A Biomechanical Model That Confirms Human Ankle Angle Changes Seen During Short Anterior Perturbations of a Standing Platform

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Abstract— Humans sway. With bipedal stance on a fixed base of support, this sway causes changes in the location of the Center of Pressure as it is projected onto this base of support. Translating the base itself also causes changes in the Center of Pressure. In this study, we have developed a two-dimensional biomechanical model that uses changes in the Anterior-Posterior Center of Pressure (APCOP) to track ankle angle changes arising from 16 mm anterior displacement perturbations of a platform on which a subject stands. The model uses the total torque generated at the ankle joint as one of the inputs, and calculates it assuming a PID controller. The necessary stiffness and damping is provided by the P and D components of the controller. This feedback system generates 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 assumes that the internal components of the net torque are controller torque and sway pattern-generating torque. The final inputs to the model are the platform acceleration and some anthropometric terms. This model of postural sway dynamics predicts sway angle and the trajectory of the center of mass; and points out the relationships among the biomechanical variables like ankle angle, torque, center of pressure, and center of mass.

I. Introduction

Good balance and mobility are necessary in order to independently perform acts of daily life and to avoid falls. Balance is a functional term that is generally defined as the ability to maintain and control the position and motion of the total Center of Mass (COM) of a body. There has been considerable study of the postural stability of humans based on analyses of Center of Mass (COM) and Center of Pressure (COP) trajectories.

Robinson has 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]. His group has shown that there is an inverse power law relationship between acceleration threshold and displacement. When this finding 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[1]**.** 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 person standing on 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

Young adults for the study were recruited from Louisiana Tech University through word-of-mouth. The Protocol for testing and the informed consent document were reviewed and approved by the Institutional Review Board (IRB) of the Overton Brooks VAMC, Shreveport, LA. The experiments were conducted at the Shreveport VA. Data from these experiments were used to develop and test the biomechanical models of this paper.

B. Equipment

An innovative Sliding Linear Investigative Platform for Assessing Lower Limb Stability with Synced Tracking, EMG and Pressure measurements (SLIP-FALLS-STEPm)[3] was used for these tests. The SLIP uses a non-contact linear motor and air bearing slides to eliminate vibration, thus obviating potential cues for movement. The FALLS system collects inputs from load cells that are placed under the platform plate to computes the AP and ML Centers of Pressure and total subject weight. This system monitors inputs from differential EMG electrodes on the right and left tibialis anterior and gastrocnemious soleus muscles, a tri-axial head accelerometer, a horizontal accelerometer on the platform, platform position (with 5 µm resolution) and motor drive voltage which is proportional to horizontal ground reaction force[2,3]. The perturbation parameters can be automatically titrated to threshold through novel LabVIEW™ routines.

C. Protocol

A 2-Alternative-Force-Choice Protocol (2AFC) was employed where a subject was forced to decide in which of two sequential intervals that a perturbation was presented.. The sequential commands “Ready”, “One”, “Two” and “Decide” were given to the subject in successive 3 to 4 s intervals during which there would be stimulus either in Interval One or Interval Two. After the prompt “Decide”, the subject pressed a telemetered switch once to signify that (s)he felt that the perturbation occurred in Interval One; and twice if in Interval Two.

Acceleration threshold was adaptively determined over 30 trials for a given fixed displacement. Perturbation displacements of 1, 4, and 16 mm were used, but only the 16 mm set is considered here. A training set of 10 trials for each displacement was given to the subject before the actual testing began at that fixed displacement. A test sequence began with the subject being instructed to stand quietly on the platform for 20s, thus providing a measure of a Quiet Standing profile. Then a series of trials was carried out for every displacement to determine the acceleration threshold with the displacement held constant throughout a $\leq$ 30 trial test set. The Parameter Estimation by Sequential Testing (PEST) algorithm was used to iterate the amplitude of the acceleration stimuli toward detection threshold.

D. Data Collection and Analysis

Perturbation parameters such as platform acceleration and position were sampled at 1000Hz and stored in a raw file that was later converted off-line into proper engineering units using batch processing. The sampled data from the four load cells was used on-line to calculate the location of the APCOP as well as medial-lateral Center of Pressure (MLCOP). These data were also stored and later converted into engineering units. These data were used for the biomechanical model of this paper. This data was also point-by-point averaged and divided into detects and non-detects for intervals One and Two, yielding four different data sets that were then used for analyzing COP profiles[3].

E. Biomechanical Model Design

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. Figure 1 shows the free body diagram of human body as an inverted pendulum on a translating platform. 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.

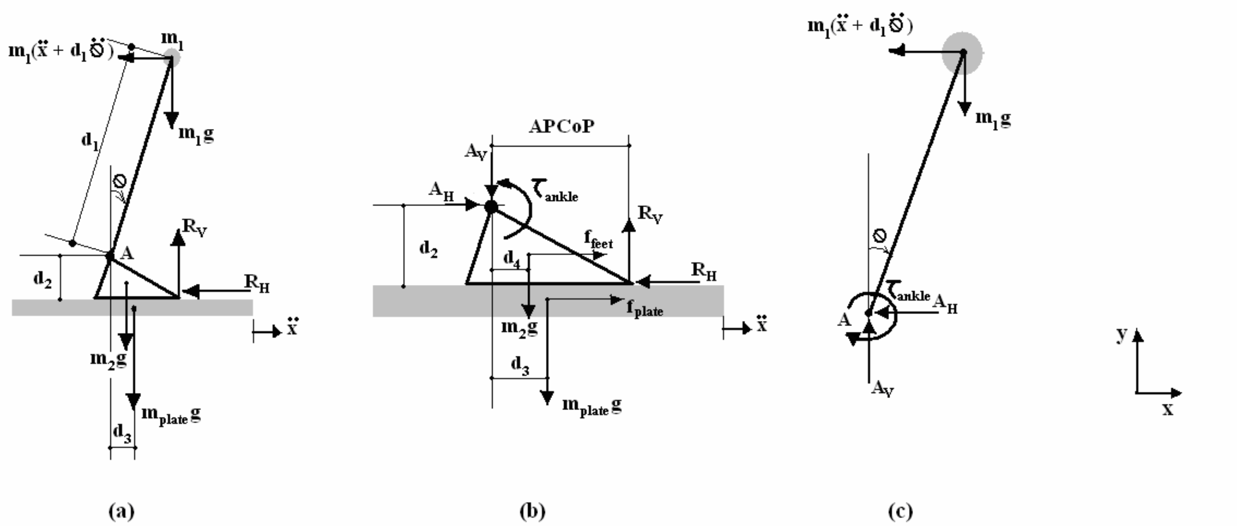


Fig. 1(a) Model of a person standing on the platform as an inverted pendulum.; Fig. 1(b) The free body diagram of both the feet together; Fig. 1(c) The free body diagram of the body excluding feet.

Table 1: Model parameters used to mimic human being standing on the platform as an inverted pendulum

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

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

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

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

As shown in the earlier works on postural stability, the balanced state of a person is decided by the two moments acting at the ankle joint in opposite directions. One is the torque generated by the subject's weight acting about the ankle, and the other is the total ankle torque. To avoid a possible collapse caused by the weight vector as well as the plat-form 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 above free body diagrams and will be one of the inputs to the model. The other input to the model will be the acceleration term shown in the right side of Equation 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), and an oscillatory sway pattern generator torque (τ_s) that generates sway patterns similar to those observed experimentally[4]. This model uses a PID controller to stabilize the system as shown by Equation 1. The P and D com-ponents of the controller are selected based on the ankle stiffness and viscous damping coefficients in such a way that the Routh criterion for stability is satisfied[5]. These P, I, D components are selected as follows:

$$P = m_1 g d_1 \theta(t) + K\theta(t)$$

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

$$D = B\dot{\theta}(t)$$

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 $2\pi f$, where $f \approx 0.5$ Hz. ∂ 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. Figure 2 gives the block diagram of the model in which the PID controller takes the error signal as an input for calculating the torque τ_c. This error signal is the difference between the reference sway angle (assumed to 0^o) and the sway angle calculated at a previous instant of time.

Another important part of the input is the sway pattern generator torque τ_s. In the past, this torque has been modeled using Gaussian noise to obtain sway patterns seen experimentally. In this study, we have modeled it using the time series record of a subject's APCOP. This not only ensures that the sway pattern will be similar to that obtained experimentally but also that multiplication of the APCOP by the weight also ensures the magnitude of this torque is at least equal to quiet standing torque. This latter torque is given by $APCOP.wt$ [6].

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.

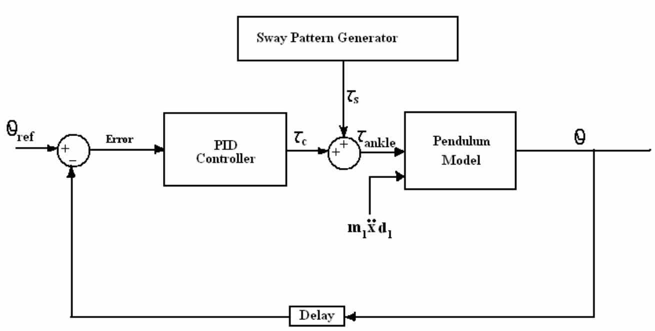


Figure 2: Block diagram of the sway model developed

A major input to the model is the time series record of the changes in the Anterior Posterior Center of Pressure (APCOP). Our data acquisition system calculates on-line the time series records of APCOP for every trial from the data collected from four force transducers in the SLIP. However, these values are calculated with respect to center of the plate. Since the ankle joint is the reference point in our model (see Fig. 1), APCOP must be measured with respect to the location of the ankle joint. We have an alternative, and completely identical (except for an offset), measure of APCOP that the STEPm system calculates off-line from the TekMat data. But, from this latter data, we do know the location of the ankle, as shown in Figure 3 below.

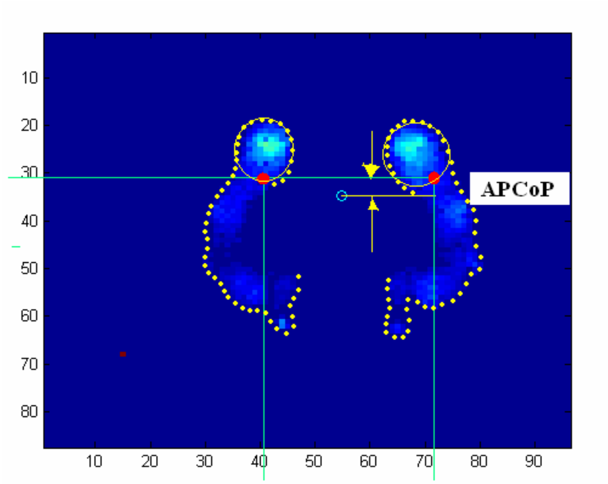


Figure 3: The x, y, pressure matrix plotted as an image at particular frame

The Tekscan system allows each trial to be saved in the form of a movie. Every 15 s trial corresponds to 750 frames in the Tekscan movie file (i.e., a sampling rate of 50 Hz). This system also provides an option of saving the location of COP and raw units (based on the pressure exerted) for each sensor element (sensel) of the HR Mat into an ASCII file. Off-line Matlab routines load the recorded movie and extract the data value of each sensel element in the form of an x, y, pressure matrix for each frame, and converts the data from sensel spacing units (5.08 mm center-to-center) to mm. The diagram in Figure 3 is the plot of such a matrix as an image at a particular frame. As seen in the image, we can easily identify the heel and then infer an aggregate ankle location. As our analysis is in 2D plane, the desired APCOP is nothing but the difference between the y-coordinates of ankle and that of COP (plotted as circle in Figure 3).

III. Results

The model in Fig. 2 was implemented in SIMULINK and the sway angle for a female elderly adult was calculated for 16mm move. The protocol for the data collection yielded 30 trials worth of data at 16 mm. Here the results are provided for only one trial.

Using the sway angle, the center of mass (COM) profile for the subject was calculated to help understand the postural behavior of the subject under short perturbations.

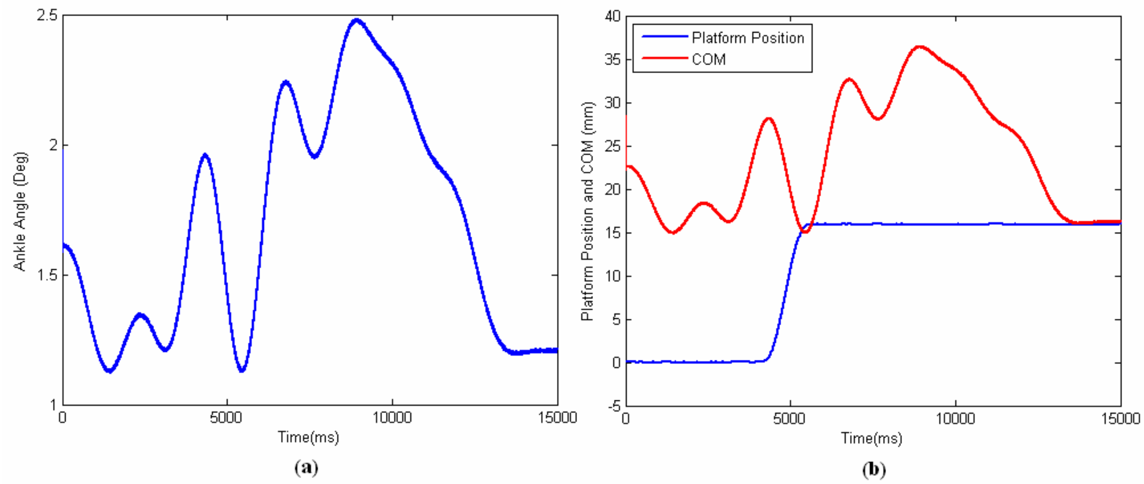


Fig. 4(a) Ankle angle changes calculated by the model for a 15s trial
Fig. 4(b) Calculated COM position along with the platform position for 16mm move

Figure 4 shows the subject's AP sway path during this trial. As the perturbation length was very short, the angle around which the subject swayed was also very small. As seen in the figure, the COM excursion increased during the movement of the platform. All of the positive values of ankle angle in Fig. 4(a) show that the subject was always leaning forward throughout the testing. This is understandable as there is always a slight forward lean associated with human standing. The forward lean ranged from 1.1° to 2.5°. Note the posterior excursion from 2° to 1.1° during the anterior move itself, a recovery from 1.1° to 2.5° just after the termination of the move, and a final transition back to 1.1°.

Figure 5 shows the net total torque produced at the ankle joint calculated using the model and the torque calculated from the measured data. This latter calculated torque was obtained by doing moment balancing on Figure 1(b) as follows,

$$\tau_{\text{calculated}} = -(APCOP.R_V - m_2 g d_4 - m_{plate} goff - R_H d_2 + m_2 \ddot{x} d_2 / 2 + m_{plate} \ddot{x}(d_2 + 10)) \qquad (2)$$

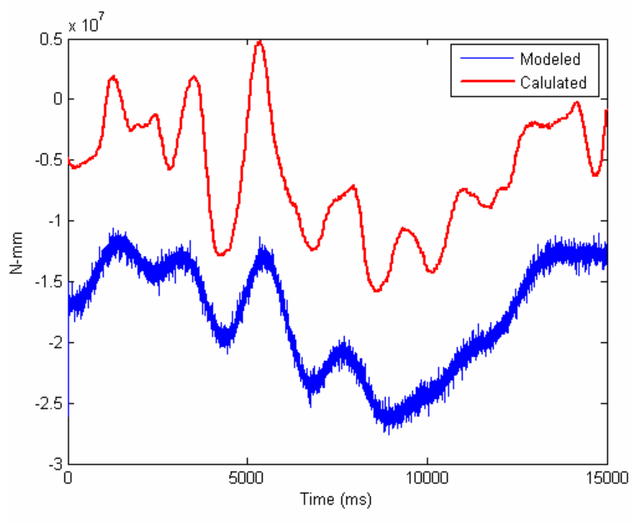
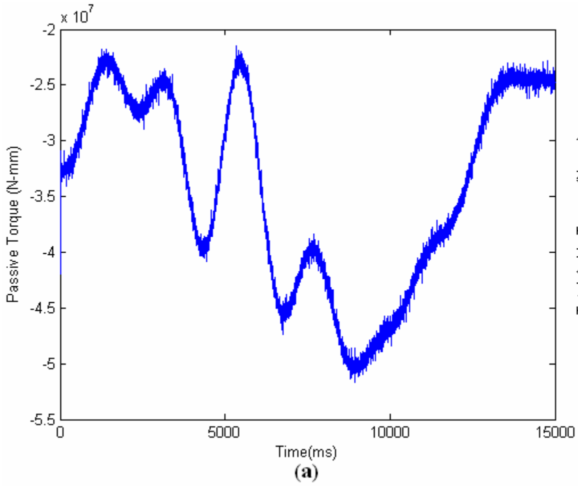


Figure 5: Modeled and calculated total torques at the ankle joint

Figure 5 shows both the ankle torques, one calculated using Eq. 2 and the one obtained from the biomechanical model with the PID controller. On average, there is an offset difference of 1N-m which is acceptable. And, the profile of both curves is similar.



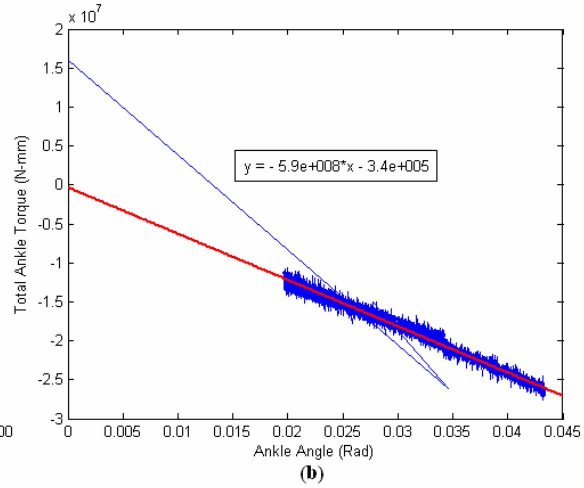


Fig 6(a) Torque produced by PID controller *Fig 6(b) Total ankle torque plotted against sway angle*

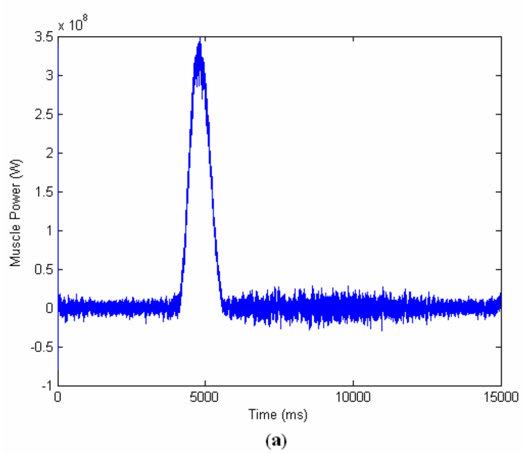
Figure 6(a) shows the controller torque generated by the PID controller to counteract a forward moment due to the weight moment arm. Figure 6(b) shows the linear relation-ship between τ_{ankle} and the ankle angle θ. The slope is equivalent to a stiffness factor. The negative slope in figure 6(b) arises from the fact that τ_{ankle} is defined in the opposite direction from ankle angle θ**.** A conclusion is that τ_{ankle} is activated opposite to the angular displacement at the ankle joint and acts as a balancing torque[4].

The torque produced not only contributes towards the input of this system, but it is also an essential contributor in calculating another biomechanical aspect, muscle power. Muscle power is the scalar product of the joint torque and a segment's angular velocity. Based on this definition, the rate of work done by the muscles at the ankle can be calculated as

$$P_{AK} = \tau_{ankle}\left(\omega_{FT} - \omega_{CM}\right) \quad (3)$$

where ω_{FT} and ω_{CM} are angular velocities of the feet and COM respectively, considering Fig.1 as 2-segment body of feet and upper body with COM at m_1.

Fig. 7 shows the rate at which work was done by the muscles at the ankle joint. As seen in the figure, a significant amount of work is done, or more power generated, during the movement of the platform. Of particular interest is that this profile is very similar to that of the perturbation velocity shown in figure 7(b).



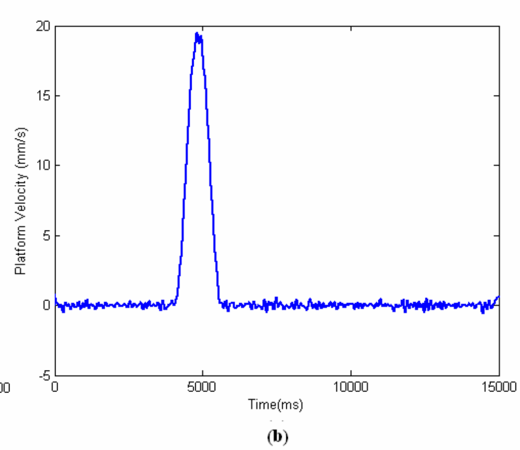


Fig. 7(a) Variations in the muscle power during a 16mm move. *Fig.7 (b) Platform Velocity*

IV. Discussion

A few models for dynamic Postural perturbation exist. These were developed by Masani, et al.[4], Johansson, et al.[5], and Sinha, et al.[6] 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. Also, instead of modeling the sway pattern generator torque using Gaussian noise, the APCOP signal is used to give more realistic pattern.

It is evident that the COP tracks the COM and oscillates on either side of it to keep the COM within the desired position between the two feet[8,9]. The same behavior of COM and COP was observed for the developed model. As seen in Figure 8, the COP is oscillating about COM to maintain balance. They are always in phase with each other.

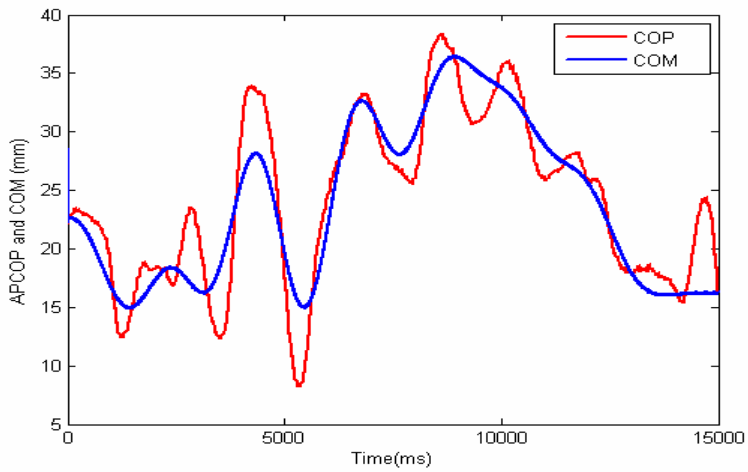


Fig. 8 COM and COP behavior during testing

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 the potential fall. To achieve this, it provides a necessary ankle stiffness and damping to the system. The sway angle calculated using this biomechanical model is very small, which is understandable since the perturbation given to platform itself is very small (16 mm). The COM and COP profiles show the behavior of the COP in keeping the subject in a controlled balanced state. This model can also prove good theoretical validation to the ankle angles observed via a 3D motion capture system.

VI. Future Work

The model initially makes assumptions about the damping coefficient as well as the undamped natural frequency of sway. These assumptions can be avoided by calculating these values dynamically. To achieve this, the viscous damping coefficient B could be considered as the function of platform velocity, and the value of damping can be calculated as a function of time. The model can also be tested against the subjects of differ-ent categories based on their age, peripheral neuropathy and their sway angle profiles can be compared. It has also been seen from the APCOP profiles that there may exist specific response pattern profiles during the movement as shown in Fig. 4(b). Support vector machines can be utilized to extract such kind of specific pattern.

VII. References

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