Human Balance Responses to Perturbations in the Horizontal Plane

Michael Fritschi[1], Herbert F. Jelinek[2], Tim McGloughlin[1], Kinda Khalaf[1], Ahsan H. Khandoker[1], and Heike Vallery[3]

***Abstract*— Human balance strategies during standing have been studied extensively. Most of these studies rely on perturbations to the feet, for example by moving platforms or treadmills, and focus on the sagittal plane. Less research has been done on reactions to perturbations to the upper body, and the direction dependence of stabilizing strategies is still an open question. Here, we describe an experiment where we apply horizontal static pulling forces to the upper body of standing human subjects in different directions by means of an overhead robotic device, the FLOAT. Based on a simplified mechanical model, we propose the normalized displacement of the center of pressure, the ΔCoP_n, as a measure of the selected balance strategy. We find that existing neuromechanical models do not fully explain responses to these static horizontal forces, because they predict too much CoP movement. Further, we found a tendency to particularly reduce CoP movement in anterior-posterior direction, indicating that reconfiguration of the body may play a larger role in this direction.**

I. INTRODUCTION

Humans rely on a fine interplay of strategies to maintain balance during standing and walking [1], [2]: The "ankle strategy" moves the *center of pressure* (CoP), the point where the line of action of the net ground reaction force intersects with the ground [3]. The "hip strategy" moves the upper body in opposite direction with respect to the lower body, changing the body's angular momentum. While ankle and hip strategy dominate balance control during stance [1], [2], foot placement dominates during walking [4], [5]. Full models of physiological human balance control exist for standing, and these have also been applied to stabilize robots [6].

To identify human balance strategies, perturbations are normally applied by treadmills or perturbation platforms. These perturbations to the feet have shown to be an efficient method to analyze and quantify human balance [7], but it remains difficult to study strategies that involve CoP movement or responses to constant perturbations.

To apply perturbations to the upper body, cable systems with winches [8], [9] or weights [10] have been used. Mergner et al. pulled subjects in an a/p direction (0° or 180° from gaze direction) on sway-referenced surfaces, and they found that at low frequencies, people would tend to lean in the opposing direction (phase-leading) [9]. This means that instead of resisting via ankle moments, humans use their body weight to generate a stabilizing moment about the CoP that counteracts the moment generated by the horizontal perturbation.

To study whether similar behavior is also observed in response to perturbations in lateral (90° and 270°) or other horizontal directions, we used the cable robot FLOAT (Lutz Medical Engineering GmbH, Rüdlingen, CH). The FLOAT's main purpose is overground gait rehabilitation, and it provides overhead support by means of a harness [11], [12]. It can transparently control forces to the person's trunk in three directions in a large workspace. Here, the robot is only used to apply static pulling forces in different horizontal directions. We also propose a measure that characterizes balance strategies based on movement of the CoP: The normalized displacement of the CoP, the ΔCoP_n.

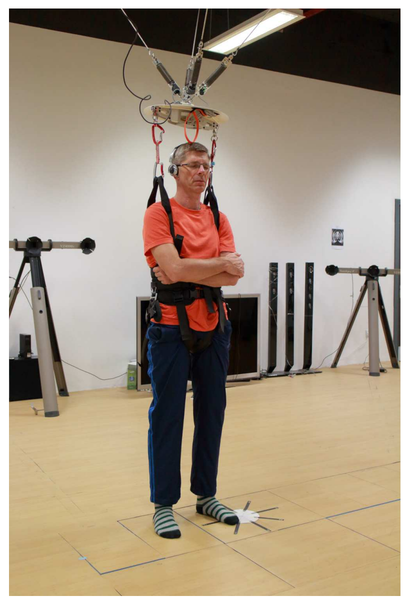

Fig. 1. Subject attached to the FLOAT, standing on force plates

II. EXPERIMENTAL METHODS

A. Experimental Hardware

The experiments were performed in the gait lab of Khalifa University. The FLOAT version installed in this lab has a workspace of $8\text{m} \times 2\text{m}$ in the horizontal plane and $2\,\text{m}$ in height. The FLOAT employs four cables, which are actuated by motorized winches and guided into the workspace by moving deflection units [11], [12]. These deflection units are not actuated, but they are moved implicitly by the forces in the cables they deflect, minimizing moving masses and thereby undesired interaction forces on the subject. The

[1]Khalifa University, Abu Dhabi, UAE michael@fritschi.de, {tim.mcgloughlin, kinda.khalaf, ahsan.khandoker}@kustar.ac.ae
[2]Charles Sturt University, Albury, AU HJelinek@csu.edu.au
[3]Delft University of Technology, NL h.vallery@tudelft.nl

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four cables meet approximately in one point. Subjects wear a harness (Hocoma AG, Volketswil, Switzerland) that is attached to this point via a spreader bar (Fig. 1). A graphical user interface (GUI) allows setting the FLOAT's supporting forces in vertical direction ($z-$axes) and both horizontal directions ($x-, y-$axis).

To measure the CoP, two force plates from KISTLER (Type: 9360AA3) were used. They are installed permanently within the lab floor (Fig. 1). Forces measured by the plates were sampled with 100Hz.

The forces commanded via the GUI were recorded with a key-logging program, together with time stamps, which were synchronized with the force recording software from KISTLER (BioWare).

B. Experimental Protocol

The Ethics & Research Committee of the Mafraq Hospital Abu Dhabi approved the experimental protocol used in this study. Five male adults, aged 27 to 42 (mean 33.2) with no history or evidence of muscular, orthopedic, or neurological motor disorders, were tested. All subjects provided consent prior to participating in this study.

Subjects stood comfortably in an upright position on the force plates, with arms crossed in front of the chest, facing straight ahead and eyes closed. They where instructed to keep their balance despite perturbing forces from the FLOAT and to avoid stepping or lifting their heels off the ground.

A perturbation of 19.62 N (equivalent to 2 kg) was presented in eight different directions in the horizontal plane, with a spacing of 45°. The *body weight support* (BWS) was held constant to 4kg over the entire experiment (Due to the working principle of the robot, this minimum vertical force is necessary). Since each angle of perturbation was presented twice, the experiment consisted of 16 trials. The order of trials was randomized. For practical reasons, we used only one axis with positive and negative force direction of the FLOAT to apply perturbations to the subject. Thus, the subject was instructed to face one of the directions (Fig. 2), which where marked on the surrounding walls. Thus, they could anticipate the line of action of the perturbing force, but they were not told the direction in advance (i.e. the sign). The orientation and the force direction to apply for a particular perturbation angle was calculated according to Tab. I.

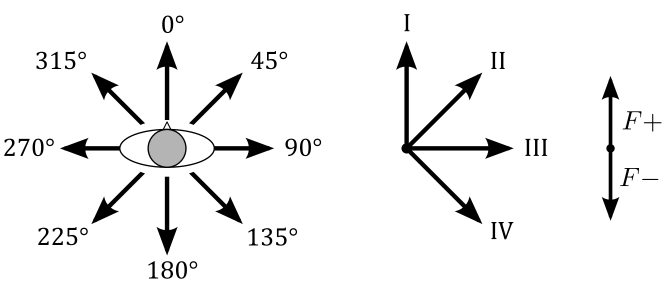

Fig. 2. Direction of perturbation relative to the subject (left). Orientation of the subject and direction of force within the experiment (center/right)

Each trial lasted 30 s. For 10 s, the subject had to stand upright without horizontal force, for 10 s with constant horizontal force, and finally for 10 s standing upright without horizontal force. After a short time ($\sim$10 s), where the subject had to change orientation according to the new perturbation, a new trial started.

TABLE I
PERTURBATIONS BY ORIENTATION AND FORCE DIRECTION

Angle	Orientation	Force	Angle	Orientation	Force
0°	I	$F+$	180°	I	$F+$
45°	IV	$F+$	225°	IV	$F+$
90°	III	$F+$	270°	III	$F+$
135°	II	$F+$	315°	II	$F+$

Before each trial, we calibrated the inter-foot distance to a constant comfortable distance of 14.5 cm (using a rectangular box) and we made sure both feet were on the force plates. To suppress visual or auditory cues, subjects were instructed to keep their eyes closed during each trial and white noise was presented via headphones.

C. Model-Based Data Analysis

To quantify balance reactions, we consider a simplified model (Fig. 3) of the person in steady state: A linear inverted pendulum of mass m moves in the plane spanned by the horizontal direction u of the perturbing force and the vertical z. The steady-state vertical and horizontal FLOAT forces, F_v and F_h, act directly on the *center of mass* (CoM). The CoP (0 in unperturbed, $0'$ in perturbed conditions) gives information on balance reactions: In a/p direction, it gives a measure of ankle moment, whereas in lateral direction, it gives an estimate of load distribution between legs.

The horizontal displacement ΔCoM of the CoM from 0 to $0'$ (Fig. 3) is found from moment equilibrium about 0:

$$\Delta CoM = \Delta CoP - \frac{F_h l}{(mg - F_v)}, \tag{1}$$

This shows that for a given height l of the CoM and for given F_h and F_v, the CoM is at a fixed offset from the CoP, in opposite direction to the perturbing force. As can be shown, this property also holds when F_h acts at a different height or with more anthropometric mass distribution, as long as the vertical displacement δl between perturbed and unperturbed conditions remains negligibly small. Thus, ΔCoP also reveals CoM movement.

To reduce variability in experimental data, we define the normalized displacement of the CoP, which is dimensionless and independent of person mass, applied forces, or geometry:

$$\Delta CoP_n := \frac{\Delta CoP(mg - F_v)}{F_h l}, \tag{2}$$

with l height of application of the perturbing force (recall that F_h is the *steady-state* value, so division by zero is impossible). Following (1): In case ΔCoP_n is zero, subjects "lean" in the ideal way to avoid a moment about 0, thereby keeping ankle moments unchanged. If ΔCoP_n is 1, subjects keep their CoM location unchanged, i.e. $\Delta CoM = 0$ in (1),

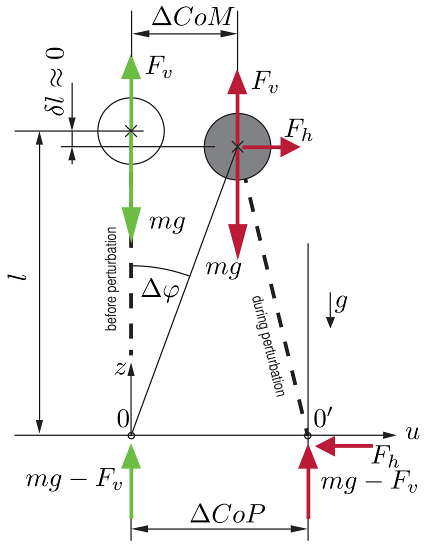

Fig. 3. Free-Body Diagrams of a simplified linear-inverted pendulum model of the person in static conditions, with unloading force F_v, and with (red, left) and without (green, right) the steady-state value F_h of the perturbation in direction u. In response, the CoP moves from 0 to $0'$.

at the expense of high ankle moments or large load imbalance between legs.

We can also predict values for ΔCoP_n with a simplified model of neural control, following [13] (their model also includes torque feedback, which is neglected here): If subjects use proportional feedback to correct posture changes $\Delta\varphi$ (Fig. 3), the moment τ_0 created by the ground reaction force about point 0 would be for small changes in angle $\Delta\varphi$:

$$(mg - F_v)\Delta CoP = \tau_0 = k_p \Delta\varphi \approx k_p \frac{\Delta CoM}{l}. \quad (3)$$

Assuming that the feedback gain k_p is related to body weight (here reduced by F_v) and to CoM height, as described in [13], we define the modified gain $k^* := k_p(mg - F_v)/l$. Then, we can calculate ΔCoP_n in function of k^* from (1-3):

$$\Delta CoP_n = \frac{k^*}{k^* - 1}. \quad (4)$$

This equation illustrates that infinitely stiff postural control with $k^* \rightarrow \infty$, i.e. a perfect "resisting" strategy, leads to a ΔCoP_n of 1. However, humans do no use infinitely stiff control. Instead, only an over-compensation of the moment created by gravity by about one third has been observed in a/p direction [13], equivalent to a value of 4/3 for k^*. Substituting this in (4), a value of 4 would be expected for ΔCoP_n. Further, given that for any values of $k^* \leq 1$, no *stable* equilibrium would result, "leaning strategies" with $\Delta CoP_n \leq 1$ and $\Delta CoM \leq 0$ could not be explained with proportional feedback on kinematic configuration alone.

From the recorded data, we calculated the CoP across both force plates. Only the CoP component in direction u of the perturbation was used for analysis. The mean values of the CoP and the ground reaction force over the first 10 s of each trial were used as the baseline to calculate ΔCoP and as the value of the vertical ground reaction force in the static case, $(mg - F_v)$, respectively. With these values, the ΔCoP_n was calculated using (2).

For statistical testing, we averaged across trials for each subject per direction, lumped the results into three groups "a/p", "diagonal", and "lateral", and calculated the group means. We only considered data of subjects who completed all trials without stepping out. Then, we conducted a non-parametric Kruskal-Wallis test to compare the mean responses (95% confidence level).

III. RESULTS

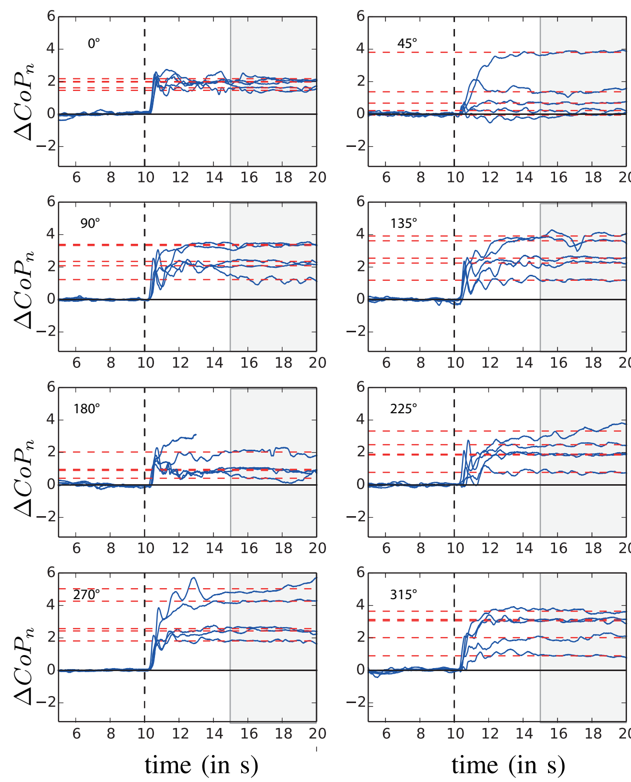

Fig. 4. The normalized displacement of the CoP, ΔCoP_n, in response to the perturbations in the different directions. For each of the five subjects, the mean values of the two trials are displayed.

Fig. 4 shows the ΔCoP_n over the relevant time of the trials (5$-$20 s) of the five subjects. The black dashed vertical line indicates the start of the perturbation, while the gray area between 15 and 20 s marks the time period where the subjects' position was judged as steady state and thus the mean value for the ΔCoP_n was calculated (red dashed lines).

One subject stepped out in response to a perturbation at 180°, which appears in the graph as a trial ending after ~13 s. The dataset of this subject is discarded in further calculations.

Fig. 5 shows the mean values of the ΔCoP_n from the remaining four subjects in a polar plot in the subject's local coordinate system. Group means for a/p, diagonal, and lateral were 1.49, 1.95, and 2.6, respectively, but differences between groups were not significant ($p = 0.17$).

IV. DISCUSSION

Although there was evidence for a leaning strategy with $\Delta CoP_n = 0$ on a few individual trials, in general a pure

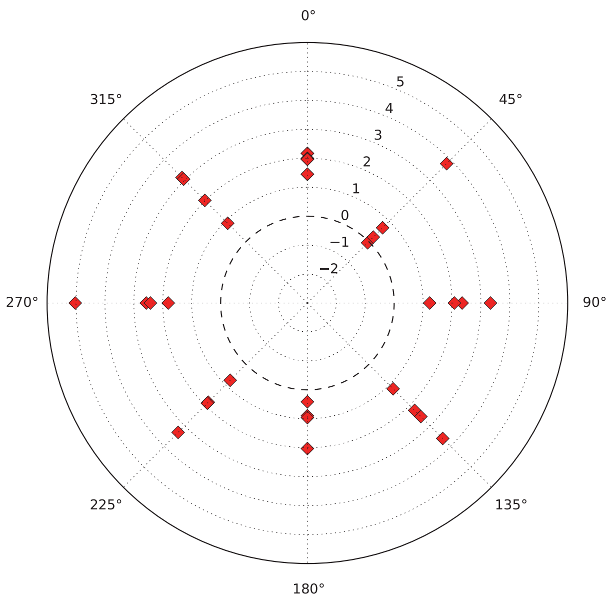

Fig. 5. Normalized CoP in response to perturbations in different directions, for the four subjects that completed all trials without stepping out.

leaning strategy was not present in any of the directions. Still, we consistently found lower values of the ΔCoP_n than the value of 4 that would be predicted by the (simplified) neural model (4) with feedback gain as in [13]. This could be explained either by a higher gain, or by yet unmodeled feed-forward compensation, in adaptation to the static horizontal forces. This is consistent with the fact that for all directions, the margin of stability decreases while the CoP moves towards the edge of the base of support, and one would expect subjects to avoid this.

Furthermore, particularly in a/p direction, where ankle moments are necessary, also energetic/fatigue considerations could play a role. The tendency for lowest values of ΔCoP_n in this direction, though not statistically significant, is consistent with this prediction.

In future experiments, perturbations with mixed quasi-static and dynamic components might be more suitable to disentangle feed-forward and feed-back mechanisms. Also, more subjects are needed to investigate direction dependence.

The 2D single inverted pendulum is a strongly simplified model. We also neglect the fact that the line of action of the FLOAT's force vector does not necessarily pass through the subject's CoM. A more precise mechanical model, e.g. based on anthropometric data and motion capture, could lead to a more reliable calculation of the CoM, and to a more accurate description of the employed balance strategies.

Further factors that could be taken into account in future studies are the effect of visual cuing (eyes open versus eyes closed), the role of anticipation in postural control [10], or subject-specific factors such as age or fitness level.

Eventually, the ΔCoP_n might be useful as a simple clinical indicator to detect pathological behavior, but it does not directly allow conclusions on the underlying reasons. Therefore, a next step could be to link the observations more closely to neuromechanical models, e.g. to reproduce pathological behavior by adjusting model parameters.

V. CONCLUSION

We proposed the normalized displacement of the CoP, the ΔCoP_n, as a fast and intuitive assessment of balance strategies in response to quasi-static horizontal forces on the upper body. We found that the observed ΔCoP_n is smaller than predicted by existing neuromechanical models for postural control, hinting at possible feed-forward mechanisms in adaptation to or anticipation of the perturbations. We also showed first (non-significant) indications that humans employ direction-dependent strategies: While subjects "resist" perturbations in lateral directions more by moving the CoP towards the perturbation, they seem to prefer reconfiguration of their body in a/p direction to prevent high ankle moments.

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REFERENCES

[1] A. D. Kuo and F. E. Zajac, "Human standing posture: multi-joint movement strategies based on biomechanical constraints," *Progress in Brain Research*, vol. 97, pp. 349–58, 1993.

[2] F. B. Horak and L. M. Nashner, "Central programming of postural movements: adaptation to altered support-surface configurations," *Journal of Neurophysiology*, vol. 55, pp. 1369–1381, 1986.

[3] P. Sardain and G. Bessonnet, "Forces acting on a biped robot. Center of pressure-zero moment point," *IEEE Transactions on Systems, Man and Cybernetics, Part A*, vol. 34, no. 5, pp. 630–637, 2004.

[4] A. L. Hof, "The 'extrapolated center of mass' concept suggests a simple control of balance in walking," *Human Movement Science*, vol. 27, pp. 112–125, 2008.

[5] M. A. Townsend, "Biped gait stabilization via foot placement." *Journal of Biomechanics*, vol. 18, no. 1, pp. 21–38, 1985.

[6] T. Mergner, "Modelling sensorimotor control of human upright stance," *Progress in Brain Research*, vol. 165, pp. 283–297, 2007.

[7] D. Engelhart, E. van Asseldonk, L. Grin, and H. van der Kooij, "Lateral balance control during walking: Foot placement and stability in response to external perturbations," in *Proc. ISPGR*, Trondheim, Norway, 2012.

[8] P. E. Pidcoe and M. W. Rogers, "A closed-loop stepper motor waist-pull system for inducing protective stepping in humans." *J Biomech*, vol. 31, no. 4, pp. 377–381, Apr 1998.

[9] T. Mergner, C. Maurer, and R. Peterka, "A multisensory posture control model of human upright stance," *Progress in Brain Research*, vol. 142, pp. 189–201, 2003.

[10] L. Mouchnino, G. Robert, H. Ruget, J. Blouin, and M. Simoneau, "Online control of anticipated postural adjustments in step initiation: evidence from behavioral and computational approaches." *Gait Posture*, vol. 35, no. 4, pp. 616–620, Apr 2012.

[11] H. Vallery and P. Lutz, "Apparatus for unloading a user's body weight during a physical activity of said user, particularly for gait training of said user," European Patent EP 2 626 051 A1, 2013.

[12] H. Vallery, P. Lutz, J. von Zitzewitz, G. Rauter, M. Fritschi, C. Everarts, R. Ronsse, A. Curt, and M. Bolliger, "Multidirectional Transparent Support for Overground Gait Training," in *Proc. ICORR*, 2013.

[13] R. J. Peterka, "Sensorimotor integration in human postural control." *J Neurophysiol*, vol. 88, no. 3, pp. 1097–1118, Sep 2002.