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Role of trunk inertia in non-stepping balance recovery

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1 Introduction

Previous research has identified two major non-stepping strategies used to recover balance following mechanical perturbations: ankle and hip strategy [1,2]. These strategies are selected depending on e.g. the perturbation magnitude, prior experience, and configuration of the support surface [2] in order to control the posture (upright trunk and leg orientation) and angular momentum [3, 4]. Following an external mechanical perturbation, both body posture and angular momentum depend, in part, on passive properties of the body, such as the amount and distribution of mass. Simple mechanical models, like the inverted pendulum (IP) [4, 5] or the double IP [6] suggest an approximately linear inverse relationship between the inertia of a perturbed body segment and the resultant acceleration and, presumably, also the segment deflection.

However, our recent perturbation experiments have yielded surprising results: when fixed-magnitude impulse-like moments were applied to the trunks of subjects with differing body mass, the resulting angular deflection of the trunk did not appear to be correlated with the body mass (or moment of inertia). It remains unclear how the use of the above-mentioned balance strategies scale with inertia. Here, we propose to investigate these (potentially naïve) conceptual assumptions by comparing these empirical results with a simulation model, focussing specifically on the hip strategy and the associated control of trunk posture in the sagittal plane. This is relevant in order to understand how (i) perturbation responses might depend on the subject's mass, and (ii) how changes in mass, through e.g. increased body mass or load carriage, might affect the person's ability to recover balance after perturbations.

2 Methods and preliminary results

In order to isolate the response of the hip strategy, we conducted perturbation experiments using a new type of wearable device that exerts controlled moments on the trunk and avoids horizontal forces on the centre of mass (CoM) that might invoke other (e.g. stepping) responses [7].

Forty discrete impulse-like (300 ms burst of magnitude 50 N m) perturbations were randomly applied to the trunk in the sagittal plane, in equal quantities in both directions: 'positive' moments resulting in hip flexion and forward trunk pitch with respect to the initial posture, and 'negative' moments resulting in hip extension and backward trunk pitch. This was repeated for 11 subjects (1 female, 10 male), and resulted in trunk pitch angles of up to 17° and horizontal centre of mass (CoM) deflections below 2 cm.

We estimated the moment of inertia of the head, arms, and trunk (HAT) about the hip joint from normalized anthropometric data [8] scaled by the height and mass of each subject, and combined with an estimate of the the CoM (mean per trial) and inertia of the borne device. For the 11 subjects, the HAT plus perturbator (HATP) moments of inertia about the hip joint ranged from 16 to 27 kgm^2.

In response to a perturbation of fixed magnitude applied to the trunk, the passive dynamics of the trunk would suggest a linear relationship between upper-body moment of inertia and maximum deflection. By simplifying the HATP as a single rigid body and considering the lower body as approximately stationary, the transfer function between the perturbing moment τ and trunk angle θ would be as follows:

$$\begin{aligned}\frac{\theta(s)}{\tau(s)} &= \frac{1}{Js^2+ds+k'} \\ &= \frac{K}{s^2+2\zeta\omega_\mathrm{n}s+\omega_\mathrm{n}^2} = \frac{K}{(s+\zeta\omega_\mathrm{n})^2+\omega_\mathrm{d}^2}\ , \end{aligned} \tag{1}$$

where $k' = (k - mgz_\mathrm{CoM})$ is the net effective stiffness of the hip, J is the moment of inertia of the HATP about the hip, d and k are the apparent viscosity and stiffness of the hip joint, m is the mass of the HATP, g is the gravitational acceleration, and z_CoM is the location of the CoM of the HATP relative to the hip joint. In addition, $K = J^{-1}$ is a static gain, $\omega_\mathrm{n} = (\frac{k'}{J})^{\frac{1}{2}}$ is the undamped natural frequency, $\zeta = \frac{d}{2(Jk')^{1/2}}$ is the damping ratio, and $\omega_\mathrm{d} = (1-\zeta^2)^{\frac{1}{2}}\omega_\mathrm{n}$ is the damped natural frequency.

The impulse response is found by taking the inverse Laplace transform of Eq. (1):

$$\theta(t) = \mathscr{L}^{-1}\left\{\frac{\theta(s)}{\tau(s)}\right\} = \frac{K}{\omega_\mathrm{d}}\mathrm{e}^{-\zeta\omega_n t}\sin(\omega_\mathrm{d}t)\ . \tag{2}$$

Thus, the peak trunk angle is expected to be proportional to $\frac{K}{\omega_\mathrm{d}}$. For rotation about the ankle, Peterka reported that

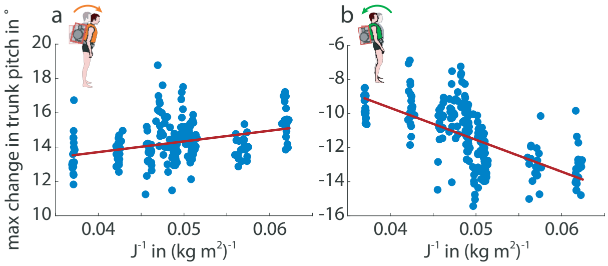

Figure 1: Maximum trunk pitch deflection for both (a) positive and (b) negative perturbations versus the inverse of the upper-body moment of inertia. Shown in each are all 20 repetitions of all 11 subjects (blue points) with linear regression model (red line).

the effective ankle joint stiffness and damping scale linearly with body moment of inertia [9]. If it is assumed that this is also the case for the trunk rotating about the hip joint, the normalized quantities ω_n, ζ, and ω_d will be unaffected by changes in body inertia. However, the amplitude of the impulse response still scales linearly by $K = J^{-1}$, which suggests that impulsive moments of fixed magnitude will cause larger and heavier persons to deflect less than their smaller and lighter counterparts.

Surprisingly, this was not found to be the case during our experiments. Our data (Fig. 1) showed that inertia was only a poor-to-moderate predictor (positive moment: $R^2 = 0.11$, $p = 1.87e^{-7}$; negative moment: $R^2 = 0.40$, $p = 8.92e^{-26}$) for the observed peak trunk rotation (appeared usually after 300 to 400 ms). A high intra-subject variability was also observed – it is hence unclear to what extent other subject-specific factors, such as muscle activation timing and magnitude, or reliance on a certain coordination strategy, might play a role in the ability to respond to a perturbation.

Based on our initial assumptions that the response would be dominated by (pseudo-)passive dynamics, we expected to see a clearer linear relationship. The results suggest that also other factors such as predictive and reactive strategies or psychological factors (like motivation or feeling of safety) should be considered to explain the observed behavior.

3 Research plan

In order to better understand which other factors might have influenced our experiments, we aim to develop a more realistic multi-body model that allows analysis of each physical and (modellable) neuromuscular parameter independently. Previous work by others has produced functional models of balance during both quiet [6, 10] and perturbed [11–16] standing. We suggest to use a simulation model with a simple level of detail, such as [17]. To test for anticipatory (e.g. using co-contraction of antagonistic muscles) or reactive strategies (e.g. by muscular reflex activity), passive and active values of muscular stiffness and damping will be considered. We intend to use optimization techniques to find the parameter sets that best characterize our experimental observations (e.g. trunk dynamics) and identify the sensitivity of the model to trunk inertia changes. The identified components will be compared to experimental data. By this, we aim to address the question how the perturbation response scales with subject mass and to what extent predictive and reactive control of joint impedance contributes to the observed variability in the data.

With this approach we will most likely not fully explain all involved mechanisms that contribute to perturbation behavior. However, we may be able to evaluate the influence of joint impedance and muscular control.

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References

[1] Lewis M. Nashner and Gin McCollum. The organization of human postural movements: A formal basis and experimental synthesis. *Behavioral and Brain Sciences*, 8(01), March 1985.

[2] F. B. Horak and L. M. Nashner. Central programming of postural movements: Adaptation to altered support-surface configurations. *Journal of Neurophysiology*, 55(6), June 1986.

[3] Hugh Herr and Marko Popovic. Angular momentum in human walking. *Journal of experimental biology*, 211(4), 2008.

[4] David A Winter. Human balance and posture control during standing and walking. *Gait & posture*, 3(4), 1995.

[5] A.L. Hof, M.G.J. Gazendam, and W.E. Sinke. The condition for dynamic stability. *Journal of Biomechanics*, 38(1), January 2005.

[6] K. Barin. Evaluation of a generalized model of human postural dynamics and control in the sagittal plane. *Biological Cybernetics*, 61(1), May 1989.

[7] Daniel Lemus, Jan van Frankenhuyzen, and Heike Vallery. Design and evaluation of a balance assistance control moment gyroscope. *Journal of Mechanisms and Robotics*, 9(5), 2017.

[8] David A Winter. *Biomechanics and motor control of human movement*. John Wiley & Sons, 2009.

[9] R. J. Peterka. Sensorimotor Integration in Human Postural Control. *Journal of Neurophysiology*, 88(3), September 2002.

[10] Arthur D. Kuo. An optimal state estimation model of sensory integration in human postural balance. *Journal of Neural Engineering*, 2(3), August 2005.

[11] A. D. Kuo. An optimal control model for analyzing human postural balance. *IEEE Transactions on Biomedical Engineering*, 42(1):87–101, January 1995.

[12] Sukyung Park, Fay B. Horak, and Arthur D. Kuo. Postural feedback responses scale with biomechanical constraints in human standing. *Experimental Brain Research*, 154(4), February 2004.

[13] Torrence D. J. Welch and Lena H. Ting. A Feedback Model Reproduces Muscle Activity During Human Postural Responses to Support-Surface Translations. *Journal of Neurophysiology*, 99(2), February 2008.

[14] Torrence D. J. Welch and Lena H. Ting. A Feedback Model Explains the Differential Scaling of Human Postural Responses to Perturbation Acceleration and Velocity. *Journal of Neurophysiology*, 101(6), June 2009.

[15] Maarten Afschrift, Ilse Jonkers, Joris De Schutter, and Friedl De Groote. Mechanical effort predicts the selection of ankle over hip strategies in nonstepping postural responses. *Journal of Neurophysiology*, 116(4), October 2016.

[16] Maarten Afschrift, Joris De Schutter, Ilse Jonkers, and Friedl De Groote. A Model of Human Non-stepping Postural Responses as the Basis for a Biomimetic Control Strategy for Robot-Assisted Balance. In Jaime Ibáñez, José González-Vargas, José María Azorín, Metin Akay, and José Luis Pons, editors, *Converging Clinical and Engineering Research on Neurorehabilitation II*, Biosystems & Biorobotics. Springer International Publishing, 2017.

[17] Alireza Sarmadi, Christian Schumacher, Andre Seyfarth, and Maziar Ahmad Sharbafi. Concerted control of stance and balance locomotor subfunctions-leg force as a conductor. *IEEE Transactions on Medical Robotics and Bionics*, 2019.