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MODEL OF LATERAL STABILIZATION FOR ONE-LEGGED BALANCE

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SUMMARY

To study human lateral stabilization strategies, we developed a simple model using pendulums and a momentum wheel representing human appendages, and we recorded appendage data from 7 subjects during one-legged standing. We demonstrated that simple pendulum models can explain some of the strategies employed by humans to stabilize and that the basic strategy people employ to stabilize themselves is to move their appendages in the direction they are falling.

INTRODUCTION

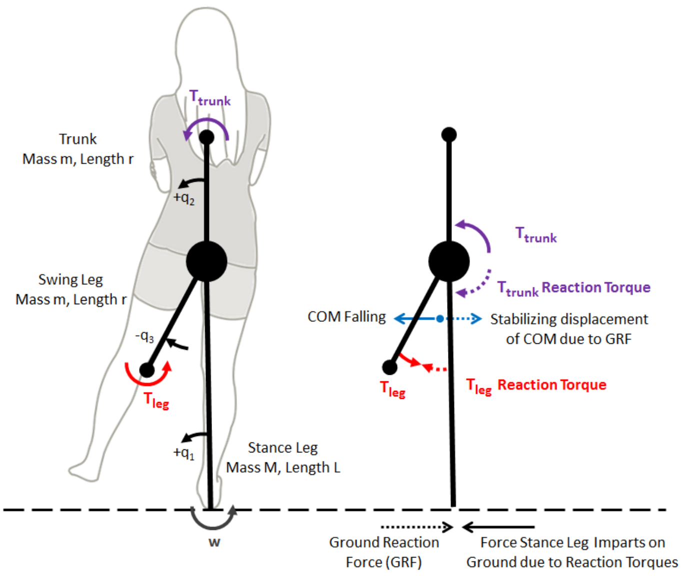
When standing on one leg, humans often move their other limbs to aid balance. The stance foot can only exert limited torques to move the body center of mass (COM) laterally, and limb movement can compensate for that limitation. This may also occur during the swing phase of walking, which is also a momentary interval of one-legged balance. Although many perform such balance quite easily, it is unclear which limbs should be moved and in what way to keep the body stabilized. Here we employ a simple balance model to examine how balance corrections might be coordinated.

The key to such balance corrections is inertia. Unlike the stance leg, body segments such as the torso, arms, and opposite leg cannot produce torques against the ground. Their inertia must instead be used to affect the COM more indirectly, as indicated by inverted pendulum models [1,2,3,5]. Most previous models have not, however, considered how multiple appendages should be coordinated, and which limbs should be prioritized. We use a multi-link model to study how feedback control should be distributed between the torso, the leg not bearing weight (termed the swing leg here), and the arms. We determine a simple control law for controlling balance through inertia and compare it against experimental observations of human balance.

MODEL AND METHODS

A simple multi-link model includes a main pelvis mass atop an inverted pendulum stance leg, and several generic appendages that may be moved to control balance (Figure 1). As an example of multi-link dynamics, we consider how different appendages representing the trunk and swing leg should be moved, when the stance leg is displaced to the left (Figure 1a). Assuming the stance leg's ankle torque is small, the only stabilization possible is from the appendages, which should both be rotated counter-clockwise. That imparts a reaction torque on the stance leg, which pivots about the foot. The pivot constraint induces a ground reaction force directed to the right, which therefore causes the body COM to be restored in that direction as well.

We developed linearized equations of motion for the model, adding a momentum wheel on top of the trunk to represent arms. As a simple demonstration of a feedback controller, we used a Linear Quadratic Regulator (LQR) design to determine the gains that would deliver the stabilizing torques. LQR calculates the optimal gain matrix for a given weighting on the states of the system and on the control of the system. We assumed that humans prefer to avoid high effort to maintain balance, and used a relatively high weighting on control effort. To simulate human behavior in quiet standing on one leg, we used the stance leg torque to introduce random process noise into the system to disturb the center of mass.



(a) (b)

Figure 1: (a) Model of one-legged standing in the lateral direction. (b) A free body diagram example to illustrate that the appendages will impart a torque in the direction of the body's fall to stabilize the body, and the resulting ground reaction force on the stance leg will move the body's center of mass towards stability.

The model yields two main predictions. One is that the torque exerted on the appendages should all be directed in the same direction as any displacement of the stance leg. For example, both torso and swing leg should be rotated counter-clockwise if the stance leg is perturbed in that direction, as in Figure 1. The second prediction is that, should one or more appendages be constrained, the gains should increase for the other appendage(s) to maintain balance. For the purposes of this demonstration, the actual values of feedback gains are immaterial, and we focus instead on the main direction of torque. These are necessarily consistent with the qualitative explanation above, and the LQR design merely assigns useful values in those directions.

We performed an experiment on 7 healthy adult subjects to compare model results to human behavior. Using spontaneous noisy fluctuations as the perturbation, we examined the effective control gains used by the subjects to maintain balance. We also examined how the addition of constraints on some appendage movements caused changes in movement of others. Subjects balanced on their left leg for 30 seconds. The movement of their left ankle was restricted with athlete tape and an Aircast ankle brace so that the subjects could only use a combination of their arms, trunk, and swing (right) leg to stabilize. Segment angles and angular velocities were captured using Xsens inertial measurement units placed on the stance leg, swing leg, trunk, and right arm. The measurements were band-pass filtered to remove slow appendage drifts and noise, and angular acceleration was obtained from differentiating the angular velocity signal. There were four different trials. In the nominal "No Constraint" condition, subjects were allowed to use their swing leg, trunk, and arms to balance. In the "Trunk Only" condition, the arms were crossed, and the feet were tied together. In the "Swing Leg Only" condition, the arms were crossed, and a stiff board was strapped to the subject's stance leg, hip, and chest, restricting movement between the trunk and stance leg. In the "Arms Only" condition, the feet were tied together and the trunk restriction was also in effect.

RESULTS AND DISCUSSION

To demonstrate that people move their appendages in the direction of the falling COM, we determined the signs of the controller gains that are associated with stance leg position. These three gains determine the direction of the trunk torque, swing leg torque, and arm torque. All three elements of model gains are negative. Since torque is the negation of the multiplication of the gains with the stance leg state, the appendages will impart a positive (counter-clockwise) torque if the stance leg is positioned such that the COM is falling in the counter-clockwise direction.

We compared the signs of the model gains to the controller gains derived from experimental data. We used measurement data from each subject's "No Constraint" condition to obtain a least squares fit between the appendage torques and angles and angular velocities. The appendage torques were estimated from linearized pendulum equations of motion. We applied an appendage torque delay of 200 ms to approximate the time delay between the stance leg state and the torque application. The median result from subject data yielded negative elements for all three elements of the experimentally calculated gains. Delays between 180 and 260 ms also gave the same results. Hence, both model and experimental data demonstrated that the torque applied by each appendage is directed in the same direction as the displacement of the stance leg

We also showed that when one appendage is constrained, the others must move more to compensate. Experimentally, we obtained the variance in appendage movement through the root square mean (RMS). We calculated RMS values from each subject's appendage angles to determine the variance for each condition. Because our trunk constraint restricted motion between the trunk and the stance leg, we calculated a relative trunk RMS value, which represented the variance in movement between the trunk and the stance leg. The mean results are shown in Figure 2. Arm RMS values for "Trunk Only" and "Swing Leg Only" were not shown because the arms were crossed and hence not valid. Nominally, relative trunk movement was higher than the arm and swing leg movement, suggesting that the trunk and stance leg are moving out of phase relative to each other. The experimental results demonstrated that there is a certain level of effort that must be met by the appendages for the human to stabilize. For example, the swing leg must increase its movement above nominal when it is the only stabilizing tool. Similarly, relative trunk movement increased when the trunk was the only free appendage and decreased when its movement was constrained. Finally, the RMS value for arm movement also increased above nominal when the arms were the only allowable appendage.

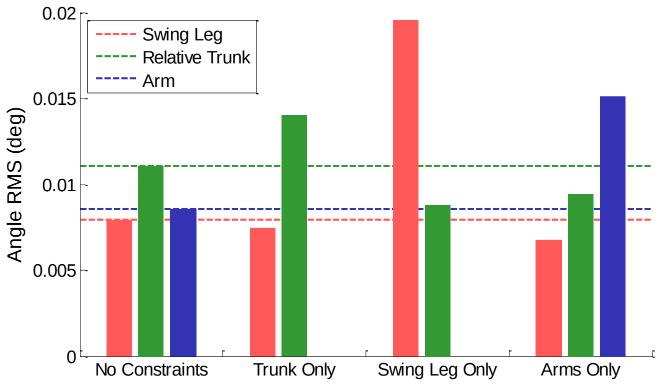


Figure 2: Mean angle RMS values during one-legged standing (N=7) to demonstrate the variance in movement exhibited by difference appendages with and without constraints.

CONCLUSIONS

We have demonstrated that simple pendulum models can explain some of the strategies employed by humans to stabilize. Humans stabilize one-legged balance by rotating various appendages in the same direction they are falling. When one appendage is constrained, the others are moved more to compensate. Although balance movements appear quite complex, they are explained by relatively simple dynamic principles.

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