A MUSCULOSKELETAL MODEL OF POSTURAL CONTROL AT THE ANKLE

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INTRODUCTION

Changes in postural stability associated with the aging process may be due to alterations in neural, skeletal and muscular characteristics. The effects of such changes are difficult to study experimentally, as they typically occur over a long time period. Therefore, many have studied postural control using analytical or numerical models, such as the simple inverted pendulum [3]. However, most models do not incorporate specific representation of the individual muscles involved in the balance process. The purpose of this paper is to describe a musculoskeletal model for studying control of sagittal plane postural sway at the ankle [3]. Development of this model will allow the study of how age-related changes in muscular properties will affect postural control.

METHODS

For initial testing of the model, experimental data were collected on a single male subject undergoing voluntarily upright sway in the sagittal plane at a frequency of ~1/3 Hz for 20 s. Center of mass motion (CoM) was measured in 3-D with an eight-camera Qualysis™ system, center of pressure (CoP) was measured with two AMTI™ force platforms, and muscle activity of the right limb soleus (SO), gastrocnemius (GA), and tibialis anterior (TA) muscles was recorded with a Delsys™ electromyography (EMG) system. Segment lengths were measured, and their masses and inertial characteristics were estimated with standard anthropometric scaling.

The musculoskeletal model was comprised of two segments representing the feet and rigid body, linked by a frictionless hinge ankle joint, confined to sagittal plane movement. The model's segment anthropometrics were matched to the subject. The equations of motion for the model were found with Autolev™, and can be expressed in general form as:

$$A(\theta)\alpha = B(\theta,\omega) + gC(\theta) + F_A + F_C$$

where θ, ω and α are vectors of segment angle, angular velocity and angular acceleration respectively; $A(\theta)$ is the inertia matrix; $B(\theta,\omega)$ is the vector describing Coriolis and centrifugal effects; $gC(\theta)$ defines the gravitational effects; and F_A is the net ankle joint torque. F_C represents foot-floor constraint forces, modeled with a series of 21 spring-damper elements along the length of the foot [1]. The ankle torque was generated by three Hill-type models representing the SO, GA, and TA muscles. Each Hill actuator consisted of non-linear series elastic and contractile components, with model parameters drawn from the literature. The experimental EMG signals were rectified, smoothed, and scaled to use as control signal shape templates for the Hill models. Muscle lengths and moment arms were computed using subject-scaled polynomial functions of joint angle modeled from SIMM™.

A variable step-size Runge–Kutta–Merson integrator was used to simulate the model motion, using initial conditions from the experimental data. A differential evolution genetic algorithm [2] was used to find the control signal scaling levels (λ), which maximized the time the model was able to remain upright. The same λ was used for SO and GA. Model performance was assessed by comparison of CoM and CoP time series with experimental data. Model sensitivity was studied by changing maximum isometric force (P_0) values in the Hill muscle models using five different literature sources (Table 1).

RESULTS AND DISCUSSION

The model was able to stay upright for the entire 20 s trial, indicating that the muscle models, control signals, and foot-floor interaction were realistic. The shape of the model's CoM and CoP motion closely matched the experimental data, with the major difference being that the model did not "lean back" as far as the subject (Figure 1).

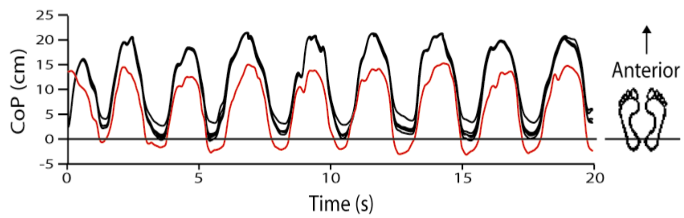

Figure 1. Experimental (thin / red) and model (Mod 1 to Mod 5; black) CoP sagittal displacement referenced to the ankle joint.

Optimized λ values (Table 1) and simulation performance were sensitive to the P_0 values used for the Hill actuators. With increases in the absolute TA P_0, the model leaned back farther as measured by its CoM minimum (r = -.94). Weaker relationships were found between the SO (r = -.57) and GA (r = -.70) P_0 values and the model CoM minimum. Although the TA P_0 values had a large range (1253 to 3114 N), the optimized λ values did not (0.72 to 0.82). The SO / GA λ values varied more (0.48 to 0.75). This suggests that the model results were dictated by the dorsiflexor TA. With a weaker TA, the model kept its CoM farther in front of the ankle joint to prevent unstoppable backward sway that would result in falling. Future work on age-related changes in muscular properties will focus on the inclusion of subject-specific parameters using MRI and ultrasound technology.

Table 1. Optimization parameters and results.

	Actuator	Mod 1	Mod 2	Mod 3	Mod 4	Mod 5
P_0 (N)*	SO	8367	8674	6976	5377	5898
	GA	5039	4854	4850	2869	3116
	TA	3114	2867	2303	2295	1253
λ	SO, GA	0.49	0.48	0.53	0.75	0.70
	TA	0.72	0.74	0.82	0.73	0.80

*Model P_0 sets [Mod 1 to Mod 5] drawn from studies in the literature.

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