM. There was a clear difference in magnitude between the CoP of with-block CoP's. Even with the small perturbations (where the CoP of the small block foot was well between the BoS boundaries) a difference in magnitude was apparent.

A surprising result was the occasional exceedance of CoP the BoS in both forward as backward perturbations. This is in contract with the theory, it is more likely that this is the result of a improper BoS measurement.

The CoP and torques measurements were averaged over the eight amplitudes for each subject. An example is shown for the a forward (A6) and a backward (A3) perturbation. These plots gave valuable insight in the CoP responses evoked by the platform translation. The plots show the left, right and mean CoP and the left, right and total torque after the onset of the perturbation. Additional information is gained with the weight-bearing and torque contribution graphs. The CoP and torque increases immediately after the onset of the perturbation. This is in accordance with the theory, which formulates that the ankle

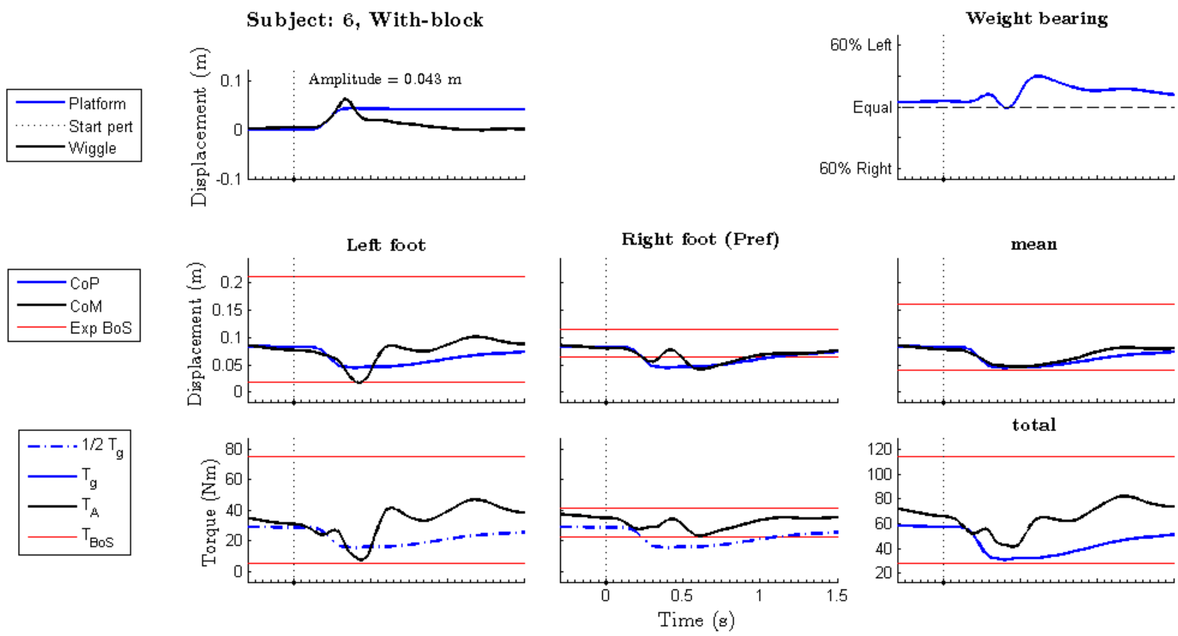

Figure 5.3: CoP and torques of subject 6 during a backward perturbation with-block. Exp BoS represents the BoS measured in preceding BoS experiment, T_g torque as a result of gravity, T_a the ankle torque. and T_{BoS} the maximal torque associated with the BoS. The CoP exceeded the BoS for the foot with the block. The onset of the perturbation is marked as zero. A change in weight-bearing, wiggling and a difference between left and right CoP's and torques are noticeable

joint creates immediately passive torque after a rotation[10]. The subjects weight-bearing was slightly on the preferred foot.

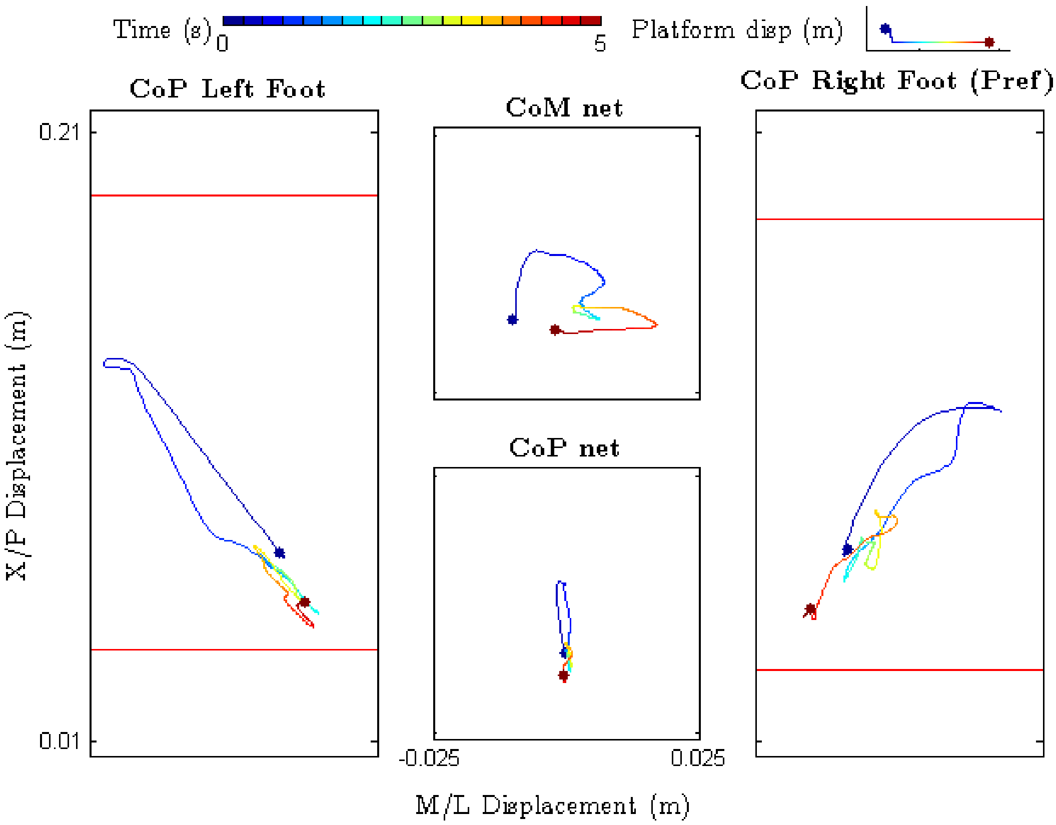

Figure 5.4: Trajectory of the CoP and CoM of subject 6 during a backward perturbation. The CoM sways forward and as a result the CoP under both feet move forward. The A/P scale is four times as large as the M/L scale. Note that the netto CoP is excursions are much smaller than CoM excursions in M/L direction, indicating a hip contribution in the frontal plane.

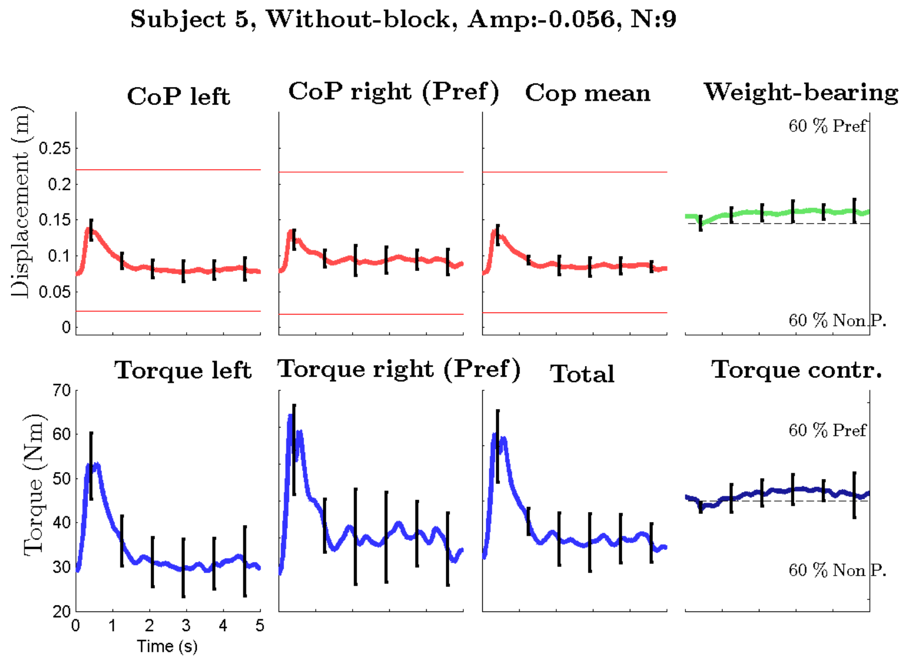

Figure 5.5: Averaged CoP and torques for perturbation amplitude A6, without-block (subject 7) The CoP of and torque of the left and right foot are identical.

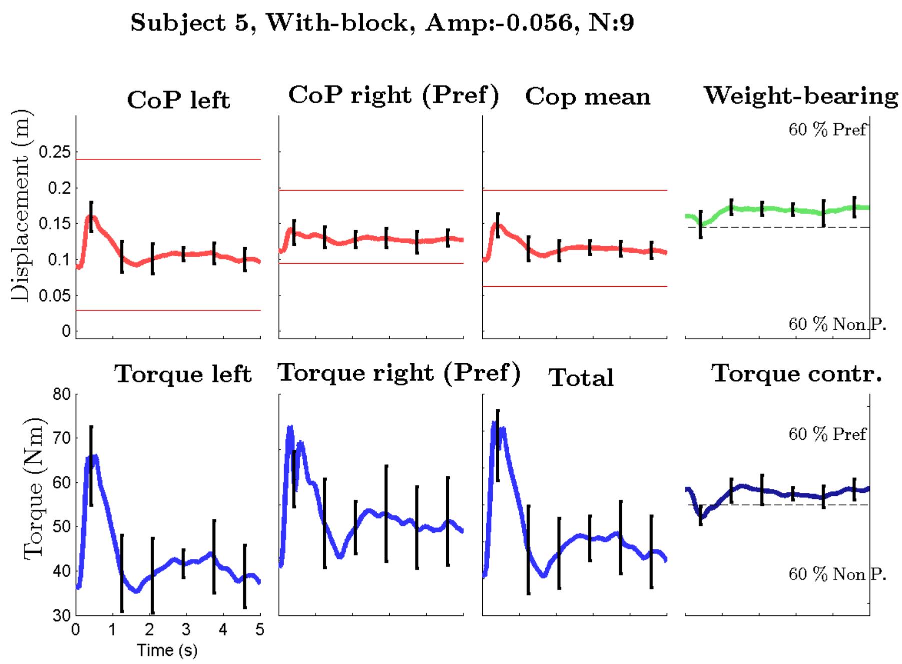

Figure 5.6: Averaged CoP and torques for perturbation amplitude A6, with-block (subject 7). The CoP of the right foot (which contained the small block) was substantially smaller than the CoP of the left foot. An peak in the torque contribution was noticeable.

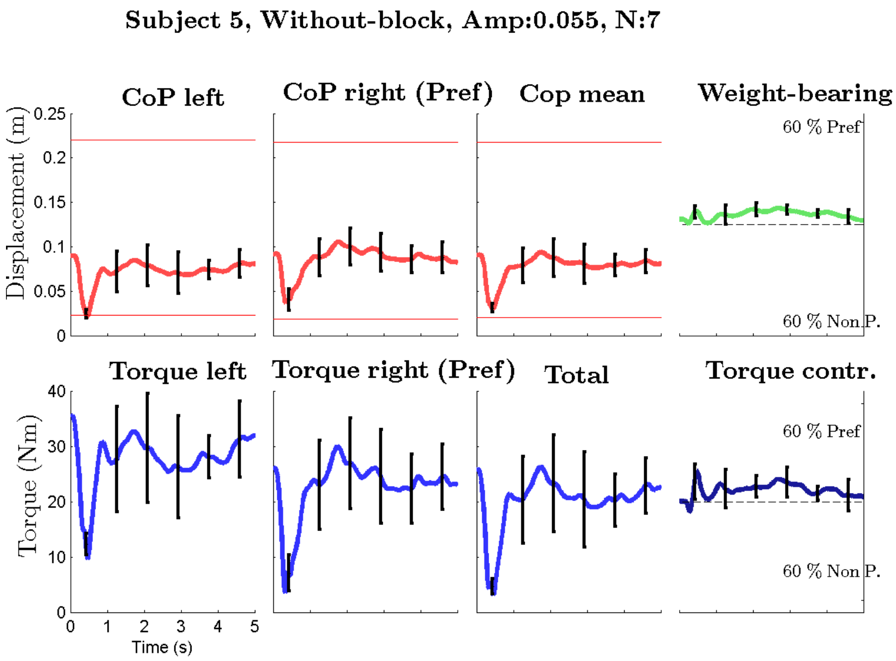

Figure 5.7: Averaged CoP and torques for perturbation amplitude A3, without-block (subject 7). The CoP are identical. Torque contribution and weight-bearing shows a slight favor for the proffered foot.

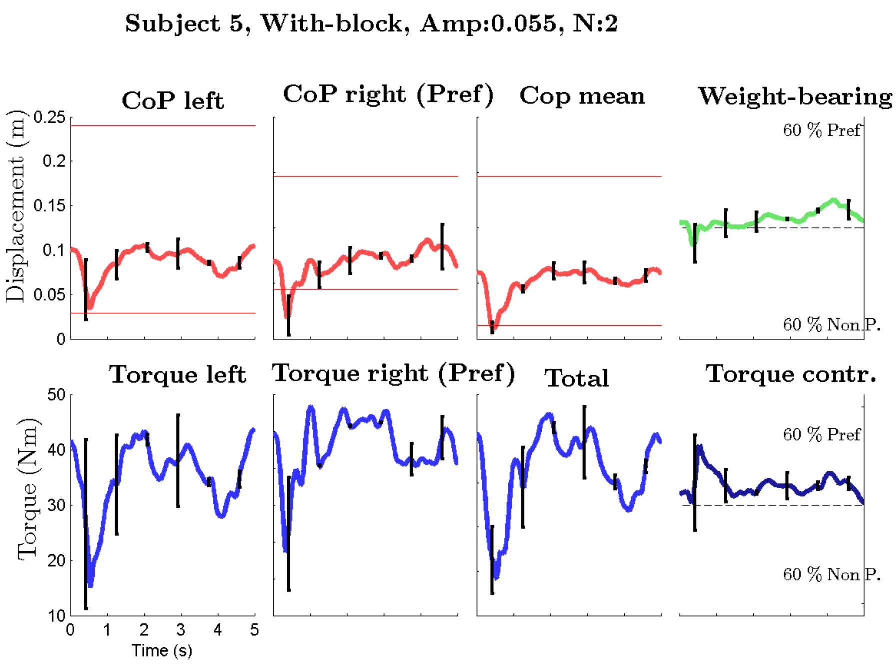

Figure 5.8: Averaged CoP and torques for perturbation amplitude A3, with-block (subject 7). Note that in this case the CoP did exceed the BoS limits (and the torque contribution shows a higher for the small block foot), the error bars are high is very small, N:2, indication for an invalid sample.

6 Discussion

The study goal was examine balance and step responses with an impaired ankle. The analysis gave insight in the underlying CoP and ankle torques. It was proven that there was a significant difference between the with-block and without-block conditions and between forward and backward perturbations. From the numerous CoP plots valuable insight was gained on CoP, torque and weight-bearing. The study failed present statical evidence on the underlying CoP and ankle torques or the regression between the amplitudes and the step incidence.

6.1 The wooden blocks approach

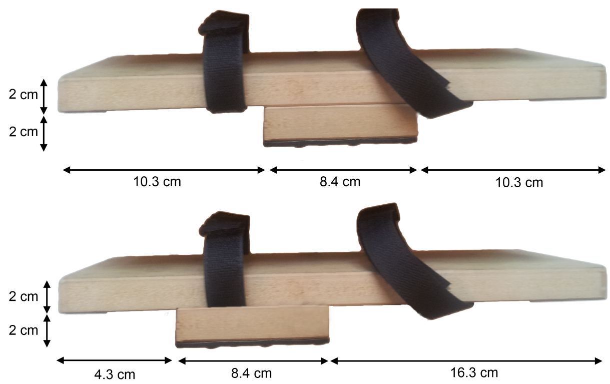

Figure 6.1: Above: the used block in the experiment. Below: the proposed block with a shifted BoS.

The wooden block approach is an unreliable method of mimicking unilateral affections. Although it does impose an asymmetry in torque, it has several shortcomings. First, the ankle torque is attenuated by reducing the BoS to a fixed length. In this manner the theoretical ankle torque is preserved when the CoP is within the BoS boundaries.

Moreover, the small block is placed halfway at the block length, figure (6.1). The ankle, where the CoM in A/P is located at quiet stance, is positioned at roughly 1/5 of distance between the heel and the toe. Thus the CoM would be outside the BoS at the small block leg. A way to overcome this problem is to rotate around the longitudinal axis.

This rotation affects the A/P torque and generate unintended M/L torques. In future studies the small block should be placed under the ankle with 0.2 directed at the ankle and 0.8 directed at the toe, as shown in the figure. An alternative should be devised with an impaired ankle torque rather than a reduced BoS. One example coulde be a torsion spring which imposes torque, linear with the rotation angle.

Another shortcoming of the blocks is the 4 cm height. This additional height increases the CoM and thereby reduces stability. To compensate for this the control subjects should wear the large blocks on both feet in future surveys.

Furthermore, the impaired ankle torque is not the only symptom for the reduced stability in PD and stroke patients. An impaired proprioception increase the CNS reaction time and therefore reduce stability.

a small deviation of the correct foot placement on the block has a major effect on the result. First, a shift in BoS will occur. This has a direct effect on the forward/backward comparison. Moreover, the wiggling would increase, which would be assigned as a step.

6.2 Protocol

BoS trials were conducted by letting subjects slowly bend forwards and backwards. The CoP excursions were not significant larger than the CoM excursions, which is an indication for small CoM angular acceleration. However, during the experiments it was observed that the subjects resorted to the knee strategy even though the hip strategy was instructed.

The original protocol design consisted of 240 perturbations per subject. The total set of translations contained both forward and backward translations, namely ten different amplitudes and had a with and without-block condition. This would result in six perturbations per subject per condition. Due to an artefact in the programming the trial consisted of only 12 perturbations, instead of 20. This resulted in 144 perturbations per subject. Furthermore, the random sorted perturbations got truncated, causing a variation in perturbation amplitudes in each trial. 10% of the perturbations got excluded as they could not met the required acceleration. Most of them were the first perturbation. The knee, hip and ankles for subject 1 during a without-block trial are presented in figure 6.3.

6.3 Subjects

A total of eight subjects volunteered in the experiment. For more significant results more subjects are required.

Subjects 5 and 6 participated besides the experimental trials also in the pilot trials. The repeated subjects had more practise than the new subjects. More practise could influence the results due to the learning and carry-over effects.

Subject 1 and subject 8 reported minor injuries. Subject 1 had an incidental ankle injury and subject 8 had more chronicle knee complaints. The ankle injury for subject 1 manifested right whereas his preferred foot was determined to be left. The subject changed his stepping for the without-block trials. It might be plausible that the with-block experiments are also effected by this change in stepping behaviour. The BoS from subject 1, appendix E, were divergent from theory, and from the other subjects BoS. The subject's left and right BoS were not assigned, the shift was apparent in all the three strategies. Subject 8 his knee complaints were in his right knee. Nevertheless, no effects on the BoS were observed.

6.4 Hydraulic Platform

A small overshoot in both forward and backward translation was noticeable in the platform amplitude.

The platform was set to start the immediately with the perturbation sequence after the platform was raised. Apparently, the hydraulic pumps were not capable in generating the required acceleration. An example of the amplitude and acceleration of the first amplitude is shown in figure 6.2

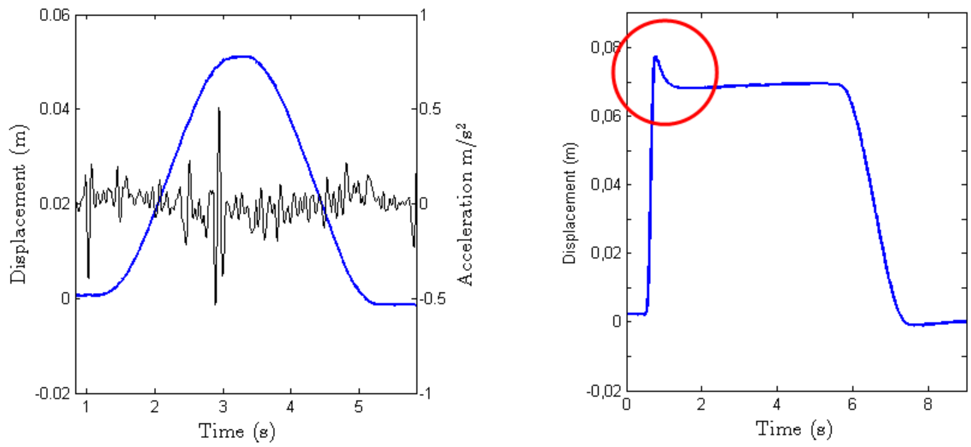

Figure 6.2: Left: An example of an excluded first translation. The platform was unable to generate the required acceleration. Right: the platform overshoot, denoted by the red circle. The overshoot occurred in all in both backward and forward translations.

6.5 Multi segment model

The body mechanics were simplified to an one segment rigid sagittal model (section2.3.2). In this manner, the corrective ankle torque could be derived. However the hip strategy is an effective balance strategy and the hip torque should therefore not be neglected (section 2.1.2). It became clear that the knee strategy was underestimated, in particular with forward perturbations. Furthermore, the expected hip strategy manifested more than expected. Ankle, knee and hip angles of the left side of the body are plotted in figure 6.3. The ankle angle is defined as the angle between the foot and the shank, the knee angle as the angle between the shank and the thigh and the hip angle as the angle between the thigh and the upper body.

To prevent a subject from using the knee and hip strategy, i.e confining the subject to the ankle strategy, the hip and knee joints could be constrained with an orthosis.

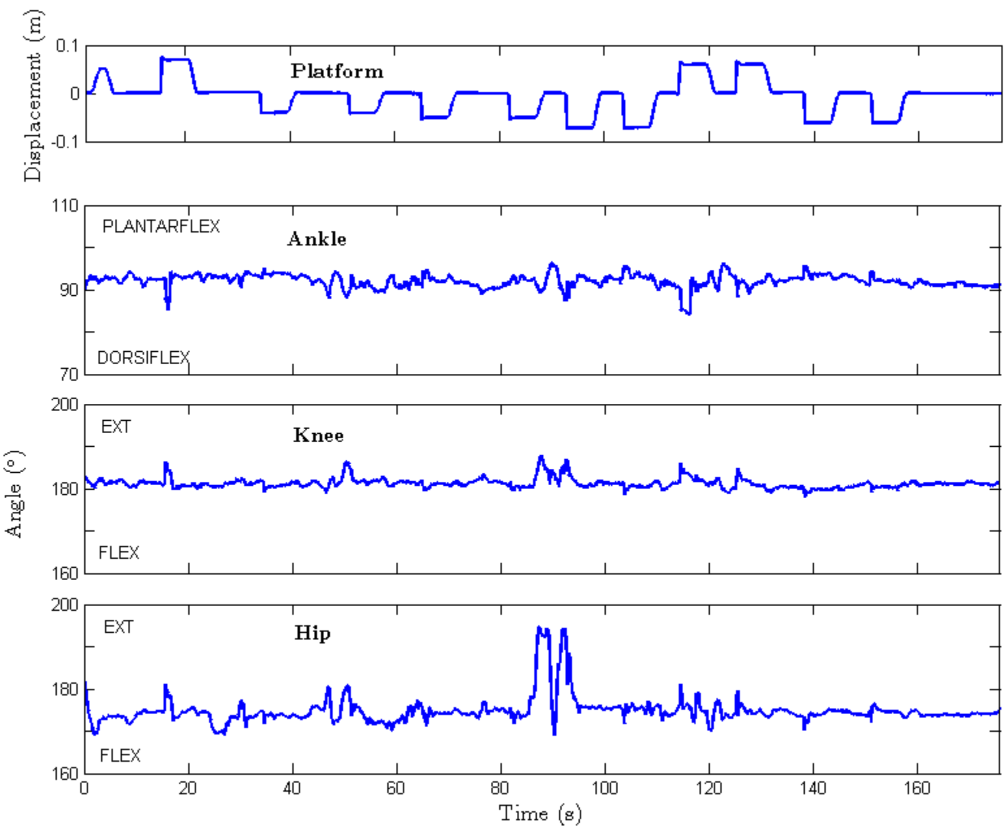

Figure 6.3: Above: the imposed perturbation (i.e. platform displacement) for subject 1, without-block. No steps were made during this trial. Below: the corresponding knee, hip and ankle angles. Although the subject was instructed to correct only with the ankle, the hip and ankle angle indicate that the subject also used additional strategies to adjust the CoM.

In order to compute the total corrective torque, the hip and knee torque have to be derived. Quantifying the hip torque is cumbersome. Not only the acceleration of the limbs has to be computed, also the internal forces and the segment center of mass has to be known. The total torque could eventually be computed by using the Newton laws, equations 2.1.

The total corrective toque can be expressed as:

$$\tau = \tau_a + \tau_k + \tau_h \tag{6.1}$$

Where τ is total corrective torque, τ_k the knee torque and τ_h the hip torque.

6.6 Electrocardiography

EMG is the measurement of the electrical signal associated with a muscle contraction. An EMG measurement could provide interesting insights in the muscles activations caused by the perturbations. In this manner, active and passive corrective torque could be differentiated (i.e the response from the CNS the imposed perturbation). The activated muscles, summed burst magnitude and reaction time could be studied and compared for the different conditions.

6.7 Subject weight and pendulum length

The measured weight of the subject by the scale and by force plate did not correspond, (table 6.1). The weight should correlate with the subjects vertical reaction force. Such a false weight could bias the $CoP - CoM$ relationship. The subjects weight was ascertained with the force plate in the static trial to have a correct $CoP - CoM$ relation,

Table 6.1: The difference between the subjects weight measured by the scale and the force plat.

#	1	2	3	4	5	6	7	8
Scale weight	65	78	105	66	85	76	88	105
Height	182	189	178	182	184	181	201	185
Weight (kg)	60.3	71.6	104.1	67.9	82.1	71.0	85.2	99.8
ℓ (cm)	107	105	97	101	106	101	161	101

6.8 Other descriptive balance methods

The present study used the peak CoM A/P velocity as subjects balance measure. Another methods may include CoM trajectory length or the peak CoM acceleration. The peak CoP and the relative distance to the BoS was evaluates for the perturb response. Another possible method that relates the contribution of the corrective torque to the balance contrition was elaborated by van Asseldonk, et al. by means of the Frequency response function [3].

6.9 Statistics

The lack of statistical basis on balance and response conclusions are twofold. The first reason was the mistake made in programming the platform perturbation sequence. This resulted not only in a reduced perturbation amount per subject, it also introduced a unequal perturbation amount per amplitude, which makes statistical tests cumbersome. The second reason for the lack of conclusions was the extent of parameters that were tried to be investigated: both forward and backward, without-block and with-block and five different amplitudes per direction. Furthermore, the responses were subdivided into three categories: stepping, weight-bearing shift and feet in place responses (as suggested in 2.1.5). During the analysis a new variable had to be introduced. The additional parameter wiggling, which had major effect on the CoP. Perturbations were randomised as much as possible (order, magnitude, recovery time and trial block sequence), which required identification and categorisation. Finally, eight subjects participated with each a designated preferred and a non-preferred foot.

Effects on balance and step responses are a broad concept. Therefore all the variables imagined were computed (for example: reaction times, sway angles and the M/L CoP). Only in the very end of the study was a clear focus on the three descriptive values that were for statistical testing (peak CoM velocity, peak CoP minus BoS, and peak CoP). Given the limit time span for the study an earlier focus on these variable was required.

Add to all (plus the weeks lost due bad luck on account of the break down of the dual force plate power supply), it could be concluded that there was a time shortage for a full statistic model.

6.10 Recommendations

A follow-up study could include a regression analysis. It is recommended to choose between either forward and backward perturbations, and have an equal amount of perturbations per amplitude. To study the CoP and torque contributions between the with-block and the without-block conditions it is advised to restrict to one amplitude. In this study the subjects CoP's and torques were averaged over the amplitudes (intra-subject). To gain a full inter-subject overview, a plot with the subsequently averaged amplitudes over the subjects could be made.

7 Conclusion

The purpose of the study was explanatory, mimicking unilateral diseases like Parkinson Disease or stroke patients. More insight was gained on the effects of a reduced impaired ankle. Significant results were gained for the step incidences comparison between forward and backward perturbations, and between the with-block and without-block conditions. The CoP was confined by the BoS. The small block CoP did not always reach the BoS limit, even though the CoM did exceed the BoS. This was compensated by the foot-size block, creating an asymmetrical ankle torque. Unfortunately, no quantification could be made on the balance (CoM velocity) and balance responses (peak CoP and CoP-BoS distance). The devised methods in the present study are proposed to be repeated in a follow-up study.

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Appendices

A Inverted pendulum mechanics

Torque and moment of inertia

Torque equals second moment of inertia times angular acceleration, which can be derived with the Newton laws.

$$\boldsymbol{\tau}_{\text{net}} = I\ddot{\theta} \tag{A.1}$$

Where $\boldsymbol{\tau}_{\text{net}}$ represents the total torque around the ankle. When only gravitation forces work on the mass the equation for the torque becomes:

$$\boldsymbol{\tau} = M \cdot g \cdot \ell \cdot \sin\theta \tag{A.2}$$

Substituting the equations for torque

$$I\ddot{\theta} = M \cdot g \cdot \ell \cdot \sin\theta \tag{A.3}$$

Considering the mass of the total body as a point mass the moment of inertia yields:

$$I = M \cdot \ell^2 \tag{A.4}$$

$$M \cdot \ell^2 \cdot \ddot{\theta} = Mg\ell \sin\theta \tag{A.5}$$

$$\ddot{\theta} = \frac{g}{\ell} \sin\theta \tag{A.6}$$

Considering only small deviations of the erect stance:

$$\ddot{\theta} = \frac{g}{\ell}\theta \tag{A.7}$$

The solution to this second order differential equations is:

$$\theta = \exp\left(\frac{g}{\ell}t\right) \tag{A.8}$$

The latter equation describes the unstable behaviour of an inverted pendulum.

Hamilton's equations

Another way to calculate the equations motion with the Hamiltonian method. This method provides a deeper inside in the classic mechanics. First de Lagrangian has to be derived:

$$L = T - V \tag{A.9}$$

L represents the Lagrangian, T the kinetic energy and V the potential energy. The Lagrangian for the inverted pendulum yields:

$$L = \frac{1}{2}M \cdot v_1^2 + \frac{1}{2}m \cdot v_2^2 - M \cdot \ell \cdot \cos\theta \tag{A.10}$$

Where x represents the platform translation, v_1 the velocity of the feet, v_2 the velocity of the center of mass. v_1 and v_1 could be derived by the derivative of the positions. M is the CoM and m the feet's mass.

$$v_1^2 = \dot{x}^2 \tag{A.11}$$

$$v_2^2 = \left(\frac{d}{dt}(x - \ell\sin\theta)\right)^2 + \left(\frac{d}{dt}(\ell\cos\theta)\right)^2 \tag{A.12}$$

Applying the trigonometric Identity the equation for v_2:

$$\sin(\theta)^2 + \cos(\theta)^2 = 1 \tag{A.13}$$

$$v_2^2 = \dot{x}^2 - 2\ell\dot{x}\dot{\theta}\cos\theta + \ell^2\dot{\theta}^2 \tag{A.14}$$

The total Lagrangian now holds:

$$L = \frac{1}{2}(M+m)\dot{x}^2 - m\ell\dot{x}\dot{\theta}\cos\theta + \frac{1}{2}m\ell^2\dot{\theta}^2 - mg\ell\cos\theta \tag{A.15}$$

The Hamilton's equations:

$$\begin{aligned} \frac{\mathrm{d}}{\mathrm{d}t}\frac{\partial L}{\partial \dot{x}} - \frac{\partial L}{\partial x} &= F \\ \frac{\mathrm{d}}{\mathrm{d}t}\frac{\partial L}{\partial \dot{\theta}} - \frac{\partial L}{\partial \theta} &= 0 \end{aligned} \tag{A.16}$$

Substitution of the Lagrangian into the Hamilton's equations:

$$(m+M)\,\ddot{x} - M\ell\ddot{\theta}\cos\theta + M\ell\dot{\theta}^2\sin\theta = F \tag{A.17}$$
$$\ell\ddot{\theta} - g\sin\theta = \ddot{x}\cos\theta \tag{A.18}$$

The equation could be linearised around the upright stance, $\theta \approx 0$:

$$(m+M)\,\ddot{x} - M\ell\ddot{\theta} + M\ell\dot{\theta}^2\theta = F \tag{A.19}$$
$$\ell\ddot{\theta} - g\theta = \ddot{x} \tag{A.20}$$

with no platform perturbations or applied forces and after linearisation the Lagrangian can be be simplified to same second order differential equation as equation 2.8:

$$\ddot{\theta} - \frac{g}{l}\theta = 0 \tag{A.21}$$

Eigenvalues

A third way to verify the inverted pendulum instability is with the polarity of the eigenvalues. [17] The inverted pendulum equation with corrective torque, $\boldsymbol{\tau} = CoP \cdot R_{\mathrm{y}}$, is:

$$\boldsymbol{\tau} - M \cdot g \cdot \ell \cdot \cos\theta = \ddot{\theta}I \tag{A.22}$$

First the inverted pendulum equation has to be translated into set of first order differential equations.

$$\dot{x} = Ax + B\boldsymbol{\tau} \tag{A.23}$$
$$x_1 = \theta, \quad x_2 = \dot{\theta} \tag{A.24}$$

$$\begin{bmatrix} \dot{x}_1 \\ \dot{x}_2 \end{bmatrix} = \begin{bmatrix} 0 & 1 \\ {}^{Mg\ell}/_{I} & 0 \end{bmatrix} \begin{bmatrix} x_1 \\ x_2 \end{bmatrix} + \begin{bmatrix} 0 \\ {}^{1}/_{I} \end{bmatrix} \boldsymbol{\tau} \tag{A.25}$$

The eigenvalues are derived with:

$$det(A - \lambda I) = 0 \tag{A.26}$$

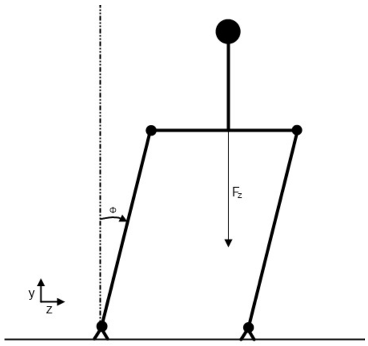


Figure A.1: The inverted pendulum model in coronal plane.

Where I represents the identity matrix.
Solving the latter equation, the following characteristic polynomials arise:

$$\lambda^2 = \frac{M \cdot g \cdot \ell}{I} \tag{A.27}$$

$$\lambda_1 = \sqrt{\frac{M \cdot g \cdot \ell}{I}} \text{ and } \lambda_2 = -\sqrt{\frac{M \cdot g \cdot \ell}{I}} \tag{A.28}$$

Considering the total body mass as a point mass, the moment of inertia yields: $I = M \cdot \ell^2$ and the eigenvalues:

$$\lambda_1 = \sqrt{\frac{g}{l}} \text{ and } \lambda_2 = -\sqrt{\frac{g}{l}} \tag{A.29}$$

Coronal model

While the CoP_l and CoP_r are almost perfectly in phase in A/P direction, in M/L direction the CoP_l and CoP_r are almost completely out of phase [9]. This is a result of the load and unload mechanism of the hip. When the load/unload mechanism is not operational the corrective ankle torque symmetrical and the corrective ankle torque yields:

$$CoP_a = CoP_l \times 0.5 + CoP_r \times 0.5 \tag{A.30}$$

B Data analysis

Digital filtering

The analog channels and VICON position data produce a high frequency noise signal. There are several ways to filter the noise signal. One way to do so is curve fitting, were the signal will be smoothed by best fit in a predetermined shape. This can be e.g a polynomial, a sum of harmonics or a spline. The shape is fitted trough the raw noisy data with the least squared fit method.

A more useful filter for the raw data is a digital filter that cuts off high frequencies. The ideal low pass filter completely eliminates all frequencies above the cutoff frequency, leaves the frequencies below the cutoff frequency unchanged. furthermore the ideal low pass filter has has no phase and amplitude distortion.

The filter used on the VICON position data is a second-order Buttersworth filter. This is a zero-phase digital filter. Butterworth filters have a shorter rise time than critical damped filter, which results in a little overshoot. Since impulsive impute is rarely seen in human movement, the butterworth filter is preferred. [?] The cutoff frequency was set on 10Hz. The filter was applied in Matlab with `filtfilt` which has as numerator and dominator arguments the output from `butter`.

The analog channel data produce a very low amplitude noise, therefore there was no need for further digital filtering. The analog data was resampled for 600 Hz to 120 Hz with `resample`, which uses a 10th order filter by default.

Marker interpolation and construction

At least three cameras need to be revisable to reconstruct the 3D map. Although the optical system uses spacial positioned six cameras, the reflective markers can disappear for a moment in time. The main cause were folds in clothes that cover the markers when making a step. The missing position can be reconstructed by interpolation. The accuracy depends on the length of time of the missing marker, done in Matlab with `interpl1` and as option `spline`. If marker data was missing at beginning or end of the trial, the data had to extrapolated. Unfortunately, this method was not rugged, occasionally resulting in marker positions above 10^5. These marker positions affects the CoM position, which becomes inherently meaningless.

At about half the trail one or more markers missing during an entire trail. These markers were reconstructed making use of the remaining markers. For example, a missing ASIS marker was reconstructed using the x and y position from the other ASIS marker and the z position form the knee marker. This method merely copies marker data and only accurate when there are no rotations.

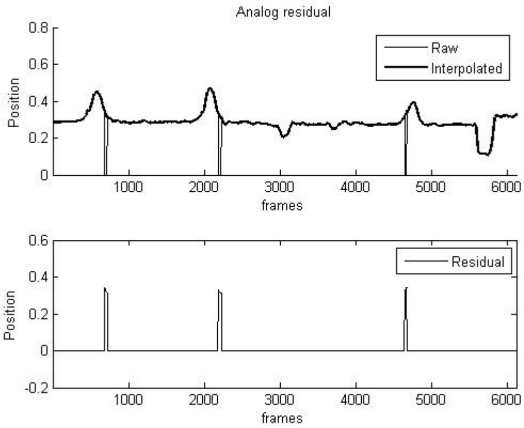


Figure B.1: Interpolation of marker data

More accurate results could be gained with the use of other visible markers during the time the markers is lost. The leg markers LTIB, LTHI are not used for analysis purposes, however they could useful for restructuring a missing knee marker. The same holds for the C7 and CLAV marker.

The x axis was directed at the screen (A/P movements), the y axis was directed to the roof and the z axis to the cupboard (M/L movements), figure 3.5.

B.0.1 coordination systems

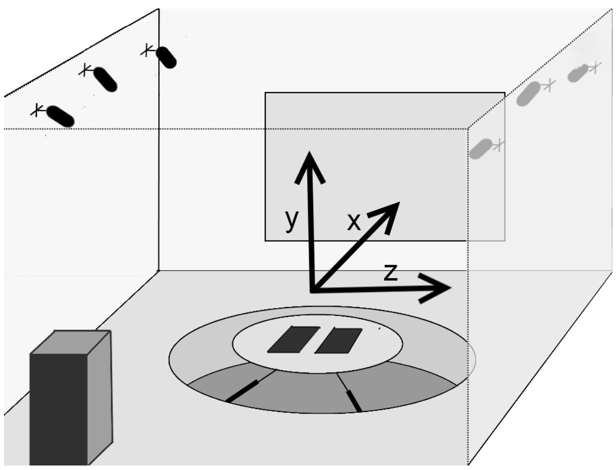


Figure B.2: The used coordination system

The optical and forceplate systems uses other coordination systems. Therefore the markers data was multiplied by the following transformation matrices. The transformation matrix for the optical system:

$$\begin{bmatrix} 0 & 1 & 0 \\ 0 & 0 & 1 \\ 1 & 0 & 0 \end{bmatrix} \begin{bmatrix} 1 \\ 2 \\ 3 \end{bmatrix} = \begin{bmatrix} x \\ y \\ z \end{bmatrix} \tag{B.1}$$

The coordination system for the force platform:

$$\begin{bmatrix} 0 & 1 & 0 \\ 1 & 0 & 0 \\ 0 & 0 & -1 \end{bmatrix} \begin{bmatrix} 1 \\ 2 \\ 3 \end{bmatrix} = \begin{bmatrix} x \\ y \\ z \end{bmatrix} \tag{B.2}$$

Muscles both A/P as M/L direction

Table B.1: Muscles involved in generating the corrective ankle and hip torque in the saggital plane.

Joint	**Movement**	**Muscle**
Ankle	Plantarflexion	Gastrocnemius
		Soleus
	Dorsiflexion	Tibialis anterior
		Extensor hallucis longus
		Extensor digitorum longus
		Peroneus tertius
	Inversion	tibialis antrior
		tibialis posterior
		extensor digitorum longus
		hallucis longus
	Eversion	peroneii
Hip	Flexion	psoas
		iliacus
		Rectus femoris
		Sartorius
	Extension	Gluteus maximus
		Semimembranosus
		Semitendinosus
	Abdduction	
	Abdduction	Adductor brevis
		Adductor longus
		Adductor magnus
		Adductor minimus
		Pectineus
		Gracilis
		Obturator externus

C Auxiliary plots

Analogue force plate data

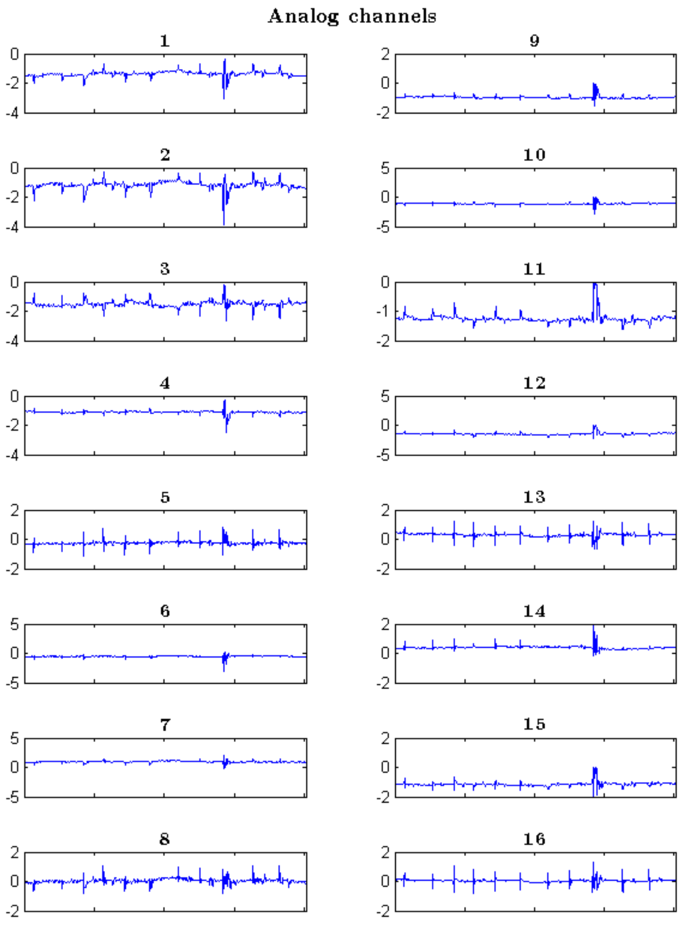


Figure C.1: The raw analog force plate channel data. Mainly used to notice .. dead. channels.

Calibrated force plata data

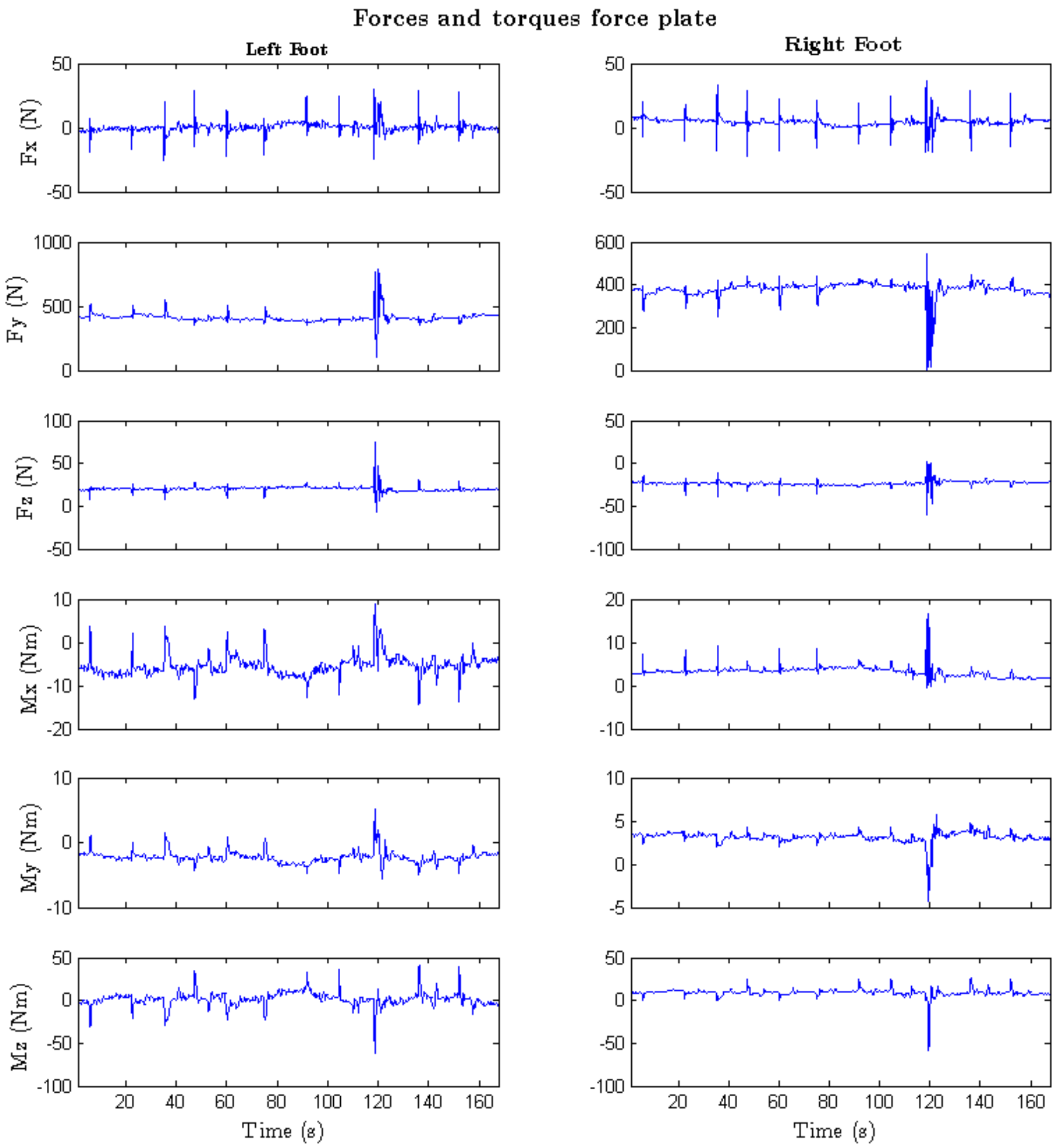


Figure C.2: The calibrated analog force plate channel data. The forces are calibrated with the static calibration matrix and the

CoP x and z direction

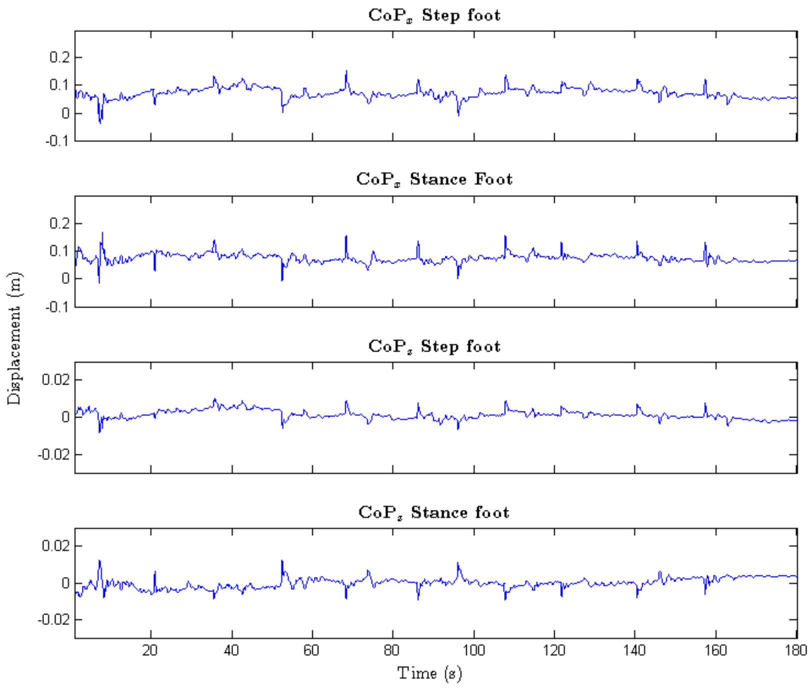


Figure C.3: Plot used to examine the $CoM_{\mathbf{X}}$ and $CoM_{\mathbf{Z}}$. The $CoM_{\mathbf{X}}$ left and right foot are in phase while the $CoM_{\mathbf{Z}}$ left and right are out phase.

Processed trail data

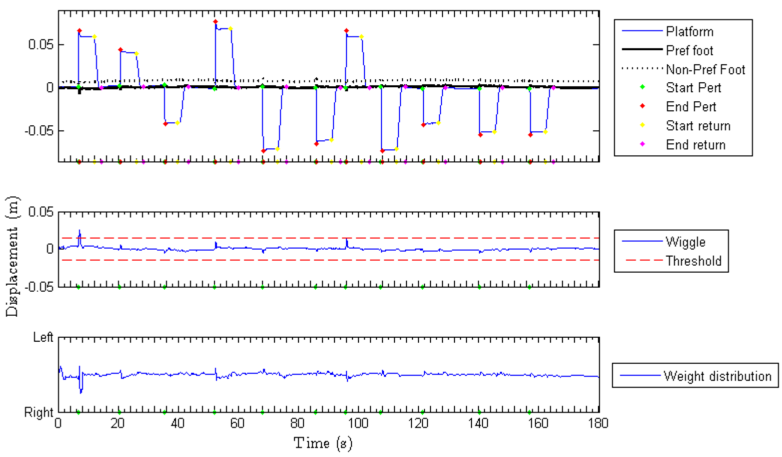


Figure C.4: This plot was used to very the assigned steps. Above: the platform displacement. Middle: the wiggle of the small block foot. Below: the subjects weight-bearing.

Recontructed 3D model

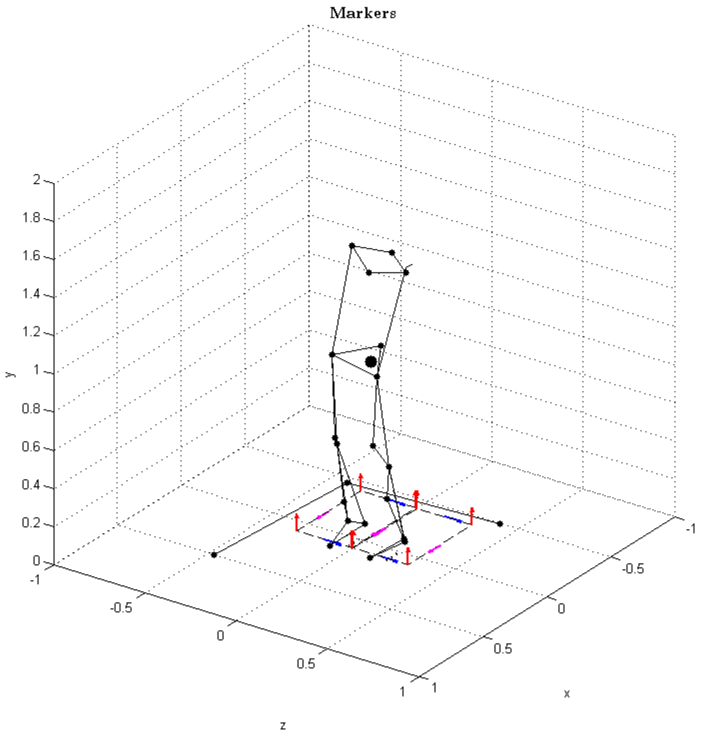


Figure C.5: A 3D reconstruction of the marker data. The small spheres indicate the subjects markers, the large sphere the subjects center of mass. The force plates measured forces are shown as arrows.

Platform perturbation amplitudes

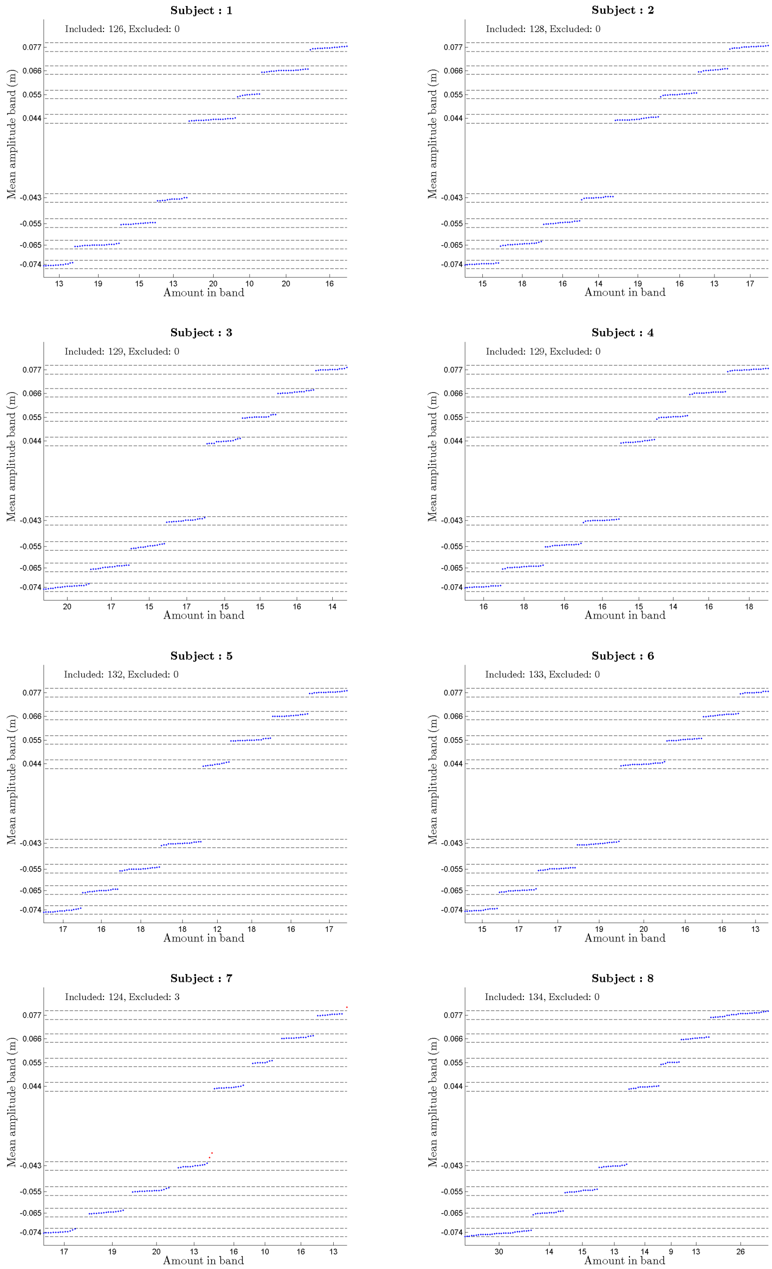
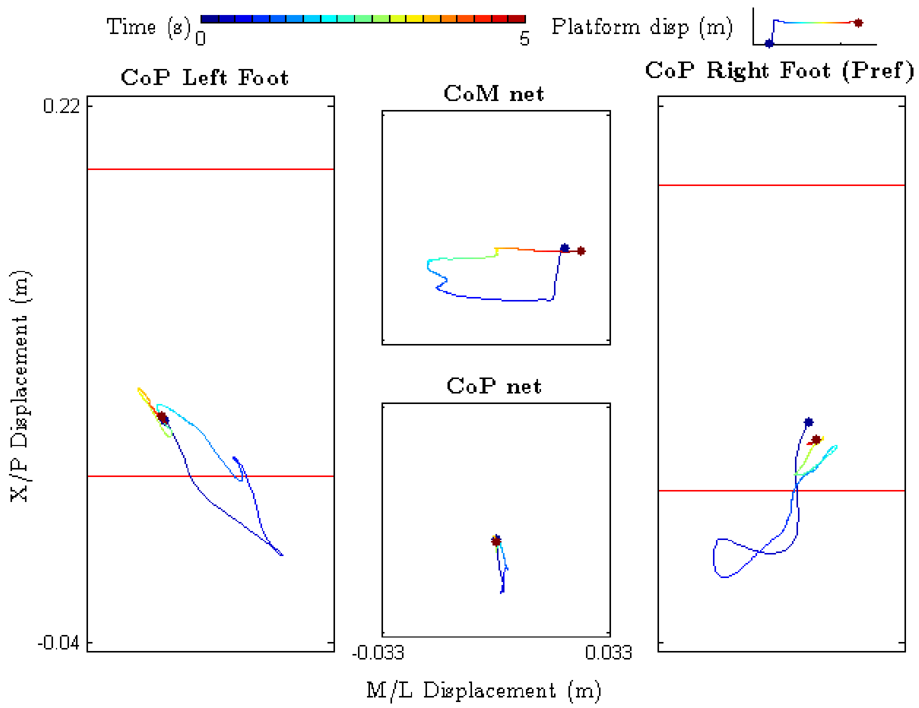


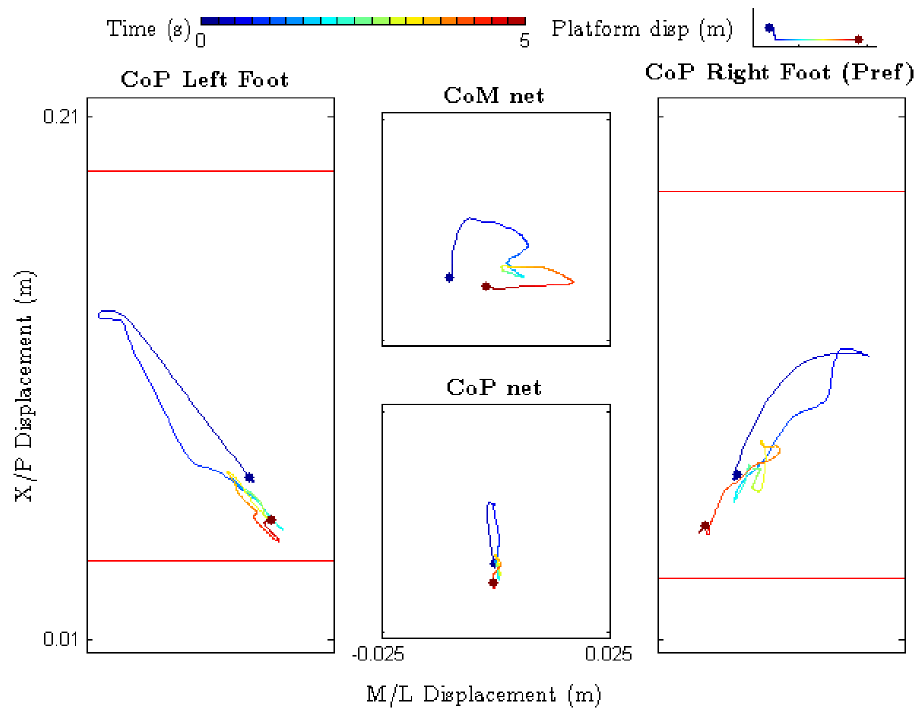
Figure C.6: The included and excluded platform translation amplitudes for all subjects. The blue point indicate an included translation while a red point indicate a excluded translation

COM and Cop trajectories of subject 6

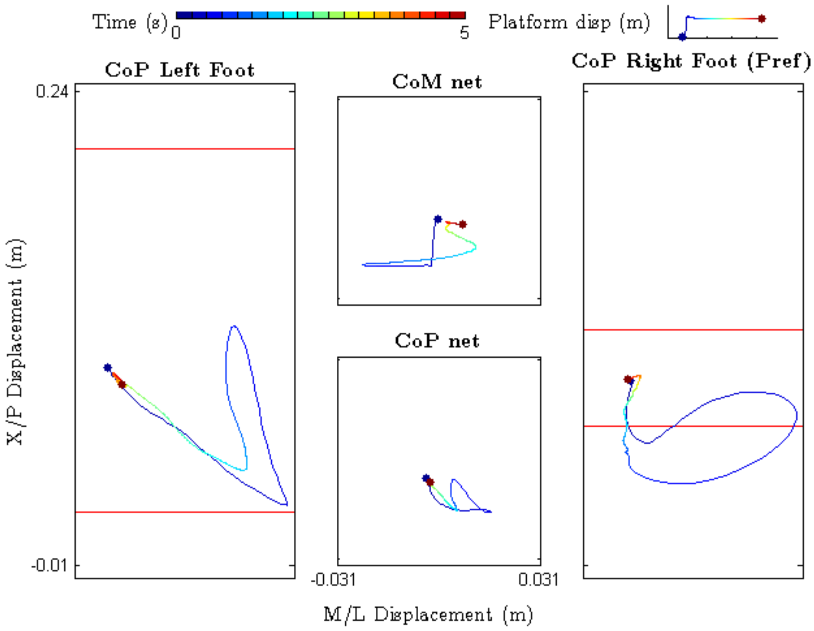
Forward perturbation, without-block



Backward perturbation, without-block



Forward perturbation, with-block



Backward perturbation, with-block

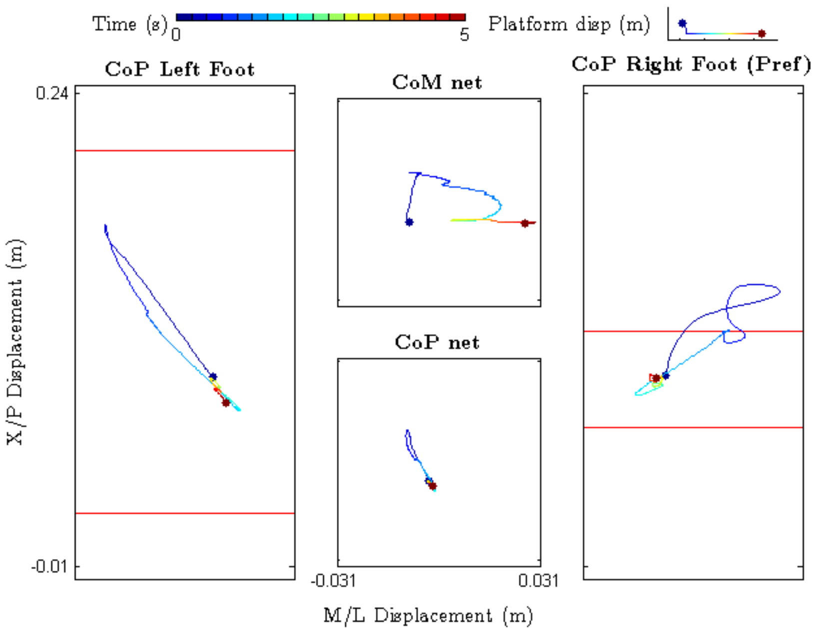
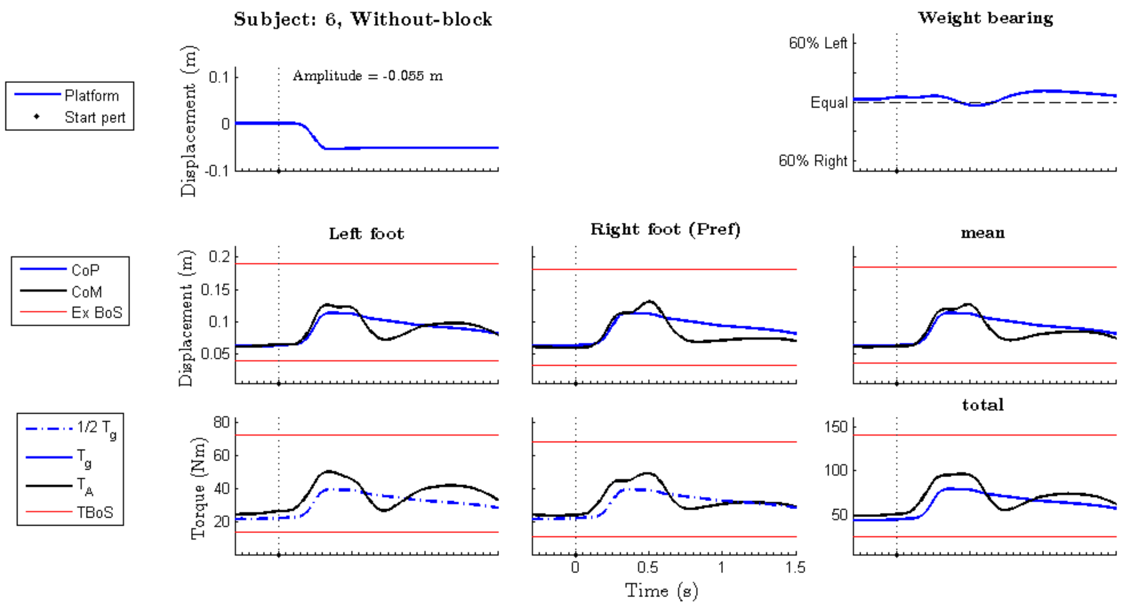
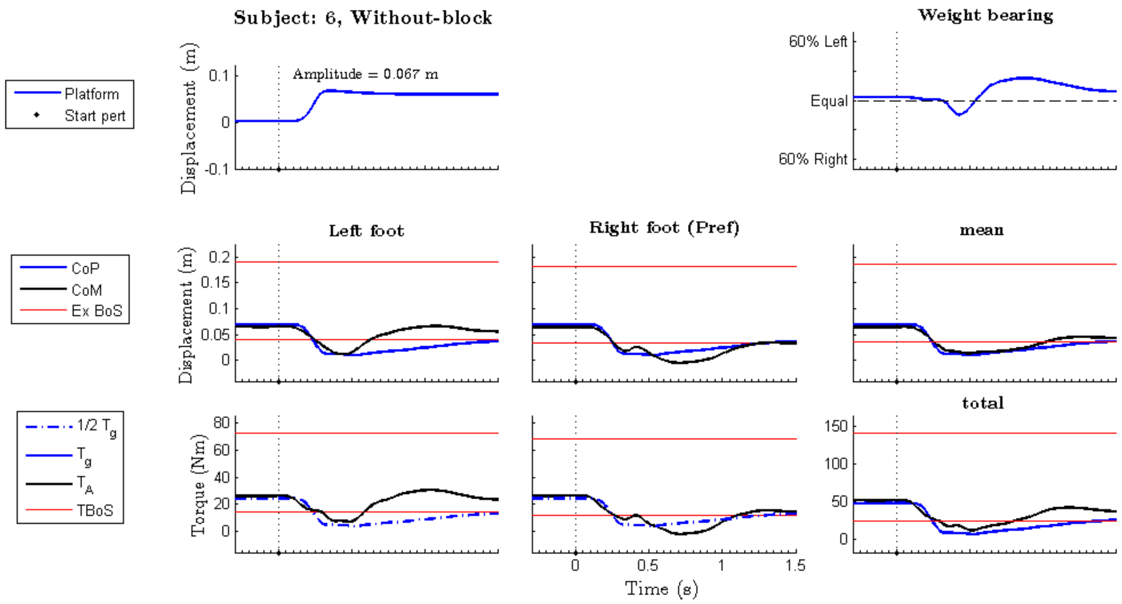
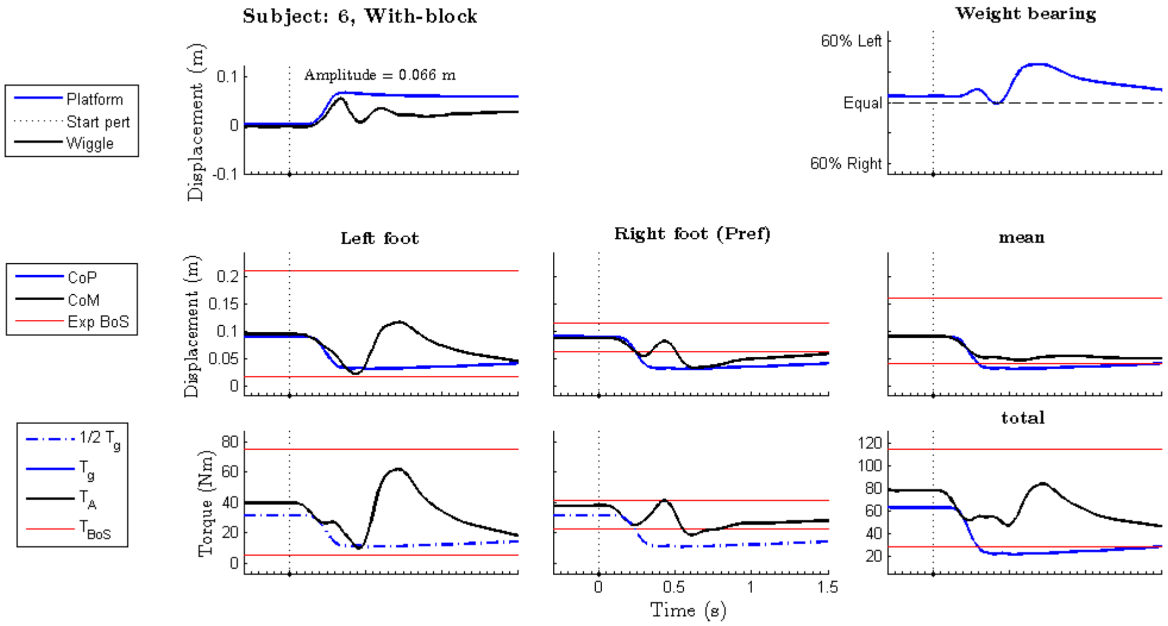


Figure C.7: The CoP and CoM trajectories of subject 6 in x and z direction between the start of the perturbation and the start of the returning motion. Left: CoP left foot. Middle top: CoM. Middle bottom: netto CoP, associated with total torque. Right: CoP right foot. The four figures show the four different conditions: with/without block and forward/backward platform translation, with an amplitude $\pm$ 0.07m.

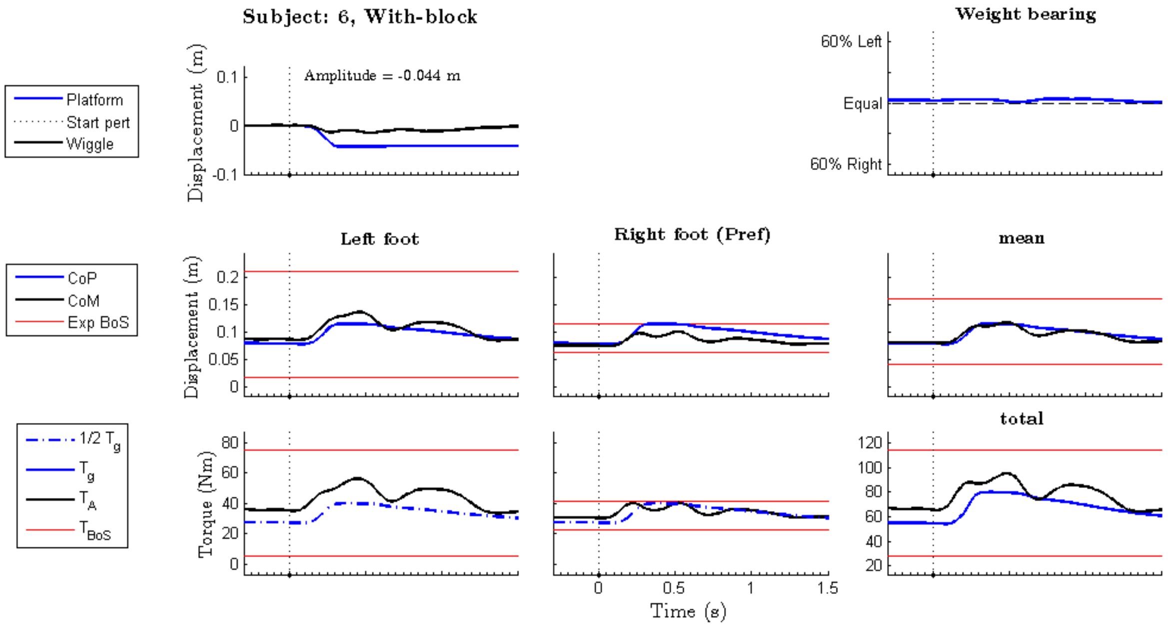
D Torques and CoP







Subject: 6, With-block
Weight bearing
Platform
Start pert
Wiggle
Amplitude = 0.066 m
Displacement (m)
60% Left
Equal
60% Right
Left foot
Right foot (Pref)
mean
CoP
CoM
Exp BoS
1/2 T_g
T_g
T_A
T_{BoS}
Torque (Nm)
total
Time (s)



Subject: 6, With-block
Weight bearing
Platform
Start pert
Wiggle
Amplitude = -0.044 m
Displacement (m)
60% Left
Equal
60% Right
Left foot
Right foot (Pref)
mean
CoP
CoM
Exp BoS
1/2 T_g
T_g
T_A
T_{BoS}
Torque (Nm)
total
Time (s)

E Base of Support

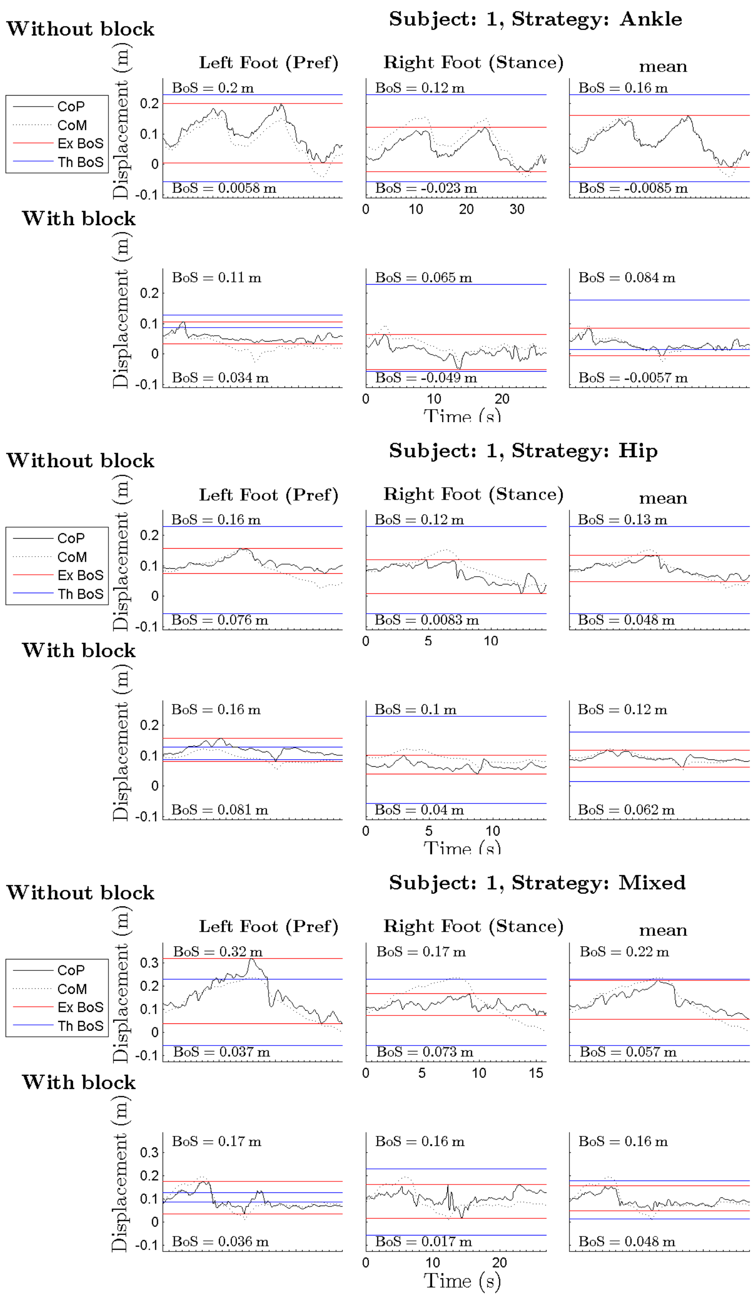


Figure E.1: The left and right foot BoS are shift. This was due to an ankle injury (right).

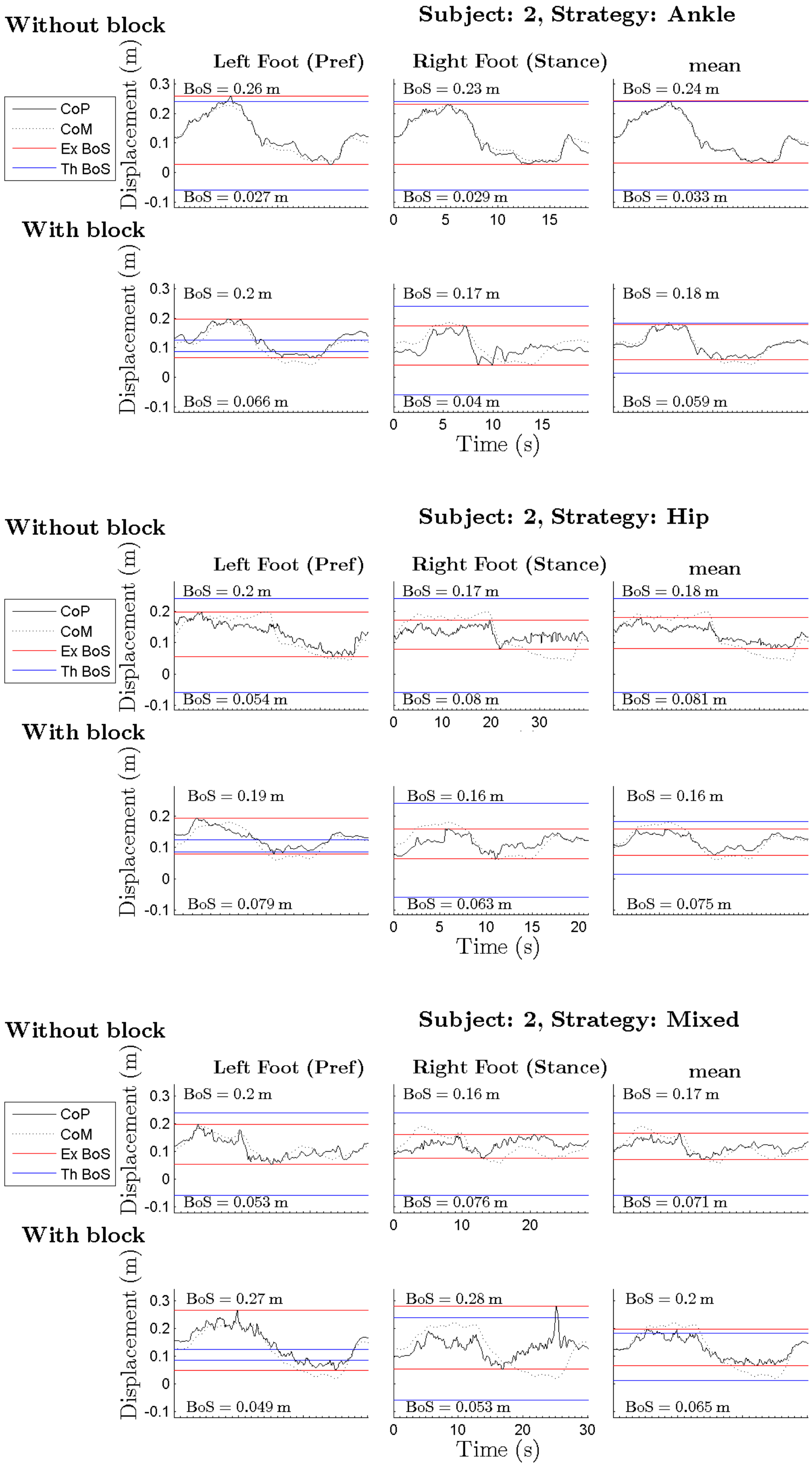
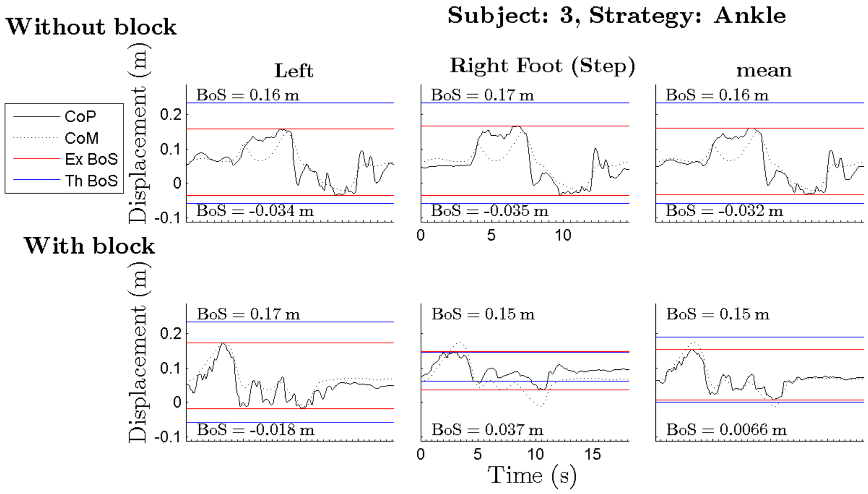
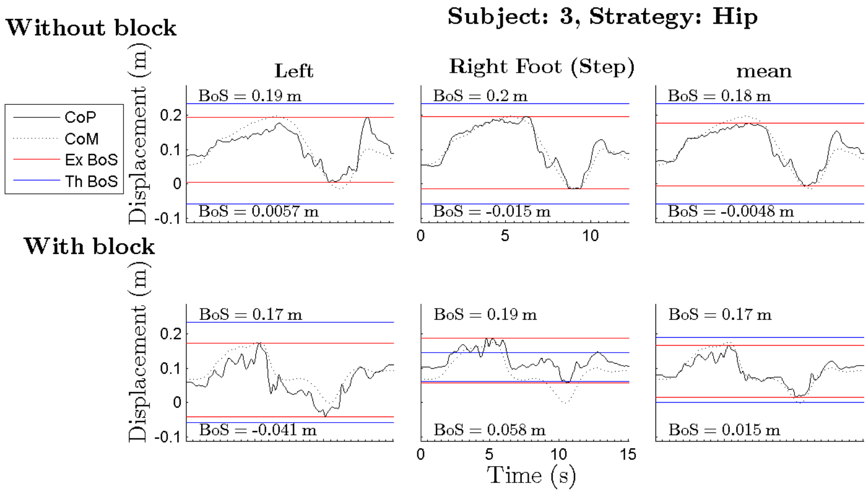
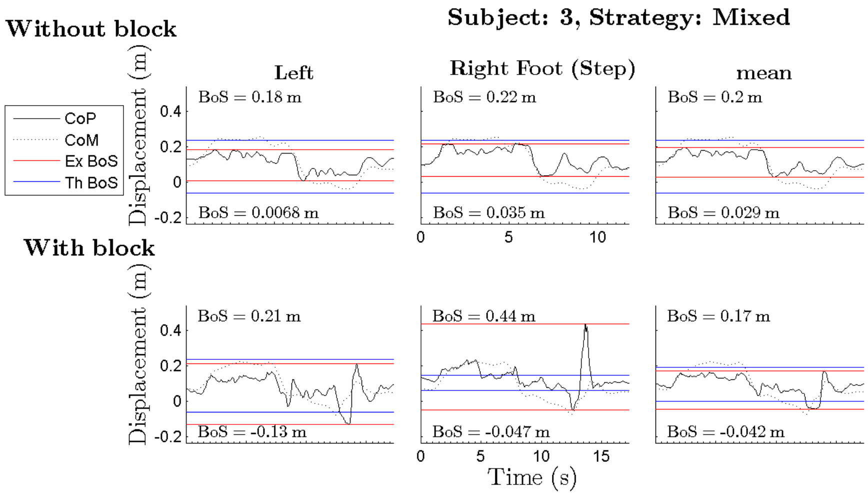


Figure E.2: Top left: the perturbation signal.


Subject: 3, Strategy: Ankle
Without block
With block
Left
Right Foot (Step)
mean
CoP
CoM
Ex BoS
Th BoS
Displacement (m)
BoS = 0.16 m
BoS = -0.034 m
BoS = 0.17 m
BoS = -0.035 m
BoS = 0.16 m
BoS = -0.032 m
BoS = 0.17 m
BoS = -0.018 m
BoS = 0.15 m
BoS = 0.037 m
BoS = 0.15 m
BoS = 0.0066 m
Time (s)


Subject: 3, Strategy: Hip
Without block
With block
Left
Right Foot (Step)
mean
CoP
CoM
Ex BoS
Th BoS
Displacement (m)
BoS = 0.19 m
BoS = 0.0057 m
BoS = 0.2 m
BoS = -0.015 m
BoS = 0.18 m
BoS = -0.0048 m
BoS = 0.17 m
BoS = -0.041 m
BoS = 0.19 m
BoS = 0.058 m
BoS = 0.17 m
BoS = 0.015 m
Time (s)


Subject: 3, Strategy: Mixed
Without block
With block
Left
Right Foot (Step)
mean
CoP
CoM
Ex BoS
Th BoS
Displacement (m)
BoS = 0.18 m
BoS = 0.0068 m
BoS = 0.22 m
BoS = 0.035 m
BoS = 0.2 m
BoS = 0.029 m
BoS = 0.21 m
BoS = -0.13 m
BoS = 0.44 m
BoS = -0.047 m
BoS = 0.17 m
BoS = -0.042 m
Time (s)

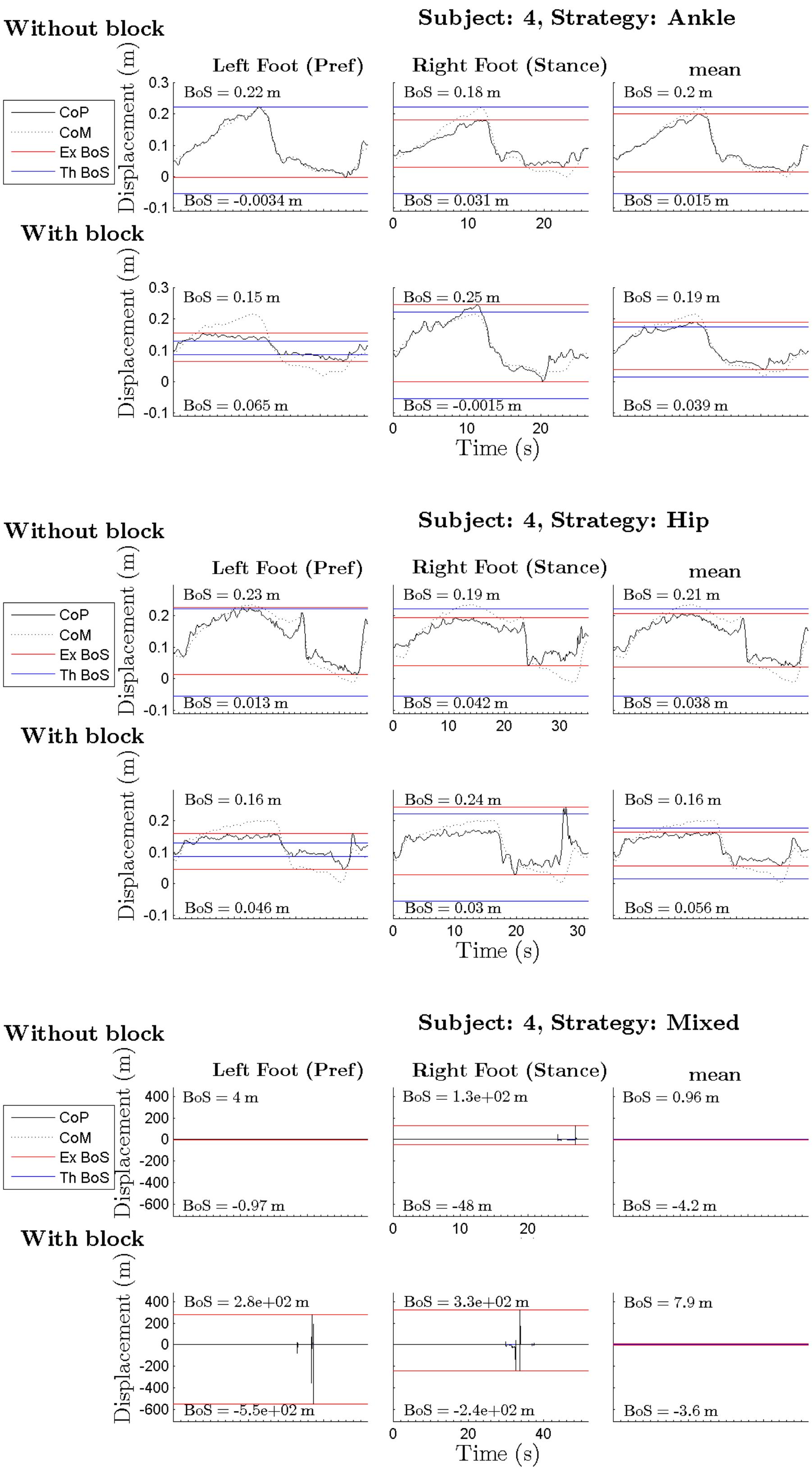
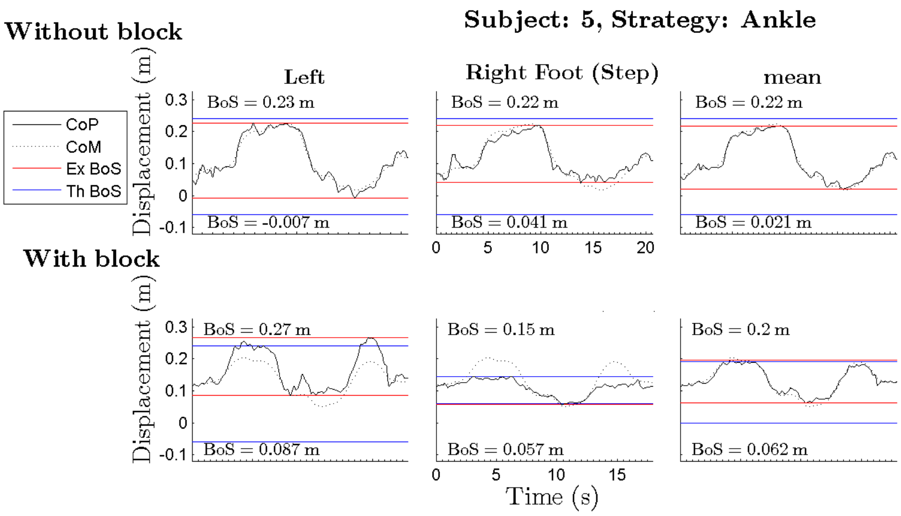
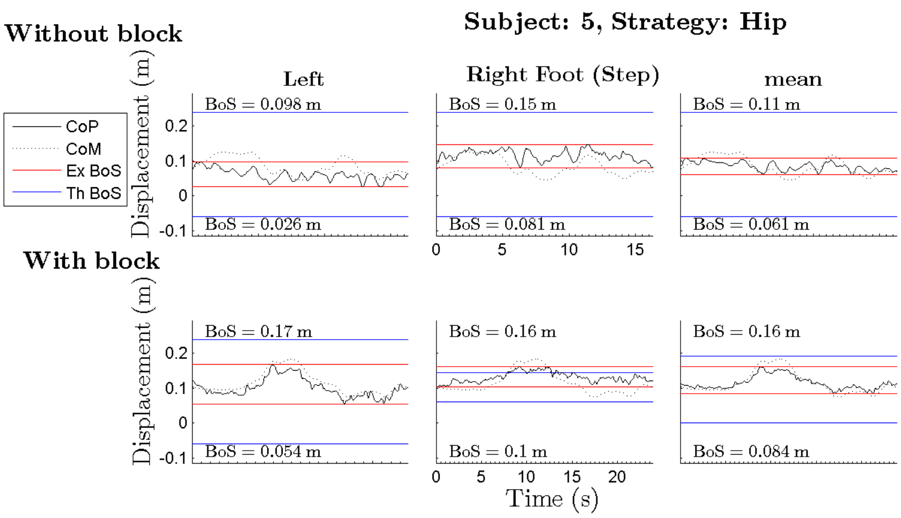
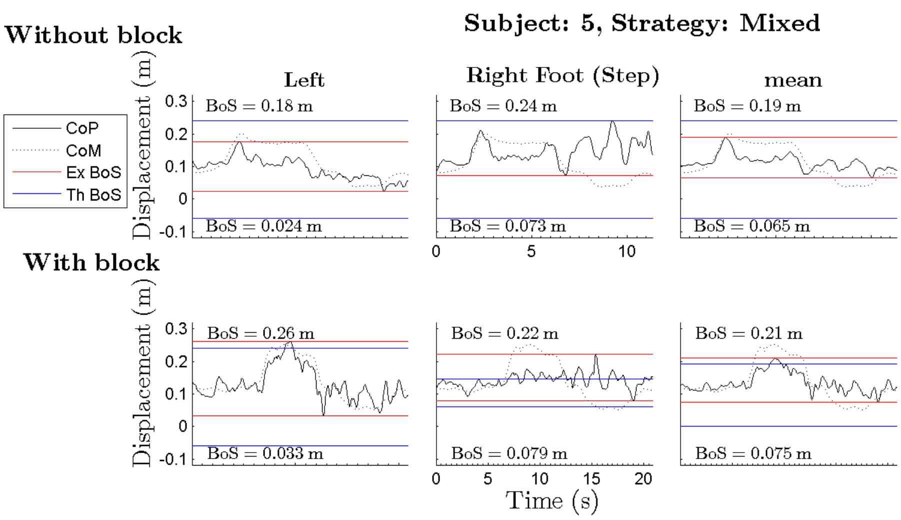
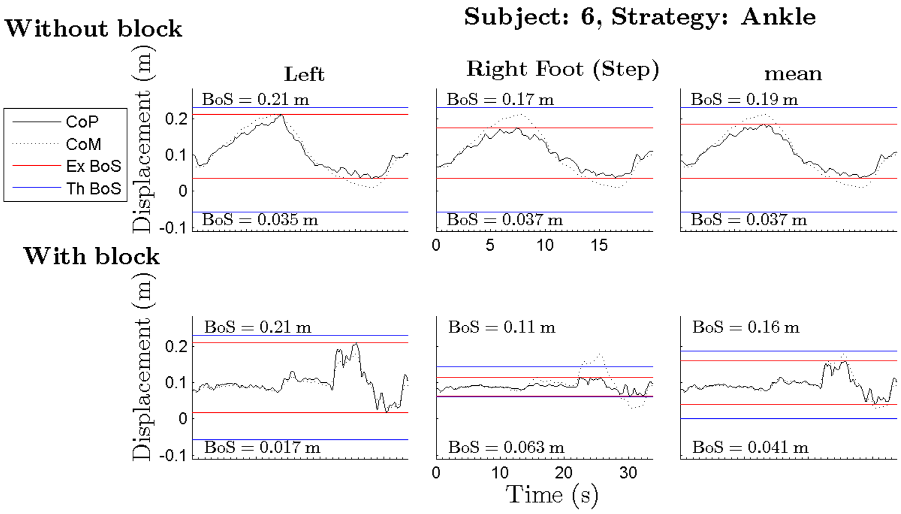


Figure E.3: The extremely high CoP are due to a step. The BoS values can be obtained just before the step.

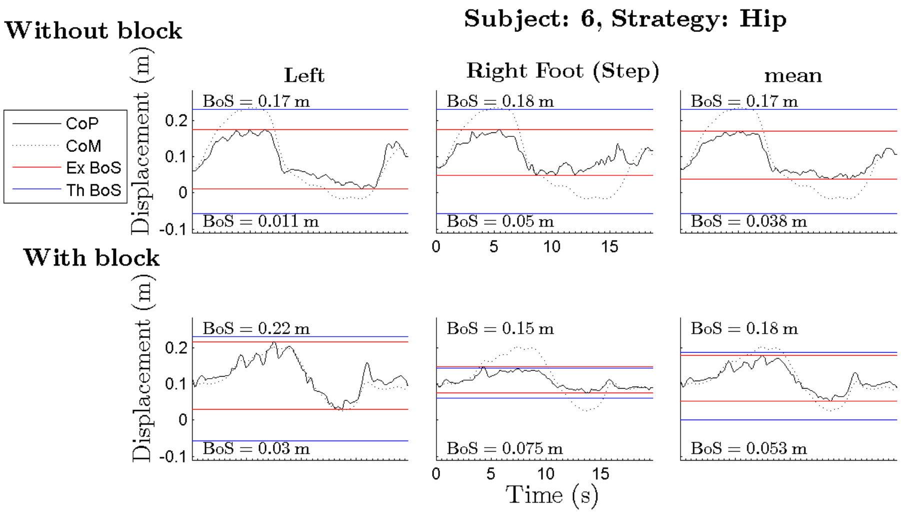

Subject: 5, Strategy: Ankle
Without block
With block
Left
Right Foot (Step)
mean
CoP
CoM
Ex BoS
Th BoS
Displacement (m)
Time (s)
BoS = 0.23 m
BoS = -0.007 m
BoS = 0.22 m
BoS = 0.041 m
BoS = 0.22 m
BoS = 0.021 m
BoS = 0.27 m
BoS = 0.087 m
BoS = 0.15 m
BoS = 0.057 m
BoS = 0.2 m
BoS = 0.062 m


Subject: 5, Strategy: Hip
Without block
With block
Left
Right Foot (Step)
mean
CoP
CoM
Ex BoS
Th BoS
Displacement (m)
Time (s)
BoS = 0.098 m
BoS = 0.026 m
BoS = 0.15 m
BoS = 0.081 m
BoS = 0.11 m
BoS = 0.061 m
BoS = 0.17 m
BoS = 0.054 m
BoS = 0.16 m
BoS = 0.1 m
BoS = 0.16 m
BoS = 0.084 m

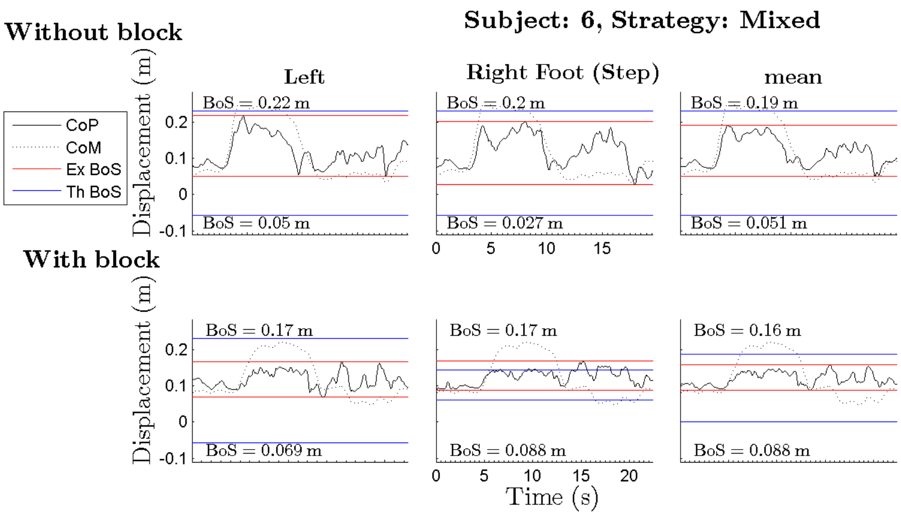

Subject: 5, Strategy: Mixed
Without block
With block
Left
Right Foot (Step)
mean
CoP
CoM
Ex BoS
Th BoS
Displacement (m)
Time (s)
BoS = 0.18 m
BoS = 0.024 m
BoS = 0.24 m
BoS = 0.073 m
BoS = 0.19 m
BoS = 0.065 m
BoS = 0.26 m
BoS = 0.033 m
BoS = 0.22 m
BoS = 0.079 m
BoS = 0.21 m
BoS = 0.075 m



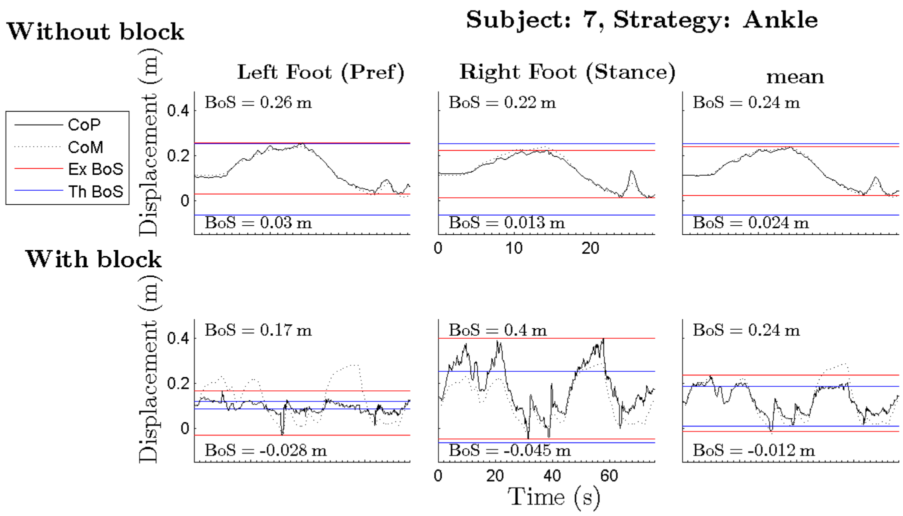
Subject: 6, Strategy: Ankle
Without block
With block
Left
Right Foot (Step)
mean
CoP
CoM
Ex BoS
Th BoS
Displacement (m)
BoS = 0.21 m
BoS = 0.035 m
BoS = 0.17 m
BoS = 0.037 m
BoS = 0.19 m
BoS = 0.037 m
BoS = 0.21 m
BoS = 0.017 m
BoS = 0.11 m
BoS = 0.063 m
BoS = 0.16 m
BoS = 0.041 m
Time (s)

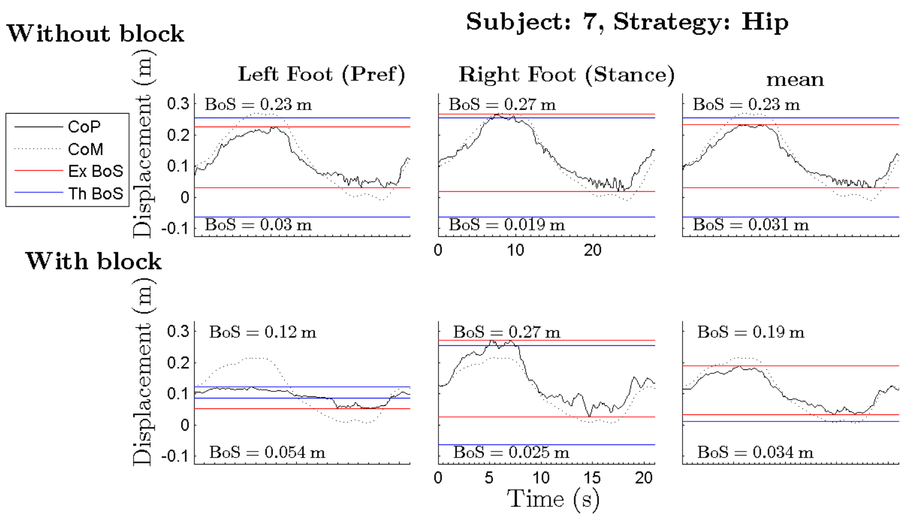


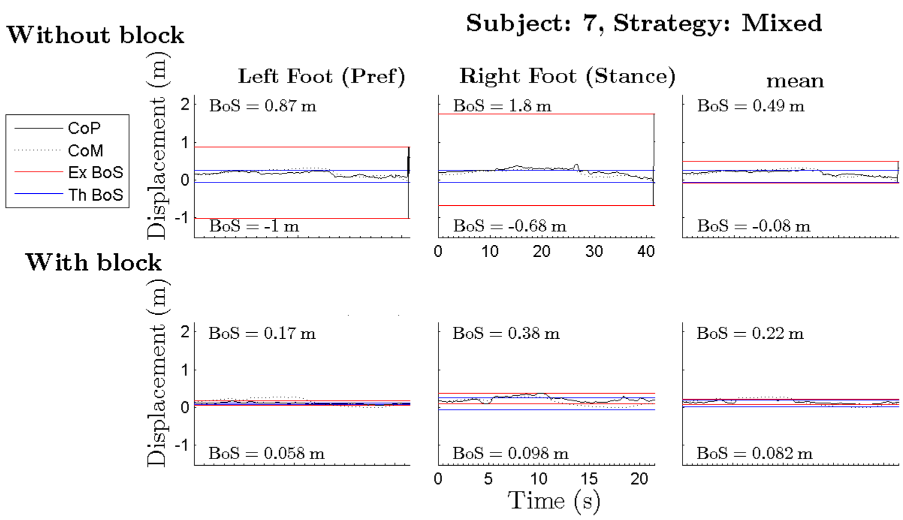
Subject: 6, Strategy: Hip
Without block
With block
Left
Right Foot (Step)
mean
CoP
CoM
Ex BoS
Th BoS
Displacement (m)
BoS = 0.17 m
BoS = 0.011 m
BoS = 0.18 m
BoS = 0.05 m
BoS = 0.17 m
BoS = 0.038 m
BoS = 0.22 m
BoS = 0.03 m
BoS = 0.15 m
BoS = 0.075 m
BoS = 0.18 m
BoS = 0.053 m
Time (s)

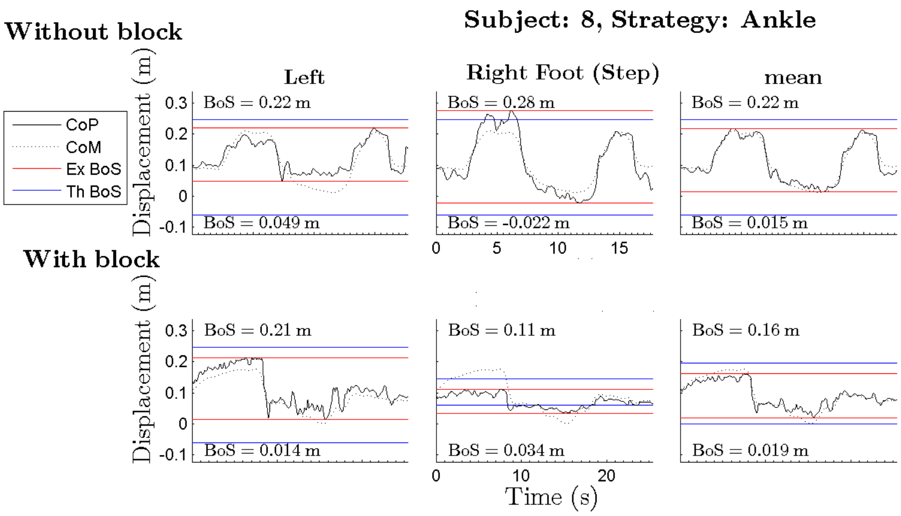


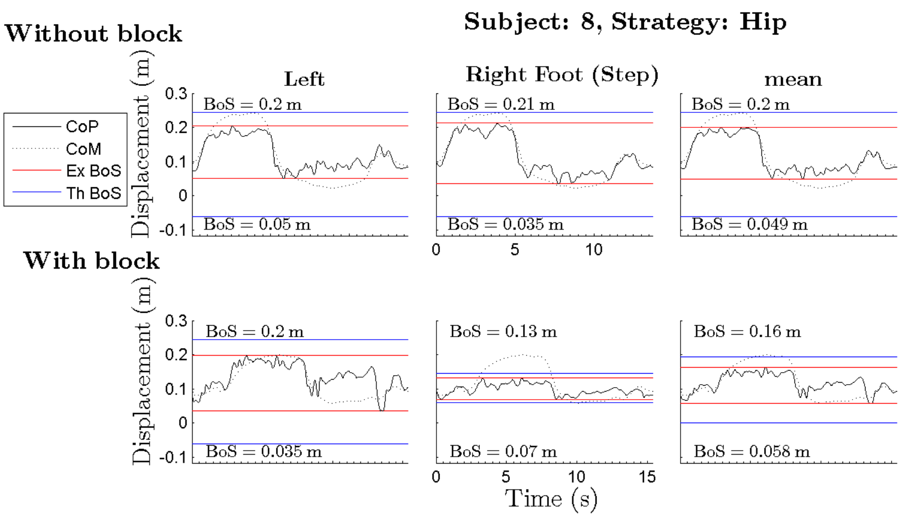
Subject: 6, Strategy: Mixed
Without block
With block
Left
Right Foot (Step)
mean
CoP
CoM
Ex BoS
Th BoS
Displacement (m)
BoS = 0.22 m
BoS = 0.05 m
BoS = 0.2 m
BoS = 0.027 m
BoS = 0.19 m
BoS = 0.051 m
BoS = 0.17 m
BoS = 0.069 m
BoS = 0.17 m
BoS = 0.088 m
BoS = 0.16 m
BoS = 0.088 m
Time (s)

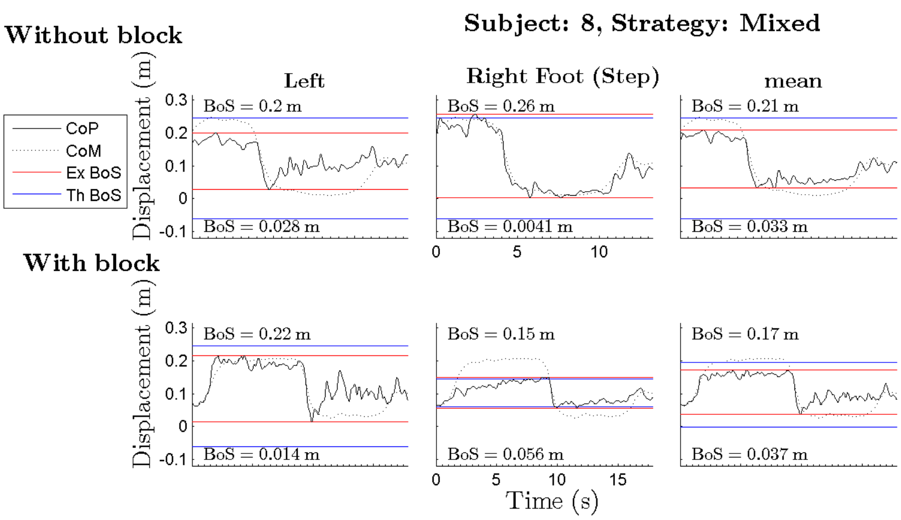

Without block
Subject: 7, Strategy: Ankle
Left Foot (Pref)
Right Foot (Stance)
mean
CoP
CoM
Ex BoS
Th BoS
Displacement (m)
BoS = 0.26 m
BoS = 0.03 m
BoS = 0.22 m
BoS = 0.013 m
BoS = 0.24 m
BoS = 0.024 m
With block
BoS = 0.17 m
BoS = -0.028 m
BoS = 0.4 m
BoS = -0.045 m
BoS = 0.24 m
BoS = -0.012 m
Time (s)


Without block
Subject: 7, Strategy: Hip
Left Foot (Pref)
Right Foot (Stance)
mean
CoP
CoM
Ex BoS
Th BoS
Displacement (m)
BoS = 0.23 m
BoS = 0.03 m
BoS = 0.27 m
BoS = 0.019 m
BoS = 0.23 m
BoS = 0.031 m
With block
BoS = 0.12 m
BoS = 0.054 m
BoS = 0.27 m
BoS = 0.025 m
BoS = 0.19 m
BoS = 0.034 m
Time (s)

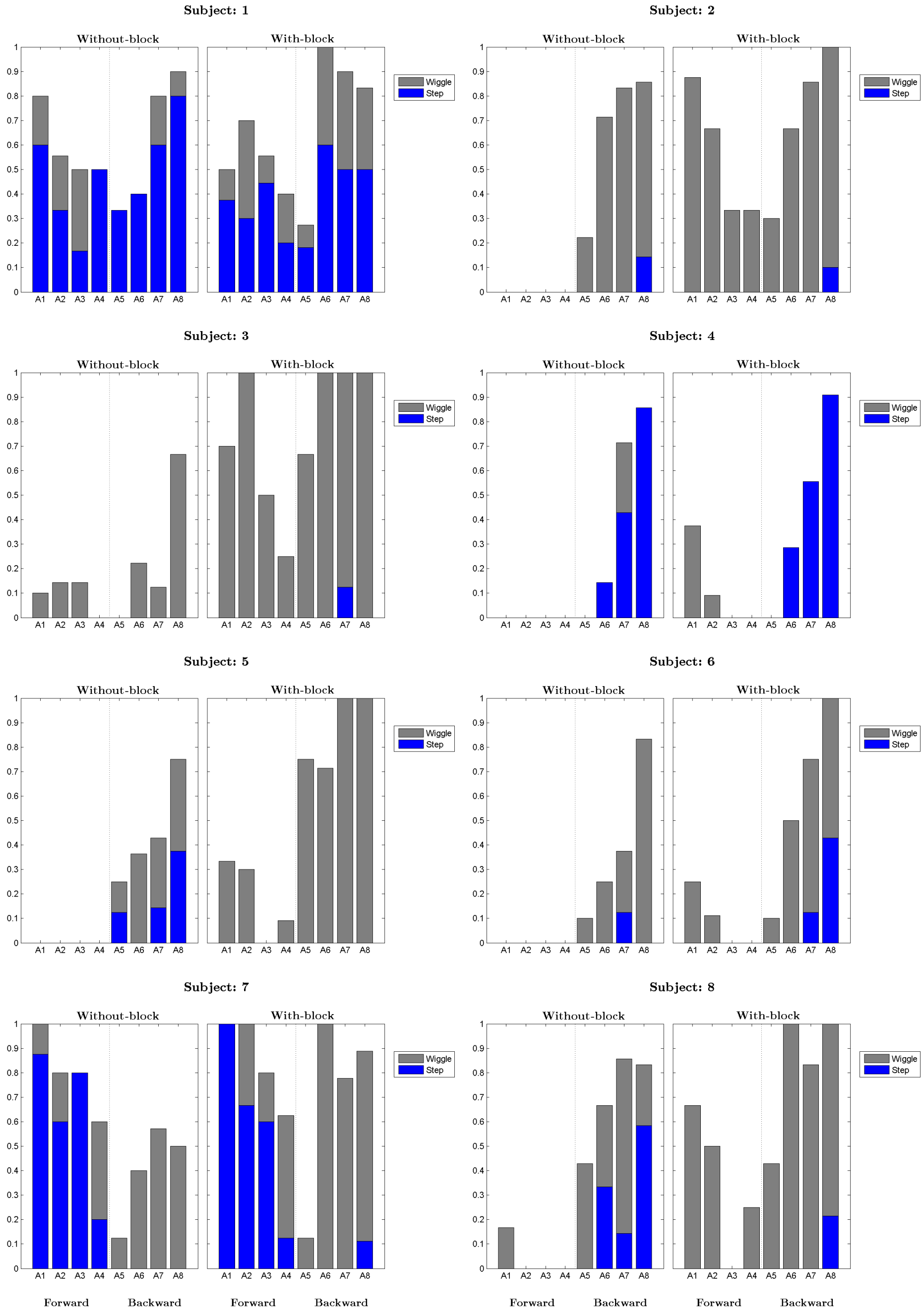

Without block
Subject: 7, Strategy: Mixed
Left Foot (Pref)
Right Foot (Stance)
mean
CoP
CoM
Ex BoS
Th BoS
Displacement (m)
BoS = 0.87 m
BoS = -1 m
BoS = 1.8 m
BoS = -0.68 m
BoS = 0.49 m
BoS = -0.08 m
With block
BoS = 0.17 m
BoS = 0.058 m
BoS = 0.38 m
BoS = 0.098 m
BoS = 0.22 m
BoS = 0.082 m
Time (s)


Subject: 8, Strategy: Ankle
Without block
With block
Left
Right Foot (Step)
mean
CoP
CoM
Ex BoS
Th BoS
Displacement (m)
BoS = 0.22 m
BoS = 0.049 m
BoS = 0.28 m
BoS = -0.022 m
BoS = 0.22 m
BoS = 0.015 m
BoS = 0.21 m
BoS = 0.014 m
BoS = 0.11 m
BoS = 0.034 m
BoS = 0.16 m
BoS = 0.019 m
Time (s)

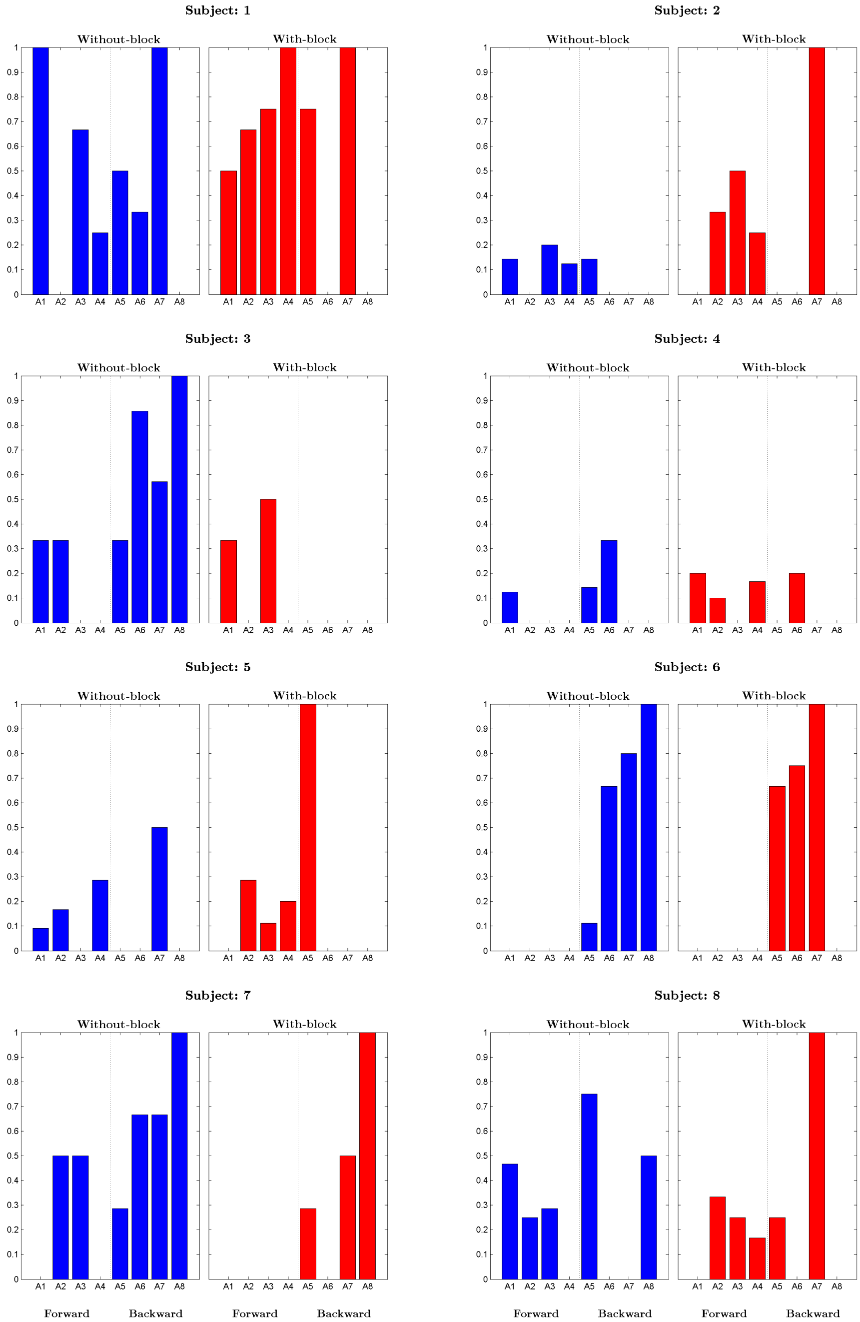

Subject: 8, Strategy: Hip
Without block
With block
Left
Right Foot (Step)
mean
CoP
CoM
Ex BoS
Th BoS
Displacement (m)
BoS = 0.2 m
BoS = 0.05 m
BoS = 0.21 m
BoS = 0.035 m
BoS = 0.2 m
BoS = 0.049 m
BoS = 0.2 m
BoS = 0.035 m
BoS = 0.13 m
BoS = 0.07 m
BoS = 0.16 m
BoS = 0.058 m
Time (s)


Subject: 8, Strategy: Mixed
Without block
With block
Left
Right Foot (Step)
mean
CoP
CoM
Ex BoS
Th BoS
Displacement (m)
BoS = 0.2 m
BoS = 0.028 m
BoS = 0.26 m
BoS = 0.0041 m
BoS = 0.21 m
BoS = 0.033 m
BoS = 0.22 m
BoS = 0.014 m
BoS = 0.15 m
BoS = 0.056 m
BoS = 0.17 m
BoS = 0.037 m
Time (s)

F Step bar charts



G Weight-bearing shift bar charts



H Protocol

Protocol

M.A. van Hirtum

July 5, 2012

NR:........ **Date: ....**
Sex: ....
Age ....

1 Accessories

- Healthy subject
- Scale
- Tapeline
- Pen
- Paper
- Safety harness
- VICON system
- Markers
- Force platform

2 Preparations

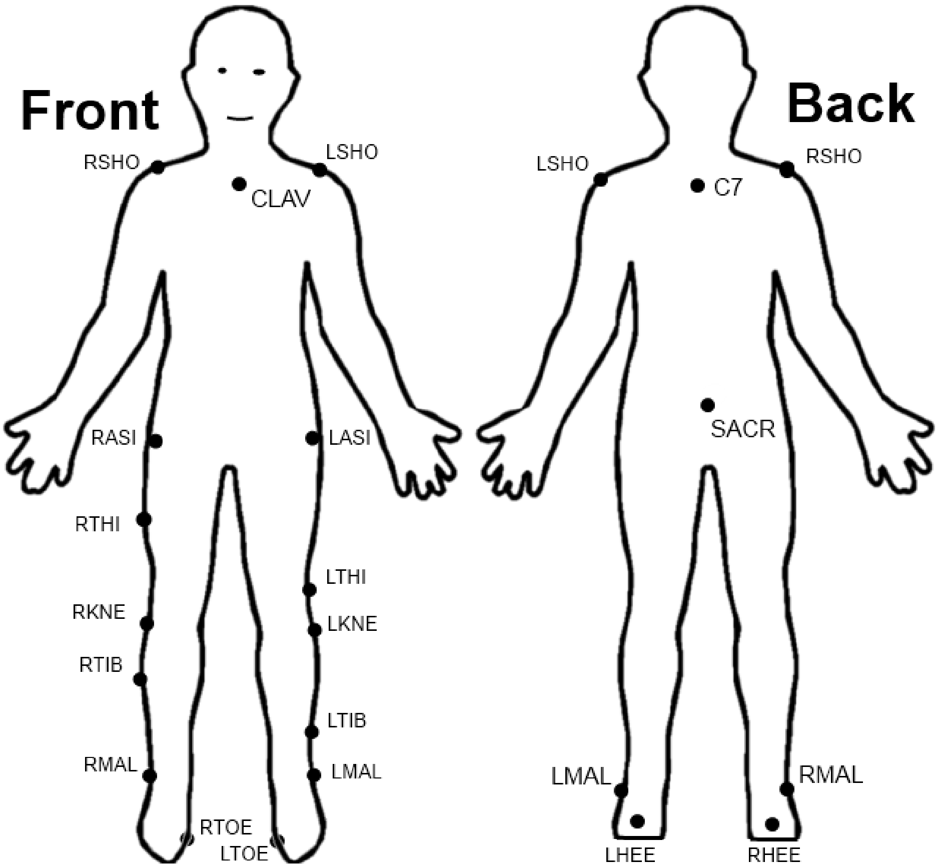
[ ] Start VICON Workstation computer
[ ] Start the VICON Data station computers
[ ] Start the force feedback notebook
[ ] Tape the platform where feet should be placed
[ ] Open database
[ ] Open Simulink model
[ ] Calibrate cameras
[ ] Calibrate force sensors
[ ] Start weight feedback
[ ] Perform a test trail with platform translations without subject

3 Prepare subject

[ ] Ask subject to change clothes and put of socks			
[ ] Explain protocol and methods			
[ ] Ask subject to sign informed consent			
[ ] Measure subjects weight with scale	Weight:	....	kg
[ ] Measure subjects height with tapeline	Length:	....	cm
[ ] Measure subjects leg length with tapeline	Leg length:	....	cm
[ ] Determine subject favourite leg	Trail 1:	[L] [R]	
	Trail 2:	[L] [R]	
	Trail 3:	[L] [R]	
	Favourite leg:	[L] [R]	

4 Marker placement

[ ] Attach all 19 markers on the subject
[ ] Place 3 markers on the platform



5 Experiment trials

Trail preparations

[ ] Make sure the emergency stop button is within reach
[ ] Fasten safety harness
[ ] Instruct subject to place feet on the platform
against the tapeline with ± 9° rotation

Trail instructions

[ ] Instruct subjects :
'Maintain balance while keeping equal weight distribution'
'Fold hands in front of the chest'
'Eyes open, fixate on the grey screen'
'Try to keep the feet on the platform or make a step'
'Try to keep the hip straight'
[ ] Assess:
If arms are folded in front of the chest
Heel/toe rises
Hip movement
Weight distribution (on force feedback notebook)
Eye fixation

- Write down the trail sequence for record perturbation trails in 'Seq'.
- When 'Condition' is 'Wooden block': attach small wooden block under preferred leg foot and large block under the other foot
- When 'Action' is [0.0x 0.0x 0.0x 0.0x 0.0x]: Run Simulink model with indicated perturb magnitudes
  'Slowly bend forward until the a step has to be made and the the same backwards'
- When 'Action' is 'Bend':
  'Slowly bend forward until the point a step has to be made,
  do the same backwards'. Instruct subject to use joints as indicated.
- Repeat trail as many times as indicated in 'Rep'
- Let the subject recover for 1 minute between trails
- Capture and save measurements
- Check for visibility markers

Seq	Check	Rep	Action	Condition
Habituation trails				
	[ ]	3	[0.04 0.04 0.05 0.05 0.05]	With blocks
	[ ]	3	[0.04 0.05 0.06 0.07 0.08]	With blocks
	[ ]	3	[0.04 0.05 0.06 0.07 0.08]	Wooden blocks
Record trails				
	[ ]	1	Bend only ankle	Without blocks
	[ ]	1	Bend only hip	Without blocks
	[ ]	1	Bend hip and ankle	Without blocks
	[ ]	1	Bend only ankle	With blocks
	[ ]	1	Bend only hip	With blocks
	[ ]	1	Bend hip and ankle	With blocks
[ ]	[ ]	3	[0.04 0.05 0.06 0.07 0.08]	Without blocks
[ ]	[ ]	3	[0.04 0.05 0.06 0.07 0.08]	Without blocks
[ ]	[ ]	3	[0.04 0.05 0.06 0.07 0.08]	Wooden blocks
[ ]	[ ]	3	[0.04 0.05 0.06 0.07 0.08]	Wooden blocks

One perturbation trail has 20 transient translation. The translations are randomized in A/P direction, randomized in magnitude (amplitudes indicated in 'Action') and randomized in resting time (3 or 12 seconds).

Viconname = Date+Viconnumber
Date = jjjjmmdd
Viconnumber = 1xx for measurment, 0xx for calibrations

6 Closing

[ ] Detach safety harness
[ ] Detach markers
[ ] Make sure all measurements are properly saved
[ ] Open curtains
[ ] Shut down VICON
[ ] Shut down force platform
[ ] Shut down force feedback computer

7 Measurement form

Name: Date:

Start time:
Stop time:

Vicon Number	**With Block**	**Signal**	**Remarks**

Vicon Number	**With Block**	**Signal**	**Remarks**

Informed consent

The undersigned declares to be aware of the purpose, method and procedure of the experiment and is informed about the possible risks associated with the experiment.
Additionally, all questions are sufficiently answered by the exterminator. Furthermore, it is clear that participation is entirely voluntary, and discontinuing is possible at any time.

Name: Date:

Signature:

I Manual

Operation manual

M.A. van Hirtum

April 21, 2012

NR: ....
Sex: ....
Age ....

Date: ....

1 VICON

1.1 Start up

- Start the VICON Workstation computer
- Start the VICON Data station computers
- Turn on screens
- Open 'VICON workstation'
- Activate camera by clicking 'System - Start'

1.2 Camera calibration

- Close curtains
- Hang safety harness away
- Lay L frame (in the cupboard) on left front corner of the platform
- Warn others in the lab before turning the off lights
- Turn off lights
- Check 'Live Monitors' for unwanted reflections
- Start static calibration (stops automatically)
- Remove L-frame
- Place marker wand on the platform
- Set 'Reference object': 'Example.cro'
- Start dynamic calibration
- Wave slowly with the wand, covering whole area
- Stop dynamic calibration
- Check results: 'Mean Residual' < 2mm, 'Static reproducibility' < 2%
- Click 'Accept'
- Remove wand
- Turn on lights

2 Marker placement

Trunk

LSHO	Left shoulder	On the shoulder bone juts out the most
RSHO	Right shoulder	Idem
C7	Cervicale 7	Instruct subject to bend his head down. C7 will be jutting out, along the spinal column at the at of the neck
CLAV	Clavicle	Placed between the to collar bones
SACR	Sacrum, wand	Between posterior superior iliac spines, S1, first sacral prominence
LASI	Left ASIS	On the lateral pelvis bone area that juts out the most
RASI	Right ASIS	Idem

Legs

LTHI	Left tight	In line with knee and
RTHI	Right tight	Idem
LKNE	Left knee	On the lateral epicondyle
RKNE	Right knee	Idem
LLEG	Left tibia	Left shin, asymmetrically to RLEG
RLEG	Right tibia	Right shin, asymmetrically to LLEG

Feet

LMAL	Left malleolus	Lateral on the ankle were the bone juts out
RMAL	Right malleolus	Idem
LMT5	Left pinky toe	On the bone right before pinky toe starts
RMT5	Right pinky toe	Idem
LTOE	Left big toe	On the bone right before big toe start
RTOE	Right big toe	Idem
LHEE	Left heel	At same height as big toe marker
RHEE	Right heel	Idem

Platform

PLFR	Platform left front
PLBA	Platform left back
PRBA	Platform right back

Segments:

HEAD	LFHD	LFHD	UPHD				
TRUNK	RSHO	LSHO	SACR	C7	CLAV	LASI	RASI
LEFTUPPERLEG	LASI	LTHI	LKNE				
RIGHTUPPERLEG	RASI	RTHI	RKNE				
LEFTLOWERLEG	LKNE	LTIB	LMAL				
RIGHTLOWERLEG	RKNE	RTIB	RMAL				
RIGHTFOOT	RMAL	RHEE	RTOE	RMT5			
LEFTFOOT	LMAL	LHEE	LTOE	LMT5			
PLATFORM	PLFR	PLBA	PRBA				

General rules of thumb:

- Subject should ware tight fitting clothes and gym shorts
- Markers should stay stationary when distal limps move
- Markers should be placed as close to the bones as possible
- When a marker falls off, the static subject has the be recalibrated

2.1 Static subject calibration

- Click ‘Trial - Capture’
- Enter ‘Time (seconds)’ : 10
- Instruct subject to stand with hands folded in front the chest
- Click capture (stops automatically)
- Click ‘Save’

2.2 Labelling

- Open static trial session
- Click ‘Trail - marker’
- Open D: \vicon\Michel\modelmichel.mkr
- Click the right marker, subsequently the right label
- Drag the time bar and check if al labels are marked
- Press F10 for unlabelled markers
- Drag the time bar and check if al labels are marked

2.3 Auto labelling

- Click ‘Autolabel - define subjects’
- Click ‘Trail - create autolabel calibration’
- Click ‘Trail - options’
- Click ‘Trail - Auto labelled’
- Drag time bar and check if everything is labelled correctly

2.4 Database

- Open database, Location: D: \vicon\Michel,
  Name: BA, Select: ‘Clinic Template.eni’
- Click ‘Create’
- ● Click green button for new classification, name: BA Michel
- ● Click yellow button for new patient, name: name subject
- ● Click grey for new session, name: current date
- Double click on session to select

2.5 Closing

- Detach safety harness
- Make sure all measurements are properly saved
- Open curtains
- Shut down VICON

3 Force platform

3.1 Start up

- Plug in force platform
- Turn on screen
- Follow procedure as indicated on the list
- Tape the platform where feet should be placed
- Start Matlab and simulink model:
  'D:\PlatformAansturing\Simulink\michel\modelPlatform'

Calibrate force sensors

- Double click on the session
- Click 'System - Analog'
- Select channels 1-6
- Set sample frequency at 120 Hz
- Click 'System - Forceplate setup', position should be: ...
- Double click on the session
- Click 'System - Calibrate analog zero levels'
- Click 'Force plate'
- Click 'Calibrate'
- Check what is calibrated
- Click 'OK'
- Click 'Trail - Capture' , 10 seconds
- Click 'Graph - Analog'
- Average levels should be $< |5|$ and std ¡ 50
- Check 'Graph - Analog'

Simulink

- Load matlab script from usb stick
  'D:\PlatformAansturing\Simulink\michel\modelPlatform'
- Start Matlab and Simulink model:
- Wipe workspace
- Double click 'program platform movement2' to alter disturbance magnitude
- Run model
- Click on 'increase status' for '..'
- Click on 'increase status' for 'Neutral'
- Click on 'increase status' for 'On'
- Click 'Manual switch' to '1' to start perturbation
- Click 'Manual switch' to '0' to stop perturbation
- Click on 'increase status' for '..'
- Click on 'increase status' for 'Neutral'
- Click on 'increase status' for 'Off'
- Wait for pomps unload!
- Click 'Stop simulation'
- Click 'program platform movement2 - edit mask - initialization' to alter signal script

3.2 Closing

- Detach safety harness
- Press Alt+x on the force platform computer
- Unplug force platform