

A MUSCULOSKELETAL MODEL OF POSTURAL CONTROL: SIMULATED AGING OF MUSCLE MECHANICAL PROPERTIES

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INTRODUCTION

Recent work has implicated age-related changes in muscular properties as detrimental to postural stability [1, 2]. However, prospective studies are usually constrained to follow subjects for only a few years, rather than over a human life span [3]. Also, the numerous age-related anatomical, physiological, and neural changes make it difficult to draw causal relationships between specific neuromuscular changes and their effect on postural control. An alternate approach is to use a musculoskeletal model, where simulated age-related changes can be applied independently and evaluated systematically.

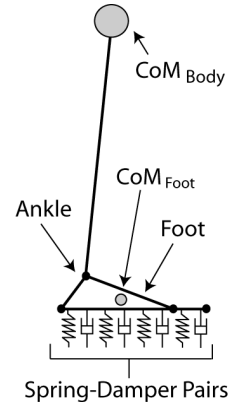
Previous modeling studies have investigated the ability of different neural controllers to regulate posture and the effects of ankle joint stiffness [4, 5]. Here we extend this work with an inverted pendulum postural dynamics model incorporating muscle mechanical properties derived from young and older adults. A feedback-based neural controller is used to activate Hill muscle models, with upright stance simulated by finding optimal controller gains. The muscle model properties are then systematically altered to investigate how age-related changes affect postural control.

METHODS

Equations of motion were derived with Autolev™ for a 2-segment sagittal plane inverted pendulum model powered by individual ankle muscles. A frictionless hinge ankle joint linked the skeletal segments; the foot-floor interaction was modeled with 21 spring-damper elements (Figure 1). A variable step-size integrator was used to integrate the model state equations to simulate upright stance.

Three Hill muscle models representing the dorsiflexors (DFs) and plantarflexors (PFs; individual gastrocnemius and soleus muscles) actuated the skeletal model. Each muscle model included a contractile component (CC) with non-linear excitation-activation, force-length (FL) and

force-velocity (FV) properties, and a series elastic component (SEC) with a non-linear force-extension relationship ($F\Delta L$). The properties were defined by 9 parameters: maximal isometric force (P_0), optimal CC length (L_0), FL width (W), SEC slack length (L_S) and stiffness coefficients (α , β), and Hill FV constants (a/P_0 , b/L_0 , ϵ). Ankle



passive stiffness was represented by a non-linear torque-angle relation. Muscle kinematical models were based on SIMM™ [6].

Two age-specific postural models were created using the average muscle parameter values from 6 young and 6 older male subjects [1]. Motor commands for postural stabilization were generated using proportional-derivative (PD) feedback control based on existing torque-generator models [4], modified to control individual muscle models. A genetic numerical optimization algorithm was used to find optimal DF and PF controller gains (K_D and K_P) to minimize center-of-mass (CoM) motion and muscular force while maintaining upright stance. A penalty was applied if the model fell.

Next, simulations were performed with the optimized young model controls while independently changing each of the nine PF muscle parameters to the mean value of the older male subjects. We focused on the PFs because they provide the majority of ankle torque during quiet stance. Here, the young controller gains were kept at their optimized levels, however in the face of a sudden muscle property alteration, it is likely that neural control would adapt to maintain stability. Therefore, a sensitivity analysis was performed in which each PF muscle parameter was altered separately and the model control gains re-optimized. Each muscle parameter was varied from optimal across 6 levels ($\pm 5\%$, $\pm 10\%$, and $\pm 15\%$). Because some older

parameters were outside this range, more optimizations were performed with parameters set to the older values.

RESULTS AND DISCUSSION

Both young and old models were able to achieve stable posture and qualitatively reproduced basic CoM and center-of-pressure (CoP) dynamics during quiet stance; however, the older model needed higher controller gains and greater muscle excitation than the younger model (Figure 2).

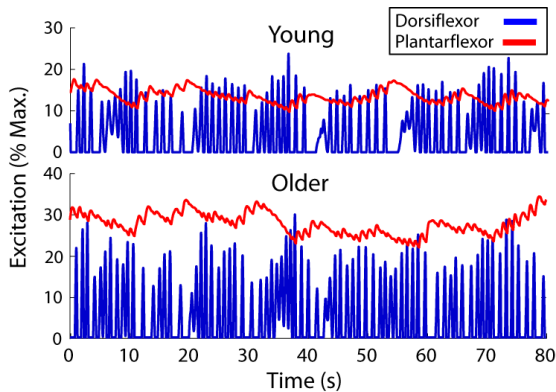


Figure 2. Optimized muscle model excitations.

Replacing all nine PF muscle parameters with older values resulted in the young optimized model promptly falling forward. When each parameter was “aged” independently, the model fell when P_0 , L_0 , and L_S were changed to older values, but remained standing when the other 6 were changed.

In the sensitivity analysis with re-optimized controller gains, the model was more robust to changes in muscle mechanical properties, and achieved stable balance in most cases. However, the model’s behavior was sensitive to changes in P_0 , L_0 , and L_S , with changes in K_P needed for optimal balance. In some cases, the model was unable to balance no matter what the control adjustments.

Changes in PF P_0 were inversely related to the optimal PF K_P (Figure 3), because weaker muscles require more excitatory drive. Nevertheless, the young model was able to remain standing even with the 42% lower older P_0 , provided the control gains were allowed to increase sufficiently. This suggests that quiet stance performance is robust to changes in P_0 (at least until a lower limit is reached).

Changes in L_0 and L_S had nonlinear effects on the controller gains (Figure 3). The PF K_P gains were close to their minimum values using the original set of young muscle mechanical properties, but increased as L_0 and L_S shifted from the young

optimum. The young model fell when the older L_0 was used. Together, this suggests that the L_0 and L_S combination in the young subjects was close to optimal for minimizing muscle excitatory drive. Conversely, the young model’s control of quiet stance was not sensitive to changes in the FV parameters or the FAL stiffness coefficients (α , β).

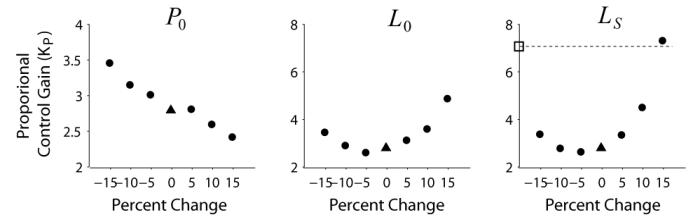


Figure 3. (●): Sensitivity analysis for K_P . (▲): Optimized young model with experimentally estimated muscle properties. (□): Model with older parameters. Older result not shown for P_0 (-42%, $K_P = \sim 5$), model fell using the older L_0 value (0.30%).

We also observed a negative relation between the mean model CoP speed and L_S . As the FL parameters were not changed in concert with L_S , decreasing L_S caused the CC to operate at longer lengths. This shifted the CC into a weaker/unstable FL position, resulting in greater control inputs and a larger CoP speed. This may explain age-related increases in CoP speed [1], as the older subject’s L_S was 20% shorter than the young.

CONCLUSIONS

We developed age-specific versions of a feedback-driven postural control model that included DF and PF muscle properties estimated from young and older adults. The model was able to balance using either set of properties, and predicted age-related changes in muscle activity consistent with experimental studies [1]. A sensitivity analysis showed that P_0 , L_0 , and L_S had the most influence on the model’s selection of neural controller gains, with the latter two properties being close to optimal when experimentally measured values were used. This is important, since our previous work [1] has shown age-related differences in L_0 and L_S .

REFERENCES

- Hasson CJ. Dissertation, Univ. of Mass. Amherst, 2009.
- Onambele GL, et al. J App Physiol **100**: 2048-56, 2006.
- Brauer SG, et al. J Gerontol A **55**: M469-476, 2000.
- Perteka RJ. J Neurophysiol **88**: 1097-1118, 2002.
- Winter DA, et al. J Neurophysiol **80**: 1211-1221, 1998.
- Delp et al. IEEE Trans Biomed Eng **37**: 757-767, 1990.

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