

# Investigating how humans control balance during stance and gait

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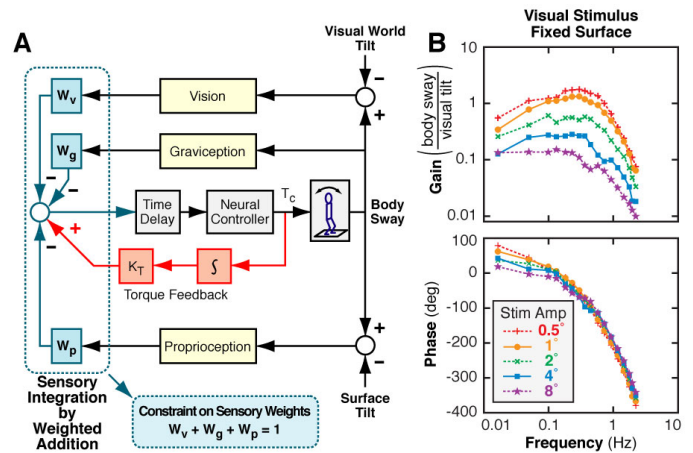
## 1 Introduction

Humans use orientation and motion information derived from sensory systems to generate corrective motor actions that resist the destabilizing effects of gravity and other external perturbations in order to maintain stable stance [1]. The primary sensory systems are the proprioceptive system (signaling orientation and motion of body segments), visual system (signaling head motion relative to the visual environment), and vestibular system (signaling head motion in space). In principle, the central nervous system could combine information from the head-mounted visual and vestibular sensors with multi-segmental proprioceptive cues in order to derive an estimate of body center-of-mass (CoM) position. In quasi-static conditions of quiet or mildly perturbed stance, stable stance is maintained if the motor actions can maintain the CoM over the base of support.

In section 2, a summary is given of methods and results from our lab that provide quantitative measures of the sensory system contributions to stance control. These results show that sensory integration is a dynamically regulated process with sensory contributions that change as a function of environmental and stimulus conditions. Section 3 shows preliminary results illustrating that the sensory integration process contributing to the control of medial-lateral (ML) body orientation during a stepping-in-place (SiP) task is similar to that used in stance control. Section 4 shows results from our preliminary investigation into the motor actions used for the control of ML dynamic balance and orientation during SiP and natural walking.

## 2 Integration of sensory cues for stance control

The relative importance of the different sensory systems for stance control was investigated by perturbing subjects with stimuli that provide conflicting orientation cues from the different sensory systems. For example, rotation of the surface on which a subject stands causes time varying changes in proprioceptive cues signaling body orientation relative to the moving surface while vestibular cues signal body orientation relative to earth vertical as do visual cues (with a stationary visual scene). In this type of experiment, subjects tend to orient toward the tilting surface. The extent to which they orient to the surface is indicative of the extent to which they are relying on proprioceptive cues for stance control. This qualitative assessment of sensory utilization



**Figure 1:** Stance control model (A) and experimental FRFs (B) calculated from CoM body sway responses to visual tilt stimuli of different amplitudes.

can be made quantitative using a model-based interpretation of the experimental results [2].

Figure 1A shows a feedback control model that accounts for sway data evoked in experiments that perturb stance using a pseudorandom rotation of the support surface and/or a visual surround. Fourier analysis is used to calculate frequency response functions (FRFs expressed as gain and phase measures as a function of frequency). Parameters of the differential equation represented by the block diagram model in Fig. 1A can be estimated by fitting the model-predicted FRF to the experimental FRF data. Important parameters include proportional and derivative gain factors of the neural controller, time delay, torque feedback parameters (torque feedback accounts for the gain decline and phase advance at frequencies  $< \sim 0.1$  Hz), and sensory weights. The sensory weights represent the contributions of proprioceptive, visual, and vestibular systems to stance control with the sum of the weights constrained to be unity.

Results have shown that sensory weights change as a function of stimulus amplitude for both surface-tilt and visual-tilt stimuli [1]. The change in sensory weights accounts for the stimulus amplitude-dependent change in FRF gains (Fig. 1C) while the overall shape of the FRFs change little with amplitude (Fig. 1B). Specifically, at larger stimulus amplitudes, subjects reduce their reliance on information from

the sensory system that is most influenced by the perturbing stimulus (i.e., subjects reduce reliance on proprioceptive cues when the surface is tilting or visual cues when the visual scene is tilting) and increase reliance on alternative sensory systems (e.g., vestibular during eyes-closed surface tilts). However, subjects are never able to completely ignore available sensory motion information. This re-weighting behavior is expected in a system that is attempting to resolve ambiguities in motion information arising from multiple noisy sensory sources [3].

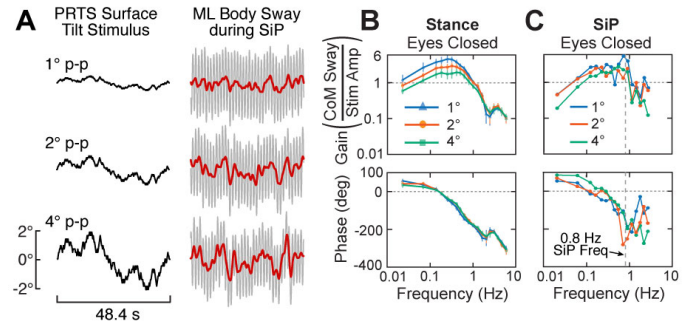
The model in Fig. 1A assumes the body sways as a single-link inverted pendulum. We have recently extended our analysis to account for the control of multi-segmental ML body sway [4]. We were able to identify a feedback model that accounts for the experimental data over a wide frequency range and at varying stance widths. Sensory re-weighting contributed to stance control in narrow stance conditions, but its contribution diminished with increasing stance width. Additionally, the complex feedback relationships that we identified between sensory detection of segmental motion and motor activation suggest that it is more appropriate to think in terms of sensorimotor re-weighting rather than sensory re-weighting.

### 3 Sensory integration for ML balance during gait

Do the sensory integration principles identified for stance control also apply to the control of balance and orientation during gait? Preliminary results show similarities between stance control and the control of body orientation during the gait-related task of SiP. Figure 2A shows ML sway during an eye-closed, metronome-paced SiP test with balance perturbed by a continuous pseudorandom ML tilt of the surface upon which the subject was stepping. When the oscillatory ML sway (gray trace) that occurs during SiP is removed by low-pass filtering (red trace), the tendency of the subject to orient to the tilting surface is revealed. This orienting behavior is similar to that which occurs during stance. A comparison of FRFs from stance and SiP experiments show similar shapes of the gain and phase curves below the SiP stepping frequency indicating that the dynamic properties of orientation control are similar during stance and SiP. The FRFs also show some evidence of gain decrease with increasing stimulus amplitude, suggesting that sensory re-weighting contributes to the control of body orientation during SiP. That is, sensory re-weighting provides a mechanism that partially limits the disturbance to body orientation caused by the perturbing stimulus.

### 4 Control mechanisms for ML dynamic balance during SiP and natural walking

The results on ML orientation control during SiP show that subjects tend to align to an off-vertical orientation in conditions when the surface on which they are stepping is tilted with respect to earth vertical. To maintain a tilted body orientation during successive step cycles requires some asym-



**Figure 2:** Pseudorandom surface-tilt stimuli and evoked body sway during SiP (A). FRFs calculated from ML CoM sway responses to stimuli at 3 different amplitudes during eyes-closed stance (B) and SiP (C).

metry in the motor actions that maintain stable ML dynamic balance during gait. For example, if a subject is biased toward maintaining a rightward-tilted orientation, the gravity torque acting on the body when the right foot is lifted off the surface (the SiP equivalent of right foot swing time) will be greater than it is if the subject were maintaining, on average, an earth-vertical orientation.

By analogy to passive dynamic walking machines, we assume that humans are also inherently unstable in the ML direction [5] and need active ML control to maintain a stable gait cycle. Possible ML control mechanisms discussed in [5] and relevant to human control include regulation of step width, generation of ankle torques, generation of reactive torques via torso motion. Step-width control is considered to be the most effective. More recently Maufroy et al. [6] investigated the regulation of step timing, which they referred to as walking phase modulation, for the control of ML stability.

Our preliminary results, using an eyes-closed SiP protocol performed on a surface that was tilted continuously according to a pseudorandom waveform, showed no evidence that step width was modulated to maintain dynamic balance. However, there was a clear modulation of step timing in sync with the pseudorandom tilt stimulus. Specifically, when the surface tilted to the right (left), the duration of the right (left) leg stance time was longer than the left (right) leg stance time. The extended stance time of the leg on the downhill side (and extended swing time of the opposite leg) effectively allowed more time for gravitational torque to act on the body to accelerate it toward the uphill side of the tilted surface. Conversely, the stance time of the uphill leg (and swing time of the opposite leg) was shortened. This asymmetric limping step rhythm appeared to contribute to dynamic ML stability while allowing the body orientation, on average, to be aligned toward the tilted surface.

To analyze our SiP results, the right- and left-leg stance times ( $R_{st}$  and  $L_{st}$ ) were measured for each step. For each step cycle, a stance-time asymmetry (STA) measure was cal-

culated using the formula:

$$STA = (R_{st} - L_{st}) / (0.5 * (R_{st} + L_{st})).$$

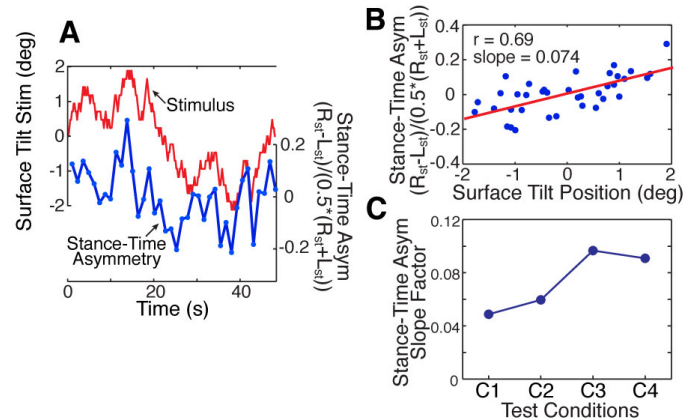
A plot of STA across the pseudorandom stimulus cycle shows an obvious relation between STA and surface tilt angle (Fig. 3A). Figure 3B shows a linear relation with significant correlation between STA and surface tilt angle. The slope factor from the linear regression is a type of sensitivity measure that indicates the extent to which step timing is modulated to control ML balance.

Additional preliminary experiments were performed to limit the contribution of other potential mechanisms for control of ML stability. The idea was that if we eliminated other mechanisms which were contributing to ML stability, then the control afforded by stance-time asymmetry (quantified by the slope factor described above) should increase in order to compensate for the loss of control afforded by these other mechanisms. We had subjects maintain constant step width by using a guide between the feet. This had only a small effect on the slope factor (Fig. 3C; condition C2 compared with unconstrained SiP in condition C1). We then had subjects maintain constant step width and step on a narrow 2 cm base that limited their ability to generate an ankle torque. This had a large effect producing a slope factor about double that in the unconstrained condition (Fig. 3C; condition C3 vs. C1) and suggesting that ankle torque is an important contributor to ML control. Finally, we limited motion of the upper body relative to the lower body to eliminate reactive torque due to torso motion in addition to fixing the step width and limiting ankle torque. This caused no further increase in the slope factor (Fig. 3C; condition C4).

Finally, a continuous pseudorandom galvanic vestibular stimulus (GVS; 2 mA peak-to-peak) was applied in preliminary experiments during natural walking and SiP to determine if modulation of step timing contributed to ML balance control in natural walking and to compare natural walking and SiP results. Experiments were performed on flat and level surfaces with vision greatly impaired by wearing goggles fogged with a translucent covering. GVS causes ML deviations of body orientation [7]. Results showed strong modulation in step timing in sync with the GVS waveform and the measured slope factors relating STA to GVS current amplitude were nearly identical for natural walking and SiP.

## 5 Conclusions and open questions

Mechanisms that control body orientation during stance appear to share properties with and may be the same as the mechanisms that control orientation during gait. Balance control mechanisms can act continuously over time during stance but gait requires the use of different control mechanisms to maintain both dynamic stability and the desired body orientation. Preliminary results demonstrated that step-timing regulation and ankle torque make important contributions to the control of ML dynamic balance and



**Figure 3:** Stance-time asymmetry (blue) measured during SiP on a pseudorandomly tilting surface (red). B. Correlation of stance-time asymmetry with surface-tilt position. C. The influence of different constraints on the contribution of step timing to balance control during SiP.

orientation during gait. It is curious that we did not identify a contribution from step-width regulation since it has been demonstrated in studies the present impulsive perturbations. It may be that step-timing regulation and ankle torque control are the first line of control when perturbations are small. Alternatively, the conditions of our experiments could have favored the use of mechanisms other than step-width modulation.

## 6 Acknowledgements

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## References

- [1] R.J. Peterka, J Neurophysiol, 88:1097-1118, 2002.
- [2] R.J. Peterka, IEEE Eng Med Biol Mag, 22:63-68, 2003.
- [3] van der Kooij and R.J. Peterka, J Comput Neurosci, 30:759-778, 2011.
- [4] A.D. Goodworth and R.J. Peterka, J Neurophysiol, 107:12-28, 2012.
- [5] A.D. Kuo, International Journal of Robotics Research, 18:917-930, 1999.
- [6] C. Maufroy, H. Kimura, and K. Takase, Auton Robot, 28:331-353, 2010.
- [7] R.C. Fitzpatrick and B.L. Day, J Appl Physiol, 96:2301-2316, 2004.