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Human-inspired balance model to account for foot-beam interaction mechanics

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***Abstract*— The locomotion and balance capabilities of bipedal robots have greatly improved in recent years. However, maintaining balance on difficult terrain still poses a significant challenge. In this paper, we examined how humans maintain mediolateral balance when standing on a narrow beam with bare feet and wearing rigid soles. Our results show that foot-beam interaction dynamics critically influence balancing behavior. Importantly, this suggests that differences in human balancing behavior across different support surfaces may not solely result from changes in their neural control strategy. They may also result from changes in foot-ground interaction. Thus, the altered foot-ground interaction dynamics must be considered to accurately capture changes in the human controller across different support surfaces. A simplified model of foot-beam interaction was added to a double inverted pendulum model for human balancing. This extended model could replicate the change in human behavior across different foot contact conditions (bare feet vs. rigid feet). A better understanding of how humans coordinate whole-body behavior across a range of conditions may inform the development of balance controllers for bipedal robots.**

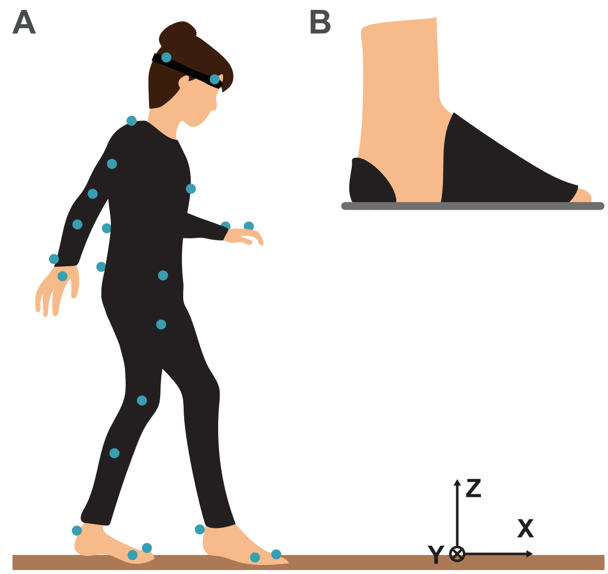

Fig. 1. Experimental Task. (A) Subjects were instructed to maintain balance on a narrow beam (3.4cm) for as long as possible without stepping off the beam. Subjects performed the task under two conditions: bare feet and “rigid feet”. (B) To simulate rigid feet on the human subjects, rigid plastic platforms were attached to the bottom of the subjects feet using Velcro straps and tape.

I. Introduction

In recent years, bipedal robots have made significant advances in their ability to balance and walk over a variety of ground conditions [1]–[4]. However, their abilities are still limited compared to those of humans. Despite low bandwidth, high noise, and long latencies in the neuromuscular system [5], humans have a remarkably robust ability to maintain balance while navigating difficult terrain [6], [7]. In fact, healthy humans are so skilled at balancing that even standing or walking on a thin wire can be learned. A better understanding of how humans coordinate whole-body behavior in such trying conditions may inform the development of balance controllers for bipedal robots [8], [9].

Prior research has shown that an inverted pendulum model can describe the behavior that humans exhibit when standing on flat, hard ground. In this model, ankle torque is used to modulate the center of mass of the system [10]–[13]. While humans often stand on this type of support surface, it is not uncommon for them to also encounter terrain with different geometry and compliance. Maintaining balance over such terrain can be challenging in part due to the limited ability to produce ankle torque. When standing on a narrow or compliant support surface, for example, humans utilize the so-called ‘hip-dominant strategy’ to maintain balance [10]–[17]. To describe this behavior, the model of the human body must be extended to a double inverted pendulum. A previous study found that the robotic controllers which predominantly utilized hip actuation best replicated human balancing on a beam [16].

A key simplifying assumption of both the single and double inverted pendulum models is that the human feet are fixed to the ground and ankle torque equals the ground reaction moment. This assumption may be sufficient for modelling balance while standing on the ground. However,

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it is unclear whether it is also appropriate when modelling human balance when standing on a narrow support surface. In the latter case, dynamics at the foot-ground interaction may become prominent and affect whole-body behavior. If significant, a competent model for human balancing should include such effects.

This study tested whether foot-ground interaction dynamics influence human balancing when standing on a narrow beam (Fig. 1A) by comparing whole-body behavior with bare feet and "rigid" feet (Fig. 1B). The width of the beam and thus the maximum range of the center of pressure were identical in both conditions. Hence, the behavior should be similar in both conditions if the dynamics at the foot-beam interaction port were minimal. However, our results showed that "rigid" human feet improved balance performance. To account for the effect of foot-ground interaction dynamics, an extension of existing double inverted pendulum models of human balance was proposed.

The remainder of the paper is organized as follows. Section II details the human experiment conducted to test whether wearing rigid soles affected balance when standing on a narrow beam. Section III examines a simplified model of human balance that accounts for foot-beam interaction and compared the simulated behavior to the human behavior reported in Section II. Section IV discusses the experimental and simulation results, and Section V summarizes important implications for robotic systems.

II. Human Balancing Experiment

The purpose of the human experiment was to characterize how wearing rigid soles influences one's ability to maintain mediolateral balance on a narrow beam. Regardless of whether the task was performed with bare feet or with rigid soles, the width of the beam was the same. Thus, we expected that balance behavior would be similar across the two conditions.

A. Methods

1) Subjects: Seven subjects (gender: 2 females, 5 males; age: $M = 29.8\text{yrs}, SD = 2.0\text{yrs}$) participated in the experiment. The experiment conformed to the Declaration of Helsinki, and written informed consent was obtained from all subjects according to a protocol approved by the ethical committee at the Medical Department of the Eberhard-Karls-Universität of Tübingen, Germany where the experiment was conducted.

2) Experimental procedure: In each trial, subjects were instructed to stand on a narrow beam (3.4cm width) for as long as possible with their feet in tandem (Fig. 1A). Subjects initially placed their left (front) foot on the beam. The trial started when they subsequently placed their right (hind) foot on the beam. The trial ended when one of their feet lost contact with the beam.

Each subject performed 5 trials without wearing footwear (Bare Feet condition) followed by another 5 trials wearing flat, rigid soles attached to the bottom of their feet (Rigid Feet condition). Note that the rigid soles were attached such that ankle inversion/eversion and plantarflexion/dorsiflexion range of motion were unimpeded (Fig. 1B).

Immediately prior to performing the two standing conditions, all subjects completed 20 trials walking across the beam in each foot condition as part of larger study [18]. Thus, all subjects were sufficiently familiar with both experimental conditions.

3) Kinematic data recording: Kinematic data were collected using a 10-camera Vicon motion capture system (Oxford, UK) at a sampling rate of 100Hz. As illustrated in Fig. 1A, the x-axis of the lab coordinate frame was aligned with the beam. Reflective markers were placed on the subjects' bodies following Vicon's Plug-In Gait marker set (Fig. 1A). For each subject, the Plug-In Gait model, which consists of 15 rigid body segments, was fit to the kinematic data using Vicon Nexus and C-Motion Visual3D software (Germantown, MD).

4) Signal processing: For each trial, we derived the following signals from the model-fitted data. The linear velocity of the whole body's center of mass in the mediolateral (i.e., y) direction at each time t, $v_{\mathrm{wb},y}(t)$ was calculated by backward finite difference on $c_{\mathrm{wb},y}(t)$, the whole body center of mass position, with $T_s = 0.01\text{s}$ as step size. The $v_{\mathrm{wb},y}$ signal was subsequently smoothed with a moving average filter.

The angular momentum of i-th body segment about the axis of the balance beam (i.e., the x-axis) at each time t, $L_{i,x}(t)$, was calculated by

$$L_{i,x}(t) = m_i(c_{i,y}(t)v_{i,z}(t) - c_{i,z}(t)v_{i,y}(t)) + j_{i,x}\omega_i(t), \tag{1}$$

where $c_{i,y}$ and $c_{i,z}$ were the positions of the center of mass in y and z direction, m_i was the mass, $v_{i,y}$ and $v_{i,z}$ were the linear velocities in y and z direction, and $j_{i,x}\omega_i$ was the x component of the angular momentum of the i-th segment about its center of mass in the lab coordinate frame, respectively.

The whole body angular momentum about the beam axis at each time t, $L_{\mathrm{wb},x}(t)$, was calculated by

$$L_{\mathrm{wb},x}(t) = \sum_{i=1}^{15} L_{i,x}(t). \tag{2}$$

The total angular momenta of the upper body segments, $L_{\mathrm{ub},x}(t)$, and lower body segments, $L_{\mathrm{lb},x}(t)$, were also calculated.

The external torque about x-axis at the foot-beam interaction point at each time t, $\tau_{\mathrm{ext},x}(t)$ was estimated by

$$\tau_{\mathrm{ext},x}(t) = \frac{L_{\mathrm{wb},x}(t) - L_{\mathrm{wb},x}(t-1)}{T_s} + mgc_{\mathrm{wb},y}(t) \tag{3}$$

where m was the mass of the subject and g was the gravitational acceleration constant.

To accommodate differences in body size across subjects, the signals of $v_{\mathrm{wb},y}$, $L_{\mathrm{wb},x}$, and $\tau_{\mathrm{ext},x}$ at time t were normalized to obtain $\hat{v}_{\text{wb},y}$, $\hat{L}_{\text{wb},x}$, and $\hat{\tau}_{\text{ext},x}$ at time $\hat{t} = \frac{t}{\left(\frac{h}{g}\right)^{1/2}}$, respectively, as follows:

$$\hat{v}_{\text{wb},y} = \frac{v_{\text{wb},y}}{h\left(\frac{h}{g}\right)^{-1/2}}, \hat{L}_{\text{wb},x} = \frac{L_{\text{wb},x}}{mh^2\left(\frac{h}{g}\right)^{-1/2}}, \hat{\tau}_{\text{ext},x} = \frac{\tau_{\text{ext},x}}{mgh} \tag{4}$$

where m and h were the respective body mass and height of each subject.

5) Dependent measures: For each subject, the dependent measures were calculated for the longest trial in each condition. Data from the first and the last 25% of each trial were omitted to minimize any possible transients or fatigue effects.

- Trial time, quantified as the amount of time the subject stood with both feet on the beam, served as the first gross measure of balance ability.
- The root-mean-square (RMS) of $\hat{v}_{\text{wb},y}$ and $\hat{L}_{\text{wb},x}$ also characterized balance proficiency.
- The correlation coefficient between the angular momenta of the upper and lower body, $\rho_{L_{\text{ub},x}L_{\text{lb},x}}$ was used to characterize the coordination between the different body segments.
- The RMS of $\hat{\tau}_{\text{ext},x}$ was used to assess foot-beam interaction torque.

6) Statistical Analysis: In order to test if wearing rigid soles affected balance performance and whole-body coordination, pairwise t-tests were conducted on the dependent measures calculated for the longest trial in each condition. The significance level was set to $\alpha = 0.05$. Statistical analyses were performed using MATLAB, Version 2016b (The Mathworks, Natick, MA).

B. Experimental Results

1) Rigid feet improved task balance ability: Subjects stood on the beam significantly longer in the rigid feet condition ($M = 236.0\text{s}, SD = 119.4\text{s}$) than in the bare feet condition ($M = 103.1\text{s}, SD = 158.3\text{s}$), ($t_6 = -2.59, p = 0.041$; Fig. 2A).

The RMS of $\hat{v}_{\text{wb},y}$ was significantly reduced in the rigid-feet condition ($M = 0.0024, SD = 0.0005$) compared to the bare-feet condition ($M = 0.0052, SD = 0.0032$), ($t_6 = 2.46, p = 0.049$; Fig. 2B). The RMS of $\hat{L}_{\text{wb},x}$ was also significantly lower in the rigid feet condition ($M = 0.0026, SD = 0.0010$) compared to the bare-feet condition ($M = 0.0078, SD = 0.0050$), ($t_6 = 3.02, p = 0.023$; Fig. 2C).

Together, these results indicate that balance performance was improved when subjects wore rigid soles.

2) Rigid feet altered whole-body coordination: Consistent with the results of [16], $\rho_{L_{\text{ub},x}L_{\text{lb},x}}$ was negative in the bare-feet condition ($M = -0.92, SD = 0.04$), indicating that the angular momenta of the upper and lower body were anti-correlated. Even though $\rho_{L_{\text{ub},x}L_{\text{lb},x}}$ was also negative in the rigid-feet condition, ($M = -0.62, SD = 0.29$), it was significantly increased (i.e., less anti-correlated) compared to the bare-feet condition ($t_6 = -3.00, p = 0.024$; Fig. 2C).

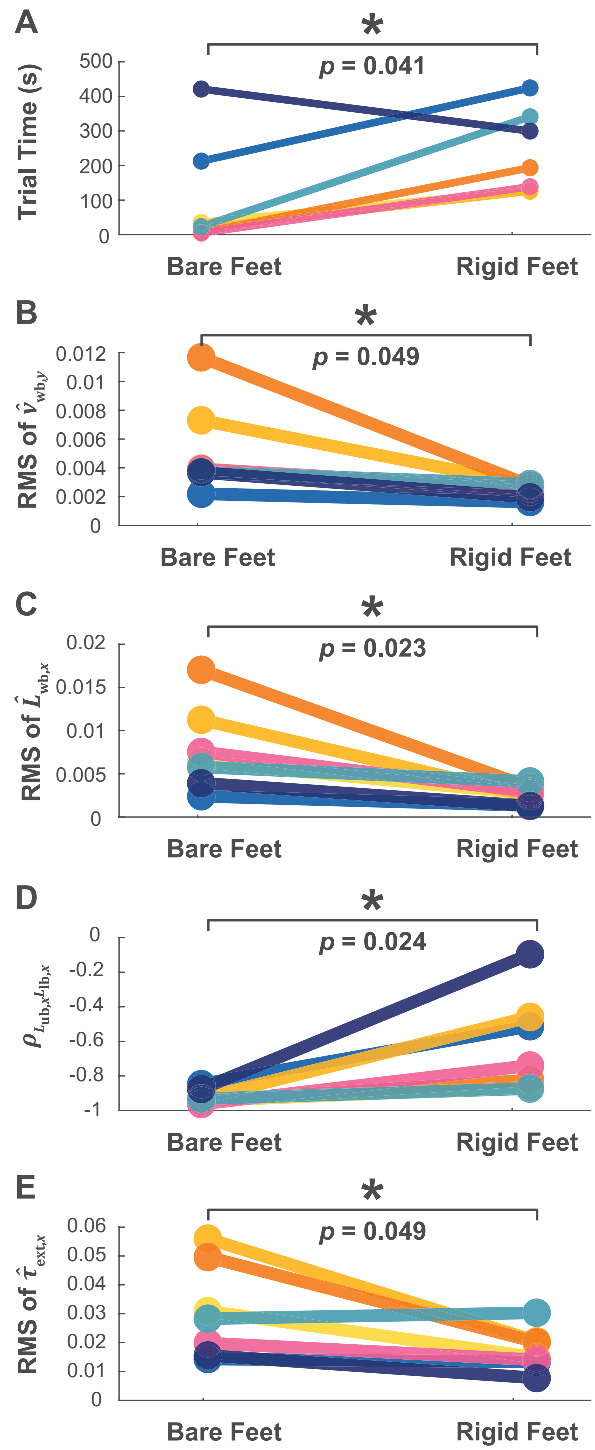

Fig. 2. Experimental Results. (A) Trial time, (B) RMS of center of mass velocity in the mediolateral direction ($\hat{v}_{\text{wb},y}$), (C) RMS of whole body angular momentum ($\hat{L}_{\text{wb},x}$), (D) correlation of upper and lower body angular momentum ($\rho_{L_{\text{ub},x}L_{\text{lb},x}}$), (E) RMS of external torque at foot-beam interaction ($\hat{\tau}_{\text{ext},x}$). Individual subjects are represented in color. An asterisk represents a significant within-subject difference in the two conditions ($p < 0.05$).

TABLE I

DOUBLE INVERTED PENDULUM MODEL PARAMETERS

Parameter	Value [unit]
Mass of lower body	28.36 [kg]
Length of lower body	0.6960 [m]
Center of mass of lower body from ankle	0.3480 [m]
Moment of inertia of lower body about its center of mass	1.145 [kg·m^2]
Mass of upper body	42.54 [kg]
Length of upper body	1.044 [m]
Center of mass of upper body from hip	0.5220 [m]
Moment of inertia of upper body about its center of mass	3.864 [kg·m^2]
Gravitational acceleration	9.810 [m/s^2]

3) Rigid feet reduced estimated foot-beam interaction torque: The RMS of $\hat{\tau}_{\text{ext},x}$ was significantly reduced in the rigid feet condition ($M = 0.0172, SD = 0.0072$) compared to the bare-feet condition ($M = 0.0306, SD = 0.0164$), ($t_6 = 2.47, p = .049$; Fig. 2C).

III. MODELING

In this section, we develop a simple model to describe the effect of altering the foot-beam interaction dynamics on overt balance behavior observed in the human experiment (Fig. 2). Previous work [16] reported that a double inverted pendulum model with the full-state linear quadratic regulator (LQR) could competently reproduce the anti-correlation between upper body and lower body angular momenta. We extended this model to now account for the influence of foot-beam interaction mechanics (Fig. 3).

A. Double inverted pendulum model

Following a general robotic manipulator equation form, the equations of motion of the double inverted pendulum can be written as

$$\mathbf{M}(\mathbf{q})\ddot{\mathbf{q}} + \mathbf{C}(\mathbf{q}, \dot{\mathbf{q}})\dot{\mathbf{q}} + \mathbf{G}(\mathbf{q}) = \boldsymbol{\tau}, \tag{5}$$

where $\mathbf{M}(\mathbf{q}) \in \mathbb{R}^{2\times2}$ is the inertia matrix, $\mathbf{C}(\mathbf{q}, \dot{\mathbf{q}})\dot{\mathbf{q}} \in \mathbb{R}^{2\times1}$ captures the Coriolis and centrifugal forces, $\mathbf{G}(\mathbf{q}) \in \mathbb{R}^{2\times1}$ is the gravitational torques, and $\boldsymbol{\tau} = [\tau_{\text{A}}, \tau_{\text{H}}]^{\text{T}}$ is the input torque vector. The relative angles $\mathbf{q} = [q_{\text{A}}, q_{\text{H}}]^{\text{T}}$ were chosen as generalized coordinates to describe the model. Subscripts A and H indicate ankle and hip, respectively. The model parameters used for simulation are listed in Table I.

B. Balancing controller

The full-state LQR was designed based on the linearized model about its equilibrium point which corresponded to the rest and upright posture ($\mathbf{q}_* = \mathbf{0}, \dot{\mathbf{q}}_* = \mathbf{0}, \boldsymbol{\tau}_* = \mathbf{0}$). The state variables were defined as $\mathbf{x} := [\mathbf{q}^{\text{T}}, \dot{\mathbf{q}}^{\text{T}}]^{\text{T}}$. Control torques, $\boldsymbol{\tau}_{\mathbf{ctl}} = [\tau_{\text{A,ctl}}, \tau_{\text{H,ctl}}]^{\text{T}}$, were determined by the optimal control law obtained by solving the following infinite-horizon optimal problem:

$$\boldsymbol{\tau}_{\mathbf{ctl}} = \arg\min_{\mathbf{u}} \int_0^\infty \mathbf{x}^T\mathbf{Q}\mathbf{x} + \mathbf{u}^T\mathbf{R}\mathbf{u} = -\mathbf{K}_{\text{LQR}}\mathbf{x}, \tag{6}$$

with the parameter matrices $\mathbf{Q}$ and $\mathbf{R}$,

$$\mathbf{Q} = \mathbf{I}_4, \mathbf{R} = \begin{bmatrix} \beta & 0 \\ 0 & \frac{1}{\beta} \end{bmatrix}, \tag{7}$$

such that $\mathbf{Q}$ equally penalize the state errors and $\mathbf{R}$ penalized ankle and hip actuation while allowing the parameter β to control the relative contribution between control actions. Changing the parameter β does not change the determinant of $\mathbf{R}$ so the relative penalty between state errors and actuation were maintained at a similar level. $\mathbf{I}_n$ is the identity matrix with dimension n. In this model, the control torques represent the joint torques generated by human neural controller.

C. Modelling foot-beam interaction

With a flat, stationary foot on a large support surface, the ankle torque and the ground reaction moment may be equal. When standing on a narrow beam, however, the ankle torque may not be directly transmitted to the ground due to the foot-beam interaction dynamics. In double inverted pendulum model, τ_{A} is the external torque applied at the foot-beam interaction port.[1] Therefore, it should be interpreted as the ground reaction moment. To describe discrepancy between the ankle control torque, $\tau_{\text{A,ctl}}$, and the applied torque at the interaction port, τ_{A}, the foot-ankle complex was represented as a torque transmission, with an efficiency factor $\eta \in [0, 1]$:

$$\tau_{\text{A}} = \eta\tau_{\text{A,ctl}} + \tau_{\text{A,pert}} \tag{8}$$

$$\tau_{\text{H}} = \tau_{\text{H,ctl}} \tag{9}$$

where $\tau_{\text{A,pert}}$ is the noise applied to the model in order to reproduce the variability in humans.

This foot-beam interaction model was introduced to describe the change in human behavior across feet conditions without altering the controller. When balancing on the ground, the feet may be regarded as an ideal transmission with $\eta = 1$. Due to foot-beam interactive dynamics, the bare feet and rigid feet could be described as an imperfect transmission with $\eta < 1$. In simulation, we tested whether humans performed better with rigid feet because they acted as more efficient transmission mechanisms than bare feet.

D. Simulation Details

The double inverted pendulum model with LQR was simulated with rest at upright posture as its initial condition. The random perturbation torque was drawn from a uniform distribution on the interval $\tau_{\text{A,pert}} \in [-10, 10]$ N·m as in the previous study [16]. The simulations were conducted using the MATLAB `ode45` function with default options. The solutions were evaluated at 100Hz using the MATLAB `deval` function to compute angular momentum, joint torques, and center of mass velocity. The dependent measures were also normalized as in human data processing (4), using the model mass and height.

[1] τ_{A} of the model is equivalent to $\tau_{\text{ext,x}}$ of the human data.

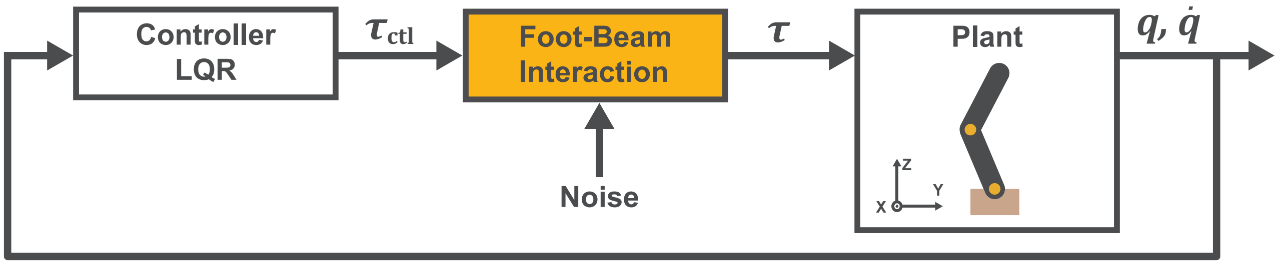

Fig. 3. Block Diagram of the Proposed Human Balance Model

In the simulation, the analysis was carried out by changing parameters β and η, as follows:

- β values from 3 to 5 (step size of 0.5);
- η values from 0.0 to 1.0 (step size of 0.05).

The pairs of parameters resulting in physically infeasible behavior (e.g., negative vertical reaction force, center-of-pressure excursion larger than beam width) or that did not reproduce human behavior (e.g., positive instead of negative correlation between upper and lower body angular momenta) were discarded.

E. Simulation Results

The RMS of the horizontal velocity of the center of mass, $\hat{v}_{\mathrm{wb},y}$, RMS of whole body angular momentum, $\hat{L}_{\mathrm{wb},x}$, correlation coefficient between upper and lower body angular momenta, $\rho_{L_{\mathrm{ub},x}L_{\mathrm{lb},x}}$, and RMS of ankle torque, $\hat{\tau}_A$, for different values of η and β are presented in Fig. 4.

1) Effect of parameter β on the model behavior: Given the range of tested parameters, the model with larger β showed smaller $\rho_{L_{\mathrm{ub},x}L_{\mathrm{lb},x}}$ and smaller RMS of $\hat{\tau}_A$. On the other hand, it was hard to find consistent trends on $\hat{v}_{\mathrm{wb},y}$ and $\hat{L}_{\mathrm{wb},x}$ across all values of β.

2) Effect of parameter η on the model behavior: Given the range of tested parameters, the model with smaller η showed smaller $\rho_{L_{\mathrm{ub},x}L_{\mathrm{lb},x}}$, larger RMS of $\hat{\tau}_A$, larger RMS of $\hat{v}_{\mathrm{wb},y}$, and larger RMS of $L_{\mathrm{wb},x}$.

IV. Discussion

The results of this study revealed that foot-beam interaction dynamics significantly influence human behavior when standing on a narrow support surface. When the contact between the foot and beam was altered by wearing rigid soles, subjects significantly improved their ability to maintain mediolateral balance. While the angular momenta of the upper and lower body were anti-correlated with both bare feet and rigid feet, there was less anti-correlation when the contact was rigid. In addition, the torque at the point of foot-beam interaction was reduced with rigid contact. This change in balance behavior could be reproduced by modeling the foot-ankle complex as a transmission between ankle control torque and the ground reaction moment. With bare feet, the transmission of ankle torque to the ground was less efficient compared to when the feet were rigid.

To reproduce the effect of wearing rigid soles in the human experiment, the model was required to show improved

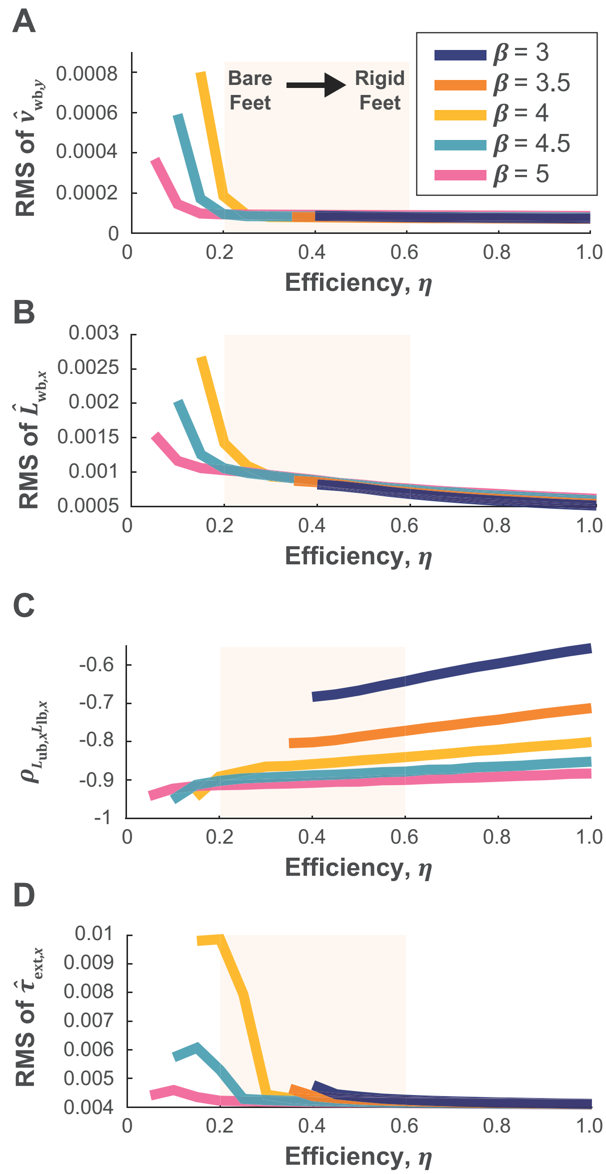

Fig. 4. Simulation Results. (A) RMS of the horizontal velocity of center of mass ($\hat{v}_{\mathrm{wb},y}$), (B) RMS of whole body angular momentum ($\hat{L}_{\mathrm{wb},x}$), (C) correlation of upper and lower body angular momentum ($\rho_{L_{\mathrm{ub},x}L_{\mathrm{lb},x}}$), (D) RMS of external torque at foot-beam interaction ($\hat{\tau}_{\mathrm{ext},x}$). Simulations with different values of the parameter β are represented with colors. The light shaded region indicates the range of $\eta \in [0.2, 0.6]$ that best represents the behavior observed in the bare feet and rigid feet conditions of the human experiment.

balance performance (i.e., reduced $\hat{L}_{\mathrm{wb},x}$ and $\hat{v}_{\mathrm{wb},y}$) with less negative $\rho_{L_{\mathrm{ub},x}L_{\mathrm{lb},x}}$ and decreased $\hat{\tau}_{\mathrm{ext},x}$. As seen in Fig. 4, changing the parameter β of the LQR in the double inverted pendulum model influenced $\rho_{L_{\mathrm{ub},x}L_{\mathrm{lb},x}}$. However, changing β in the model could not reproduce the effect of adding rigid feet observed in the human experiment. The parameter β determines the relative contribution between the joint torques at the hip and ankle. A large value of β puts a large penalty on the ankle torque. As a result, hip torque is largely used to balance, causing anti-phase motion between the two links and negative $\rho_{L_{\mathrm{ub},x}L_{\mathrm{lb},x}}$. On the other hand, a small value of β allows more ankle torque relative to hip torque, resulting in less negative or even positive $\rho_{L_{\mathrm{ub},x}L_{\mathrm{lb},x}}$. Although decreasing β could replicate the reduced anti-correlation in the rigid feet condition, it also increased $\tau_{\mathrm{ext},x}$. In the human experiment, $\hat{\tau}_{\mathrm{ext},x}$ was decreased in the rigid feet condition. Hence, it is likely that the change of the mechanical interface was responsible for the altered human behavior, not a change in the controller. Moreover, it was previously found that wearing rigid soles had an immediate effect on balance during beam walking and practice with the rigid soles had no effect on subsequent performance with bare feet [18], which similarly suggests that subjects did not adapt their neural control strategy. The same control strategy with different values of parameter β may instead account for differences in behavior across individual subjects.

By considering the ankle-foot complex as a transmission mechanism with a parameter η, we could meet the requirement without modifying the controller. In the model, η was considered smaller (i.e., lower torque transmission efficiency) in the bare feet condition compared to the rigid feet condition. For an arbitrary control torque, $\tau_{\mathrm{A,ctl}}$, a smaller η yields a smaller applied torque, τ_{A}. Over time, however, a smaller η results in a greater accumulation of state errors, which causes an increase in control torque $\tau_{\mathrm{A,ctl}}$. Thus, simulation of the model with a smaller η actually resulted in greater τ_{A}, as well as worse balance performance (Fig. 4D). Because τ_{H} was unaffected by η, a smaller η reduced the contribution of ankle torque relative to hip torque, resulting in a more negative $\rho_{L_{\mathrm{ub},x}L_{\mathrm{lb},x}}$. An important question for future work is how the torque transmission loss, which is currently lumped into a single parameter η, can explained by physiological reasoning.

V. Conclusions

This study showed that including a highly-simplified model of foot-beam interaction dynamics could reproduce several features of human beam balancing. Importantly, it demonstrated that the difference in human behavior between the bare feet and rigid feet conditions could be explained by accounting for foot-beam interaction dynamics rather than a change in the controller. In addition to advancing our basic understanding of human balance, the development of competent, yet simplified models of human neuromechanical control, such as the one presented here, may provide valuable and fundamental insight for improving balance control in bipedal robots.

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