Gait & Posture 53 (2017) 241–247

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Full length article

Compliant support surfaces affect sensory reweighting during balance control

I.M. Schut[a,b,*,1], D. Engelhart[a,1], J.H. Pasma[b], R.G.K.M. Aarts[c], A.C. Schouten[a,b]

[a] Laboratory of Biomechanical Engineering, Institute for Biomedical Technology and Technical Medicine (MIRA), University of Twente, P.O. Box 217, 7500 AE Enschede, The Netherlands
[b] Department of Biomechanical Engineering, Delft University of Technology, Mekelweg 2, 2628 CD Delft, The Netherlands
[c] Department of Mechanical Automation, University of Twente, P.O. Box 217, 7500 AE Enschede, The Netherlands

ARTICLE INFO

Article history:
Received 4 April 2016
Received in revised form 30 January 2017
Accepted 3 February 2017

Keywords:
Balance control
Proprioception
Sensory reweighting
System identification
Compliant support surface

ABSTRACT

To maintain upright posture and prevent falling, balance control involves the complex interaction between nervous, muscular and sensory systems, such as sensory reweighting. When balance is impaired, compliant foam mats are used in training methods to improve balance control. However, the effect of the compliance of these foam mats on sensory reweighting remains unclear.

In this study, eleven healthy subjects maintained standing balance with their eyes open while continuous support surface (SS) rotations disturbed the proprioception of the ankles. Multisine disturbance torques were applied in 9 trials; three levels of SS compliance, combined with three levels of desired SS rotation amplitude. Two trials were repeated with eyes closed. The corrective ankle torques, in response to the SS rotations, were assessed in frequency response functions (FRF). Lower frequency magnitudes (LFM) were calculated by averaging the FRF magnitudes in a lower frequency window, representative for sensory reweighting.

Results showed that increasing the SS rotation amplitude leads to a decrease in LFM. In addition there was an interaction effect; the decrease in LFM by increasing the SS rotation amplitude was less when the SS was more compliant. Trials with eyes closed had a larger LFM compared to trials with eyes open.

We can conclude that when balance control is trained using foam mats, two different effects should be kept in mind. An increase in SS compliance has a known effect causing larger SS rotations and therefore greater down weighting of proprioceptive information. However, SS compliance itself influences the sensitivity of sensory reweighting to changes in SS rotation amplitude with relatively less reweighting occurring on more compliant surfaces as SS amplitude changes.

1. Introduction

Human balance control during stance is continuously challenged by the gravitational field. To maintain an upright posture and prevent falling, balance control involves the complex interaction of nervous, muscular and sensory systems. The central nervous system (CNS) receives feedback about the body orientation from three main sensory systems: the visual, proprioceptive and vestibular system. For each sensory system, the feedback is compared to its reference. The CNS integrates this information and generates an 'error', representing deviations of body orientation from upright stance. The error signal of each sensory system is weighted in relation to its reliability; the CNS prefers reliable over less reliable sensory information within an adaptive weighting process termed sensory reweighting [1–3]. Subsequently, the neural controller (NC) generates with a time delay, a corrective, stabilizing torque by selective activation of muscles. This stabilizing torque (together with a torque caused by the intrinsic dynamics of the muscle properties) keeps the body in upright position.

In elderly and in people with neurological, sensory or orthopedic disorders, balance control might be impaired, leading to postural instability and falls [4,5]. People with impaired balance control often undergo functional balance training that is specifically oriented to improve steadiness while standing on compliant surfaces like foam mats [6,7]. It is assumed that sensory

* Corresponding author at: Department of Biomechanical Engineering, Delft University of Technology, Mekelweg 2, 2628 CD, Delft, The Netherlands.
E-mail address: i.m.schut@tudelft.nl (I.M. Schut).
[1] I.M. Schut and D. Engelhart contributed equally to this work.

http://dx.doi.org/10.1016/j.gaitpost.2017.02.004
0966-6362/

reweighting will be trained using these foam mats, since proprioceptive information is disturbed by the compliant support surface [8]. However, the effect of the compliance of these foam mats on sensory reweighting and balance control remains unclear due to a causality problem. The compliant support surface, i.e. the surface in contact with the feet, might have an effect on sensory reweighting induced by the compliance itself, but on the other hand also might have an effect on sensory reweighting provoked by support surface rotations induced by the compliance. System identification techniques in combination with specifically designed external disturbances provide a way to disentangle cause and effect in balance control. By externally exciting the system with an unique input that is not related to the internal signals of the system, a causal relation between the external disturbances and output signals can be created. This 'opens' the closed loop and generates informative data about a dynamic system such as balance control [9].

In this paper we investigated the effect of compliant support surfaces, comparable to foam mats, on sensory reweighting of proprioceptive information in balance control using system identification techniques, independent of the effect caused by the change in support surface rotation amplitude. Previous studies showed that increasing the amplitude of support surface rotations result in a decrease of the proprioceptive weight (i.e. down weighting), since the proprioception becomes less reliable [1,3,10]. Therefore, we hypothesize that increasing the amplitude of the support surface rotations lowers the reliability of the proprioceptive information and thus results in down weighting of proprioceptive information [11]. Due to the compliance effect of the support surface, an increase in compliance might affect this sensory weighting. If a compliance effect is present, sensory reweighting due to increasing support surface rotation will change for different levels of compliance.

2. Methods

2.1. Subjects

Eleven healthy young volunteers (age:20–30 years, 8 women, weight: 67.4 ± 8.2 kg, height: 1.85 ± 0.09 m), without any history of balance disorders, musculoskeletal injuries or neurological disorders, participated in this study and gave written informed consent prior to participation. The study was performed according to the principles of the Declaration of Helsinki and approved by the Medical Ethics Committee of Medisch Spectrum Twente, Enschede, the Netherlands.

2.2. Apparatus

Fig. 1A shows the Bilateral Ankle Perturbator (BAP, Forcelink B. V., Culemborg, The Netherlands) consisting of two pedals with similar SS rotations around the ankle axis [10]. The BAP was used to mimic compliant support surfaces and to evoke sensory reweighting by rotations of the support surface (SS). The control scheme (Fig. 1B) shows the control of each compliant support surface of the BAP and the implementation in the human balance control scheme as a torque controlled device; the input is a disturbance torque resulting in a disturbance rotation amplitude as output. For each leg, the reference torque ($T_{r,L/R}$) comprises a disturbance torque ($T_{d,L/R}$) and an exerted ankle torque by the human ($T_{h,L/R}$), and is translated via the BAP dynamics to a SS rotation amplitude ($\theta_{ss,L/R}$). The BAP dynamics consist of a virtual inertia (I_v), a virtual damping (B_v) and an adjustable virtual stiffness, from here on called the SS stiffness(K_{SS}). A high SS stiffness mimics stiff (i.e. less compliant) support surfaces and a low SS stiffness mimics more compliant support surfaces.

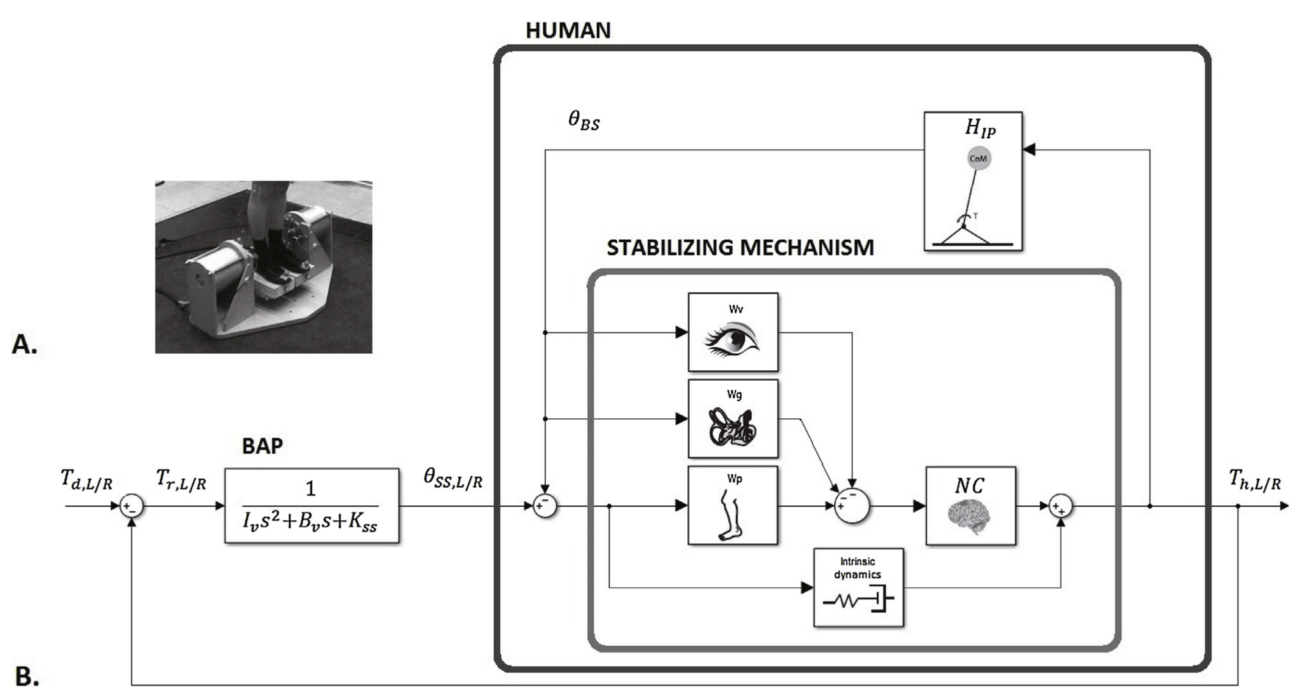

Fig. 1. (A) The Bilateral Ankle Perturbator (BAP) consists of two pedals, each driven by an electromotor. The two pedals together form the support surface (SS) which can be controlled using a SS stiffness such that it mimics standing on foam. Rotating the SS with specific SS rotation amplitudes around the subjects ankle can evoke sensory reweighting of proprioceptive information. (B) The control scheme of each support surface of the Bilateral Ankle Perturbator (BAP) is shown in combination with the human. The BAP dynamics include a virtual inertia $\boldsymbol{I_v}$, a virtual damping $\boldsymbol{B_v}$ and a virtual stiffness (i.e. support surface stiffness $\boldsymbol{K_{ss}}$). The human is represented by the dynamics of the rigid body ($\boldsymbol{H_{IP}}$) and the stabilizing mechanism. The stabilizing mechanism contains the vestibular ($\boldsymbol{W_g}$), visual ($\boldsymbol{W_v}$) and proprioceptive system ($\boldsymbol{W_p}$), the neural controller (NC) (including a time delay) and intrinsic dynamics due to the muscle properties.

2.3. Disturbance signal

A 20-s multisine disturbance was generated with frequencies in the range of 0.05–10 Hz containing 41 logarithmically distributed frequencies. Signal power was constant up to 3 Hz after which signal power decreased exponentially with frequency. The disturbance torque (T_d) was divided into two and applied to both pedals simultaneously (Fig. 2A). Previous studies found that in normal stance, humans lean slightly forward and exert a total ankle torque of approximately 40 Nm [12]. Therefore, an additional torque of 20 Nm was added to the disturbance signal for each pedal of the SS, resulting in normal stance of the subjects when standing on the SS.

2.4. Procedures

Subjects were asked to stand on the BAP, without shoes and their arms crossed over their chest. Subjects wore a safety harness to prevent falling, which did not constrain movements or provided support in any way.

The experiment consisted of eleven trials, each containing 9 repetitions of the disturbance signal resulting in trials of three minutes (9 times 20 s). In the first nine trials, performed with eyes open, a combination of three levels of SS stiffness and three levels of SS rotation amplitude were applied.

The level of SS stiffness was chosen such that a high level ($K_{ss,H}$ = 700 Nm/rad) is comparable to the required stiffness to maintain an upright stance [13], a low level ($K_{ss,L}$ = 100 Nm/rad) is the minimum stiffness on which subjects are able to maintain balance (as determined in a pilot experiment) and a medium level ($K_{ss,M}$ = 300 Nm/rad) is set between the two extremes. Virtual damping and inertia were set to low values (i.e. B_v of 24.2, 76.7 and 117.1 Nms/rad and virtual inertia I_v of 0.2 kgm^2).

Sensory reweighting of proprioceptive information was evoked by applying different levels of disturbance torques (T_d), equal for all subjects. Three levels of disturbance torques were applied to generate three levels of rotations of the SS around the ankle axis (i.e. SS rotations (θ_{ss})). These levels differed for each level of simulated SS stiffness, such that the resulting SS rotations had a

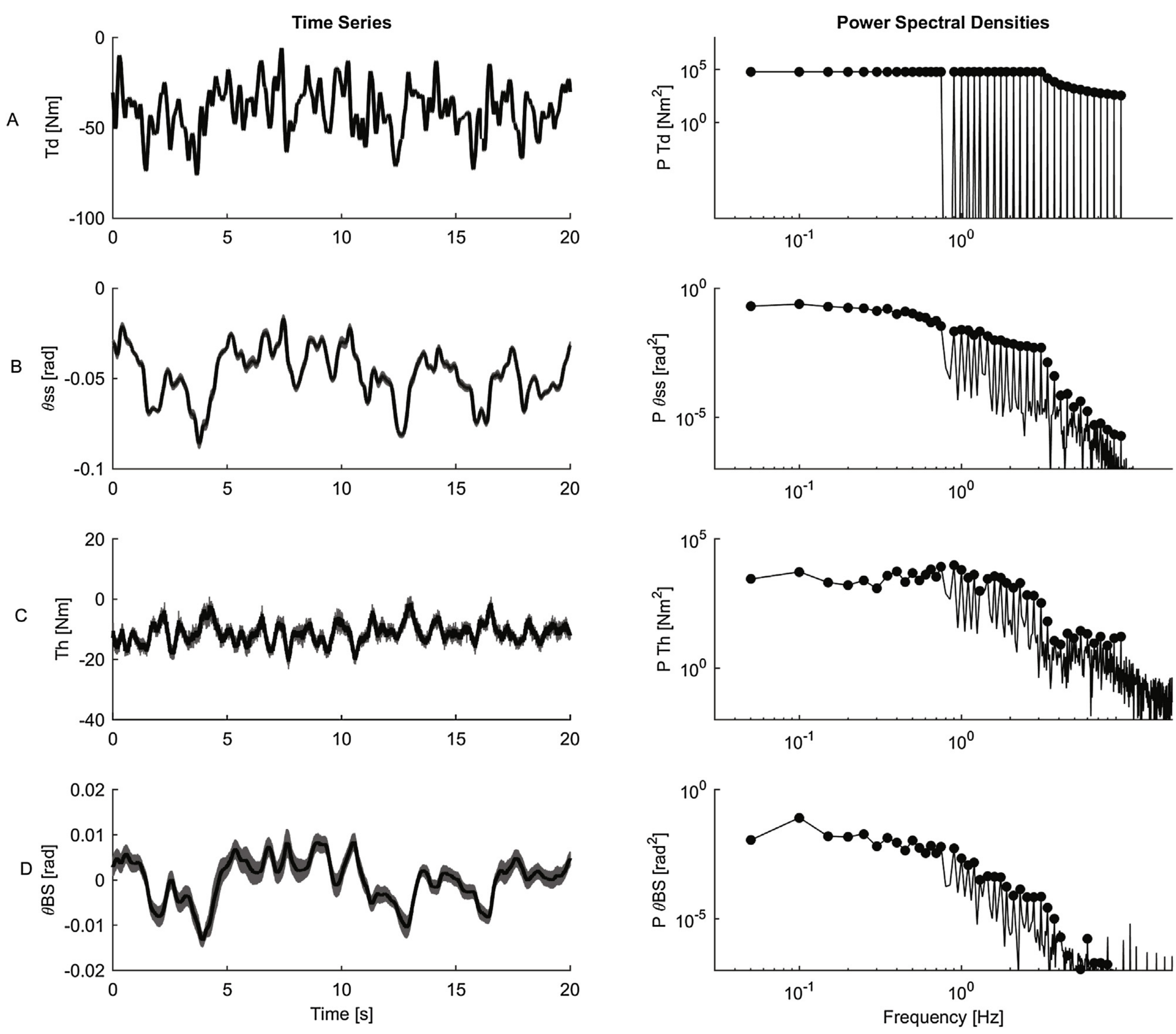

Fig. 2. (Left) Time series of a typical subject for the trial with medium support surface (SS) stiffness and medium SS rotation amplitude, in which the disturbance torque ($\boldsymbol{T_d}$), SS rotation amplitude ($\boldsymbol{\theta_{ss}}$), corrective ankle torque ($\boldsymbol{T_h}$) and body sway angle ($\boldsymbol{\theta_{BS}}$) are shown. The mean of the time series is displayed in black and the standard errors in grey. For display purposes, signals were filtered with a phase preserving fourth order low pass digital Butterworth filter with cut-off frequency of 20 Hz. (Right) The associated power spectral densities are shown with the excited frequencies as dots.

peak-to-peak amplitude of approximately 0.03 ($\theta_{ss,L}$), 0.07 ($\theta_{ss,M}$) and 0.13 rad ($\theta_{ss,H}$). These SS rotation amplitudes were comparable to previous sensory reweighting studies [1,3].

After the nine randomly performed trials with eyes open, the trial with medium SS rotation amplitude ($\theta_{ss,M}$) combined with medium SS stiffness ($K_{ss,M}$) and high SS stiffness ($K_{ss,L}$) were performed with eyes closed resulting in a total of eleven trials. The combination of medium SS rotation amplitude ($\theta_{ss,M}$) with low SS stiffness ($K_{ss,L}$) was not included since pilot experiments indicated that most subjects were unable to perform this condition.

2.5. Data recording and processing

The applied torques to both support surfaces were summed to result in the disturbance torque (T_d). The angles of both SS rotations (i.e. BAP motor angles) were measured and averaged to give the SS rotation amplitude (θ_{ss}). Total corrective ankle torque (T_h) was obtained by the summation of the recorded torques (i.e. BAP motor torques) of both support surfaces. Two draw wire potentiometers (Celesto SP2-25, Celesto, Chatsworth, CA, United States) were attached to the right upper leg and the subjects' trunk, measuring the translations of the upper and lower body segments. All signals were recorded at a sample frequency of 1 kHz and processed in Matlab (The MathWorks, Natick, MA).

The height of the Center of Mass (CoM) was calculated according to the equations of Winter et al. using the measured distance between the ground and ankle joint (lateral malleolus), the ankle and hip joint (greater trochanter), the hip and shoulder joint (acromion), and the subject's height[14]. Body sway angle (θ_{BS}), i.e. angle of the Center of Mass (CoM) with respect to vertical was calculated based on potentiometer data and the height of the CoM.

2.6. Data analysis

The time series were segmented into data blocks of 20 s (i.e. the length of the disturbance signal) after which the first cycle was discarded because of transient effects, resulting in eight data blocks. Subsequently, for each subject and each trial, data were transformed to the frequency domain using the Fourier transform and averaged across the eight data blocks in the frequency domain. Cross spectral densities (CSD) between disturbance torque and body sway were calculated according to

$$\Phi_{T_d\theta_{BS}}(f) = \frac{1}{N} \cdot T_d(f) \cdot \theta_{BS}{}^*(f) \quad (1)$$

in which $T_d(f)$ and $\theta_{BS}{}^*(f)$ represent the Fourier transform of the disturbance torque and body sway. The asterisk indicates the complex conjugate. Similarly, CSDs were calculated between disturbance torque and corrective ankle torque and disturbance torque and SS rotation amplitude. CSDs were used to calculate the frequency response functions (FRFs) (Eqs. (2) and (3)). Two FRFs were estimated using closed-loop system identification methods [1,9]:

$$^{ss}\hat{S}_{Th}(f) = \frac{\Phi_{T_dT_h}(f)}{\Phi_{T_d\theta_{ss}}(f)} \quad (2)$$

$$\hat{H}_{IP}(f) = \frac{\Phi_{T_d\theta_{BS}}(f)}{\Phi_{T_dT_h}(f)} \quad (3)$$

The FRFs were only evaluated on the excited frequencies in the disturbance signal (f) up to 3 Hz. Eq. (2) is the FRF of the torque sensitivity function $\left({}^{ss}\hat{S}_{Th}(f)\right)$ to the disturbance, which describes the dynamic relation between the proprioceptive disturbances (θ_{SS}) and the torque exerted by the ankles (T_h) in terms of amplitude (magnitude) and timing (phase) as function of stimulus frequency [3]. The FRF of the sensitivity function contains dynamics of the rigid body and the stabilizing mechanism which comprises the visual (W_v), proprioceptive (W_p) and vestibular system (W_g), neural controller (NC) (including a time delay) and the intrinsic dynamics (Fig. 1B). The estimated FRF of the sensitivity function $\left({}^{ss}\hat{S}_{Th}(f)\right)$ was normalized for the gravitational stiffness, i.e. participants mass and the distance from the ankles to the CoM multiplied by the gravitational acceleration (mgl_{CoM}), which influences the FRF magnitude. A change in the FRF magnitude implies a relative change of responsiveness to the proprioceptive perturbations, i.e. sensory reweighting. The effects of SS stiffness and SS rotation amplitude on the FRF magnitude are most pronounced at the lower frequencies as the influence of sensory reweighting is most evident at low frequencies where system dynamics are dominated by sensory influences and are minimally affected by other factors such as inertia [1,3,10]. Therefore, the FRF magnitudes were averaged over the five lowest frequencies (0.05–0.25 Hz) resulting in a low frequency magnitude (LFM).

$\hat{H}_{IP}(f)$ (Eq. (3)) is the estimated FRF of the rigid body dynamics, describing the relation between the body sway (θ_{BS}) and the torque exerted by the ankles (T_h) (Fig. 1B). To check whether this FRF was constant across all the trials for each subject and to validate the identification, the experimentally obtained FRF was compared to the theoretical transfer function of the rigid body dynamics ($H_{IP}(s)$), which is representative of an inverted pendulum and can be described with a moment of inertia (I_{IP}), and a gravitational stiffness (mgl_{CoM}):

$$H_{IP}(s) = \frac{1}{I_{IP}s^2 - mgh} \quad (4)$$

In which s denotes the Laplace operator, with $s = i2\pi f$, m is the mass of the subject, h the subjects height of the CoM relative to the ankle and g the gravitational acceleration[2]. Both the inertia and the CoM are derived according to Winter et al. [14].

2.7. Statistical analysis

Linear mixed models were used to statistically compare LFMs of the conditions with different levels of SS rotation amplitude and SS stiffness. SS rotation amplitude, SS stiffness and their interaction were included as covariates and set as fixed effects. The subject was included as a random effect to correct for differences in SS rotation amplitude due to variations in subjects corrective ankle torque and to take the measurement repetitions into account. For illustration purposes, regression lines were plotted between individual LFM and SS rotation amplitude for all levels of SS stiffness, together with the means and standard errors of both LFM and SS rotation amplitudes. To investigate the effect of closing the eyes, similar linear mixed models were used with eyes condition as additional covariate and fixed effect. For all tests, the significance level (α) was set at 0.05. All analyses were performed with SPSS version 22.0 (SPSS, Chicago, IL).

3. Results

3.1. Time series

Fig. 2 shows the time series as response to the disturbances of a typical subject in the medium K_{ss} – medium θ_{ss} condition with eyes open. The standard error of the SS rotation amplitude is relatively low. The responses of body sway angle and corrective ankle torque are slightly more variable. The power spectral densities corresponding to the time series show that the excited frequencies are present in all signals.

3.2. Rigid body dynamics

Fig. 3 shows the mean FRF and standard error of the FRF of the rigid body dynamics ($\hat{H}_{IP}$) of one typical subject, averaged over all trials with eyes open. Other subjects showed similar results. In addition, the theoretical FRF (H_{IP}) according to the transfer function described in the section 'Data Analysis' which is based on the subject's anthropometries, is shown. The experimental FRF shows (up to 3 Hz) similar characteristics as the theoretical FRF although the magnitude is somewhat lower. Also, the standard error of the experimental FRF is low.

3.3. Sensitivity function

Fig. 4 shows the estimated FRFs for each level of SS stiffness describing the sensitivity functions ($\left({}^{ss}\hat{S}_{Th}(f)\right)$), as an average across subjects with the SS rotation amplitude (in trials with eyes open). Fig. 5A shows, for all trials, the corresponding mean LFM and SS rotation amplitude averaged over all subjects with corresponding standard error of both SS rotation amplitude and LFM. In addition the regression line as function of the SS rotation amplitude is shown for each SS stiffness. The SS rotation amplitude is given as Root Mean Square (RMS) to eliminate the effect of outliers on the peak-to-peak amplitude. Due to the torque exerted by the human, the RMS of the SS rotation amplitude of all trials deviated from the three desired SS amplitude levels. Statistical analysis showed a significant interaction effect of SS rotation amplitude and SS stiffness on the LFM ($p < 0.001$) as shown in the slope of the regression lines. The slope of the regression line becomes smaller as the SS stiffness decreases, i.e. less sensory reweighting for a given change in SS amplitude. For each level of SS stiffness, linear mixed models showed a significant main effect of SS rotation amplitude ($p < 0.001$) on the LFM; by increasing SS rotation amplitude the LFM decreases.

Looking at the effect of closing the eyes, mean LFM and standard errors are shown in Fig. 5B. Linear mixed models showed a significant effect of closing the eyes ($p < 0.001$) on the LFM. Closing the eyes resulted in a higher LFM. In addition, there was a significant effect of SS rotation amplitude ($p = 0.005$) on the LFM similar to the previous trials. There was no significant effect of stiffness ($p = 0.088$) and no interaction effect between SS stiffness and SS rotation amplitude ($p = 0.811$).

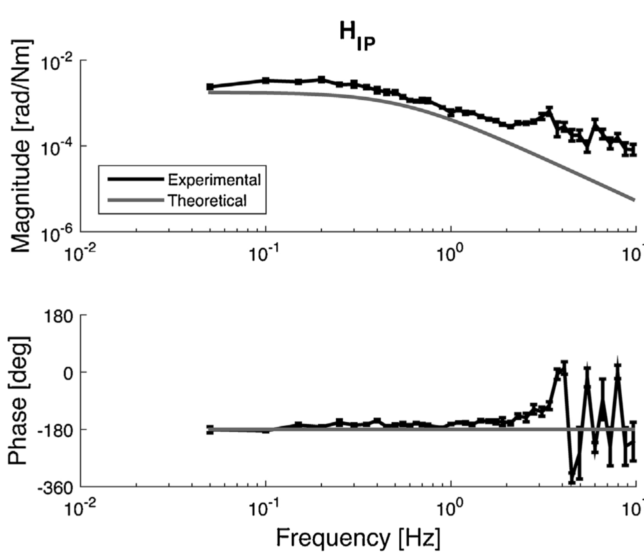

Fig. 3. Frequency Response Function (FRF) of the rigid body dynamics ($\boldsymbol{H_{IP}}$) for one typical subject is shown. The mean and standard error in the nine experimental trials with eyes open are show in black. The theoretical transfer function of the inverted pendulum, based on the body mass and height of the subjects Centre of Mass, is shown in grey.

4. Discussion

4.1. Sensory reweighting

As in previous studies, sensory reweighting is most pronounced at low frequencies where system dynamics are dominated by sensory influences and are minimally affected by other factors such as inertia [1,3,10]. In this study, we found a comparable effect of the SS rotation amplitude on the LFM; by increasing the SS rotation amplitude, the LFM decreased. This indicates that the stabilizing mechanism was down weighting the proprioceptive information accompanied by up weighting the vestibular and/or visual information as proprioceptive information was less reliable.

In addition, the trials with eyes closed showed a significant increase in LFM compared to eyes open conditions. This implies that closing the eyes (i.e. eliminating visual information) results in an up weighting of the proprioceptive information [1,3,10].

4.2. The effect of compliant support surfaces

Standing on compliant support surfaces, such as foam mats, is believed to disturb the proprioceptive information of the ankles by producing a time varying SS angle and making this information less reliable[7]. Due to the existence of the interaction effect of SS rotation amplitude and SS stiffness, the independent effect derived from SS stiffness is hard to interpret.

As for the interaction effect of SS stiffness and SS rotation amplitude, results showed that by decreasing SS stiffness, the slope of the regression line becomes smaller. This means that sensory down weighting of proprioceptive information as a response to increasing SS rotation amplitude is less when standing on more compliant support surfaces. Thus, sensory reweighting is relatively less when the proprioceptive information is already perturbed by a compliant SS independent of the effect of changes in SS rotation amplitude. The overall results of this study imply that when balance control is trained using foam mats, two different effects should be kept in mind. A foam mat with higher compliance (lower stiffness) leads to larger SS rotations and thus down weighting of proprioceptive information. However, the amount of down weighting for a given amount of SS rotation will also depend on the compliance of the foam mat itself.

4.3. Methodological considerations

The experimental FRF shows (up to 3 Hz) similar characteristics as the theoretical FRF although the magnitude is somewhat lower; humans behave more or less as an inverted pendulum. Also, the standard error of the experimental FRF of the rigid body dynamics is low, indicating that the rigid body dynamics do not change over the conditions within a subject. This implies that the changes in the FRF of the human (${}^{ss}\hat{S}_{Th}(f)$) are solely due to changes in the stabilizing mechanism.

In this study we applied disturbance torques, which resulted in SS rotation amplitudes of approximately 0.03, 0.07 and 0.13 rad peak-to-peak. However, the SS rotation amplitude could not be fully controlled due to the closed loop nature of the experimental setup where SS rotation amplitude was determined not only by the applied disturbance torque (which was controlled), but also by the

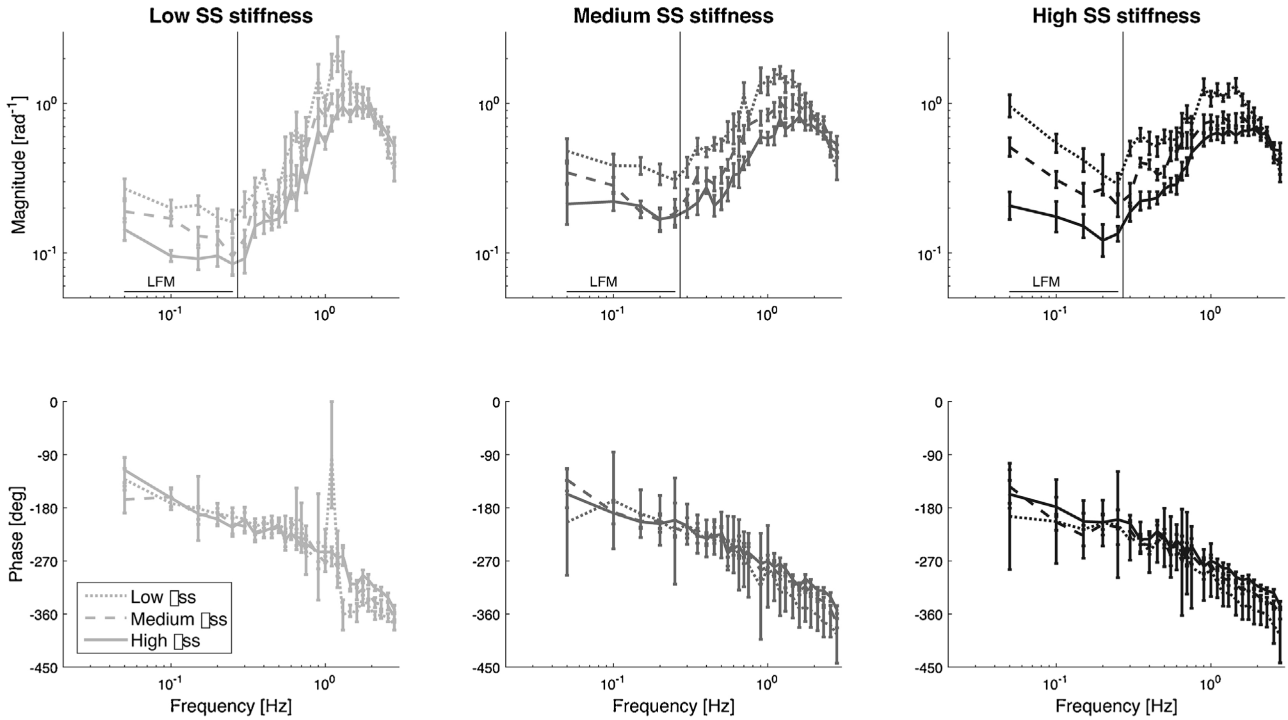

Fig. 4. Mean and standard error of the Frequency Response Functions (FRF) of the sensitivity function (${}^{ss}S_{Th}$), averaged over all subjects, normalized by gravitational stiffness. For each level of support surface (SS) stiffness (low, medium and high $\boldsymbol{K_{ss}}$), the effect of SS rotation amplitude (low, medium and high $\boldsymbol{\theta_{ss}}$) on the FRF is shown. The vertical line indicates the lower frequency window over which statistical analysis is performed. The lower frequency magnitude (LFM) was calculated by averaging the magnitude of the five lowest frequency magnitudes from 0.05 to 0.25 Hz.

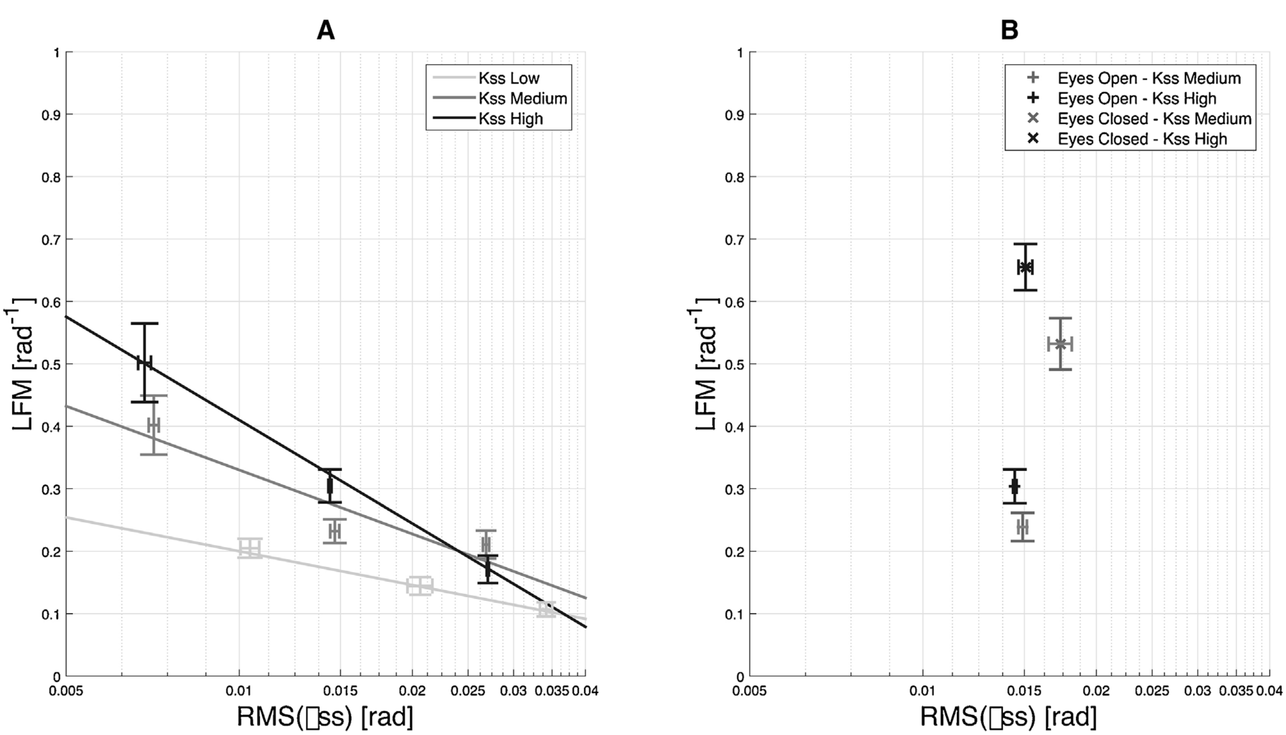

Fig. 5. (A) Mean and standard errors of both lower frequency magnitude (LFM) of the sensitivity function (${}^{ss}S_{Th}$) and support surface (SS) rotation amplitudes, averaged over all subjects for each level of SS stiffness ($\boldsymbol{K_{ss}}$). In addition, three regression lines, fitted on all individual data, are plotted for each level of SS stiffness. (B) Mean and standard errors of both LFM and SS rotation amplitudes, averaged over all subjects as function of SS rotation amplitude for the condition with medium (***KssMedium***) and high (***KssHigh***) SS stiffness combined with medium SS rotation amplitude with eyes closed and open.

torque applied by the subject (which was not experimentally controlled). In contrast to the levels of stiffness, it was therefore not possible to compare the LFM within one level of SS rotation amplitude due to the variabilities within each level of SS rotation amplitude. Therefore, data were statistically analyzed with linear mixed models in which the real SS rotation amplitude was used as an independent variable.

Based on the study of Peterka [1], the assumption was made that changes in the LFM are caused by changes of sensory reweighting. However, the neural controller also might have changed, thereby influencing the LFM.

5. Conclusion

In this study we investigated the effect of compliant support surfaces by changing the level of stiffness on sensory reweighting of proprioceptive information in human balance control during stance using closed loop system identification techniques. Results indicate sensory down weighting of proprioceptive information occurs with increasing SS rotation amplitude, but this sensory down weighting is also affected by the level of SS stiffness resulting in relatively less sensory reweighting on more compliant surfaces. When balance control is trained using compliant foam mats, therapists should be aware that the compliance of the foam mat will influence the relative amount of sensory reweighting and thus perhaps the training effect. It may be advantageous to use foam mats with different compliances to provide more comprehensive exercises of the sensory reweighting mechanism.

Declaration of conflicting interest

The authors whose names are listed above certify that they have NO affiliations with or involvement in any organization or entity with any financial interest (such as honoraria; educational grants; participation in speakers' bureaus; membership, employment, consultancies, stock ownership, or other equity interest; and expert testimony or patent-licensing arrangements), or non-financial interest (such as personal or professional relationships, affiliations, knowledge or beliefs) in the subject matter or materials discussed in this manuscript.

Acknowledgements

This research is supported by the Dutch Technology Foundation STW (NeuroSIPE #10737 BalRoom) which is part of the Netherlands Organisation for Scientific Research (NWO), and which is partly funded by the Ministry of Economic Affairs.

References

[1] R.J. Peterka, Sensorimotor integration in human postural control, J. Neurophysiol. 88 (3) (2002) 1097–1118.
[2] M. Doumas, R.T. Krampe, Adaptation and reintegration of proprioceptive information in young and older adults' postural control, J. Neurophysiol. 104 (4) (2010) 1969–1977.
[3] J.H. Pasma, T.A. Boonstra, S.F. Campfens, A.C. Schouten, H. Van der Kooij, Sensory reweighting of proprioceptive information of the left and right leg during human balance control, J. Neurophysiol. 108 (4) (2012) 1138–1148.
[4] J.H. Pasma, D. Engelhart, A.C. Schouten, H. Van der Kooij, A.B. Maier, C.G.M. Meskers, Impaired standing balance: the clinical need for closing the loop, Neuroscience 267 (2014) 157–165.
[5] D. Engelhart, J.H. Pasma, A.C. Schouten, C.G.M. Meskers, A.B. Maier, T. Mergner, H. Van der Kooij, Impaired standing balance in elderly: a new engineering method helps to unravel causes and effects, J. Am. Med. Dir. Assoc. 15 (3) (2014) e1–6.
[6] P. Page, Sensorimotor training: a 'global' approach for balance training, J. Bodyw. Mov. Ther. 10 (2006) 77–84.
[7] M. Moghadam, H. Ashayeri, M. Salavati, J. Sarafzadeh, K.D. Taghipoo, Saeedi, Reliability of center of pressure measures of postural stability in healthy older adults: effects of postural task difficulty and cognitive load, Gait Posture 33 (4) (2011) 651–655.
[8] A. Shumway-Cook, F.B. Horak, Assessing the influence of sensory interaction on balance – suggestion from the field, Phys. Ther. 66 (10) (1986) 1548–1550.
[9] H. Van der Kooij, E. Van Asseldonk, F.C.T. Van der Helm, Comparison of different methods to identify and quantify balance control, J. Neurosci. Methods 145 (1-2) (2005) 175–203.
[10] A.C. Schouten, T.A. Boonstra, F. Nieuwenhuis, S.F. Campfens, H. Van der Kooij, A bilateral ankle manipulator to investigate human balance control, IEEE Trans. Neural Syst. Rehabil. Eng. 19 (6) (2011) 660–669.
[11] M. Patel, P.A. Fransson, D. Lush, S. Gomez, The effect of foam surface properties on postural stability assessment while standing, Gait Posture 28 (4) (2008) 649–656.
[12] D.A. Winter, A.E. Patla, S. Rietdyk, M.G. Ishac, Ankle muscle stiffness in the control of balance during quiet standing, J. Neurophysiol. 85 (6) (2001) 2630–2633.
[13] M. Casadio, P.G. Morasso, V. Sanguineti, Direct measurement of ankle stiffness during quiet standing: implications for control modelling and clinical application, Gait Posture 21 (4) (2005) 410–424.
[14] D.A. Winter, A.E. Patla, K.S. Rietdyk, M.G. Ishac, Biomechanics and Motor Control of Human Movement, vol. second edition, John Wiley and Sons, Waterloo, 1990.