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# Sensory contributions to stabilization of trunk posture in the sagittal plane



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## ABSTRACT

Trunk stabilization is required to control posture and movement during daily activities. Various sensory modalities, such as muscle spindles, Golgi tendon organs and the vestibular system, might contribute to trunk stabilization and our aim was to assess the contribution of these modalities to trunk stabilization. In 35 healthy subjects, upper-body sway was evoked by continuous unpredictable, force-controlled perturbations to the trunk in the anterior direction. Subjects were instructed to either 'maximally resist the perturbation' or to 'relax but remain upright' with eyes closed. Frequency response functions (FRFs) of admittance, the amount of movement per unit of force applied, and reflexes, the modulation of trunk extensor activity per unit of trunk displacement, were obtained. To these FRFs, we fitted physiological models, to estimate intrinsic trunk stiffness and damping, as well as feedback gains and delays. The different model versions were compared to assess which feedback loops contribute to trunk stabilization. Intrinsic stiffness and damping and muscle spindle (short-delay) feedback alone were sufficient to accurately describe trunk stabilization, but only with unrealistically low reflex delays. Addition of muscle spindle acceleration feedback or inhibitory Golgi tendon organ feedback yielded realistic delays and improved the model fit, with a significantly better model fit with acceleration feedback. Addition of vestibular feedback did not improve the model fit. In conclusion, muscle spindle feedback and intrinsic mechanical properties are sufficient to describe trunk stabilization in the sagittal plane under small mechanical perturbations, provided that muscle spindles encode acceleration in addition to velocity and position information.

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## 1. Introduction

Trunk stabilization can be defined as maintaining control over trunk posture and movement, in spite of the disturbing effects of gravity and mechanical perturbations. In many daily-life activities, the trunk resembles an inverted pendulum that is continuously perturbed by small disturbances, due to reaction forces caused by for example breathing (Smith et al., 2005) or arm movements (Hodges et al., 1999). Accurate control of trunk posture is required, for example, to allow hand movement with some degree of precision (Kaminski et al., 1995; Pigeon et al., 2000). How this stabilization is achieved is to a large extent unknown. Here, we specifically focus on control of an upright, unsupported trunk posture under low-force perturbations, a situation that is representative for many daily activities.

To adapt the degree of precision with which upright trunk posture is controlled, feedback gains appear to be modulated rather than the level of co-contraction (Willigenburg et al., 2010). This suggests dominance of feedback control in this regard and information providing feedback on trunk posture could be derived from several sensory modalities. Manipulation of sensory inputs has demonstrated that proprioceptive, vestibular and visual information can all contribute (Andreopoulou et al., 2015; Goodworth and Peterka, 2009; Maaswinkel et al., 2014; Willigenburg et al., 2012), but such manipulations cannot elucidate whether and how much these modalities contribute in conditions that are representative for daily activities.

We have recently developed a system identification technique, based on unpredictable force disturbances that cause small-amplitude perturbations of trunk posture (<2.5° lumbar rotations), to identify contributions of different sources of feedback to trunk stabilization (van Drunen et al., 2013; van Drunen et al., 2014). This method relies on fitting a physiological model encompassing various sensory feedback loops to frequency responses of trunk

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displacement and trunk muscle electromyography (EMG) to low-amplitude force disturbances. Using this technique, trunk position and velocity encoded by muscle spindles (MS) were found to be the dominant sources of feedback for stabilization of the trunk (van Drunen et al., 2013; van Drunen et al., 2014). However, in these experiments, subjects were required to keep their hands on their head; a posture which likely entailed activity of shoulder and trunk muscles, which would increase the contribution of intrinsic muscle stiffness and damping to stabilization and, consequently, may have masked contributions of other sources of feedback.

Previous studies using system identification have indicated that several other feedback mechanisms may play a role in stabilization of posture in addition to position and velocity feedback by muscle spindles. Inclusion of spindle acceleration feedback (de Vlugt et al., 2006; Schouten et al., 2008a) and of excitatory feedback by Golgi tendon organs (GTO) (Mugge et al., 2010; Mugge et al., 2009) improved the model fit to measured arm admittance. In addition, visual and vestibular feedback were shown to contribute to frontal plane trunk stabilization (Goodworth and Peterka, 2009). However, other studies showed that the visual contribution to control of upright sitting without external perturbations or under anterior directed perturbations is not significant (Maaswinkel et al., 2015; Maaswinkel et al., 2014).

The aim of the present study was to assess the additional contribution to MS velocity and position feedback of acceleration feedback by MS, inhibitory and excitatory force feedback by GTO, and acceleration feedback by the vestibular organ to trunk postural control in an upright, unsupported posture with eyes closed and arms crossed over the chest, during low force perturbations. To this end, models representing various combinations of the mentioned feedback loops in addition to intrinsic mechanical parameters and MS position and velocity feedback were fitted to experimental data. Subsequently, the variance accounted for of trunk displacement and of trunk muscle EMG was assessed. Finally, the effect of feedback parameters on trunk stabilization in selected models was assessed.

## 2. Methods

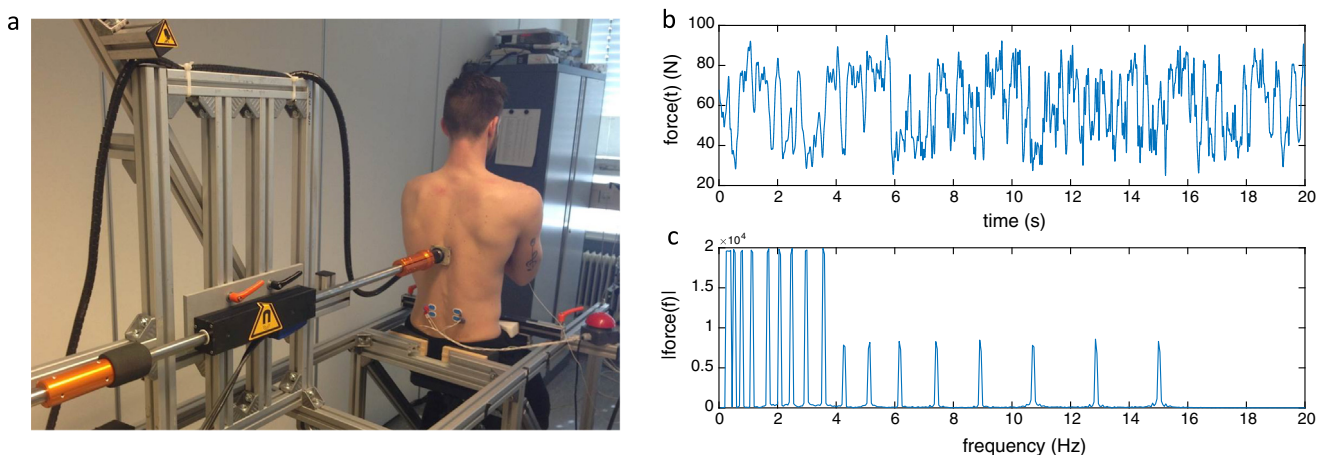
Thirty-five participants (16 females, 19 males; body mass: 70 SD 13 kg; height 1.75 SD 0.09 m; age 36 SD 17 yrs) volunteered for this experiment. Participants did not report any disorders or use of medication that could affect balance control, nor any low-back pain in the 12 months prior to the experiments. All

participants signed an informed consent and the protocol had been approved by the Ethics committee for Movement Sciences at the Vrije Universiteit Amsterdam (ECB 2015-18).

For the measurement, participants were seated in a kneeling-seated posture with their pelvis restrained (Fig. 1). Force perturbations were applied in the anterior direction at the level of the 10th thoracic vertebra by a magnetically driven linear actuator (Servotube STB2510S Forcer and Thrustrod TRB25-1380, Copley Controls, USA), with a thermoplastic patch ( $40 \times 40 \times 3$  mm) placed between the subject and the actuator to improve force transfer and comfort. Subjects were instructed to keep their eyes closed and arms crossed in front of the chest and to 'maximally resist the perturbation' by minimizing flexion/extension excursions (resist task), or to 'relax but remain sitting upright' (relax task). Tasks were performed in random order and repeated three times. Each trial had a duration of 50 s. The perturbation force increased linearly from 0 to 60 N over 3 s, which was followed by a crested multi-sine signal (Pintelon and Schoukens, 2001), containing power at 8 paired frequencies, which were logarithmically distributed within a bandwidth of 0.2–15 Hz, with a peak amplitude of 35 N superimposed on the 60 N baseline preload.

The actuator's target force, displacement and the contact force between the actuator and subject were recorded at 400 samples/s. EMG of the lumbar part of the Longissimus muscle was measured bilaterally (REFA, TMSi, Netherlands), using single-use, adhesive electrodes (Blue Sensor N, Ambu A/S, Ballerup, Denmark) placed over the muscle bellies 3 cm lateral to interspinous space between L4 and L3. The EMG signals were sampled at 2000 samples/s. The Longissimus muscle was chosen because of the high coherence between this muscle's activity and thorax displacement (van Drunen et al., 2013). EMG signals were band-pass filtered 5–400 Hz (2nd order Butterworth), rectified and averaged between left and right muscles.

All data processing and system identification were performed using custom-made software (The Mathworks, Inc., Natick, MA, United States). Closed loop identification (van Drunen et al., 2013; van Drunen et al., 2014) was used to describe the translational low-back admittance and reflexes as frequency response functions (FRFs). The admittance FRF describes the actuator displacement relative to the contact force, representing the inverse of low-back mechanical impedance, or in other words the resistance against the perturbation as a function of frequency. The reflex FRF describes the EMG amplitude modulation of the lumbar part of the Longissimus muscle relative to the actuator displacement. Coherences for admittance and reflex FRFs were



**Fig. 1.** (a) Experimental set-up, showing the linear actuator applying an anterior directed force to the participant's trunk and the EMG electrodes recording lumbar longissimus muscle EMG. (b) Time series of the perturbation force. (c) Fourier transform of the perturbation force, showing a single peak for each pair of closely spaced input frequencies.



**Table 2**

Parameters that were estimated by model fitting of frequency response functions of admittance and reflexes, with lower and upper bounds used in the estimation procedure.

Parameter	Lower bound	Upper bound
m, effective mass (kg)	10	100
$B_C$ , contact damping ( $\text{Nsm}^{-1}$ )	0	12,000
$K_C$ , contact stiffness ( $\text{Nm}^{-1}$ )	5	60,000
$B_{\text{int}}$ , intrinsic damping ( $\text{Nsm}^{-1}$ ) <sup>a</sup>	5	2000
$K_{\text{eff}}$ , intrinsic stiffness – mg, gravity ( $\text{Nm}^{-1}$ ) <sup>a</sup>	–2500	12,000
$K_A$ , MS acceleration gain ( $\text{Ns}^2 \text{m}^{-1}$ ) <sup>a</sup>	–1000	1000
$K_V$ , MS velocity gain ( $\text{Nsm}^{-1}$ ) <sup>a</sup>	0	6000
$K_P$ , MS position gain ( $\text{Nm}^{-1}$ ) <sup>a</sup>	0	12,000
$\tau_{\text{MS}}$ , MS delay (ms)	5	70
$K_{\text{GTO}}$ , GTO gain (dimensionless) <sup>a</sup>	–6 <sup>b</sup>	6
$\tau_{\text{GTO}}$ , GTO delay (ms)	10	100
$K_{\text{VES}}$ , VES gain ( $\text{Ns}^2 \text{m}^{-1}$ ) <sup>a</sup>	0	5000
$\tau_{\text{VES}}$ , VES delay (ms)	80	200

MS = muscle spindle; GTO = Golgi tendon organ; VES = vestibular.

<sup>a</sup> Indicates that parameters were estimated separately for the relax and resist tasks.

<sup>b</sup> Since negative GTO gains often resulted in model instability, we re-estimated the models containing GTO feedback with a lower boundary on GTO gain of 0.

of 7 ms. With inclusion of  $MS_A$ , MS delays were estimated at median values from 36 to 39 ms (Fig. 4).

Inclusion of inhibitory GTO feedback increased reflex VAFs relative to the reference model, but not relative to models including  $MS_A$  (Table 3 and Fig. 3). Inclusion of GTO feedback affected estimates of MS delays in a similar way as inclusion of  $MS_A$ , although estimates were slightly lower with medians from 23 to 36 ms (Fig. 4). Simultaneous identification of  $MS_A$  and inhibitory GTO feedback led to very low GTO delays (Fig. 4).

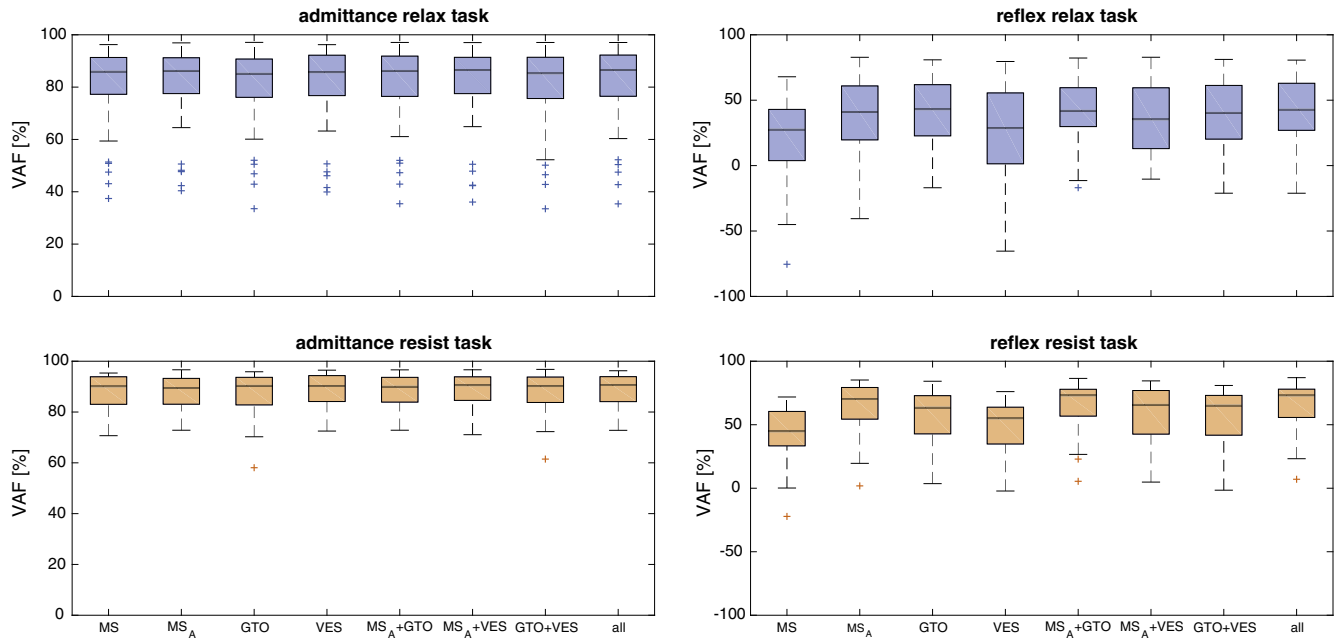
Inclusion of VES feedback improved only the VAF of the admittance in the resist task and these effects were smaller than 1% (Table 3 and Fig. 3). In absence of  $MS_A$  or GTO feedback, VES feedback led to MS delays close to the lower boundary of 5 ms (Fig. 4). In addition, delay estimates for vestibular feedback approached upper boundaries of 200 ms when VES was combined with  $MS_A$  (Fig. 4).

Estimates of other model parameters are shown in [supplementary material](#). Predictably, inclusion of inhibitory GTO feedback

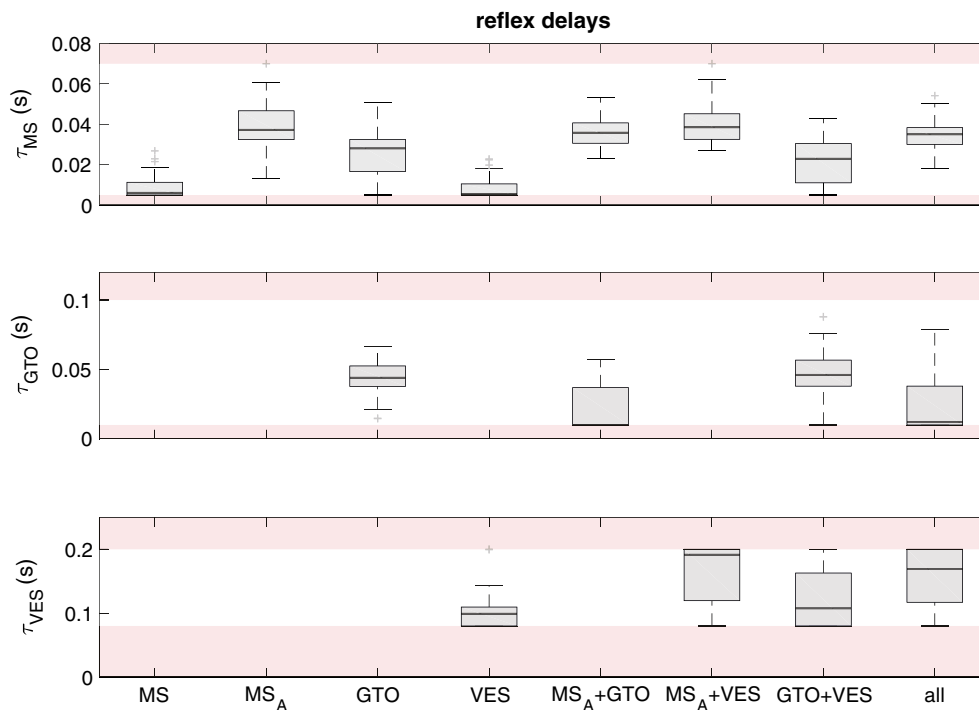
**Table 3**

Overview of significant differences in VAF between models. A + symbol marks a significantly higher VAF of the model in the left column than the model in the first row; a – symbol marks a significantly lower VAF of the model in the left column than the model in the first row and an empty cell indicates no significant difference.

		MS	$MS_A$	GTO	VES	$MS_A$ + GTO	$MS_A$ + VES	GTO + VES	All
$MS_A$	Adm. relax	+		+	+				–
	Adm. resist				–	–			–
	Reflex relax	+			+				
	Reflex resist	+		+	+		+	+	
GTO	Adm. relax		–			–	–	–	–
	Adm. resist				+	+	–		–
	Reflex relax	+			+				
	Reflex resist	+	–		+	–	–		–
VES	Adm. relax		–			–	–		–
	Adm. resist	+	+	–				+	
	Reflex relax		–	–		–	–	–	–
	Reflex resist		–	–		–	–	–	–
$MS_A$ + GTO	Adm. relax	+		+	+			+	–
	Adm. resist		+	–				–	
	Reflex relax				+			+	
	Reflex resist			+	+		+	+	
$MS_A$ + VES	Adm. relax	+		+	+			+	
	Adm. resist			+				+	
	Reflex relax	+			+				
	Reflex resist	+	–	+	+	–			–
GTO + VES	Adm. relax			+		–	–		–
	Adm. resist				–	+	–		–
	Reflex relax	+			+	–			–
	Reflex resist	+	–		+	–			–
All	Adm. relax	+	+	+	+	+		+	
	Adm. resist	+	+	+				+	
	Reflex relax	+			+			+	
	Reflex resist	+		+	+		+	+	



**Fig. 3.** Boxplots of Variance Accounted For (VAF) in the time series of trunk displacement (admittance, left) and EMG amplitude modulation (reflex, right) for the relax (top) and resist (bottom) tasks for all models. Symbols indicate median (red line), interquartile range (blue box), extremes (black whiskers) and outliers (red+). Outliers are defined as larger than  $q_3 + 1.5(q_3 - q_1)$  or smaller than  $q_1 - 1.5(q_3 - q_1)$ , where  $q_1$  and  $q_3$  are the 25th and 75th percentiles (marked by the blue boxes), respectively. See Table 2 for model properties and Table 3 for significant differences between models. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)



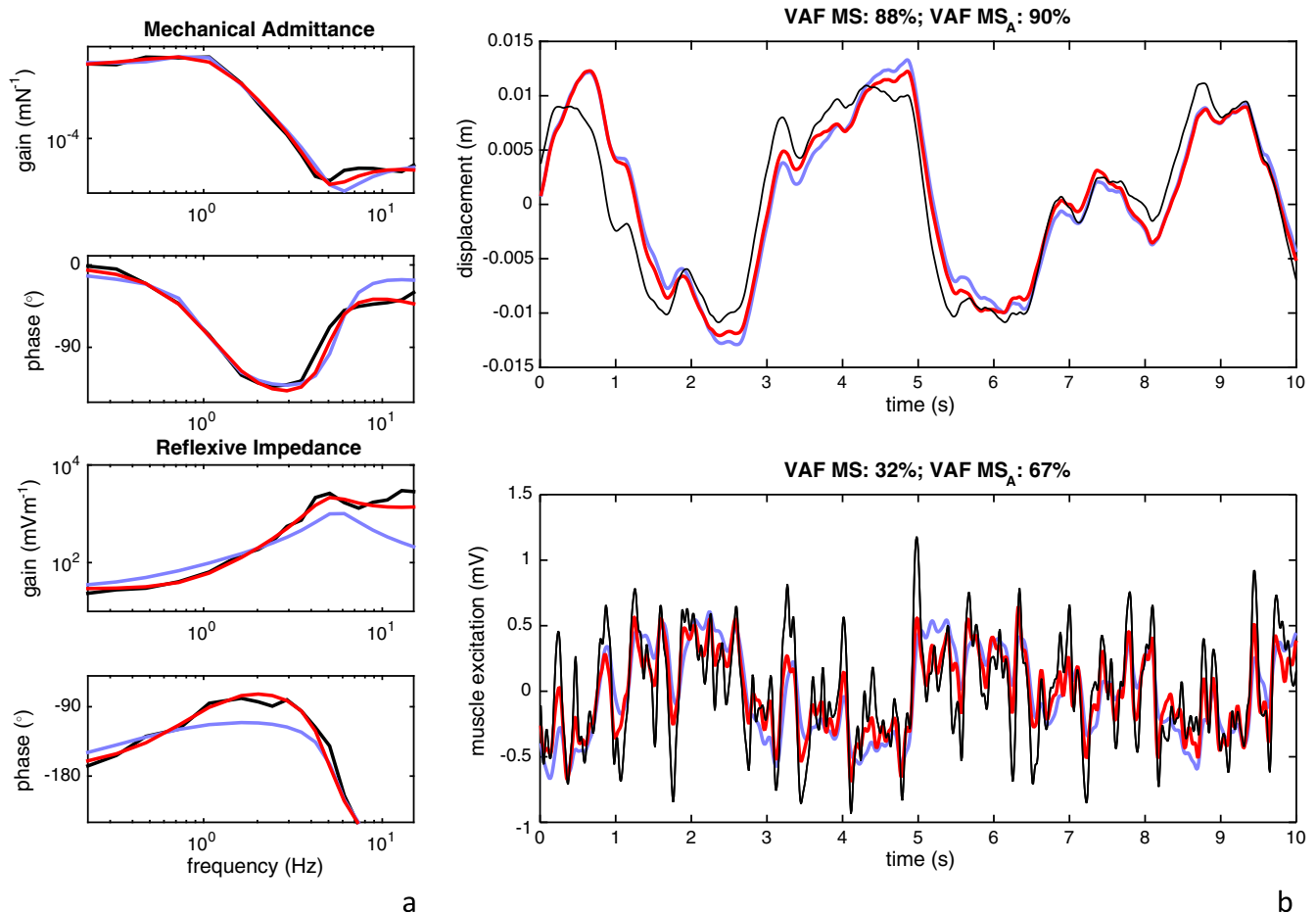
**Fig. 4.** Boxplots of estimates of reflex delays for all models. Symbols indicate median (red line), interquartile range (blue box), extremes (black whiskers) and outliers (red+). Outliers are defined as larger than  $q_3 + 1.5(q_3 - q_1)$  or smaller than  $q_1 - 1.5(q_3 - q_1)$ , where  $q_1$  and  $q_3$  are the 25th and 75th percentiles (marked by the blue boxes), respectively. The shaded areas indicate the boundraies used in the optimization. See Table 2 for model properties. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

caused increases of estimates of MS gains, most notably of MS position feedback (Fig. S.1). Parameters reflecting intrinsic trunk properties and contact dynamics were fairly consistent between models (Fig. S.2). Estimates of effective mass were generally

moderately correlated to, but lower than, anthropometric estimates of upper body mass (Fig. S.3).

In conclusion, substantial improvements in VAFs, without yielding unrealistic delay estimates, were achieved with inclusion of





**Fig. 5.** Typical example of the empirical and modeled FRFs (a) and reconstructed time series (b) of trunk displacement and muscle excitation for the relax task for a single subject. Empirical data (black line), optimized MS model (blue line) and optimized MS<sub>A</sub> model (red line). For comparability to model data empirical time series were Fourier transformed and inverse transformed. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

MS<sub>A</sub> or inhibitory GTO feedback, with the former providing the highest VAFs. Models combining MS<sub>A</sub>, GTO and VES feedback did not lead to clear further improvements in VAF or yielded unrealistic reflex delays. An illustration of the goodness of fit of the MS<sub>A</sub> model is given in Fig. 5.

Comparison of FRFs based on median, 25th and 75th percentiles of the acceleration feedback gain shows that acceleration feedback gain had strong effects on reflex gain and phase above 1 Hz (Fig. 6). However, this did not strongly affect mechanical admittance, due to the inertia of the trunk and the low-pass filtering properties of the activation dynamics ( $H_{act}$ ). While the effect of acceleration feedback on the admittance FRFs was limited, inclusion of acceleration feedback substantially affected estimates of MS feedback delays (Fig. 4). As described above, without acceleration or force feedback, MS delay estimates tended towards the lower boundary of 5 ms, while the median estimate of the MS<sub>A</sub> model was 37 ms. Using this median delay of the MS<sub>A</sub> model in a model without acceleration feedback did appreciably increase admittance around 1 Hz, particularly in the resist task (Fig. 6). This suggests that with feedback delays of around 30 ms, the high-frequency response provided by acceleration feedback is useful to attenuate oscillations in this frequency range.

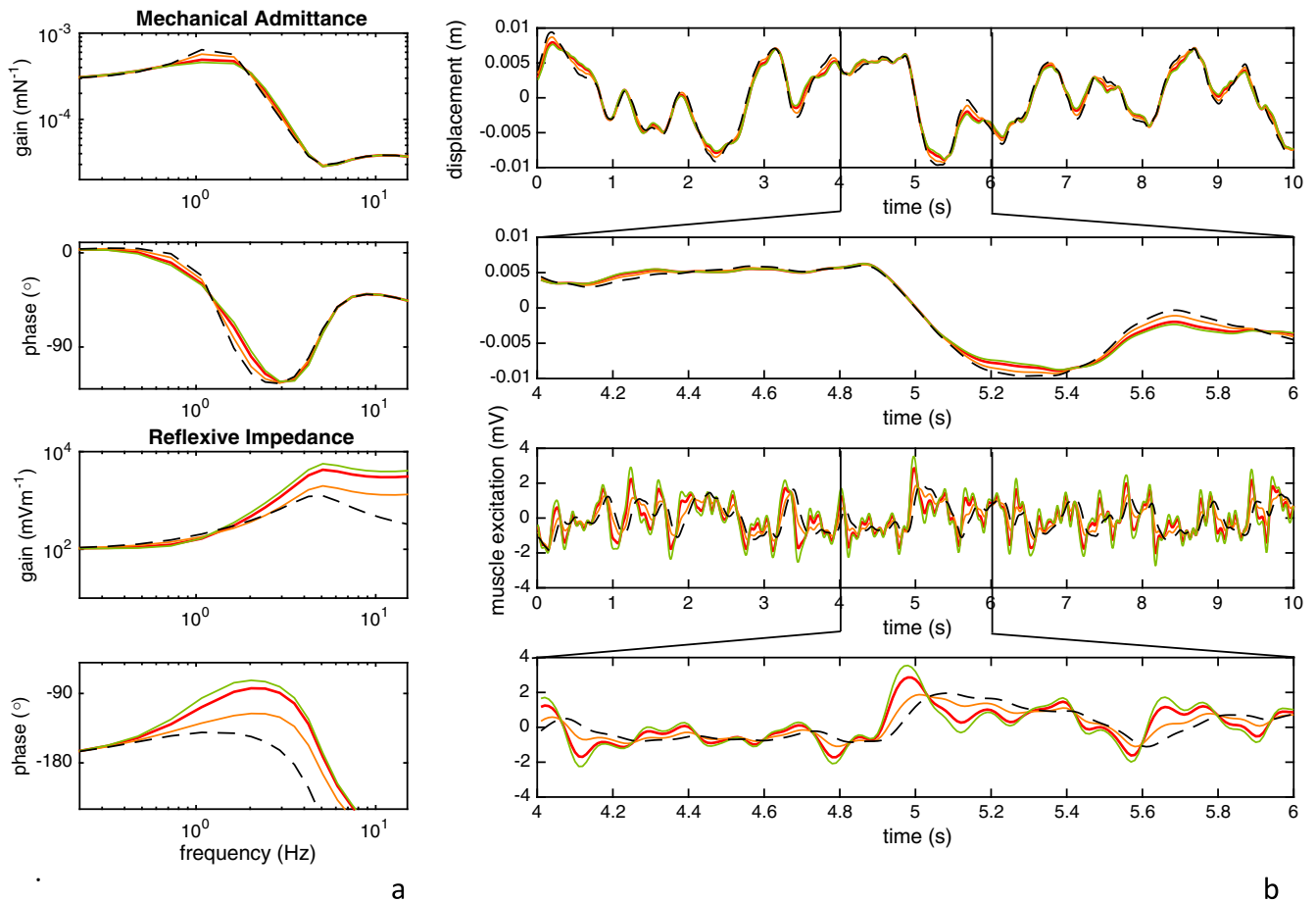
#### 4. Discussion

The aim of this study was to assess contributions of MS acceleration feedback, GTO force feedback, and VES acceleration feedback,

in addition to MS velocity and position feedback, to the stabilization of unsupported upright trunk posture under low-amplitude perturbations. The results indicate that in addition to MS feedback of position and velocity, MS feedback of acceleration (MS<sub>A</sub>) plays a role in trunk stabilization. No indications were found for contributions of excitatory GTO and VES feedback. Inhibitory GTO feedback did improve model fit but to a lesser extent than MS<sub>A</sub> feedback.

Measurements of MS afference support the notion that MS may encode acceleration (Dimitriou and Edin, 2008; Henatsch, 1971; Schafer and Kijewski, 1974) and it has been shown that acceleration feedback would simplify control of upright standing (Insperger et al., 2013). Furthermore, muscle activation patterns following perturbations of whole-body balance were in line with predictions of a control model including acceleration feedback (Welch and Ting, 2008, 2009). The presence of MS<sub>A</sub> was supported here by significantly better VAFs of models with than without MS<sub>A</sub>. The better fit obtained with MS<sub>A</sub> concurs with our previous work (Table 2 in van Drunen et al. (2014)) and was now proven to be significant. Previously, we may have missed effects of MS<sub>A</sub>, due to low statistical power and a high intrinsic stiffness given the posture in which subjects were tested.

Excitatory GTO feedback was found to cause instability of trunk posture. In contrast, excitatory GTO feedback has been reported to contribute to stabilization of arm posture. However, the most pronounced contribution was found when the hand interacted with an object that provided considerable resistance against movement (Mugge et al., 2009). For the trunk, this might resemble control of a supported trunk posture, as when leaning against a chair back,



**Fig. 6.** (a) Frequency response functions of the admittance and reflexes derived from MS<sub>A</sub> models for the resist task. (b) Time domain representations of the model predicted trunk displacement and modulation of muscle excitation for the resist task. In all graphs, thick red lines represent predictions based on median estimates of all model parameters, thin orange and green lines represent model predictions based on the 25th and 75th percentile estimates of acceleration feedback gains, respectively with all other estimates at median values, and black dashed lines represent predictions of an MS model (zero acceleration feedback gain) with the median delay (37.3 ms) of the MS<sub>A</sub> model. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

but this may not generalize to unsupported postures. GTO feedback may also contribute indirectly to position, velocity or acceleration feedback, as it can resolve ambiguities in MS afference that arise from tendon compliance leading to independency between MS length and joint angle (Dimitriou and Edin, 2010; Kistemaker et al., 2013). This aspect of GTO feedback would not be captured by the models applied here.

Inhibitory GTO feedback increased reflex VAFs by a better fit at higher frequencies, but it did not improve admittance VAFs and improvements were smaller than for MS<sub>A</sub> feedback. When lying prone, GTO feedback in the triceps surae muscle is inhibitory; when standing this inhibition is reduced (Faist et al., 2006; Horslen et al., 2017). Possibly the upright posture and the average forward loading in our test conditions minimized GTO gains. Inhibitory GTO feedback acts as a high-pass filter on force output of the muscle (Schouten et al., 2008b). The high-pass filtering effect of inhibitory force feedback requires an increase of MS (or VES) gains proportional to the GTO gain to keep average muscle force constant, as was reflected in the parameter estimates (Fig. S.1). Acceleration feedback also increases the high-frequency content of the muscle force, given the higher frequency content of acceleration compared to velocity and position. Therefore, it cannot be excluded that GTO and acceleration feedback are simultaneously present. The similarity of their effect on muscle force may make it impossible to obtain reliable identification of both simultaneously.

No evidence for a contribution of VES acceleration feedback to trunk stabilization was found in the present study. In contrast,

system identification by Goodworth and Peterka (2009) indicated an important contribution of VES feedback to frontal trunk stabilization, which increased with perturbation amplitude. However, subjects were placed on a moving support surface, while surface instability has been shown to lead to increased VES feedback in frontal plane trunk control (Andreopoulou et al., 2015). Here all tests were performed on a rigid support surface, under small amplitude perturbations, conditions that were chosen as representative for functional tasks. In addition, differences between frontal plane and sagittal plane stabilization may account for this disparity, since our recent work suggested very limited contributions of VES feedback to sagittal plane stabilization even on a moving surface (van Drunen et al., 2015; van Drunen et al., 2016). Finally, Goodworth and Peterka (2009) modelled the VES contributions as velocity feedback, whereas we modelled the VES contribution as acceleration feedback. The vestibular system encodes linear acceleration by otolith afferents and angular velocity by semicircular canal afferents (Angelaki and Cullen, 2008). In previous studies using platform induced perturbations, we found that head rotations were negligible relative to upper body rotations, while head translations were more substantial (van Drunen et al., 2015; van Drunen et al., 2016). This suggests limited input to the semicircular canals compared to the otoliths, especially when considering the detection thresholds of these sub-systems (Sadeghi et al., 2007; Yu et al., 2012).

In the present study, subjects had their eyes closed and visual feedback was ruled out. Previously using the same set-up as used



in the current study no differences were found in FRFs between eyes open and eyes closed conditions (Maaswinkel et al., 2015). However, that study used a resist task only and a visual contribution in a relax task can thus not be excluded. However, removal of visual feedback did not increase trunk sway in upright unsupported and unperturbed sitting (Maaswinkel et al., 2014). Consequently, a substantial contribution of visual feedback in tasks such as tested here appears unlikely.

When no acceleration or GTO feedback is present, feedback delays of position and velocity feedback need to be very short to achieve the admittance observed in the present study. The median feedback delay estimated for the MS model was 7 ms, while including acceleration or GTO feedback it was estimated at 37 and 28 ms, respectively. The latter values are in line with previous system identification estimates (Goodworth and Peterka, 2009; van Drunen et al., 2013; van Drunen et al., 2014). However, EMG responses in trunk extensor muscles after transient perturbations were found to occur at two latencies ranging from 9–20 and 30–50 ms, respectively (Dimitrijevic et al., 1980; Tani et al., 1997; Zedka et al., 1999). So, while shorter delays than estimated with the MS<sub>A</sub> and GTO models cannot be ruled out, the delays estimated with the MS (and MS + VES) model appeared too short. With a reflex delay of around 30 ms, the high-frequency response provided by positive acceleration or negative force feedback are needed to attenuate perturbations with a frequency content between 1 and 2 Hz.

In the models used, trunk movement was described in terms of translation, since we previously showed that the effective rotation point was not well defined and inconsistent over subjects and tasks (van Drunen et al. 2013). However, the destabilizing effect of gravity was taken into account, as lumped with the intrinsic stiffness. The simplification to a translational model thus ignores only the height of the upper body centre of mass and the arm of the actuator above the centre of rotation. The actuator used to perturb trunk posture may have provided some tactile information used to control trunk posture, but this was unlikely to completely efface the effects of other sensory inputs (Maaswinkel et al., 2014). As in many daily life activities, pressure distribution on the seat may have provided a source of sensory information that was not taken into account. Finally, we addressed stabilization in the sagittal plane only. A strength of this study was that we tested a large group of participants with a wide age range (19–67 years). Hence, the results can be considered representative for healthy adults. The results may not generalize to individuals with disorders, for example, changes in sensory weighting for trunk control have been shown in patients with vestibular disorders (Goodworth and Peterka, 2010) and in patients with back pain (Claeys et al., 2011; Willigenburg et al., 2013).

In conclusion, the results of the present study indicate that in addition to MS feedback of position and velocity, MS feedback of acceleration plays a role in sagittal plane trunk stabilization. Despite the finding that effects of changes in acceleration gains on kinematics were limited due to the low-pass filtering properties of the series elasticity in the trunk muscle-joint system and of trunk inertia, they were deemed relevant, since at perturbation frequencies between 1 and 2 Hz position and velocity feedback appeared too slow. This would imply functional relevance of acceleration feedback, since admittance is maximal in this frequency range.

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## Conflict of interest

The authors have no conflict of interest to report.

## Appendix A. Supplementary material

Supplementary data associated with this article can be found, in the online version, at <http://dx.doi.org/10.1016/j.jbiomech.2017.07.016>.

## References

- Andreopoulou, G., Maaswinkel, E., Cofre Lizama, L.E., van Dieën, J.H., 2015. Effects of support surface stability on feedback control of trunk posture. *Exp. Brain Res.* 233, 1079–1087.
- Angelaki, D.E., Cullen, K.E., 2008. Vestibular system: the many facets of a multimodal sense. *Annu. Rev. Neurosci.* 31, 125–150.
- Bobet, J., Norman, R.W., 1990. Least-squares identification of the dynamic relation between the electromyogram and joint moment. *J. Biomech.* 23, 1275–1276.
- Claeys, K., Brumagne, S., Dankaerts, W., Kiers, H., Janssens, L., 2011. Decreased variability in postural control strategies in young people with non-specific low back pain is associated with altered proprioceptive reweighting. *Eur. J. Appl. Physiol.* 111, 115–123.
- de Vlugt, E., Schouten, A.C., van der Helm, F.C.T., 2006. Quantification of intrinsic and reflexive properties during multijoint arm posture. *J. Neurosci. Methods* 155, 328–349.
- Dimitrijevic, M.R., Gregoric, M.R., Sherwood, A.M., Spencer, W.A., 1980. Reflex responses of paraspinal muscles to tapping. *J. Neurol. Neurosurg. Psychiatry* 43, 1112–1118.
- Dimitriou, M., Edin, B.B., 2008. Discharges in human muscle receptor afferents during block grasping. *J. Neurosci.* 28, 12632–12642.
- Dimitriou, M., Edin, B.B., 2010. Human muscle spindles act as forward sensory models. *Curr. Biol.* 20, 1763–1767.
- Faist, M., Hoefer, C., Hodapp, M., Dietz, V., Berger, W., Duysens, J., 2006. In humans lb facilitation depends on locomotion while suppression of lb inhibition requires loading. *Brain Res.* 1076, 87–92.
- Goffe, W.L., 1996. SIMANN: a global optimization algorithm using simulated annealing. *Stud. Nonlin. Dyn. Econom.* 1, 169–176.
- Goodworth, A.D., Peterka, R.J., 2009. Contribution of sensorimotor integration to spinal stabilization in humans. *J. Neurophysiol.* 102, 496–512.
- Goodworth, A.D., Peterka, R.J., 2010. Influence of bilateral vestibular loss on spinal stabilization in humans. *J. Neurophysiol.* 103, 1978–1987.
- Henatsch, D., 1971. Pro und Contra zur Beschleunigungsempfindlichkeit der Muskelspindeln. *Bull. Schweiz. Akad. Med. Wiss.* 27, 266–281.
- Hodges, P.W., Cresswell, A.G., Thorstensson, A., 1999. Preparatory trunk motion accompanies rapid upper limb movement. *Exp. Brain Res.* 124, 69–79.
- Horslen, B.C., Inglis, J.T., Blouin, J.S., Carpenter, M.G., 2017. Both standing and postural threat decrease Achilles tendon reflex inhibition from tendon electrical stimulation. *J. Physiol.*
- Inspersperger, T., Milton, J., Stepan, G., 2013. Acceleration feedback improves balancing against reflex delay. *J. R. Soc. Interface/R. Soc.* 10, 20120763.
- Kaminski, T.R., Bock, C., Gentile, A.M., 1995. The coordination between trunk and arm motion during pointing movements. *Exp. Brain Res.* 106, 457–466.
- Kistemaker, D.A., Van Soest, A.J., Wong, J.D., Kurtzer, I., Gribble, P.L., 2013. Control of position and movement is simplified by combined muscle spindle and Golgi tendon organ feedback. *J. Neurophysiol.* 109, 1126–1139.
- Maaswinkel, E., van Drunen, P., Veeger, H.E., van Dieën, J.H., 2015. Effects of vision and lumbar posture on trunk neuromuscular control. *J. Biomech.* 48, 298–303.
- Maaswinkel, E., Veeger, H.E., Dieën, J.H., 2014. Interactions of touch feedback with muscle vibration and galvanic vestibular stimulation in the control of trunk posture. *Gait Posture* 39, 745–749.
- Mugge, W., Abbink, D.A., Schouten, A.C., Dewald, J.P., van der Helm, F.C., 2010. A rigorous model of reflex function indicates that position and force feedback are flexibly tuned to position and force tasks. *Exp. Brain Res.* 200, 325–340.
- Mugge, W., Schuurmans, J., Schouten, A.C., van der Helm, F.C., 2009. Sensory weighting of force and position feedback in human motor control tasks. *J. Neurosci.* 29, 5476–5482.
- Pigeon, P., Yahia, L.H., Mitnitski, A.B., Feldman, A.G., 2000. Superposition of independent units of coordination during pointing movements involving the trunk with and without visual feedback. *Exp. Brain Res.* 131, 336–349.
- Pintelon, R., Schoukens, J., 2001. *System Identification: A Frequency Domain Approach*. John Wiley & Sons, New York, USA.
- Sadeghi, S.G., Chacron, M.J., Taylor, M.C., Cullen, K.E., 2007. Neural variability, detection thresholds, and information transmission in the vestibular system. *J. Neurosci.* 27, 771–781.

- Schafer, S.S., Kijewski, S., 1974. The dependency of the acceleration response of primary muscle spindle endings on the mechanical properties of the muscle. *Pflügers Arch.* 350, 101–122.
- Schouten, A.C., De Vlugt, E., Van Hilten, J.J.B., Van der Helm, F.C.T., 2008a. Quantifying proprioceptive reflexes during position control of the human arm. *IEEE Trans. Biomed. Eng.* 55, 311–321.
- Schouten, A.C., Mugge, W., van der Helm, F.C.T., 2008b. NMCLab, a model to assess the contributions of muscle visco-elasticity and afferent feedback to joint dynamics. *J. Biomech.* 41, 1659–1667.
- Smith, M., Coppieters, M.W., Hodges, P.W., 2005. Effect of experimentally induced low back pain on postural sway with breathing. *Exp. Brain Res.* 166, 109–117.
- Tani, T., Yamamoto, H., Ichimiya, M., Kimura, J., 1997. Reflexes evoked in human erector spinae muscles by tapping during voluntary activity. *Electroencephalogr. Clin. Neurophysiol.* 105, 194–200.
- van der Helm, F.C., Schouten, A.C., de Vlugt, E., Brouwn, G.G., 2002. Identification of intrinsic and reflexive components of human arm dynamics during postural control. *J. Neurosci. Methods* 119, 1–14.
- van Drunen, P., Koumans, Y., Van der Helm, F.C., van Dieën, J.H., Happee, R., 2015. Modulation of intrinsic and reflexive contributions to low-back stabilization due to vision, task instruction, and perturbation bandwidth. *Exp. Brain Res.* 233, 645–749.
- van Drunen, P., Maaswinkel, E., van der Helm, F.C., van Dieën, J.H., Happee, R., 2013. Identifying intrinsic and reflexive contributions to low-back stabilization. *J. Biomech.* 46, 1440–1446.
- van Drunen, P., Maaswinkel, E., van der Helm, F.C.T., van Dieën, J.H., Happee, R., 2014. Corrigendum to “Identifying intrinsic and reflexive contributions to low-back stabilization” [*J. Biomech.* 46(8) (2013) 1440–1446]. *J. Biomech.* 47, 1928–1929.
- van Drunen, P., van der Helm, F.C., van Dieën, J.H., Happee, R., 2016. Trunk stabilization during sagittal pelvic tilt: from trunk-on-pelvis to trunk-in-space due to vestibular and visual feedback. *J. Neurophysiol.* 115, 1381–1388.
- Welch, T.D., Ting, L.H., 2008. A feedback model reproduces muscle activity during human postural responses to support-surface translations. *J. Neurophysiol.* 99, 1032–1038.
- Welch, T.D., Ting, L.H., 2009. A feedback model explains the differential scaling of human postural responses to perturbation acceleration and velocity. *J. Neurophysiol.* 101, 3294–3309.
- Willigenburg, N.W., Kingma, I., Hoozemans, M.J., van Dieën, J.H., 2013. Precision control of trunk movement in low back pain patients. *Hum. Mov. Sci.* 32, 228–239.
- Willigenburg, N.W., Kingma, I., van Dieën, J.H., 2010. How is precision regulated in maintaining trunk posture? *Exp. Brain Res.* 203, 39–49.
- Willigenburg, N.W., Kingma, I., van Dieën, J.H., 2012. Precision control of an upright trunk posture in low back pain patients. *Clin. Biomech.* 27, 866–871.
- Yu, X.J., Dickman, J.D., Angelaki, D.E., 2012. Detection thresholds of macaque otolith afferents. *J. Neurosci.* 32, 8306–8316.
- Zedka, M., Prochazka, A., Knight, B., Gillard, D., Gauthier, M., 1999. Voluntary and reflex control of human back muscles during induced pain. *J. Physiol.* 520, 591–604.