EFFECTS OF AGING-RELATED LOSSES IN MUSCLE STRENGTH ON THE FEASIBLE REGION FOR BALANCE RECOVERY

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INTRODUCTION

Backward falls are a concern for older adults. They are the hardest falls to prevent and can lead to serious injury, including hip fracture. It is therefore important to understand the factors that adversely influence the ability to prevent a backward fall. One such factor may be aging-related losses in muscle strength. Muscle strength decreases with older age, and weaker lower extremity muscles have been associated with lesser balance control among older adults [1]. Conceivably, aging-related losses in muscle strength might impair the ability to support the weight of the body and stop its downward motion following a recovery step. Yet in one study, older adults who fell following an induced trip were stronger than those who did not [2]. At present, it remains unknown to what extent aging-related losses in muscle strength affect the ability to recover from a backward balance loss. This study used a mathematical modeling approach, along with a modification of the concept of the feasible region for balance recovery [3], to investigate the effects of aging-related losses in muscle strength on the ability to restore static balance after a recovery step from a backward balance loss.

METHODS

A six-link, sagittal-plane model was developed to simulate the balance recovery motions of young and older adults after touchdown of a backward recovery step. The links of the model represented the front and rear feet, rear leg, rear thigh, head-arms-torso, and front thigh-and-leg. The rear foot, representing the foot that took the recovery step, was fixed to the ground and the front foot was constrained to slide along the ground. The rear limb was actively controlled through a set of 10 Hill-type musculotendon actuators that included uniarticular flexor and extensor actuators across the ankle, knee, and hip and biarticular rectus femoris, hamstrings, and gastrocnemius actuators. The front limb of the model moved passively. Angle-dependent passive joint moments were used to enforce the anatomical range of motion at each joint.

Each musculotendon actuator consisted of a contractile element (*CE*), a passive elastic (*PE*) element, and a series elastic (*SE*) element, acting between assumed points of origin and insertion. The *CE* was controlled by a dimensionless neural excitation signal and incorporated time-dependent excitation-activation dynamics, a length-tension relationship, and a force-velocity relationship. The *PE* and *SE* elements were modeled as nonlinear springs. A set of 11 parameter values determined the characteristics of each musculotendon actuator.

Two sets of musculotendon parameter values were derived to represent the respective lower extremity strength characteristics of young and older adults. Parameter values for young adults were derived from published sources. Aging-related losses in muscle strength were simulated by decreasing the *CE* maximum isometric force by 25%, decreasing the ratio of Type II to Type I fibers by 30%, increasing the *CE* deactivation time constant by 20%, and increasing *PE* and *SE* stiffness by 8%.

Feasible regions for balance recovery were determined for the models of young and older adults using repeated optimizations. In each optimization, the backward velocity of the body center of mass (*COM*) and the downward velocity of the hips were initialized to 15% body height/s, representative of the state at step touchdown for successful recoveries from a forward slip during a sit-to-stand [4]. Either the initial COM horizontal position (X_{COM}) or the initial hip height (Z_{HIP}) was chosen. Then, either the maximum or minimum value of the other variable was found from which static balance could be restored. The simulated annealing algorithm was used for this optimization. "Control" variables were the initial joint angles and velocities, initial *CE* activation levels, and the neural excitation signals to the actuators. Initial angles and velocities were constrained to anatomical ranges. Neural excitation signals were parameterized by the start time, magnitude, and stop time of five periods of constant excitation. These periods of constant excitation were connected by linear changes in excitation. The cost function to be minimized was of the form:

$$I_{COST} = kf_0 + \sum g_m + \sum h_n$$

where the f_0 was an initial X_{COM} or Z_{HIP}, the k was +1 to minimize f_0 and -1 to maximize f_0, the g_m were penalty values associated with violating the continuous constraints on the ground reaction forces during the movement, and the h_n were penalty values associated with violating the constraints on the final state. The penalty functions required the six-link model to perform the body-stabilizing motions during the simulation in a manner that young and older adults could in real life and to bring itself to a statically stable, near-stationary state at the end of the simulation. A fourth-order Runge-Kutta method with adaptive step sizes was used to integrate the equations of motion of the model during the forward dynamic simulations. The duration of each simulation was 1 second.

RESULTS AND DISCUSSION

As hypothesized, the feasible region for balance recovery was smaller in area for older adults than for young (Figure 1). The differences between regions were primarily in the range of X_{COM} from which static balance could be restored. The feasible region for older adults did not extend as far anteriorly or posteriorly as that for young adults, with the largest differences observed at lower hip heights. The forward boundary for older adults at a Z_{HIP} of 50% body height (Bh) was 25% foot length (fl) less anterior than that for young adults and the rear boundary for older adults at a Z_{HIP} of 45% Bh was 12% fl less posterior than that for young adults. In contrast, there were negligible differences in the predicted ranges of Z_{HIP} from which young and older adults could restore static balance at X_{COM} common to both regions.

Although a low hip height at step touchdown was previously found to be a primary predictor of backward falls [4], the present results suggest that typical aging-related losses of muscle strength do not notably impair the ability to recover balance at low hip heights. Instead, aging-related losses of muscle strength appear to reduce the range of stepping foot placements for which balance recovery is feasible. Older adults are less able than young to recover from a backward balance loss if they take a very short or very long step.

CONCLUSIONS

Aging-related losses in muscle strength can impair the ability to recover from a backward balance loss if too short or too long a recovery step is taken. However, if a medium-length step is taken, older adults can show the same ability to recover balance as young adults. Therefore, the present results suggest that training of the stepping response might prove to be more effective than strength training in preventing backward falls by older adults.

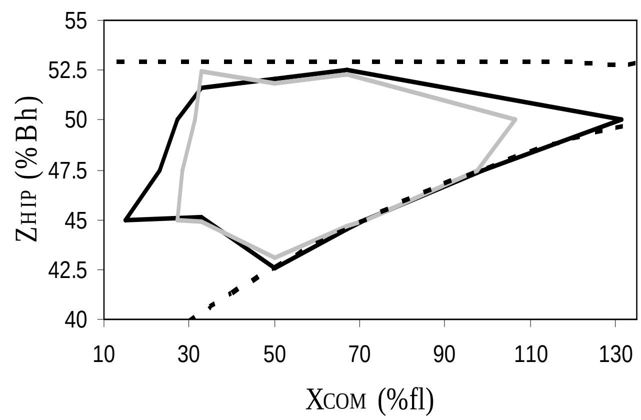

Figure 1: The feasible regions of balance recovery showing the initial states from which young (black line) and older (gray line) adults could restore static balance. The dashed lines represent the anatomical constraints on the initial state. X_{COM} = center of mass position anterior to the rear heel. Z_{HIP} = hip height. fl = foot length. Bh = body height.

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