The influence of the velocity of support surface rotations on sensory reweighting during standing balance.

MASTER THESIS

Emily Zoetbrood

FACULTY OF ENGINEERING TECHNOLOGY
Department of Biomechanical Engineering

December 24, 2020

EXAMINATION COMMITTEE
Dr. ir. Alfred Schouten
Ronald van ‘t veld, Msc
Dr. ir. Bert-Jan van Beijnum

DOCUMENT NUMBER
BW-777

Preface

In front of you is the report 'The influence of the velocity of support surface rotations on sensory reweighting during standing balance'. This report is the master thesis of Emily Zoetbrood, written to graduate the master Biomedical Engineering in the Neural Motor Systems track at the University of Twente. From the start of March till the end of November 2020 I conducted research, designed a protocol, performed measurements, analysed the measurements and wrote this report. This study was documented as an article, starting after this preface. The initial goal of this study was quite different than the final result, partly due to the global Corona pandemic. The influences of this pandemic on this study will be elaborated in the section below.

First of all, I would like to thank my supervisors for their guidance, constant feedback and support during this study. Secondly, I would like to thank all my subjects for their time and effort. Next, I would like to thank M. A. Ouwens for his help with Labview. He was always there to answer any of my questions and help me to debug the program. I would also like to thank Louise for her daily online motivational support and the lunch breaks. Last but not least, I would like to thank my dear friend Daan: he made room for the mBAP at his house when the university closed so I could continue researching. He also spend many hours standing on the mBAP, so I could test and improve the protocol. He also supported me throughout this study.

I hope you enjoy reading.

Emily Zoetbrood
16 November 2020

The effect of the Corona pandemic

This study started in the first week of March 2020, before the global Corona pandemic reached the Netherlands. This pandemic had a lot of consequences for this research because the initial goal could not be achieved. The initial goal was to develop and evaluate a protocol to measure the proprioceptive, visual and vestibular contributions to balance control in healthy participants. The measures and restrictions of the Dutch government or the University of Twente guided this research into a different direction or gave limitations on the experiments. The biggest consequence for this study was the closure of the University of Twente for several weeks at the start of this study. Therefore, the experiments had to be performed at home, which had some limitations:

1. There was no opportunity to put the subjects in a harness to prevent them from falling: this is necessary to measure subjects with their eyes closed. Without a harness, this is too dangerous. The subject's vision was fixed to reduce the visual contributions but prevent falls.
2. The original plan was to use EMG sensors, but these were not allowed to be used outside of the University of Twente. Thus, no EMG sensors were used.
3. Only one potentiometer was available, although the plan was to use two.

Not only the number of potentiometers was the problem. The possibility to mount the potentiometer on different heights was a problem. It was attached to the windowsill with tape, resulting in only one possible height. To be able to measure taller subjects, a plastic container was used to adapt the height of the potentiometer. If the experiments could have been executed at the university, better equipment would have been available to do more precise measurements. After measuring the first four subjects, the potentiometer broke down and could not be replaced in time because of the lock-down.

The guidelines of the Dutch government stated that people must keep 1.5-meter distance between each other. The measurement protocol was therefore adjusted to make sure that the subjects and researcher were able to keep that distance. The subjects would attach the belt and potentiometer themselves, with only visual inspection of the researcher. Because of the distance between the subject and the researcher, the length of the subjects could not be measured. If the length of the subject was not known, the length was estimated. Another result of the 1.5-meter distance was the foot placement. Under normal circumstances, the researcher would mark the foot placement of the subject so it would be the same each trial. This is impossible while staying 1.5-meters away. The subjects were instructed to align their ankle joint with the rotating axis of the mBAP. Specifically their malleolus medialis. Although these guidelines, there is the possibility that the subject had a different foot placement each trial because of this.

The last limitation was the age of the subjects: the corona virus is more dangerous for older people than younger people. Therefore, the decision was made to only measure young people, with an age between 20 and 30 years, for extra safety.

Due to these measures, it was not possible to develop and evaluate a protocol. As a result, a different goal has been chosen for this research and only the influence of the velocity of the support surface rotations on sensory reweighting during standing balance is investigated.

University of Twente, Enschede
Department of Biomechanical Engineering

The influence of the velocity of support surface rotations on sensory reweighting during standing balance.

Emily Zoetbrood[1]

[1] *Student Biomedical Engineering, Neural Motor Systems track*

Abstract

Multiple sensory organs contribute to the control of balance. The signals from these sensors are combined in the brain, where reliable information is favoured over unreliable information: a process called sensory reweighting. Support surface rotations can be used to investigate sensory reweighting during balance. Typically, a pseudorandom ternary stimulus (PRTS) is applied. With the increase of the amplitude of the PRTS, proprioceptive sensory information is downweighted. However, by increasing amplitude, the velocity is indirectly increased. The aim of this study is to investigate the influence of the velocity of support surface rotations on sensory reweighting during standing balance.

Four healthy subjects stood on a rotating support surface, following a PRTS signal, for nine trials. A base trial, four trials with an increasing amplitude but the same duration as the base trial and four trials with an increasing amplitude but the same velocity as the base trial. For every trial, body sway and ankle torque were measured. The RMS, sensitivity functions and coherence of the ankle torque and body sway were obtained. With an increase of amplitude and the same duration, the RMS increased, and the sensitivity functions mostly decreased. With an increase of amplitude and the same velocity, the RMS mostly increased, and the sensitivity functions fluctuated. In conclusion, both PRTS amplitude and PRTS velocity have an influence on sensory reweighting and should be considered.

Keywords

Standing balance — BAP — PRTS — Sensory reweighting — Support surface rotations

1. Introduction

Impaired balance is defined as having difficulties maintaining an upright position in daily life activities, for example walking, reaching or rising from a chair [1, 2]. This can have several causes, for example problems with mobility, balance or loss of muscle strength [1–6]. When several of these causes co-occur, impaired balance becomes symptomatic [2]. Roughly, one third of the elderly, aged 65 or older, fall at least once a year [1, 4, 7]. These falls can result in serious injuries, which may lead to decreased mobility and eventually loss of independence [1, 2, 4, 6, 7]. Targeted inventions improving standing balance are necessary to prevent falling. These inventions require detection of the underlying cause of impaired balance at an early stage [1].

Keeping standing balance is challenging because of gravity and the structure of the human body. About two-thirds of our weight is in our upper body, which balances on our legs, while our feet provide us with a narrow base of support [8]. The body's centre of mass (CoM) needs to stay above this narrow base of support to maintain balance [2]. This causes demands on the postural and balance control systems. These systems include the following: the sensory system (made up of the proprioceptive, visual and vestibular systems), the central nervous system (CNS) and the muscular and skeletal systems [8–10].

The sensory system is the main source of sensory information in the human body [10, 12]. The proprioceptive system consists of muscle, joint and cutaneous receptors. It provides us with information about the state of the effector system and information about our environment. The visual system collects information about our environment and gives the CNS the orientation and movement of the body in this environment. The vestibular system, however, provides information about the body orientation in the inertial frame of reference. The CNS integrates all this information to determine the appropriate action plan [3, 8–10, 12, 13]. The musculoskeletal system responds by regulating the body's posture and movement [8].

A simplified scheme of balance control is given in figure 1 [11]. The most simplified way to describe the dynamics of the human body is by an inverted pendulum in a closed loop system [1, 13, 14], with the mass of the human rotating around the ankle joint. The neuromuscular controller analyses the input signal from three sensors (proprioceptive, visual and vestibular) and sends out a signal towards the muscles to achieve a torque. The three different sensors have their

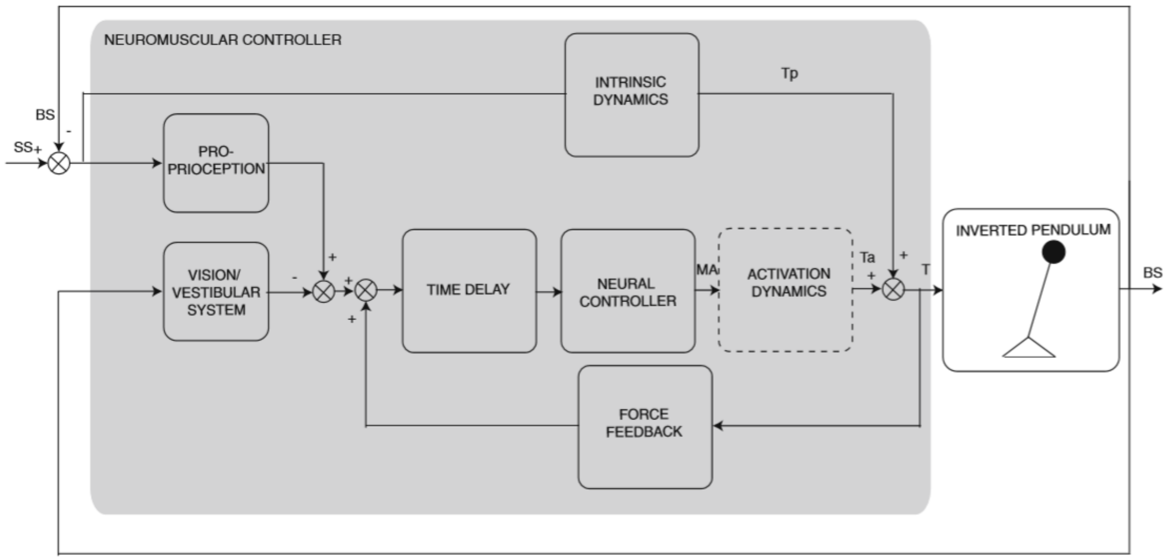


Figure 1. A basic model of balance control. The human body is described by the dynamics of an inverted pendulum, resulting in a measured body sway (*BS*). The human body is controlled by a neuromuscular controller (grey box), consisting of an active and a passive part. The passive part consists of the intrinsic dynamics representing the intrinsic properties of the muscles and produces a passive torque (T_p). The active part consists of the neural controller with a time delay and activation dynamics producing an active torque (T_a). The active and passive torque together is the ankle torque (T). The neural controller receives feedback from the sensory systems and force feedback. These sensory systems each have a weight factor. The measured body sway and the support surface rotation (*SS*) represent proprioception. [11]

own weighting parameters based on noise. These weighting parameters enable the verification of the input and partial compensation for the deteriorated systems, a compensation strategy called sensory reweighting [1–3, 5, 6, 8, 10–13]. This strategy states that the nervous system prefers reliable sensory information of one sensory system over less reliable sensory information of other sensory systems within a continuous dynamically weighting process. Generally, an increasing disturbance, or noise, on a sensor means a smaller weighting parameter [2, 6]. Each sensory system deteriorates with advanced age or injuries, making sensory reweighting necessary.

The elderly rely more on visual and proprioceptive information than young people and are therefore less capable of reweighting sensory information [1, 6]. To maintain standing balance, the elderly rely more on the hip strategy, meaning movement around the hip joint, while younger people rely more on ankle strategy, meaning movement around the ankle joint to maintain standing balance. If conditions become more challenging, young people will respond by adapting their balance strategy to the hip strategy. The elderly are less able to adapt to environmental changes because they already rely more on the hip strategy.

Sensory reweighting can be investigated with posturography or with system identification techniques with disturbances of proprioceptive information of the ankle [6]. Posturography eliminates or disturbs sensory information with external disturbances. Because changes in CoP and CoM movement are affected by all systems involved in standing balance and other compensation strategies, it is difficult to investigate sensory reweighting with posturography. Only the changes in CoP and CoM movement are included in the conclusions, while the contribution of the other underlying systems to these changes are not considered. System identification techniques identify the contribution of each individual system necessary in maintaining upright stance and thus allow the investigation of the contributions of each sensory system regardless of changes in the underlying systems involved in standing balance and compensation strategies used. Mechanical or sensory disturbances are used to disturb the specific underlying systems. Combining the disturbances with the body response provides a description of the balance control system. By applying sensory disturbances with increasing disturbance amplitudes over trials, resulting in less reliable sensory information with each increasing amplitude, it is possible to investigate sensory reweighting. However, by increasing the disturbance amplitudes, the velocity of the disturbance signal is also increased, while the influence of the increased velocity is not investigated.

To be able to measure the balance parameters, the subject should be perturbed to measure the body sway [8, 15, 16]. These perturbations can be translations or rotations. Translations are used to investigate the neural controller and rotations are used to investigate sensory reweighting. Since this study investigates sensory reweighting, rotational perturbations are needed. A way to rotational perturb a subject is with a Bilateral Ankle Perturbator (BAP). The BAP exists of two, independent, rotating pedals driven by a motor [16]. Force transducers measure the applied torque and indirectly the subject's momentum. The body kinematics are recorded with a motion capture system and muscle activity with surface EMG. It is therefore possible to investigate the effect of support surface perturbations on the generation of moments during balance control (proprioception).

However, most of these devices are big and laboratory bound. The University of Twente is currently working on a mobile BAP (mBAP). This mobile BAP weighs only 25 kilograms, creating a lot of new, not laboratory bound opportunities. The mBAP can come to patients with impaired balance control instead of them coming to the laboratory. With the BAP, both ankles can be perturbed separately, while with the mBAP both ankles are perturbed simultaneously. For a reliable measurement, it is important that the subject does not know the pattern of the perturbations since the subject should not be able to predict the perturbations [10]. Therefore, a pseudo random ternary signal (PRTS) is often used [6, 10, 11, 15, 16]. The advantage is that the signal seems random for the subject, but the disturbance is the same each trial. In most studies where a BAP and PRTS are used, the amplitude of the disturbance signal is altered but the duration of the signal is not, resulting in not only a change in amplitude but also a change in velocity. The influence of the change in velocity has not been investigated yet.

1.1 Aim

The aim of this study is to investigate the influence of the velocity of the support surface rotations on sensory reweighting during standing balance.

In previous research, the amplitude of the support surface rotations were increased while indirectly increasing the velocity. In this study, the amplitude of the disturbance signal will be increased while maintaining the same velocity.

1.2 Hypotheses

It is expected that with a constant velocity, but increasing amplitude, the subject is more able to withstand the higher perturbations than with the different velocities. When the amplitude of the disturbance signal increases, the proprioceptive information is expected to be downscaled. It is expected that the increasing of the amplitude in the trials with the same velocity has a smaller influence on the downscaling of the proprioceptive information than on the trials with the different amplitudes.

2. Method

2.1 Subjects

Four subjects volunteered in this research (1 male, mean ± 1 SD: Age = 25.5 ± 1.7 year, length = 173 ± 7 cm, weight = 80 ± 25 kg). Exclusion criteria were: any medical history of balance impairment or an age less than 20 and greater than 30 years. All participants gave written informed consent to participate in this study. The ethics committee Natural Sciences and Engineering Sciences of the Faculty of Engineering Technology of the University of Twente approved this study.

Figure 2. The set-up with the mobile Bilateral Ankle Perturbator. The subject will stand on the mBAP with their bare feet or with socks on. The subject stands with their arms crossed in front of their chest. The mBAP rotates around the ankle. The torques and weights are measured underneath the plates. The draw wire is hardly visible, but attached to the black belt. The subjects are instructed to look at the branch of the tree outside the window and align their ankle with the rotating axis of the mBAP.

2.2 Apparatus and recording

Figure 2 shows the experimental setup of this study. A mobile Bilateral Ankle Perturbator (University of Twente, Enschede, the Netherlands) was used to disturb the proprioceptive information of the participants. The weight and the applied torques of the subjects on the left and right support surface (SS) of the mBAP were measured with a sample frequency of 1000 Hz and were stored for further analysis. A positive signal meant an increase in weight and forward rotation. The body sway in anterior-posterior direction was measured using a draw wire potentiometer (Sentech SP2, Celesco, Chatsworth, CA, United States) with a sample frequency of 1000 Hz. A positive signal meant a forward rotation. The subject had a belt wrapped around their torso at the height of their belly button, the assumed height of the CoM. With a small magnet, the potentiometer was attached to the belt buckle. LabVIEW 2020 (National Instruments, TX, USA) was used to communicate with the mBAP and the potentiometer. LabVIEW would send out a certain voltage which corresponds to an angle. With these voltages the rotation angles were derived. The data was analysed using MATLAB 2019a (The MathWorks Inc., MA, USA).

2.3 Procedure

During all experiments, the subjects stood on the mBAP with their bare feet or with socks on and their arms crossed in front of their chest. They were instructed to align their ankle (specifically their malleolus medialis) with the rotating joint of the mBAP and keep their eyes open and look at a branch of a tree outside of the window. The support surface rotated following a pseudorandom ternary sequence (PRTS). This PRTS is a 4 stage shift register with feedback according to Peterka [10]. An 80 stage signal with a time increment of 0.25s was generated, resulting in a signal with a period time of 20s, see Figure 3. An increasing signal means a forward rotation of the SS platform. A decreasing movement means a backwards rotation. The subjects performed nine trials with different perturbation amplitudes and different durations:

- *Base trial:* This trial has a peak-to-peak ratio of 2° and a consistent trial duration of 2 minutes.
- *Same duration:* Four trials were increased or decreased to create different peak-to-peak amplitude (0.5°, 1°, 2°, 4°, 8°), while keeping the duration the same. Indirectly, changing the velocity of each trial.
- *Same velocity:* Four trials (with peak-to-peak ratios of 0.5°, 1°, 4°, 8°) were accelerated or delayed to have the same velocity as the base trial and the same amplitudes as the same duration trials.

Table 1 gives an overview of the trials. The trials with the same duration are the trials used in previous research [6,10,11]. The trials with the same velocity are the trials specific designed for this research. Both are measured to compare the results. Figure 3B shows the continuous velocity of the disturbance signal with a peak-to-peak ratio of 2. The velocity is the maximum velocity, zero or the negative maximum velocity. Table 1 shows the maximum velocity.

All trials consisted of 6 complete cycles of the perturbation signal. Before each trial, the participant was given five seconds to get accustomed to the perturbation. After each trial, the subject was given a five-minute rest before the next trial. The order of the trials was randomized for each subject, while making sure that every subject had a different order, reducing the influence of the order of trials on the outcome of the research [6]. The subjects were not informed of the amplitude or duration of the trial, preventing influence of knowledge on the result.

2.4 Data analysis

Stimulus response

The body sway was calculated from the potentiometer data, resulting in the segment angle of the leg relative to the vertical, see equation A.6 in appendix A and representing the angle of the CoM relative to the vertical. The ankle torques were obtained from the recorded torques on the SS. The relevant information about the body sway and ankle torque is below 10 Hz, therefore, the data was filtered with a second-order Butterworth filter, cut-off at 10 Hz.

Table 1. Trial overview. The first trial is the base trial. The next four trials have the same duration were the amplitude is increased or decreased, but maintaining the duration, resulting in different velocities. The last four trials have the same velocity that are adjusted to maintain the same velocity. This velocity is the maximum velocity.

Amplitude (°)	Period time (s)	Duration (s)	Velocity (°/s)
2	20	120	0.9
0.5	20	120	0.2
1	20	120	0.4
4	20	120	1.8
8	20	120	3.6
0.5	5	30	0.9
1	10	60	0.9
4	40	240	0.9
8	80	480	0.9

Body sway descriptors

A description of the stimulus response is given by the root mean square (RMS) averaged over the six cycles and the time series [6, 10]. This is calculated for both the body sway and the ankle torque. The RMS of the disturbance signal is also calculated for comparison.

Sensitivity functions

To describe and obtain a non-parametrical description of the human balance control, the sensitivity functions of the output of the human balance control (body sway and ankle torque) to the perturbation were obtained by estimating Frequency Response Functions (FRFs). Therefore, the perturbation, ankle torque and body sway were Fast Fourier Transformed (FFT) to the frequency domain. Next, the Power Spectral Densities (PSD) and Cross Spectral Densities (CSD) were computed within seven frequency bands 0-0.1 Hz, 0.1-0.3 Hz, 0.3-0.7 Hz, 0.7-1.4 Hz, 1.4-2.2 Hz, 2.2-3.1 Hz and 3.1-4.1 Hz [6]. The PSD and CSD were then averaged over the six cycles and the subjects. The FRFs were estimated using the indirect approach [15]:

$$^{SS}S_x(f) = \Phi_{SS,x}(f) \cdot [\Phi_{SS,SS}(f)]^{-1} \tag{1}$$

In which $\Phi_{SS,x}(f)$ represents the CSD of the SS rotation and x, which represents the ankle torque (T) or body sway (BS). $\Phi_{SS,SS}(f)$ represents the PSD of the SS rotation. The FRF magnitude and the FRF phase represent the amplitude ratio and the relative delay, respectively, between the FRF rotation and the ankle torque and body sway. A magnitude of 1 and phase of 0° indicates that the subject‘s body was perfectly oriented to the support surface and relies completely on the proprioceptive information, while a magnitude of zero indicates that the body remained oriented to earth-vertical independent of the surface orientation and relies completely on the vestibular information [11, 14].

The coherence for each frequency band was estimated

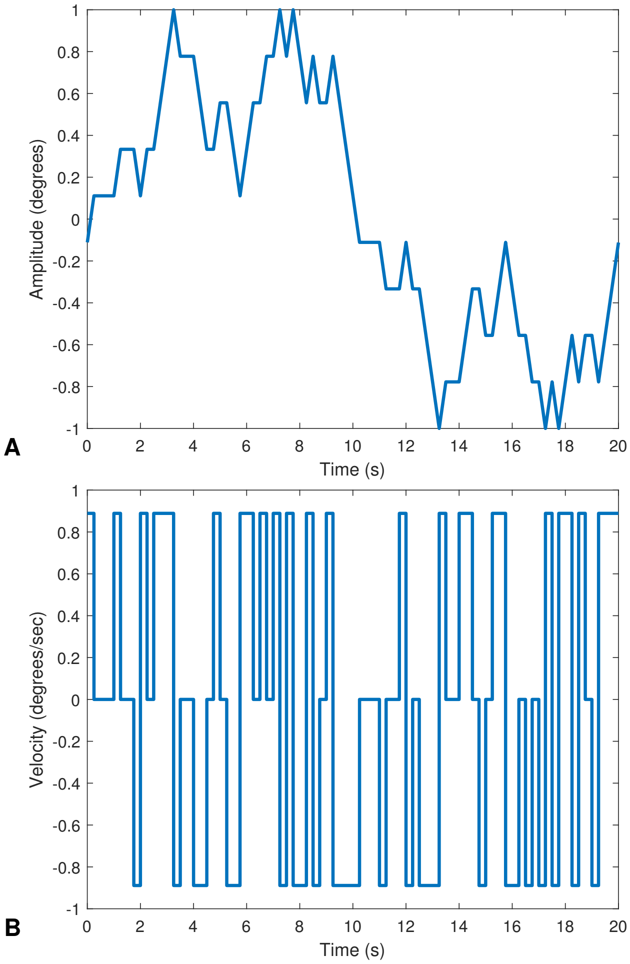


Figure 3. The PRTS signal. Figure A shows the disturbance signal of the base trial. Figure B shows the velocity of the disturbance signal of the base trial. The velocity is $0.9°/s$, $0°/s$ or $-0.9°/s$.

with the following formula:

$$\gamma_x^2(f) = |\Phi_{SS,x}(f)|^2 \cdot [\Phi_{SS,SS}(f) \cdot \Phi_{x,x}(f)]^{-1} \tag{2}$$

In which $\Phi_{SS,x}(f)$ represents the CSD of the SS rotation and x, which represents the ankle torque (T) or body sway (BS). $\Phi_{x,x}(f)$ represents the PSD of x, which represents the ankle torque (T) or body sway (BS). Values of the coherence vary from 0 to 1, with 0 indicating that there is no linear correlation between the stimulus and response, and 1 indicating a perfect linear correlation with no noise. Values less than 1 occur in practice either because there is noise in the system or there is a non-linear relation between stimulus and response [10].

2.5 Statistical Analysis

The statistical analysis was performed using IBM SPSS Statistics 25 (IBM, NY, VS). A repeated measures ANOVA is used to test the significant differences in sensitivity functions between the amplitudes with the same duration and same velocity. For the statistical analysis, only the frequency bands of which the coherence is greater than 0 are used. The frequency band is included as covariate to adjust for the differences due to frequencies. The significance was set at 0.05.

Before the results of the repeated measures ANOVA could be interpreted, it is important to first check whether the data is spheric or not. Meaning, that the variance of the difference scores between the conditions must be equal. Thus, a Mauchly's Test of Sphericity was performed. If the P-value is greater than 0.05, then the data can be assumed to be spherical. If the data is assumed to be spherical, the sphericity assumed results of the repeated measures ANOVA will be used. If the data is not assumed to be spheric, the results of the Greenhouse-Geisser test will be checked. If this is below 0.75, then the results of the Greenhouse-Geisser test will be used. Is the Green-house-Geisser test above 0.75, the results of the Huynh-Feldt test will be used. [17]

3. Results

Here the group averaged results are shown, results of the individual participants are shown in appendix C.

3.1 Body sway descriptors

Figure 4 shows the stimulus response for the trials. The stimulus response is averaged over the trials and the subjects. Figure 5 shows the RMS between the subjects averaged over the cycles. Figure 5A shows the RMS of the disturbance signal and shows an increase in RMS with an increase of amplitude. The change in velocity had no influence on the RMS. The RMS doubles with the amplitude. Figure 5B shows the RMS of the ankle torque of all trials for all trials. Figure 5C shows the RMS of the body sway for all trials.

Same duration

Figure 4A shows the stimulus response of the trials with the same duration for both the ankle torque and body sway. The stimulus response increases with an increase in amplitude for both the ankle torque and body sway. Figure 5B and 5C show the RMS of the ankle torque and body sway respectively. Yellow and green represent the trials with the same duration. The RMS of the ankle torque increases with an increase of amplitude. The RMS of the body sway stays practically the same for the smaller trials (0.5 till 2°) with an increase in amplitude. For the other trials, the RMS of the body sway increases with an increase in amplitude.

Same velocity

Figure 4B shows the stimulus response of the trials with the same velocity for both the ankle torque and body sway. The stimulus response increases with an increase in amplitude for both the ankle torque and body sway. Figure 5B and 5C show the RMS of the ankle torque and body sway respectively. Blue and green represent the trials with the same velocity. The RMS of ankle torque stays the same for the first two trials (0.5 and 1°) with an increase in amplitude. For the base trial the RMS of the ankle torque decreases with an increase in amplitude. For the two largest trials (4 and 8°) the RMS of the ankle torque increases with an increase in amplitude. The RMS of the body sway increases with an increase with an increase in amplitude.

The RMS of the ankle torque is larger for the trials with the same velocity than for the trials with the same duration, except for the amplitude of 0.5°. The RMS of the body sway is larger for the trials with the same velocity than for the trials with the same duration, except for the amplitude of 4°.

3.2 Sensitivity functions

Figure 6 shows the mean sensitivity functions of the frequency bands of the ankle torque and body sway to the disturbances. Figure 6A shows the mean sensitivity functions of the trials with the same duration. Figure 6B shows the mean sensitivity functions of the trials with the same velocity. Because of the small duration of 5s of the trial with amplitude 0.5°, there is no frequency in the frequency band 0-0.1 Hz. Only the frequency bands of 0.1-0.3 Hz, 0.3-0.7 Hz, 0.7-1.4 Hz, 1.4-2.2 Hz and 2.2-3.1 Hz are used in the statistical analysis of the magnitude because the coherence is (almost) zero at the first and last frequency band.

Same duration

Table 2 shows the results of the repeated measures ANOVA for the sensitivity functions of the trials with the same duration for both the ankle torque and body sway. For the ankle torque, there is a significant decrease in sensitivity with an increase of amplitude (P=0.000). For the smaller amplitudes (0.5° till 2°) of the body sway, the sensitivity decreases with an increase of amplitude. With an amplitude of 4° the sensitivity increases and with an amplitude of 8° it decreases again. These fluctuations are significant (P=0.019).

The mean difference of the magnitude of the ankle torque between the trials with the same duration 2.53.

Table 2. The average and standard deviation of the magnitude of the ankle torque and body sway for the trials with the same duration of the second till sixth frequency band. The P-value was calculated with a repeated measures ANOVA with a significance of 0.05.

Amplitude (°)	Magnitude of	
	ankle torque (Nm/°)	body sway (°/°)
0.5	16.15 ± 9.35	0.43 ± 0.38
1	13.58 ± 6.19	0.16 ± 0.12
2	11.62 ± 5.34	0.12 ± 0.09
4	9.36 ± 4.49	0.15 ± 0.10
8	6.05 ± 3.38	0.13 ± 0.08
P-value	0.000	0.019

Same velocity

Table 3 shows the results of the repeated measures ANOVA for the sensitivity functions of the trials with the same velocity for both the ankle torque and body sway. For the two smallest amplitudes (0.5 and 1°) the magnitude of the ankle torque decreases with an increase of the amplitude. With an amplitude of 2° the magnitude increases with an increase in amplitude. For the two largest amplitudes (4 and 8°) the magnitude practically stays the same with an increase in amplitude. These fluctuations are significant (P=0.029). For the four smallest amplitudes (0.5 till 4°) the sensitivity of the body sway decreases with an increase in amplitude. For the trial with an amplitude of 8° the sensitivity increases with an increase in amplitude. These fluctuations are not significant (P>0.05).

For the smaller frequencies (0.5 and 1°) the magnitude of the ankle torque of the trials with the same velocity is lower than the magnitude of the ankle torque of the trials with the same duration. For the larger frequencies (4 and 8°) the magnitude of the ankle torque of the trials with the same velocity is larger than the magnitude of the ankle torque of the trials with the same duration.

The mean difference of the magnitudes of the ankle torque between the trials with the same velocity is 1.71.

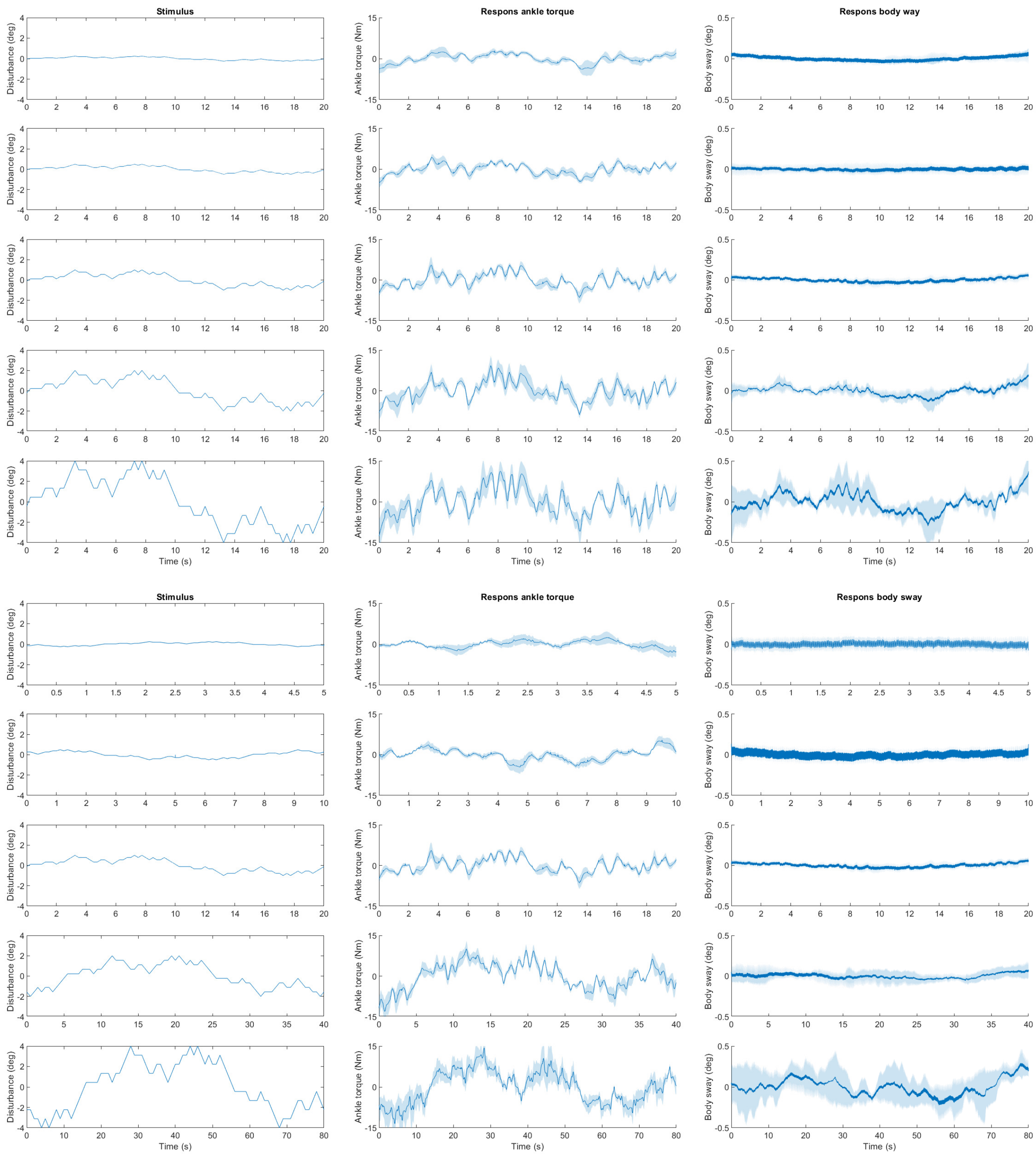
Table 3. The average and standard deviation of the magnitude of the ankle torque and body sway for the trials with the same velocity of the second till sixth frequency band. The P-value was calculated with a repeated measures ANOVA with a significance of 0.05.

Amplitude (°)	Magnitude of	
	ankle torque (Nm/°)	body sway (°/°)
0.5	13.25 ± 10.11	0.41 ± 0.35
1	9.28 ± 4.01	0.37 ± 0.45
2	11.62 ± 5.34	0.12 ± 0.09
4	11.96 ± 7.34	0.09 ± 0.07
8	11.79 ± 8.52	0.16 ± 0.13
P-value	0.029	0.094

4. Discussion

Multiple sensory organs contribute to the control of balance. The signals from these sensors are combined in the brain, where reliable information is favoured over unreliable information: a process called sensory reweighting. In this study, support surface rotations following a PRTS signal, were used to investigate sensory reweighting during standing balance. Four subjects performed 9 trials. A base trial, four trials with an increasing amplitude but the same duration as the base trial and four trials with an increasing amplitude but the same velocity as the base trial. For every trial, the body sway and ankle torque were measured. The stimulus response, RMS, sensitivity functions and coherence of the ankle torque and body sway were obtained.

The shape of the stimulus response functions of the trials with the same duration matches the shape of the stimulus response functions of that found by Peterka [10]. The results show that the RMS of the body sway increases with an increasing amplitude. This matches the results found by Pasma et al. and Peterka [6, 10]. The results show that the velocity has an influence on the RMS of the body sway since the RMS of body sway changes with a change in velocity. The RMS



A

B

Figure 4. The stimulus response of the ankle torque and body sway. The stimulus response is averaged over the trials and the subjects. The light blue areas are the standard deviations between the subjects. The base trial is shown in both figure A and B in the middle. Figure A shows the stimulus response of the trials with the same duration. Only the amplitude of the disturbance signal is adjusted, with the velocity indirectly adjusted. Figure B shows the stimulus response of the trials with the same velocity. Note the scaling of the time axis in figure B. The amplitude of the disturbance signal is adjusted and the duration of the signal, keeping the same velocity.

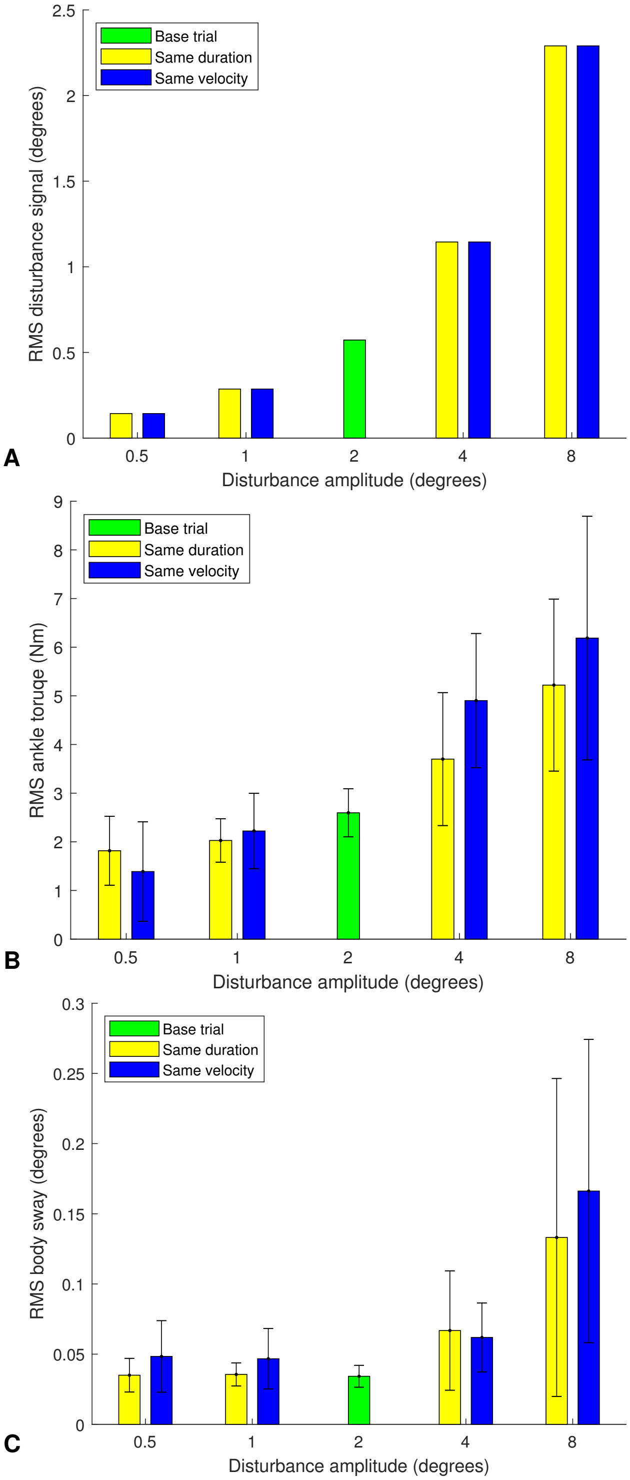


Figure 5. The root mean square (RMS) averaged over the subjects. Figure A shows the RMS of the disturbance signal. Figure B shows the RMS and standard deviation of the ankle torque. Figure C shows the RMS and standard deviation of the body sway. Green shows the base trial with an amplitude of 2°. Yellow shows the trials with the same duration of 20s. Blue shows the trials with the same velocity of 0.9°/s.

of the ankle torque fluctuates with an increase in amplitude. There is a large standard deviation caused by the subjects. As shown in the figures in appendix C the difference in response between the subjects is very large.

The shape of the magnitude of the sensitivity functions of the ankle torque with the trials of the same duration and same velocity matches the results found by Pasma et al. and Peterka [6, 9, 10]. With an increase in amplitude, the magnitude decreases. Above 1 Hz, the magnitude increases, to decrease for higher frequencies. As expected, with an increasing amplitude, the magnitude decreases. Meaning that the subjects rely more on their vestibular information than their proprioceptive information with an increasing amplitude. In the trials with the same velocity the difference between magnitudes with increasing amplitude is smaller than with the trials with the same duration. Indicating that increasing both the velocity and the amplitude makes for more unreliable proprioceptive information.

The phase of both the ankle torque and body sway of the trials with the same duration matches the results found by Pasma et al. [6]. The sensitivity functions did not always decrease with an increasing amplitude, against the expectations. As expected, the difference in magnitude of the ankle torque between the amplitudes of the trials with the same velocity is smaller than the difference in magnitude of the trials with the same duration. All the sensitivity functions of the trials with the same velocity for the ankle torque are smaller than the differences between the trials with the same duration.

The coherence is (almost) zero for the frequency bands of 0-0.1 Hz and 3.1-4.1 Hz, indicating that there is no linear correlation between the stimulus and the response. For the other frequency bands, the coherence is larger. The trials with the same duration show a coherence peak at frequency band 0.7-1.4 Hz. The trials with the same velocity show a coherence peak at 0.1-0.3 Hz. The coherence of the ankle torque is larger than the coherence of the body sway for all trials. Peterka and Pasma et al. found a coherence between 0.2-0.4 and 0.8 [9, 10]. Which is larger than found in this study. This is possibly caused by the averaging over the frequency bands. Figure C.1 shows the sensitivity functions of all the trials without frequency bands. The coherence is much larger here and matches the results found by Peterka and Pasma et al. [9, 10].

The mean difference of the magnitudes of the ankle torque between the trials with the same duration is 2.53, while the mean difference of the magnitudes of the ankle torque between the trials with the same velocity is 1.71. This suggests that the influence of the amplitude is about twice the influence of the velocity.

4.1 Limitations

The first and main limitation was the draw wire potentiometer. Only one potentiometer was available. Therefore, the body sway is measured less accurately. After the four subjects, the results coming from the potentiometer were not completely satisfying. In appendix B the investigation into this problem is shown. Unfortunately, during this investigation, the potentiometer broke down. There was no possibility to get a new potentiometer in the short amount of time left for this study. The height of the potentiometer was also estimated. The potentiometer was taped to the windowsill. If a subject was taller, a plastic container was taped between the windowsill and the potentiometer, which may have resulted in the potentiometer not always being at the same height as the subject's belly button. There was a high variance in waist circumference of the subjects, resulting in different belts for the subjects for the attachment the potentiometer. The used non-elastic belts were hard to mount on the proper position and could also slightly move, causing a difference in the extraction of the draw wire. All these things may have influenced the body sway performance. These may explain why the body sway is not always significant or the bigger variances in the body sway.

In this study, the subjects had fixed vision. Because of the fixed vision, there is a small visual feedback. Thus, the results not only show the proprioceptive feedback but also a small visual feedback. With closed eyes, there is no visual feedback. Peterka investigated the difference in results between fixed vision and the eyes closed [10]. It showed that the RMS of the body sway larger is when the subjects have their eyes closed. Meaning, that with the eyes closed, the subject has a larger body sway and thus more difficulty maintaining upright stance.

Only four subjects were used in this study. These subjects were young. Because the mBAP will mostly be used for the elderly, these results will not represent that population. The elderly also rely more on the hip strategy than the young, this difference in tactics is not taken into consideration.

For this research, the disturbance signals were altered to the middle velocity of the trial with an amplitude of $2°$. There is the possibility that this velocity is not a representation of the outside world. It would be interesting to investigate the influence of the other velocities with increasing amplitudes. The change in velocity is only indirectly investigated in this research, since the change in amplitude also has an influence.

One subject mentioned that standing on the mBAP made them stand with more distance between their feet than natural. Because it does not feel normal, it could have influenced the results. One subject had the feeling that they were disturbed more forwards than backward, while the disturbance was for both sides equally. One subject mentioned that their sport was horse-riding and that they learned to respond to perturbations with their core muscles and hips. The body sway of this subject (3) was larger than the body sway of the other subjects.

4.2 Recommendations

For further research there are some recommendations. First, repeat this measurement with the correct settings for the potentiometer, as mentioned in appendix B. This will increase the accuracy of the body sway. The second recommendation is to use two potentiometers. One of the potentiometers can measure the leg angle and the other potentiometer can measure the hip angle. With two potentiometers, the accuracy of the body sway will increase, and it is possible to make a difference between the ankle strategy and hip strategy. Third, it should be possible to adjust the height of the potentiometer easier and more precise. When the potentiometer is adjusted to a vertical slider, the height can be precisely adjusted to the height of the CoM of the subject.

Fourth, this study should be repeated with more subjects. Not only more subjects but also subjects from different age groups. More subjects result in more realistic results. Different age groups can provide insight into the influence of the velocity on the results related to age.

In addition, it can be interesting to investigate the influences of other velocities of the disturbance signal. Adjusting the PRTS to the higher velocities can result in long experiments. The number of durations should therefore be taken into consideration. Or it can be interesting to only investigate the influence of the velocity by taking one amplitude and adjusting the duration and thus the velocity. It would also be interesting to investigate the pure influence of the velocity. The subjects perform trials with the same amplitude but different velocities.

Finally, to investigate the influence of velocity on sensory reweighting more closely, the use of a model will be interesting. With the help of a model, the difference between the weighting parameters can be investigated closely.

5. Conclusion

This study showed that the subjects responded different to the change in velocity or amplitude of the disturbance signal. The RMS of both the ankle torque and body sway were larger for the trials with the same velocity than for the trials with the same duration. The difference in magnitude of the ankle torque between the amplitudes of the trials with the same velocity in smaller than the difference in magnitude of the trials with the same duration.

In conclusion, both the velocity and the amplitude of the disturbance signal have an influence on sensory reweighting. The data suggest that the influence of the amplitude is larger than the influence of the velocity. But increasing both the amplitude and the velocity has a bigger influence than only changing the amplitude.

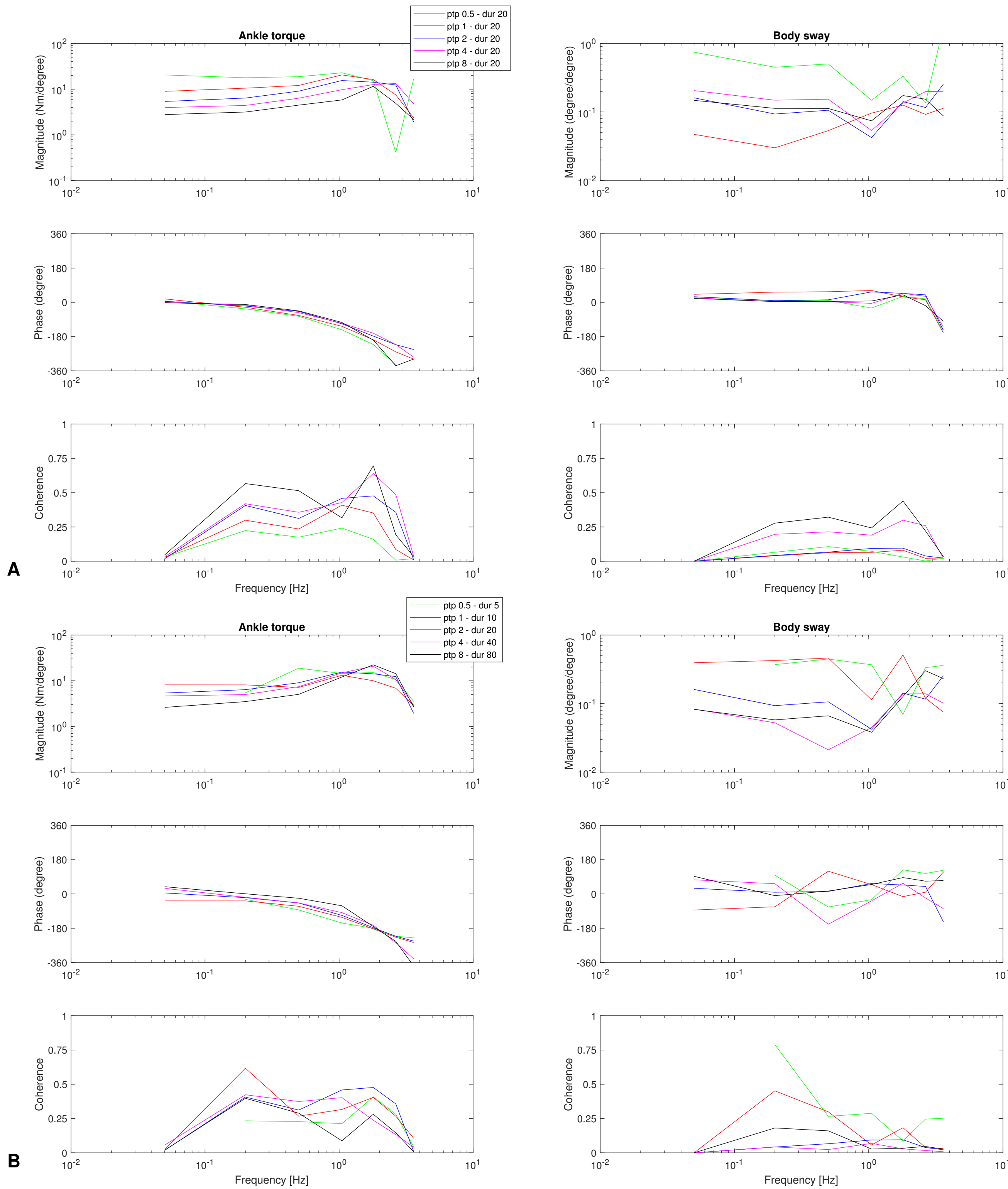


Figure 6. Sensitivity functions with frequency bands. The sensitivity functions are averaged over the trials and subjects. The base trial is shown in both figure A and B in the middle. Figure A shows the sensitivity function of the trials with the same duration. Figure B shows the sensitivity function of the trials with the same velocity. Note that the trial with an amplitude of 5° (green line) starts at a later frequency than the other lines because of the small sample size of this trial.

References

[1] Jantsje H. Pasma, Denise Engelhart, Alfred C. Schouten, Herma Van der Kooij, Andrea B. Maier, and Carel G. M. Meskers. Impaired standing balance: the clinical need for closing the loop. *Neuroscience*, 267:157–165, 2014.

[2] Denise Engelhart, Jantsje H. Pasma, Alfred C. Schouten, Carel G. M. Meskers, Andrea B. Maier, Thomas Mergner, and Herman van der Kooij. Impaired standing balance in elderly: a new engineering method helps to unravel causes and effects. *Journal of the American Medical Directors Association*, 15(3):227–e1, 2014.

[3] J. H. J. Allum, M. G. Carpenter, F. Honegger, A. L. Adkin, and B. R. Bloem. Age-dependent variations in the directional sensitivity of balance corrections and compensatory arm movements in man. *Journal of Physiology*, 542(2):643–663, 2002.

[4] Curtis S. Florence, Gwen Bergen, Adam Atherly, Elizabeth Burns, Judy Stevens, and Cynthia Drake. Medical costs of fatal and nonfatal falls in older adults. *Journal of the American Geriatrics Society*, 66(4):693–698, 2018.

[5] Isabella K. Wiesmeier, Daniela Dalin, and Christoph Maurer. Elderly use proprioception rather than visual and vestibular cues for postural motor control. *Frontiers in Aging Neuroscience*, 7:97, 2015.

[6] Jantsje H. Pasma, Denise Engelhart, Andrea B. Maier, Alfred C. Schouten, Herman van der Kooij, and Carel M. Meskers. Changes in sensory reweighting of proprioceptive information during standing balance with age and disease. *Journal of Neurophysiology*, 114(6):3220–3233, 2015.

[7] Anne Shumway-Cook, Marcia A. Ciol, Jeanne Hoffman, Brian J. Dudgeon, Kathryn Yorkston, and Leighton Chan. Falls in the medicare population: incidence, associated factors, and impact on health care. *Physical Therapy*, 89(4):324–332, 2009.

[8] David A. Winter, Aftab E. Patla, and James S. Frank. Assessment of balance control in humans. *Medical Progress through Technology*, 16(1-2):31–51, 1990.

[9] Jantsje H. Pasma, Tjitske A. Boonstra, S. Floor Campfens, Alfred C. Schouten, and Herman Van der Kooij. Sensory reweighting of proprioceptive information of the left and right leg during human balance control. *Journal of Neurophysiology*, 108(4):1138–1148, 2012.

[10] Robert J. Peterka. Sensorimotor integration in human postural control. *Journal of Neurophysiology*, 88(3):1097–1118, 2002.

[11] Jantsje H. Pasma, Joost Van Kordelaar, D. De Kam, V. Weerdesteyn, Alfred C. Schouten, and Herman Van Der Kooij. Assessment of the underlying systems involved in standing balance: the additional value of electromyography in system identification and parameter estimation. *Journal of Neuroengineering and Rehabilitation*, 14(1):97, 2017.

[12] Robert J. Peterka and Franklin O. Black. Age-related changes in human posture control: sensory organization tests. *Journal of Vestibular Research*, 1(1):73–85, 1990.

[13] Jantsje H. Pasma, Tjitske A. Boonstra, Joost van Kordelaar, Vasiliki V. Spyropoulou, and Alfred C. Schouten. A sensitivity analysis of an inverted pendulum balance control model. *Frontiers in Computational Neuroscience*, 11:99, 2017.

[14] Robert J Peterka. Simplifying the complexities of maintaining balance. *IEEE Engineering in Medicine and Biology Magazine*, 22(2):63–68, 2003.

[15] Herman van der Kooij, Edwin van Asseldonk, and Frans C. T. van der Helm. Comparison of different methods to identify and quantify balance control. *Journal of Neuroscience Methods*, 145(1-2):175–203, 2005.

[16] Alfred C. Schouten, Tjitske A. Boonstra, Frank Nieuwenhuis, S. Floor Campfens, and Herman van der Kooij. A bilateral ankle manipulator to investigate human balance control. *IEEE transactions on neural systems and rehabilitation engineering*, 19(6):660–669, 2011.

[17] Ruben Geert van den Berg. SPSS Repeated Measures ANOVA II. `https://www.spss-tutorials.com/spss-repeated-measures-anova-example-2/`, 2020. Accessed: 2020-12-07.

Appendices

A Body sway calculation

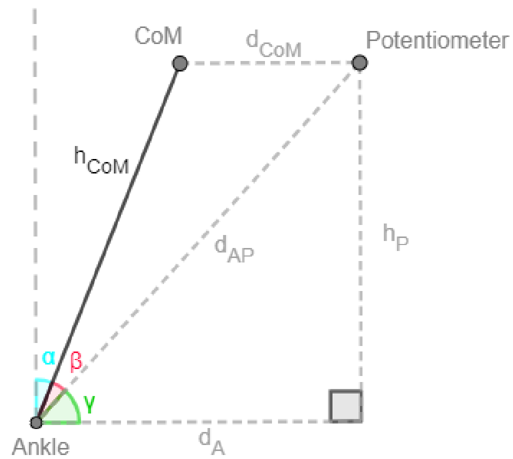


Figure A.1. The simplified sketch of the experimental setup.

Figure A.1 shows a simplified sketch of the experimental setup for the calculation of the body sway. The body sway is the angle between the leg segment and the vertical (α). The body sway is calculated using the angles between the ankle and the floor (eq. A.1).

$$\alpha = \frac{\pi}{2} - \beta - \gamma \tag{A.1}$$

With β being the angle between the leg segment and the line between the ankle joint and the potentiometer, and γ being the angle between the floor and the line between the ankle joint and the potentiometer. The angle between the leg segment and the line between the ankle joint and the potentiometer (β) can be calculated with the cosine rule (eq. A.2) and rewriting it (eq. A.3).

$$d_{CoM}^2 = h_{CoM}^2 + d_{AP}^2 - 2 \cdot h_{CoM} \cdot d_{AP} \cdot \cos(\beta) \tag{A.2}$$

$$\beta = \cos^{-1}\left(\frac{h_{CoM}^2 + d_{AP}^2 - d_{CoM}^2}{2 \cdot h_{CoM} \cdot d_{AP}}\right) \tag{A.3}$$

In which h_{CoM} represents the height of the CoM in m. d_{CoM} is the distance between the CoM and the Potentiometer, calculated with the voltage of the wire in m. d_{AP} represents the distance between the ankle joint and the potentiometer in m, calculated with the Pythagoras theorem (eq. A.4).

$$d_{AP} = \sqrt{d_A^2 + h_P^2} \tag{A.4}$$

In which d_A represents the horizontal distance between the ankle joint and the potentiometer constant as 0.43 m and h_P represents the height of the potentiometer in m.

The angle between the floor and the line between the ankle joint and potentiometer (γ) is calculated with the tangent of the height of the potentiometer divided by the horizontal distance between the ankle joint and the potentiometer (eq. A.5).

$$\gamma = \tan^{-1}\left(\frac{h_P}{d_A}\right) \tag{A.5}$$

In which d_A represents the horizontal distance between the ankle joint and the potentiometer in m and h_P represents the height of the potentiometer in m.

Together this results in the equation for the body sway as shown in equation A.6:

$$\alpha = \frac{\pi}{2} - \cos^{-1}\left(\frac{h_{CoM}^2 + d_A^2 + h_P^2 - d_{CoM}^2}{2 \cdot h_{CoM} \cdot \sqrt{d_A^2 + h_P^2}}\right) - \tan^{-1}\left(\frac{h_P}{d_A}\right) \tag{A.6}$$

B Potentiometer investigation

In paragraph 3.1 the stimulus response of the trials is given. As mentioned, there was an interesting standard deviation to the body sway. One explanation was that this was caused by the programming in Labview. The outputs of the mBAP and the outputs of the potentiometer were measured over the same voltage width of $-10V$ and $10V$. The potentiometer only needs voltage between $0V$ and $5V$. After the measurements of the four subjects, the influence of the voltage settings was investigated with a wooden stick attached to the mBAP (Fig. B.1). With this set-up the potentiometer would follow the exact same path as the mBAP would do. Figure B.2 shows the ‘body sway’ response of the wooden stick. In the right column, the response is shown with the old settings ($-10V$ and $10V$) and in the middle column with the new settings ($0V$ and $5V$). The new settings are clearly more accurate. However, the potentiometer broke down after this little experiment, so it was not possible to measure subjects with the new settings. Figure B.3 shows the RMS of the body sway with both the old and the new settings. The first notable thing is that the body sway with the old settings is increasing while the body sway with the new settings is decreasing.

Figure B.1. The set-up with the stick.

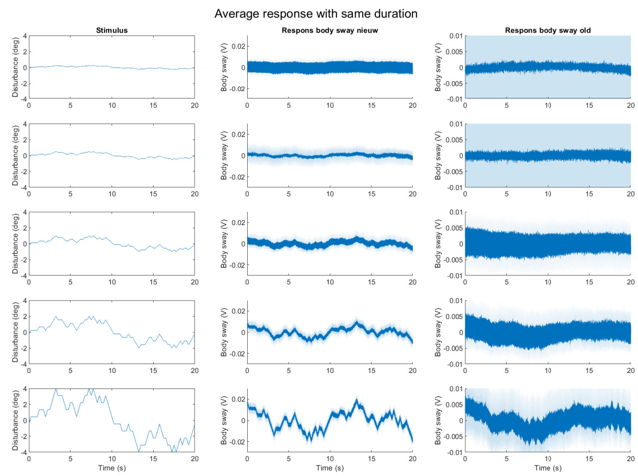


Figure B.2. The stimulus response function of the potentiometer with the stick in the old and new settings.

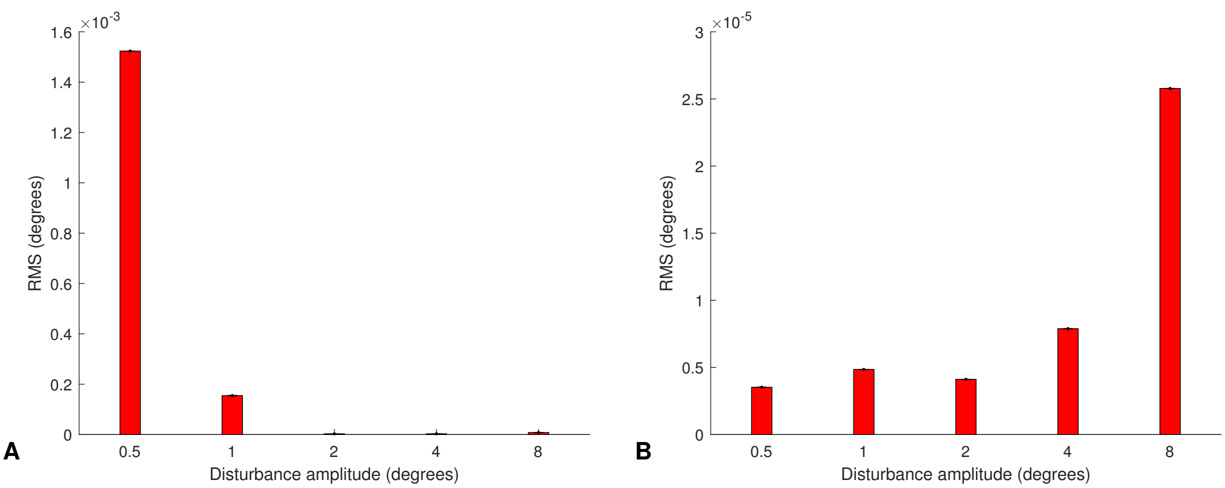


Figure B.3. The RMS of the body sway of the stick. Figure A shows the RMS of the stick with the old settings. Figure B shows the RMS with the new settings. The settings of figure A are used in this study. Note the large difference in RMS scale.

C Figures results

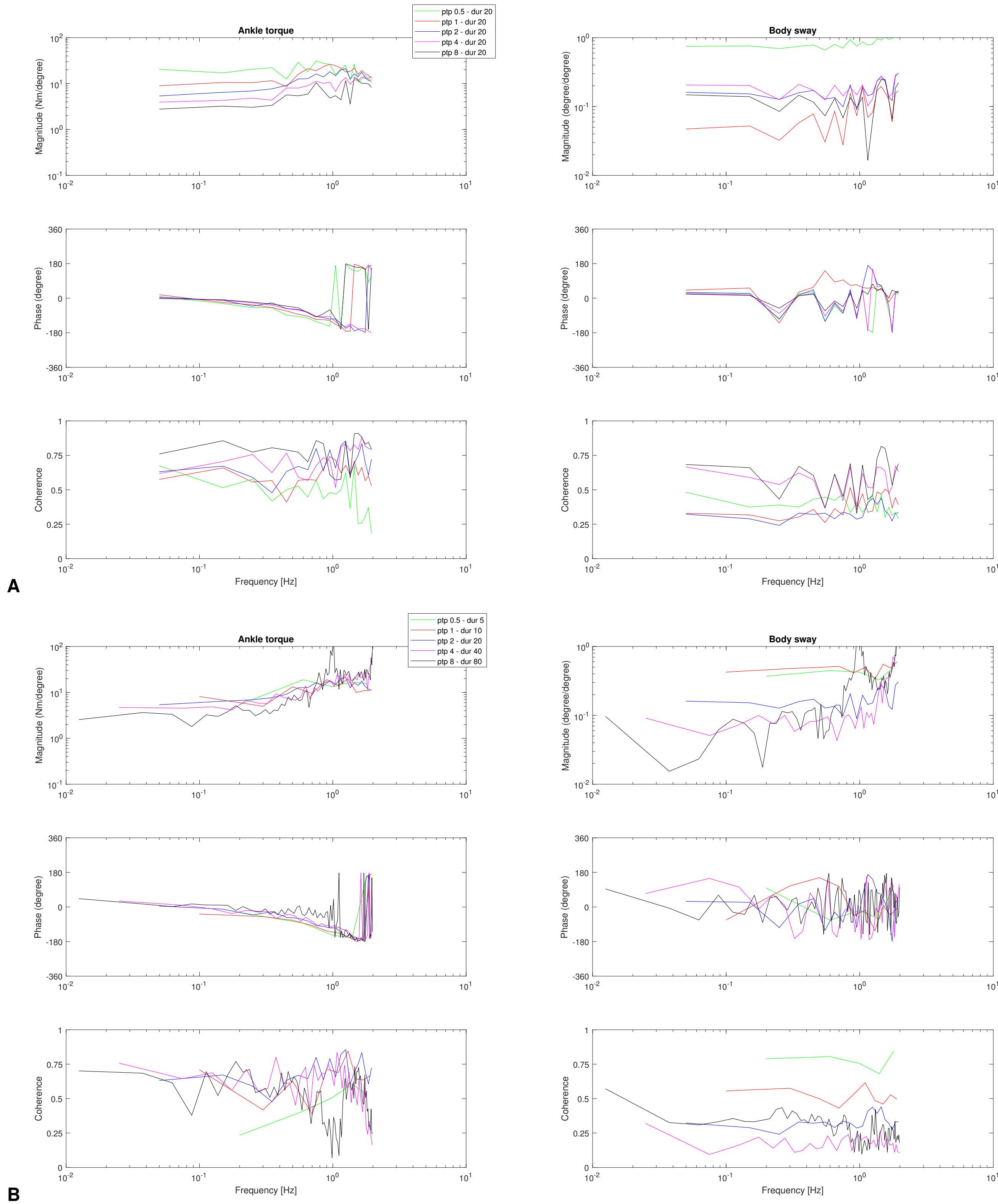
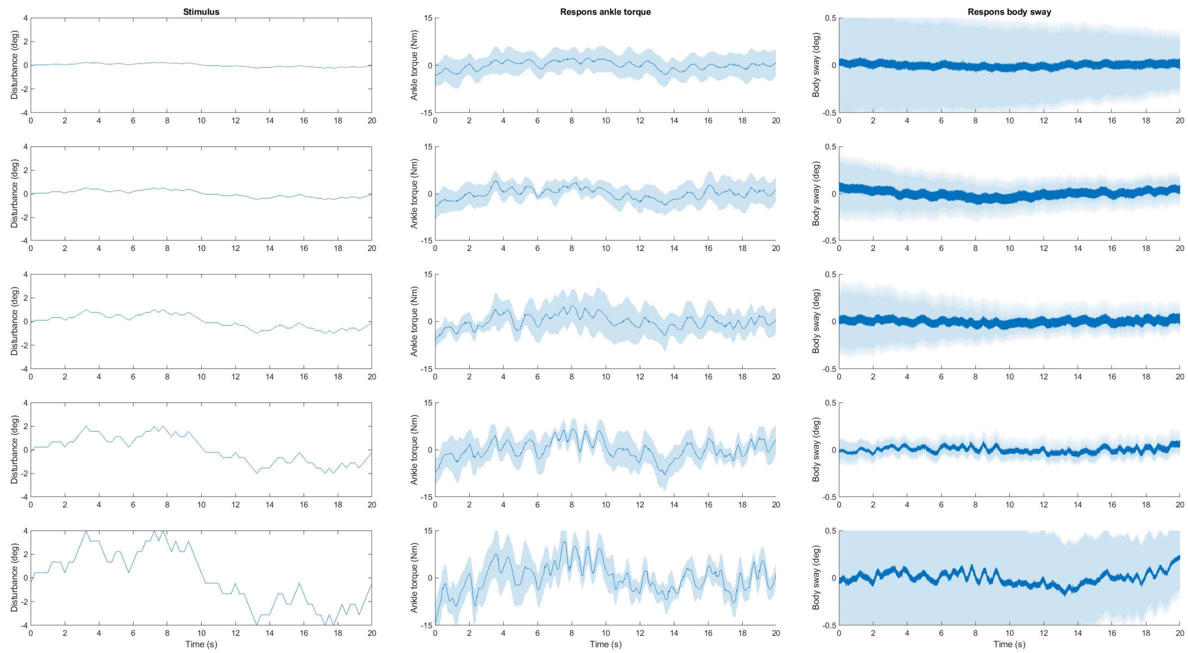
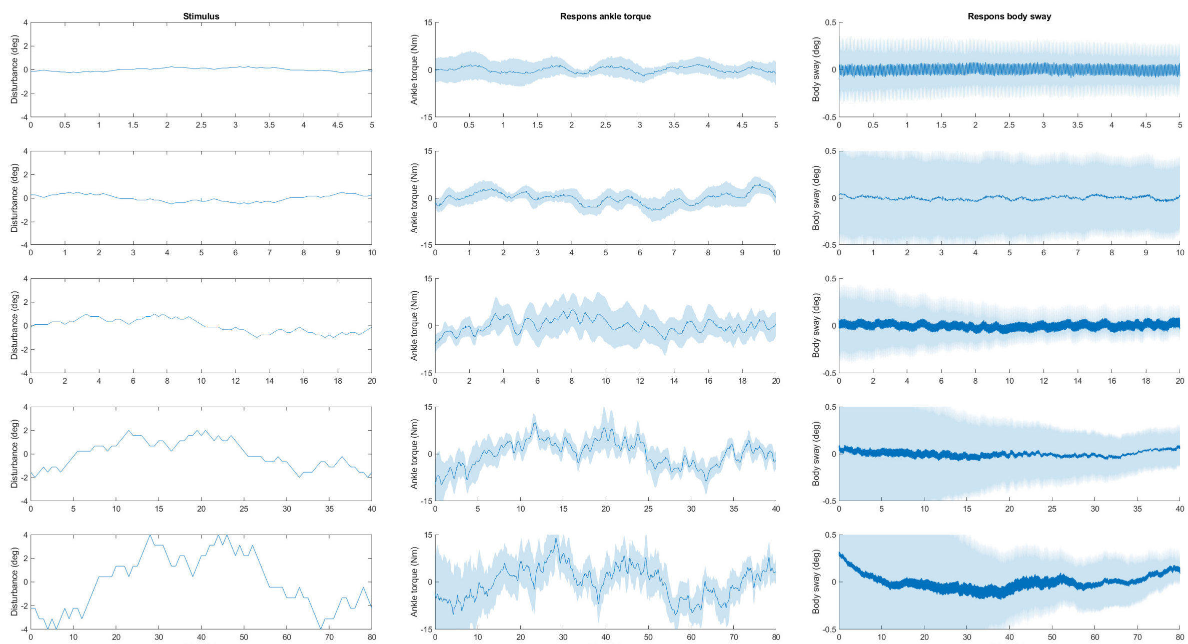


Figure C.1. Sensitivity functions without frequency bands. The sensitivity functions are averaged over the subjects. Figure A shows the sensitivity function of the trials with the same duration. Figure B shows the sensitivity function of the trials with the same velocity.

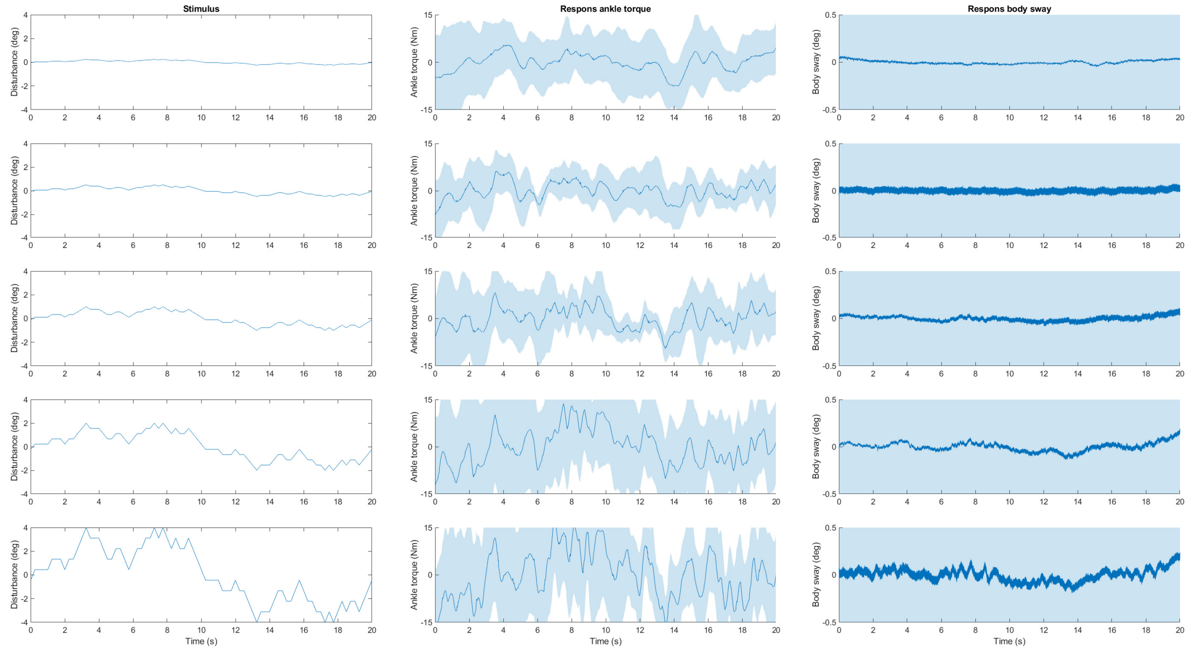


A

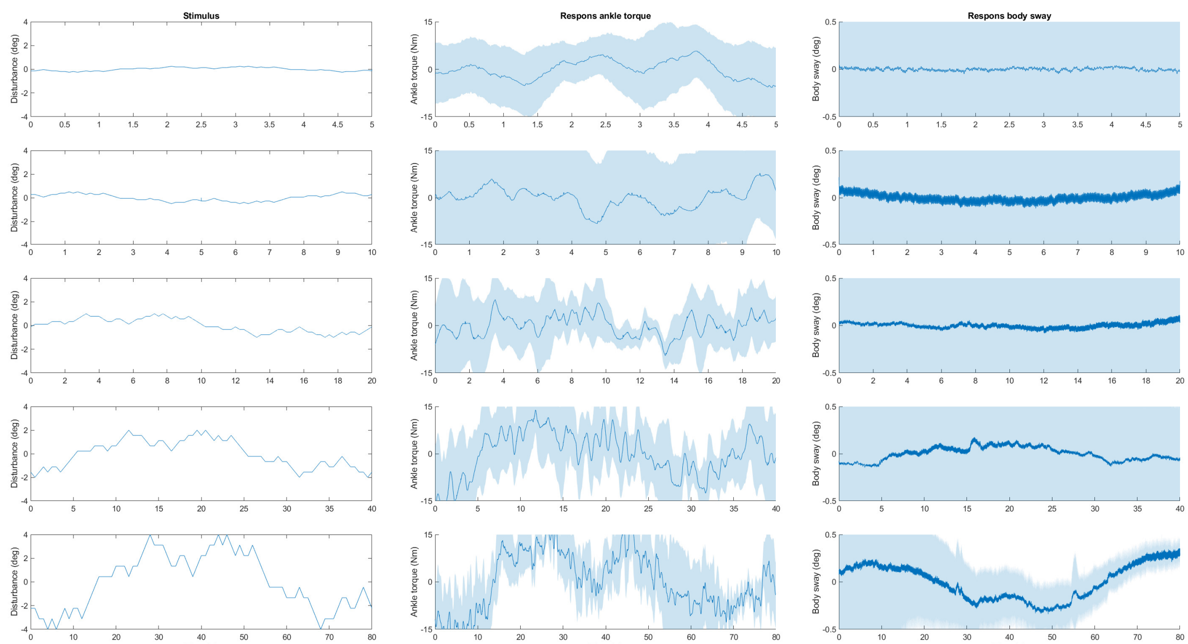


B

Figure C.2. The stimulus response of the ankle torque and body sway of subject 1. The stimulus response is averaged over the trials. The light blue areas are the standard deviations. Figure A shows the stimulus response of the trials with the same duration. Figure B shows the stimulus response of the trials with the same velocity.

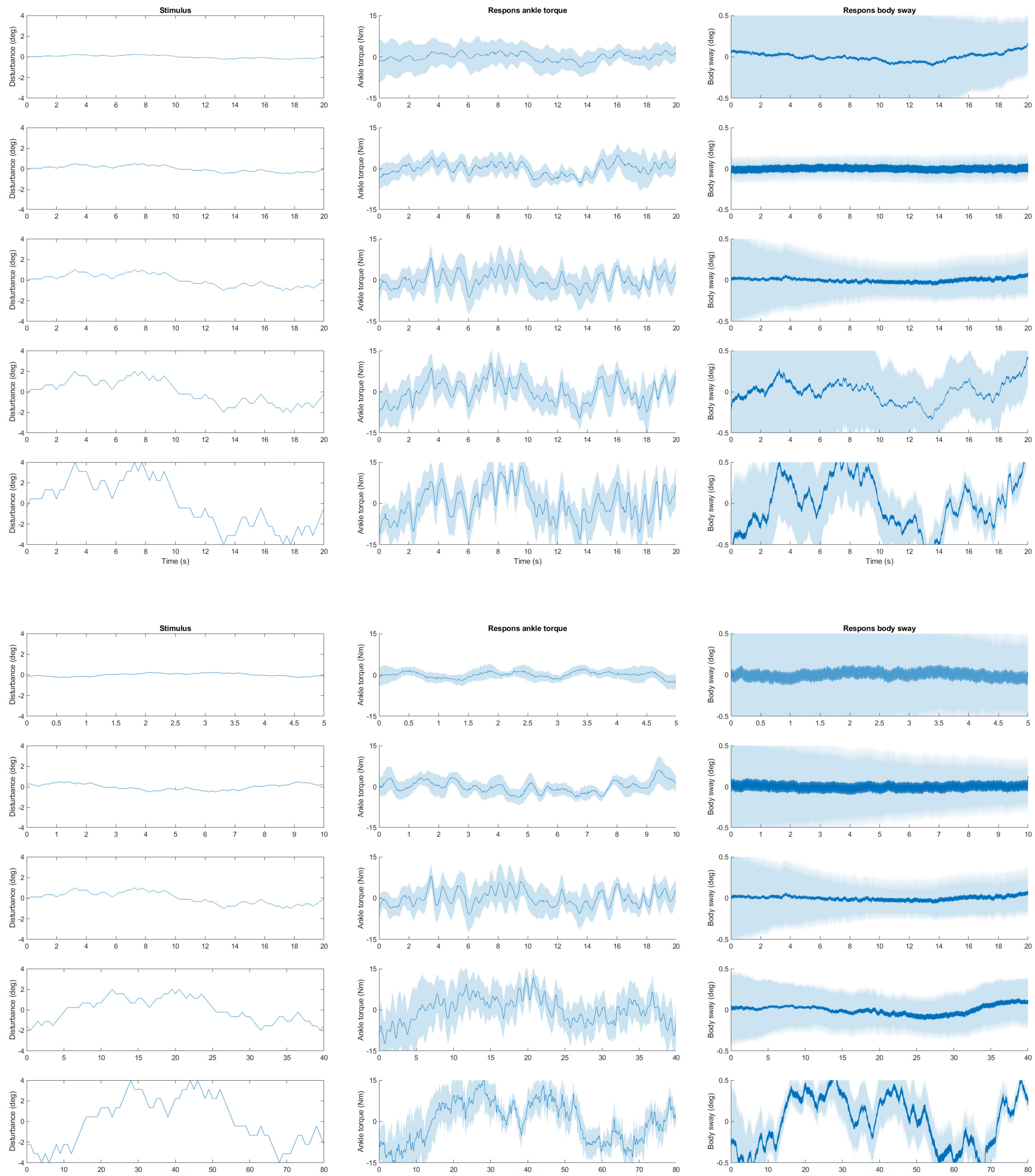


A



B

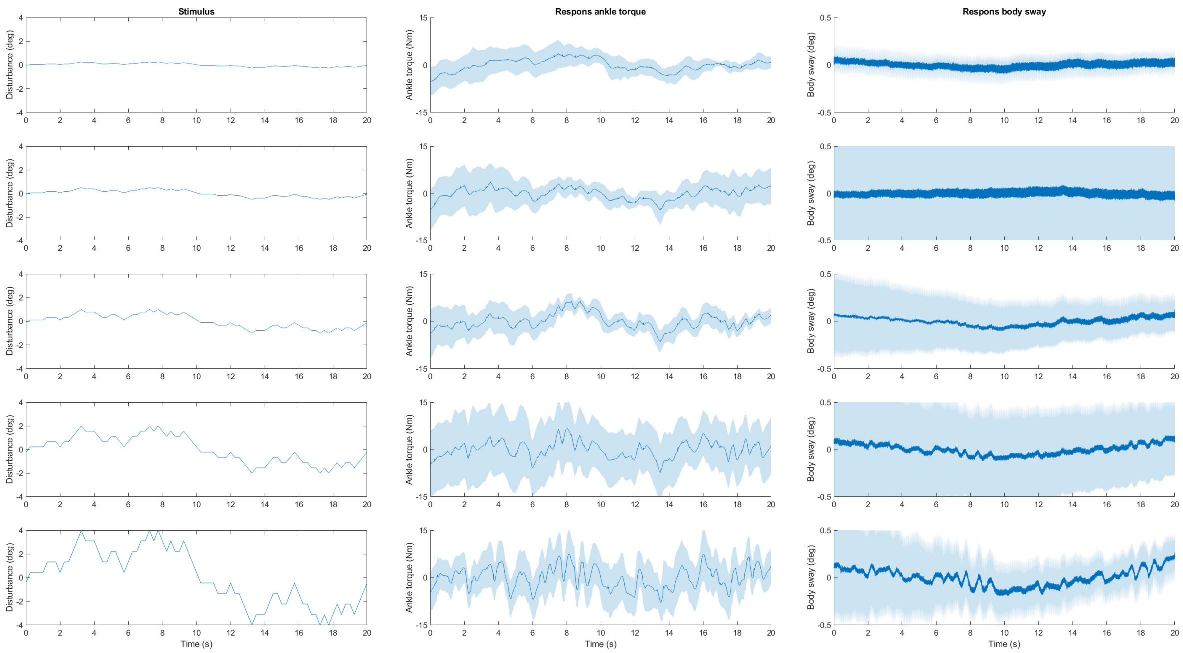
Figure C.3. The stimulus response of the ankle torque and body sway of subject 2. The stimulus response is averaged over the trials. The light blue areas are the standard deviations. Figure A shows the stimulus response of the trials with the same duration. Figure B shows the stimulus response of the trials with the same velocity.



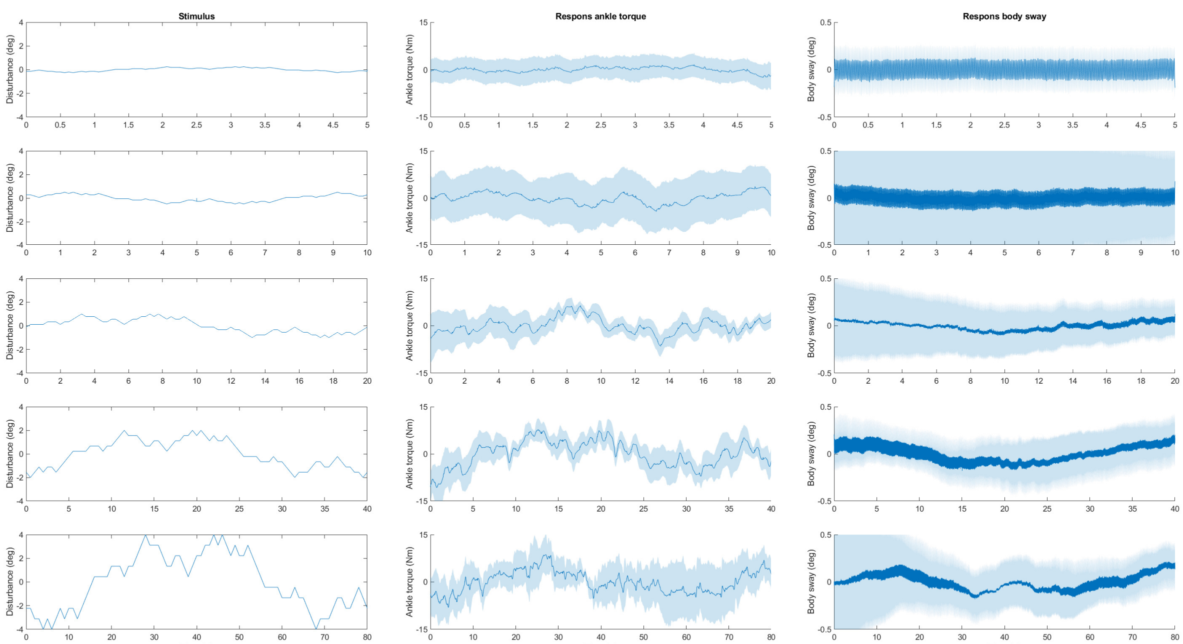
A

B

Figure C.4. The stimulus response of the ankle torque and body sway of subject 3. The stimulus response is averaged over the trials. The light blue areas are the standard deviations. Figure A shows the stimulus response of the trials with the same duration. Figure B shows the stimulus response of the trials with the same velocity.



A



B

Figure C.5. The stimulus response of the ankle torque and body sway of subject 4. The stimulus response is averaged over the trials. The light blue areas are the standard deviations. Figure A shows the stimulus response of the trials with the same duration. Figure B shows the stimulus response of the trials with the same velocity.

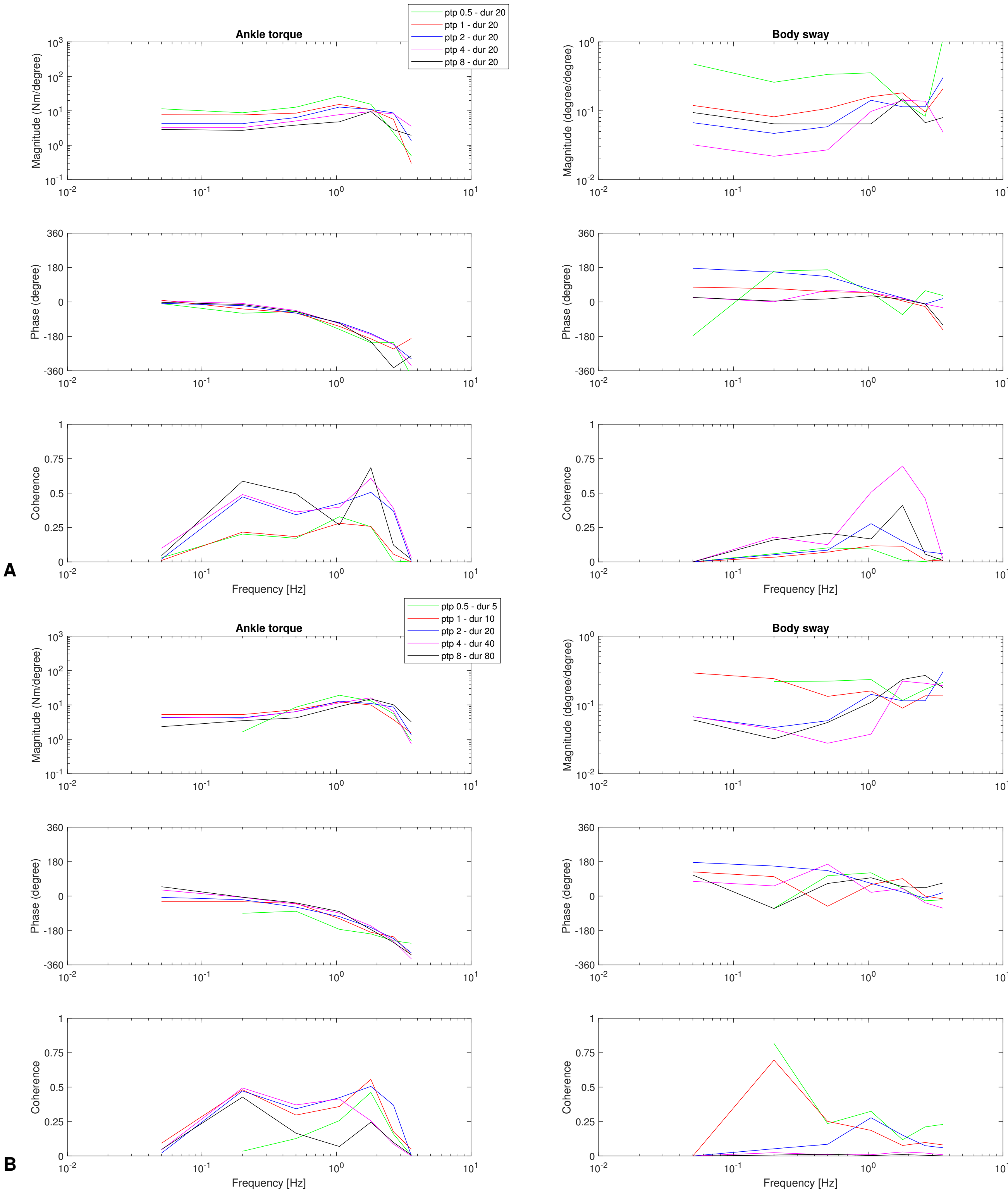


Figure C.6. Sensitivity functions with frequency bands of subject 1. Figure A shows the sensitivity function of the trials with the same duration. Figure B shows the sensitivity function of the trials with the same velocity.

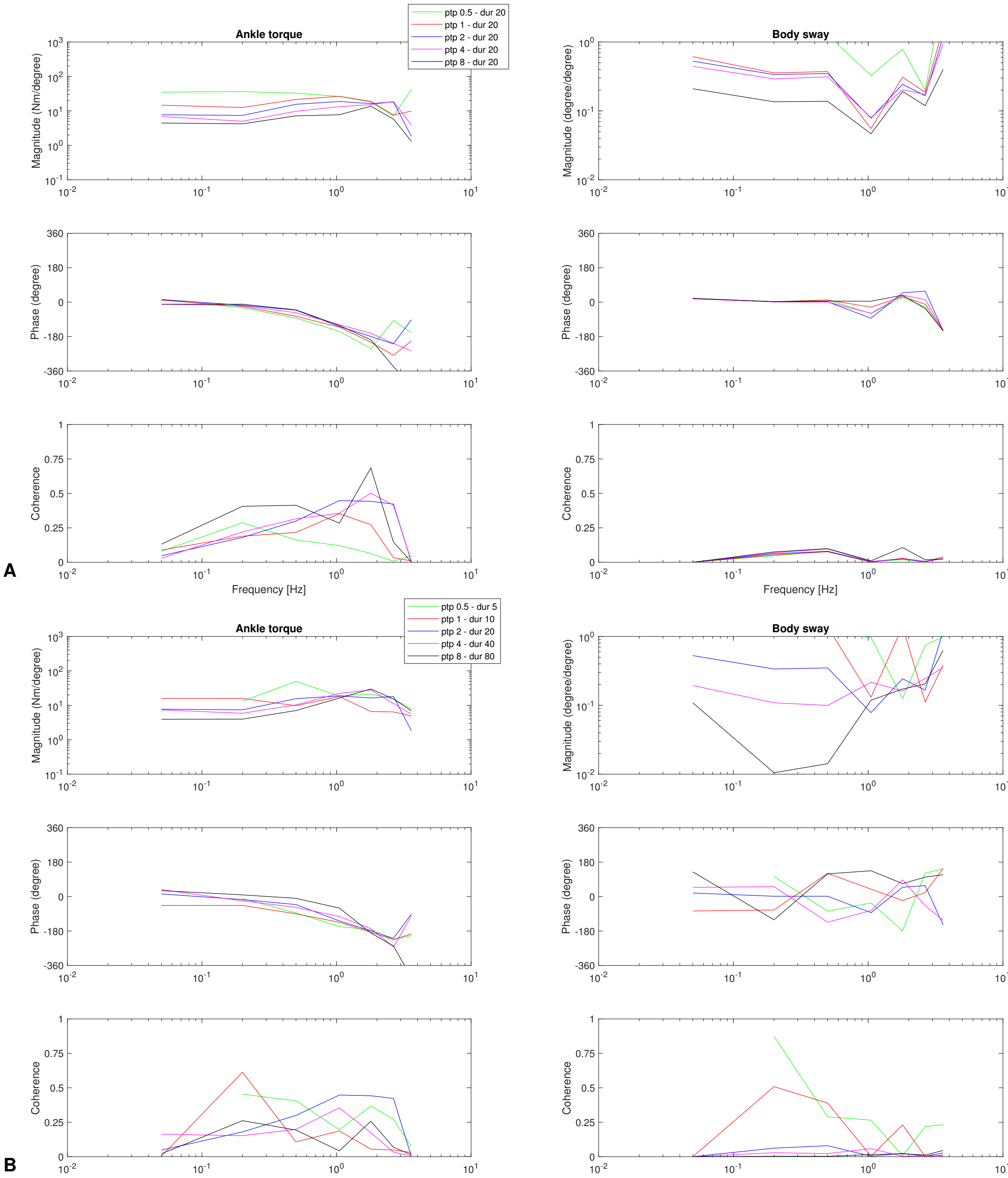


Figure C.7. Sensitivity functions with frequency bands of subject 2. Figure A shows the sensitivity function of the trials with the same duration. Figure B shows the sensitivity function of the trials with the same velocity.

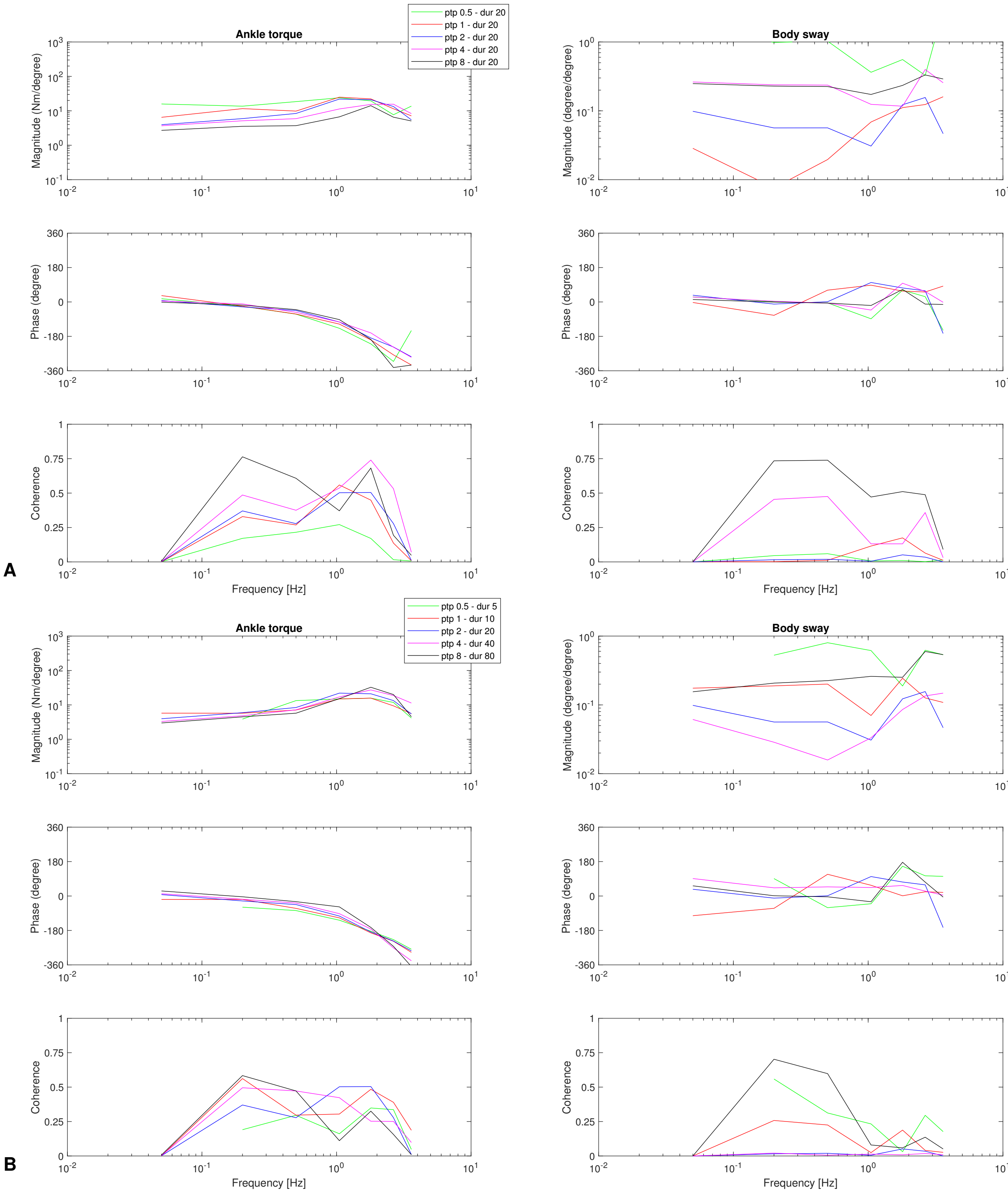


Figure C.8. Sensitivity functions with frequency bands of subject 3. Figure A shows the sensitivity function of the trials with the same duration. Figure B shows the sensitivity function of the trials with the same velocity.

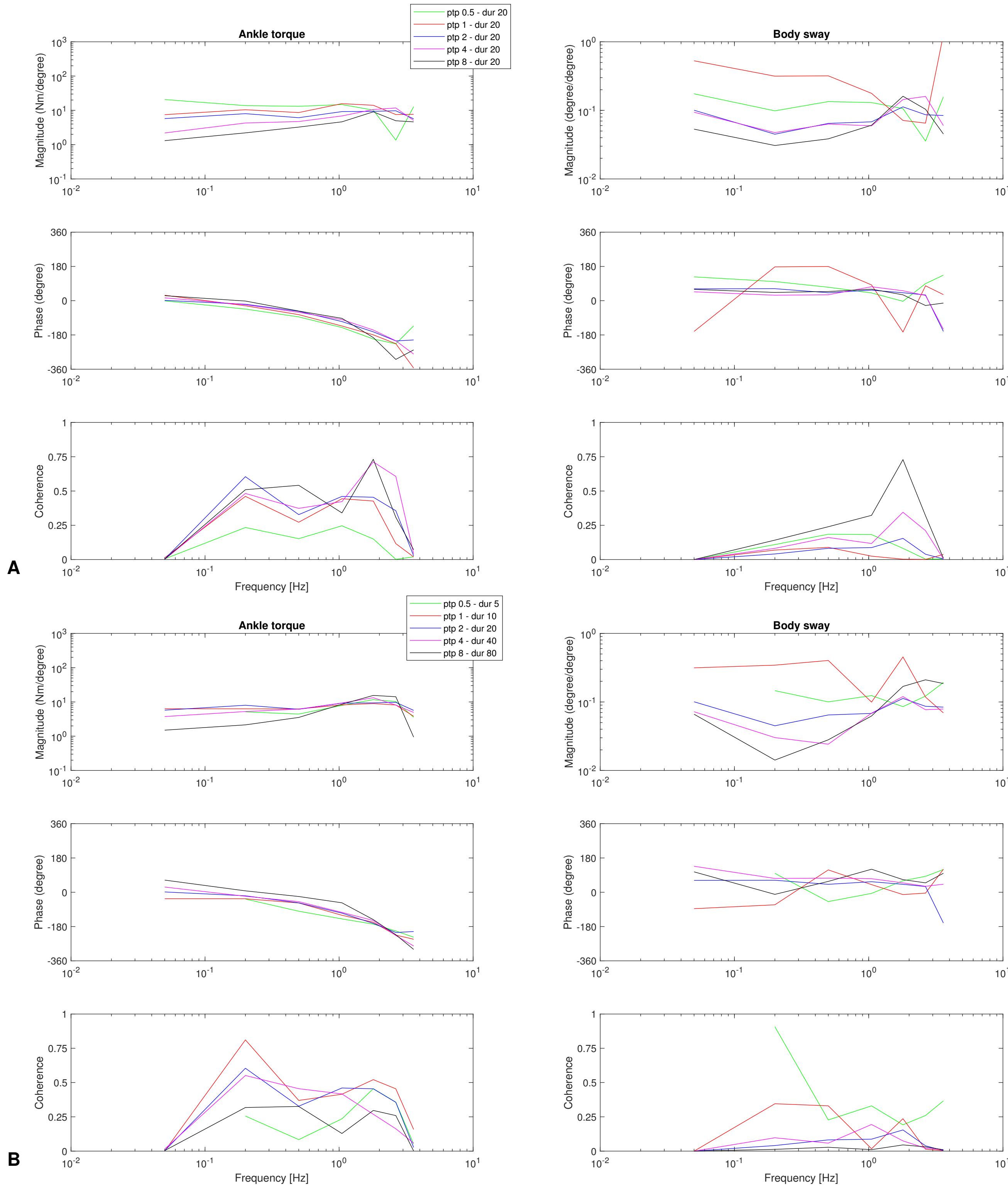


Figure C.9. Sensitivity functions with frequency bands of subject 4. Figure A shows the sensitivity function of the trials with the same duration. Figure B shows the sensitivity function of the trials with the same velocity.