Contributors to Postural Stabilization : A Modeling-Simulation Study

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ABSTRACT

The neuro-physiological mechanisms involved in postural stabilization are not well understood. Human body mechanically resembles an inverted pendulum that is inherently unstable. Active and passive mechanisms at muscle level, as well as other visual and vestibular processes, are attributed to stability. The available evidence suggests that muscle stiffness alone is insufficient to stabilize body sway, and must rely on active mechanisms of stabilization that are unlikely to have a reflex nature due to the intrinsic delays in the reflex pathways and the low-pass characteristics of the muscle response. The role played by the central nervous system in active control of stance thus remains an open and intriguing question. In this study we present simulation results to support an active-passive model of postural stabilization. Besides expanding our understanding of the postural stabilization process, the insight gained would be useful to promote intervention techniques for therapists and clinicians working with fall-prone individuals.

KEYWORDS: *postural stabilization, biomechanical model, muscle stiffness*

1. INTRODUCTION

Postural stabilization refers to maintenance of posture amid perturbations to the base of support (BOS), defined as the area under the two feet when in contact with the ground. The BOS thus defines a stability boundary whereby the standing posture is preserved as long as the center of mass (COM) of the whole body is maintained over the BOS. A fall becomes imminent if the COM travels outside of the BOS. In the literature a two dimensional view of the postural stabilization problem is usually considered. Investigators [1] have reported that standing humans when perturbed by forward-backward translation of a moving support surface and instructed not to move their feet, typically respond by moving in the sagittal plane, using a combination of "hip" and/or "ankle" strategies, whereas larger disturbances trigger grasping and/or stepping responses.

Biomechanical models of varying complexity have been extensively used in literature to study coordination of postural and voluntary movements. Control of balance in human upright standing is particularly well suited for modeling, and is also a popular experimental paradigm [2,3,4,5]. Modeling in this experimental paradigm (disturbances to posture in the sagittal plane) is particularly convenient, as the dynamics can be linearized without significant impact on the simulation accuracy. For example, Iqbal and Pai [3] have used a four-segment biomechanical model to study the role of knee joint in postural stabilization. Pai and Iqbal [6] have similarly used an inverted pendulum model over moving foot to investigate slipping, sliding, and falling behavior.

The role and contribution of the central nervous system (CNS) in the postural stabilization process has been a topic of intense debate among researchers [7,8,9,10,11]. Some researchers [11] have postulated that passive stiffness at the ankle joint plays a predominant role in postural stability. This view is strengthened by the consideration that latencies in the motor servo loop and low pass muscle characteristics tend to limit the effectiveness of active mechanisms. However, as Morasso and Schieppati [8] point out, the observed value of ankle stiffness by, e.g. [12,13], falls well short of that predicted by the stiffness-only models. Therefore active CNS involvement in the postural stabilization process can not be ruled out. It can be hypothesized that the massive computational power allows CNS to make predictive assessment of proprioceptive feedback and issue timely intervention commands to counter any perturbations. Alternatively, the data base of postural reactions, gathered and perfected through years of experience and stored in the CNS, combined with timely processing of information, allows CNS to issue just the right commands to allay the effects of perturbations.

The aim of this paper is to employ a modeling-simulation paradigm to study the relative contribution of active and passive mechanisms towards postural stability. In the modeling scenario active and passive mechanisms translate into feedforward and feedback commands issued to a biomechanical model. These can then be varied to find the optimum mix to support execution of a postural movement between two given end points. The task objective may be defined as minimization of residual dynamics and position errors at movement termination. A quantitative criterion (or a cost function) can then be employed to evaluate the movement, and parameter optimization methods can be used to obtain the set of parameters most suited to the task.

This paper is organized as follows: Model formulation and optimization methods are discussed in Section 2,

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followed by simulation results in Section 3, and discussion and conclusions in Section 4.

2. MODEL FORMULATION

The human body is modeled as a multi-segment structure comprising skeletal, muscular, and sensory subsystems as shown in Figure 1. More specifically, a four-segment planar rigid-body biomechanical model connected by single-degree-of-freedom joints is used for sagittal plane mechanics. The segments represent a bilateral symmetrical arrangement of the feet, legs, thighs, and head-arm-trunk (HAT) with stationary foot segment. The model parameters, i.e., the segment length, mass, center of gravity and moment of inertia, are based on gross average anatomical proportions of the human body [14]. The length of the stationary foot segment defines the base of support (BOS) in the anterior-posterior direction. The leg, thigh, and HAT each have a single-rotational degree of freedom. The CNS commands and active force generation in the muscle are represented by torques applied at the ankle, knee, and hip joints. The joint torques each comprise a feedforward and a feedback component, where the feedback component represents the action of passive viscoelasticity in the muscle tendon and the CNS response to proprioceptive feedback from muscle spindle and Golgi tendon organ (GTO).

1.1 *Model Dynamics*

In this section we state the dynamic equations for the four-segment rigid body biomechanical model as shown in Figure 1. To that effect, define the following variables

$$\theta \triangleq [\theta_1, \theta_2, \theta_3]^T \tag{1}$$

where $\theta \in \mathbb{R}^{3\times 1}$ represents the segment angle vector measured from the horizontal. Then the model equation can be derived from Euler-Lagrangian dynamics (see Appendix) which is formulated as a matrix differential equation given by

$$D(\theta)\ddot{\theta} + H(\theta,\dot{\theta})\dot{\theta} + G(\theta) = E \cdot \tau \tag{2}$$

with the initial conditions $\theta(0) \triangleq \theta_i, \dot{\theta}(0) \triangleq v_i$, where $D \in \mathbb{R}^{3\times 3}$, $H \in \mathbb{R}^{3\times 3}$, and $G \in \mathbb{R}^{3\times 1}$; $E \in \mathbb{R}^{3\times 3}$ is the input matrix given as

$$E \triangleq \begin{bmatrix} 1 & -1 & \\ & 1 & -1 \\ & & 1 \end{bmatrix} \tag{3}$$

The total torque τ is comprised of the feedback (τ_{fb}) and feedforward (τ_{ff}) torques, i.e., $\tau = \tau_{fb} + \tau_{ff}$, where the feedforward torques are represented as a sigmoidal variation between the static torques supporting end points given by

$$\tau_{ff} = \tau_i + (\tau_f - \tau_i) \cdot \frac{1-e^{-ct}}{1+e^{-ct}}, \quad c \in \mathbb{R}_+, \quad t \geq 0 \tag{4}$$

where the static initial and final torques are computed as

$$G(\theta_i) = E \cdot \tau_i, \quad G(\theta_f) = E \cdot \tau_f \tag{5}$$

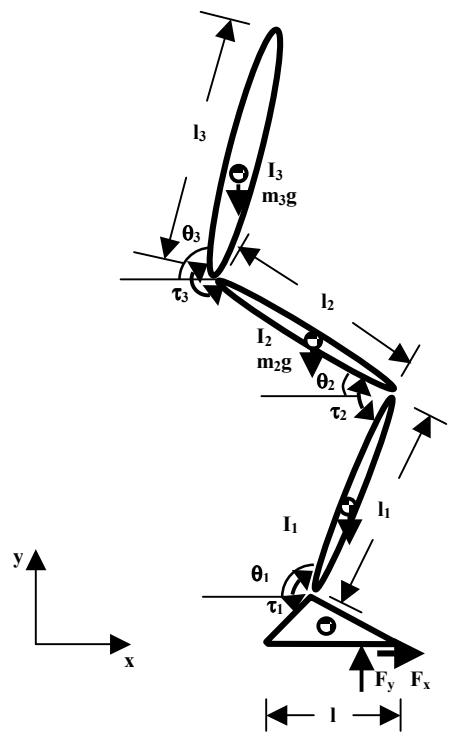

Figure 1. A multi-segment biomechanical model

The feedback torque τ_{fb} is represented as sum of three components: a position dependent component, a velocity dependent component, and a nonlinear component that represents physiological restraints on the joint range of motion. The feedback torque is mathematically given as

$$\tau_{fb} = -K_1(\theta - \theta_{ref}) - K_2\dot{\theta} - K_3 \cdot g(\phi) \tag{6}$$

where the reference segment angle θ_{ref} is given by a sigmoidal variation between the initial and final angles in the trajectory:

$$\theta_{ref} = \theta_i + (\theta_f - \theta_i) \cdot \frac{1 - e^{-ct}}{1 + e^{-ct}} \tag{7}$$

and

$$\phi = \begin{bmatrix} 1 & 0 & 0 \\ 1 & -1 & 0 \\ 0 & -1 & 1 \end{bmatrix} \theta - \begin{bmatrix} \pi/2 \\ 0 \\ 0 \end{bmatrix} \tag{8}$$

are the local angles at the joints needed to represent physiological bounds on mobility.

Let $\phi_{low} = \left[\frac{\pi}{6}, 0, 0\right]$ and $\phi_{high} = \left[\frac{\pi}{3}, \frac{2\pi}{3}, \frac{\pi}{3}\right]$ be the lower and upper bounds for ϕ; then the function $g(\cdot)$ is a nonlinear function of ϕ and is defined as

$$g(\phi) = e^{(\phi - \phi_{sat})} - 1 \tag{9}$$

where $\phi_{sat} \in [\phi_{low}, \ \phi_{high}]$ is given as

$$\phi_{sat} = \begin{cases} \phi_{low}, & \phi \le \phi_{low} \\ \phi, & \phi_{low} < \phi < \phi_{high} \\ \phi_{high}, & \phi \ge \phi_{high} \end{cases} \tag{10}$$

1.2 *Movement Simulation and Optimization*

The four-segment biomechanical model was built and simulated on the computer in MATLAB-SIMULINK® environment. The inputs to the simulation model included static posture at the start and the termination of movement. Model outputs included time histories of all kinematic and dynamic variables. A comprehensive cost function was constructed to provide quantitative evaluation of the simulation run. The components of the cost function included the following: (a) position errors from the reference trajectory, (b) terminal position errors, (c) residual velocities and accelerations, and (d) the overall jerk (time derivative of acceleration) encountered in the execution of movement.

These components were suitably weighted and summed up in the cost function. An optimization routine was employed to determine muscle synergies and system parameters for optimal system performance. Additionally, active and passive components of joint moment were individually varied to arrive at an optimal combination for postural stability. Due to the highly irregular shape of the cost function, a probabilistic optimization scheme (Simulated Annealing) was employed to determine the system parameters for optimum performance. The optimization variables included: (a) initial segment velocities, (b) the delay and the slope constant for joint torque profiles, and (c) the feedback gains representing linear proprioceptive feedback and nonlinear physiological restraints on joint motion.

The optimization results represented the selection of system parameters and muscle synergies to execute a skilled postural movement.

3. SIMULATION RESULTS

We chose a typical movement in the sagittal plane where the start and end positions were given as

$$\theta_i = \left[\frac{\pi}{2}, \frac{\pi}{2}, \frac{\pi}{2}\right] \tag{11}$$

$$\theta_f = [0.58, \ 0.33, \ 0.55] \cdot \pi \tag{12}$$

The optimization routine was used to select the parameter values (initial segment velocities and feedback gains) for the movement. The delay and slope constant were kept constant during optimization. The resulting values of the optimization variables are given as:

$$\dot{\theta}_0 = [-0.046, \quad 0.049, \quad -0.014] \text{ rad/sec} \tag{13}$$

$$K_1 = 12.9 \text{ Nm/rad}, \ K_2 = 3.1 \text{ Nm/rad} \cdot \text{sec},$$

$$K_3 = 1915 \text{ Nm/rad} \tag{14}$$

Figures 2-7 respectively show the movement trajectories, the segment velocities, COM trajectory, the feedforward torques, the feedback torques, and the total torques.

As seen from the plots, the movement is not very smooth. Specifically, the initial part of the movement has induced oscillations which result from a high value of feedback gain K_3. Lowering K_3 did damp out the oscillations but resulted in small knee extension (3 degrees) beyond physiological limits. Another probable cause for non-smooth movement could be that optimization was performed over a limited number of variables. We believe that a smoother movement would have resulted if more variables were included in the optimization process.

4. CONCLUSIONS

In this paper computer simulation results have been provided to support an active-passive model of postural stabilization. We found that stabilization by passive ankle stiffness alone requires artificially high values (above 1500 Nm/rad). However, active regulation of muscle tone by the CNS coupled with moderate amounts of passive stiffness (400-500 Nm/rad) can provide effective regulation amid disturbances

at the base of support. We also found that even small latencies in the reflex loop compromised postural stability. Predictive estimation of proprioceptive feedback was shown to improve the delay tolerance.

In conclusion, a good understanding of the CNS control of postural stabilization process continues to elude the researchers. This study shows that stiffness-only models take an overly simplistic view of complex stabilization problem. Whereas, moderate passive stiffness combined with active regulation of muscle tone can effectively stabilize human body against perturbations.

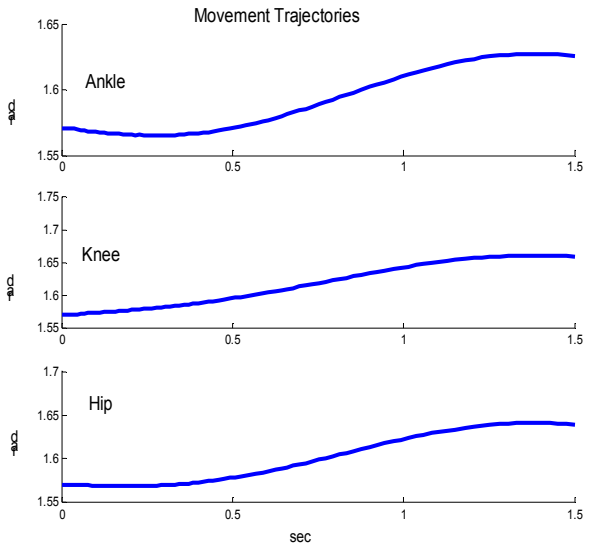

Figure 2. Movement trajectories for the four-segment biomechanical model.

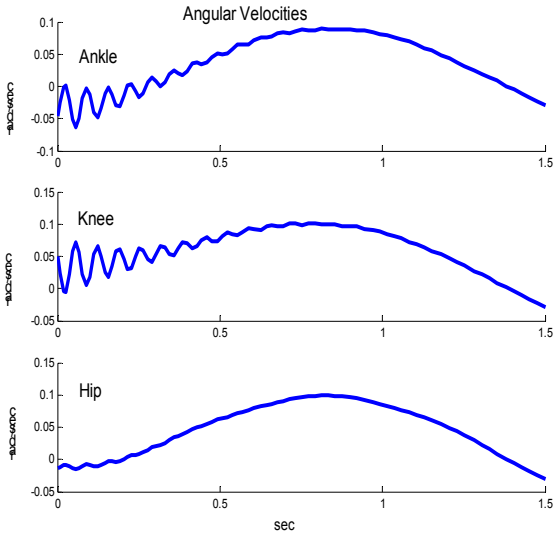

Figure 3. Segment velocities for the four-segment biomechanical model.

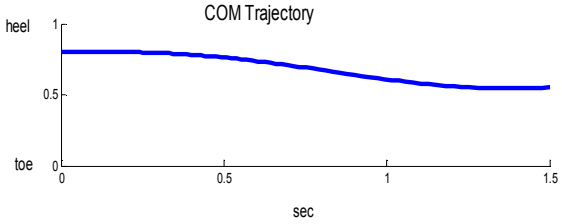

Figure 4. Center of mass (COM) trajectories for the four-segment biomechanical model.

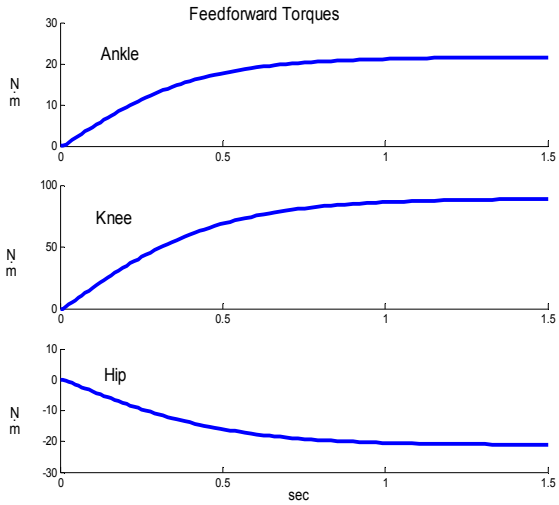

Figure 5. Feedforward torques for the four-segment biomechanical model.

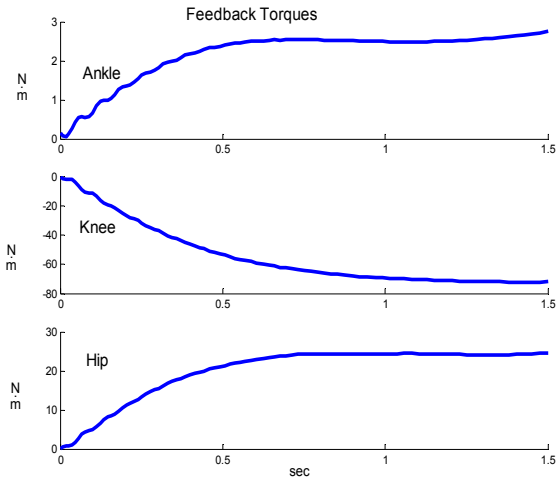

Figure 6. Feedback torques for the four-segment biomechanical model.

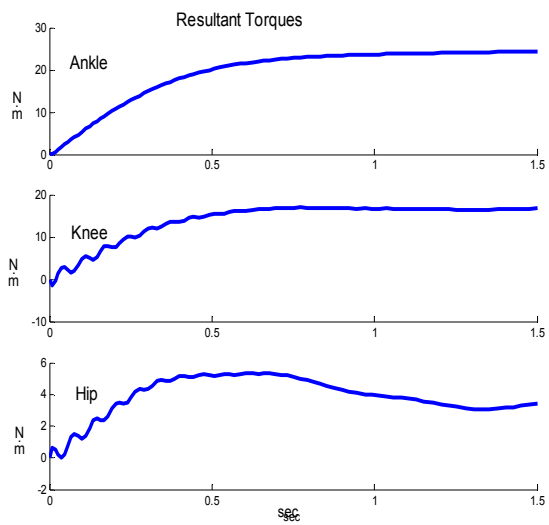

Figure 7. Resultant torques for the four-segment biomechanical model.

5. APPENDIX : DYNAMIC EQUATIONS

Let $\theta_1, \theta_2, \theta_3$ be segment angles in inertial reference frame; $\dot{\theta}_1, \dot{\theta}_2, \dot{\theta}_3$ be segment velocities; $\ddot{\theta}_1, \ddot{\theta}_2, \ddot{\theta}_3$ be segment accelerations; τ_1, τ_2, τ_3 be net joint moments; d_{11}, d_{22}, d_{33} be components of the inertia matrix; f_1, f_2, f_3 be gravitational moment components; m_1, m_2, m_3 be segment mass vector; I_1, I_2, I_3 be segment moments of inertia vector; l_1, l_2, l_3 be segment lengths; k_1, k_2, k_3 be position of the segment centers of mass; l_f, m_f be foot mass and length respectively; *a, b, c* be ankle-heel, ankle height, ankle-foot COM respectively; *g* be gravitational acceleration; τ_{ff}, τ_{fb} be feedforward and feedback moments; and τ_i, τ_f be initial and final values of moments.

Then the dynamic equations of the four-link model can be derived using Euler-Lagrangian dynamics. These equations are given as

$$d_{11}\ddot{\theta}_1 + d_{12}[\ddot{\theta}_2 \cos(\theta_1 - \theta_2) + \dot{\theta}_2^2 \sin(\theta_1 - \theta_2)]$$
$$+ d_{13}[\ddot{\theta}_3 \cos(\theta_1 - \theta_3) + \dot{\theta}_3^2 \sin(\theta_1 - \theta_3)] + g f_1 \cos\theta_1 = \tau_1 - \tau_2$$

$$d_{22}\ddot{\theta}_2 + d_{12}[\ddot{\theta}_1 \cos(\theta_1 - \theta_2) - \dot{\theta}_1^2 \sin(\theta_1 - \theta_2)]$$
$$+ d_{23}[\ddot{\theta}_3 \cos(\theta_2 - \theta_3) + \dot{\theta}_3^2 \sin(\theta_2 - \theta_3)] + g f_2 \cos\theta_2 = \tau_2 - \tau_3$$

$$d_{33}\ddot{\theta}_3 + d_{13}[\ddot{\theta}_1 \cos(\theta_1 - \theta_2) - \dot{\theta}_1^2 \sin(\theta_1 - \theta_3)]$$
$$+ d_{23}[\ddot{\theta}_3 \cos(\theta_1 - \theta_3) - \dot{\theta}_3^2 \sin(\theta_1 - \theta_3)] + g f_3 \cos\theta_3 = \tau_3$$

where

$$d_{11} = m_1 k_1^2 + (m_2 + m_3) l_1^2 + I_1$$

$$d_{22} = m_2 k_2^2 + m_3 l_2^2 + I_2, \; d_{33} = m_3 k_3^2 + I_3$$

$$f_1 = m_1 k_1 + (m_2 + m_3) l_1, \; f_2 = m_2 k_2 + m_3 l_2$$

$$f_3 = m_3 k_3, \; d_{12} = f_2 l_1, \; d_{13} = f_3 l_1, \; d_{23} = f_3 l_2$$

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