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A Biomechanical Model of Human Ankle Angle Changes Arising From Short Peri-Threshold Anterior Translations of Platform on Which a Subject Stands

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Abstract

This study modeled ankle angle changes during small forward perturbations of a standing platform. A two-dimensional biomechanical inverted pendulum model was developed that uses sway frequencies derived from quiet standing observations on a subject's Anterior Posterior Center of Pressure (APCoP) to track ankle angle changes during a 16 mm anterior displacement perturbation of a platform on which a subject stood. This model used the total torque generated at the ankle joint as one of the inputs, and calculated it assuming a PID controller. This feedback system generated a simulated ankle torque based on the angular position of the center of mass (CoM) with respect to vertical line passing through the ankle joint. This study also assumed that the internal components of the net torque were only a controller torque and a sway-pattern-generating torque. The final inputs to the model were the platform acceleration and anthropometric terms. This model of postural sway dynamics predicted sway angle and the trajectory of the center of mass. Knowing these relationships can advance an understanding of the ankle strategy employed in balance control.

Keywords

Sway Frequencies; Biomechanical Model; Ankle Angle; APCoP; Center of Mass

I. INTRODUCTION

GOOD balance and mobility are necessary to perform acts of daily living independently and to avoid falls causing injuries and/or hospitalization. The trajectories of the Center of Mass (CoM) and Center of Pressure (CoP) provide measures of stability and are important parameters when modeling the human postural control system. Robinson et al. have shown that differences exist in the APCoP patterns between short anterior perturbations made near the psychophysical detection threshold that are correctly detected and those that are not [1],[2]. When this finding

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is represented in the frequency domain, the frequency response curve of the system mimics a classical Second Order Linear Differential Equation (SOLDE), with inertial, stiffness and spring constants [2]. This suggests that a biomechanical model with the characteristics of a second order system can be developed to model the biomechanical responses to short anterior perturbations made at psychophysical threshold.

The model basically consists of two parts, one is PID controller and the other is the transfer function for the inverted pendulum model of a person standing on the platform. The values of the mass moment of inertia (J) and stiffness (K) that are used in the model depend on anthropometric data that will differ between subjects. The coefficient of viscous damping B is calculated from the assumed value of damping coefficient ζ. The computed ankle angle can be then compared to that actually observed using motion analysis techniques during the same experiments.

II. METHODOLOGY

A. Subjects, Equipment and Protocols

The data for the study was colleted from young adults at Louisiana Tech University and Clarkson University, with Institutional Review Board (IRB) approved protocols from both the universities.

An innovative Sliding Linear Investigative Platform for Assessing Lower Limb Stability with Synced Tracking, EMG and Pressure measurements (SLIP-FALLS-STEPm) was used for these tests [3].

Two different protocols were used to collect data from the SLIP-FALLS-STEPm system. One is a 2-Alternative Forced Choice Protocol (2AFC) that makes the subject report the interval in which (s)he felt the movement [1],[2]. Another is Single-Interval Adjustment Matrix (SIAM) Protocol in which the subject reports whether (s)he felt the move or not instead of reporting where (s)he felt it [4].

B. Data Collection and Analysis

The 2AFC data was sampled at 1000Hz and stored in a raw file that was later converted off-line into proper engineering units using batch processing. The newer SIAM protocol performs this conversion on-the-fly [5].

C. Biomechanical Model Design

This section systematically explains the model design procedure.

1) Free Body Diagrams—The sway dynamics of a person standing on a moving platform can be approximated with the classical model of an inverted pendulum with a moving base. Fig. 1 shows the free body diagram of the human body as an inverted pendulum on a translating platform and Table I explains all the terms used. While the inverted pendulum model is a classic example of a non-linear system, we show here that postural control via such a model in our experimental situation using small perturbations is well described by a linear system.

For the pendulum model shown in Fig.1, the equation for moment balance is:

$$J\ddot{\theta} - m_1 g d_1 = \tau_{ankle} - m_1 \ddot{x} d_1 \tag{1}$$

Where J is the moment of inertia of a body around the ankle joint, calculated as $m_1 d_1^2$.

Earlier works on postural stability have shown that the balanced state of a person is decided by the two moments acting at the ankle joint in opposite directions [6]. One is the torque generated by the subject's weight acting about the ankle, called as gravitational torque and the other is the total ankle torque. To avoid a possible collapse caused by the weight vector as well as platform movement, this counteracting total torque is produced at the ankle joint by passive and active elements. This torque is shown as τ_{ankle} in the free body diagrams (Fig. 1) and is used as one of the inputs to the model. The other input to the model is the acceleration term shown in the right side of Eq. 1.

The ankle torque τ_{ankle} is a combination of two torques. One is the torque generated by the central nervous system (CNS) that uses muscle elements as well as passive elements present at the ankle joint (τ_c). The second is an oscillatory sway pattern generator torque (τ_s) that generates sway patterns similar to those observed experimentally [7].

The conceptual block diagram of the model is shown Fig.2.

Fig. 2 gives the block diagram of the model in which the PID controller takes the error signal as an input for calculating torque τ_c. This error signal is the difference between the reference sway angle (assumed to 0°) and the sway angle calculated at a previous instant of time. The blocks PID controller and the Sway Pattern Generator are explained as follows.

2) Design of Controller Mechanism—To ensure the presence of a controller mechanism, this model uses a PID controller to stabilize the system. The P and D components of the controller are based on the ankle stiffness and viscous damping coefficients in such a way that the Routh criterion for stability is satisfied [6]. These P, I, D components are selected as follows:

$$P = m_1 g d_1 \theta(t) + K\theta(t) \tag{2a}$$

$$I = \partial \int_{t_0}^{t} \theta(t) \tag{2b}$$

$$D = B\,\dot{\theta}(t) \tag{2c}$$

Thus, the value of θ at a particular instant of time determines the controller component values. K is the stiffness coefficient. K will be equal to $J\omega^2$ where ω is the undamped natural frequency of the sway, calculated as 2πf, and f is the frequency at which a person sways. It is calculated by doing Fourier Transform of subject's quiet standing APCoP and getting the frequency at which the maximum energy of the signal is present. ∂ is the neuromuscular delay occurring in sending the signal to muscles from the CNS. B is the viscous damping coefficient and is equal to, $B = 2\zeta\sqrt{JK}$ where ζ is assumed to be equal to 0.5.

3) Design of Sway Pattern Generator—The first step in designing the sway pattern generator or the oscillator for our model was to extract the frequencies present in the APCoP signal. This could be done separately for each of the 30 trials in the 2AFC test, but the pre-test length is only 4 seconds. However, we had available to us 20 s of data from each subject during which they were standing quietly on the platform. Fourier transforming this QS data yields the three frequencies, f_1, f_2 and f_3 where the maximum power was present. We used these frequencies f_1, f_2 and f_3 to form the oscillator (Eq. 3) for the system and generate the sway using the following equation,

$$\widehat{s}(t) = A_1 \sin(2\pi f_1 t + \varphi_1) + A_2 \sin(2\pi f_2 t + \varphi_2) + A_3 \sin(2\pi f_3 t + \varphi_3) \tag{3}$$

Note that the accurate calculation of phase angles ϕ_1, ϕ_2 and ϕ_3 is very important to be able to generating a pattern similar to that observed experimentally. Cross-Correlation techniques were used to obtain the phases ϕ_1, ϕ_2 and ϕ_3 . A subject's captured QS sway profile was thus used in this to get the phases and to ensure that the generated sway will be similar to the captured one. The amplitudes, A_1, A_2 and A_3, were selected in such a way that the sway generated would be within the same range of the captured sway, i.e., APCoP. The combination of controller and sway pattern generator torques gives the total torque τ_{ankle} that, along with the perturbation term $m_1\ddot{x}d_1$, acts as an input to the model as shown in the Fig. 2. The delay includes propagation and muscle contraction delay.

III. Results

The model was implemented in SIMULINK and the sway angle time series for a healthy female mature adult was calculated for a series of thirty trials of a 16 mm move. Fig. 3 shows the raster plot for the ankle angle calculated for 10 trials with the offset of 1.5. These trials were categorized into Interval 1(Fig. 3a) and Interval 2 (Fig. 3b) based on in which interval the perturbation was given.

Note that, for both the intervals, a solid black line represents the ankle angle calculated for the trial taken at threshold acceleration. All the ankle angles above this threshold trial are Detects, and all the ankle angle profiles below are Non-Detects.

Fig. 4 shows the net total torque produced at the ankle joint calculated using the model for one of the trials and calculated from the measured data as well as the torque produced by the passive elements of the joint. The calculated torque shown was obtained by doing moment balancing (Eq. 4) on Figure 1b.

$$\tau_{\text{calculated}} = -\left(APCoP.R_V - m_2 g d_4 - m_{plate}\, g o f f - R_H d_2 + m_2 \ddot{x} d_2/2 + m_{plate}\ddot{x}(d_2 + 10)\right) \tag{4}$$

The controller torque generated by the PID controller counteracts a forward moment due to the weight moment arm. The slope is equivalent to a stiffness factor. The negative slope in Fig. 4b arises from the fact that τ_{ankle} is defined in an opposite direction from ankle angle θ. A conclusion is that τ_{ankle} is activated to oppose an angular displacement at the ankle joint and acts as a balancing torque [7].

IV. DISCUSSION

A few models for dynamic postural perturbation exist. These were developed by Masani, et al. [7], Johansson, et al. [8], and Sinha, et al. [9]. The model developed here takes into account all the possible torques introduced from the dynamic nature of our testing, and uses a Second Order Linear Differential Equation to determine resultant changes in sway angle. The use of a PID controller in the model ensures the presence of necessary stiffness and damping to make the system stable.

A. Frequencies to be used in a model

The newness lies in how we modeled the design of the sway pattern generator. We generated a modeled sway by using frequencies taken from a subject's quiet standing response. Then we correlated this modeled sway with the subject's APCoP for a given trial to calculate the phase differences for the equation of the oscillator (Eq. 3). Finding the frequencies f_1, f_2 and f_3 used in the Eq. 3 was a critical task as these frequencies should be close enough to simulate subject's actual sway seen experimentally. Initially, only a single frequency was used to simulate the subject's sway. However, with only one frequency, the results did not match a subject's response. Hence, we can assume that there are multiple frequencies present in a subject's sway.

We then used at least three frequencies (extracted from subject's QS data) to re-generate the subject's response using simple harmonic motion.

B. CoG and CoP Behavior

The model's validity can be seen by observing the calculated Center of Gravity (CoG) profile. CoG is the horizontal distance that CoM travels with respect to the ankle joint. It is evident that the CoP tracks the CoG and oscillates on either side of it to keep the CoG within the desired position between the two feet [11], [12]. The same behavior of CoG and CoP was observed for the developed model. As seen in Fig. 5, the CoP is oscillating about CoG to maintain balance. They are always in phase with each other. This behavior was observed for the data collected by the protocols, 2AFC as well as SIAM.

V. CONCLUSION

The model designed helps us to understand the behavior of the total torque generated at the ankle joint when the system is perturbed by anterior moves. This ankle torque acts as a balancing torque to avoid a potential fall. To achieve this, it provides a necessary ankle stiffness and damping to the system. The sway angle change calculated using this biomechanical model is very small, which is understandable since the perturbation given to the platform itself is very small (16 mm for 2AFC). The design of the sway pattern generator showed the presence of multiple frequencies in human sway. The CoG and CoP profiles showed the behavior of the CoP in keeping the subject in a controlled balanced state.

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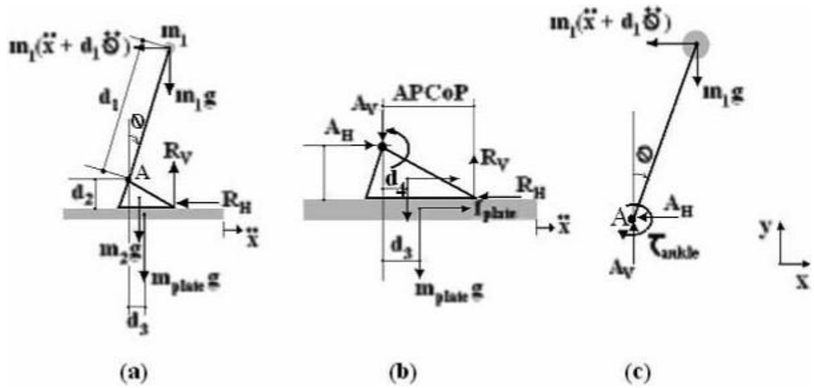

Fig. 1.
(a) Model of a person standing on the platform as an inverted pendulum; (b) The free body diagram of both the feet together; (c) The free body diagram of the body excluding feet.

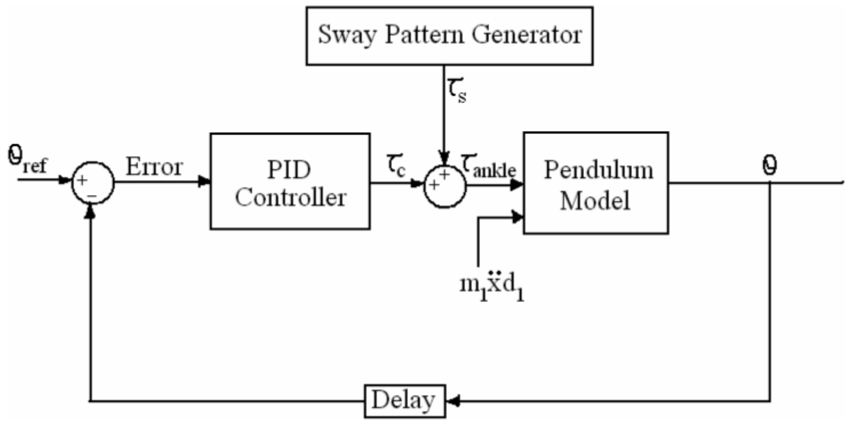

Fig.2.
Block diagram of a Designed Model

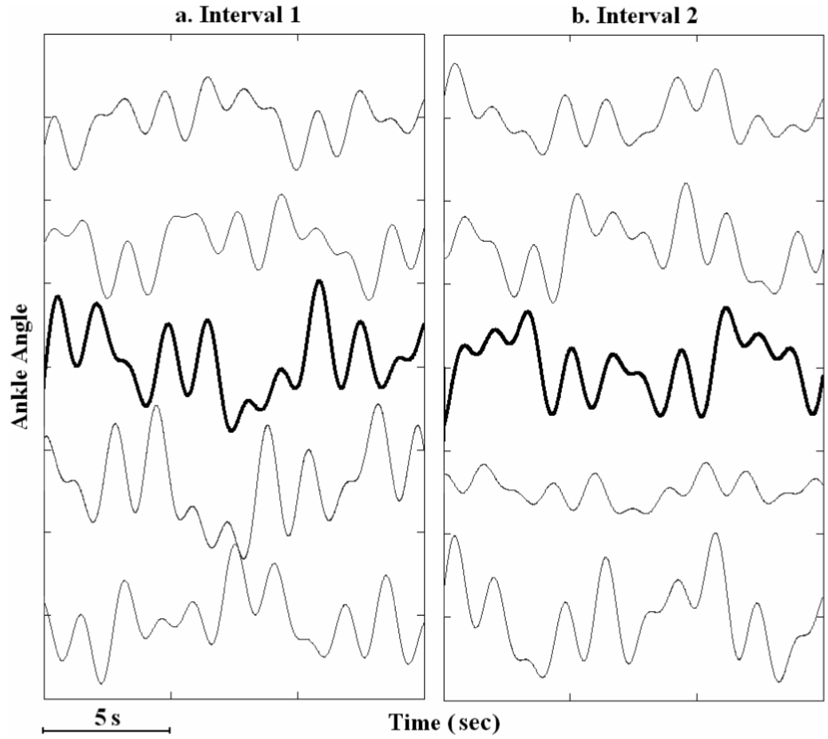

Fig. 4.
(a) Ankle Angle changes plotted for Interval One Trials with offset of 1.5; (b) Ankle Angle changes Plotted for Interval Two Trials with the offset of 1.5

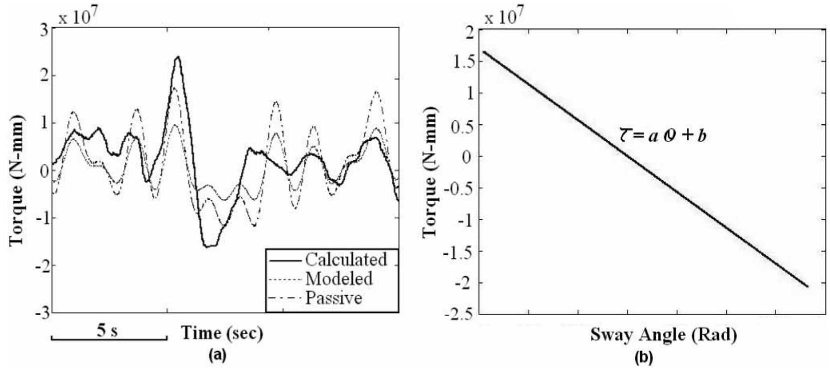

Fig. 4.
(a) Various Torques Produced at the Ankle Joint; (b) Passive torque Plotted against the Ankle Angle

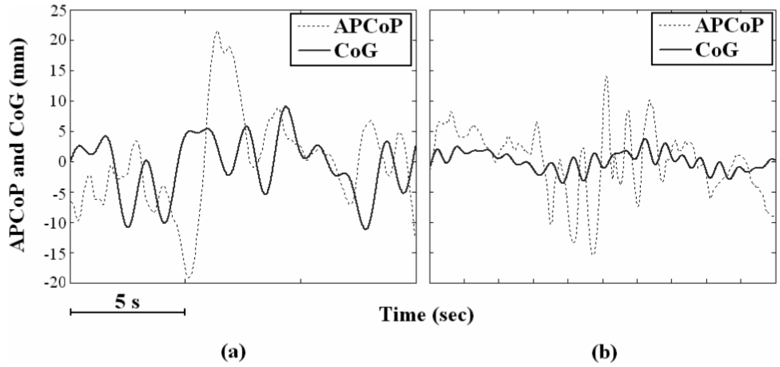

Fig. 5.
(a) CoP and CoG Behavior for 2AFC Protocol; (b) CoP and CoG Behavior for SIAM Protocol

TABLE I

PARAMETERS USED TO MODEL HUMAN STANDING ON THE PLATFORM AS AN INVERTED PENDULUM

m_1	Mass of a body excluding feet
m_2	Mass of the feet
m_{plate}	Mass of the platform
A	Ankle joint
Θ	The ankle angle with respect to vertical line
d_1, d_2, d_3, d_4	Anthropometric measurements for the subject
$\ddot{x}$	Platform acceleration
A_H, A_V	Horizontal and vertical forces acting at the ankle joint
R_H, R_V	Platform shear force and vertical ground reaction force respectively
τ_{ankle}	The total torque produced at the ankle joint
APCoP	Anterior posterior center of pressure
f_{feet}, f_{plate}	Forces generated at feet and plate respectively due to the linear acceleration of the platform
off	Horizontal distance between ankle joint A and center of the platform.