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Age-Related Differences in Muscle Power During Single-Step Balance Recovery

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The purpose of this study was to investigate age-related differences in muscle power during a surrogate task of trip recovery. Participants included 10 healthy young men (19–23 years old) and 10 healthy older men (65–83). The task involved releasing participants from a forward-leaning posture. After release, participants attempted to recover their balance using a single step of the right foot. Muscle power at the hip, knee, and ankle of the stepping limb were determined from the product of joint angular velocity and joint torque. Muscle powers during balance recovery followed a relatively consistent pattern in both young and older men, and showed effects of both lean and age. Interestingly, the effects of age did not always involve smaller peak power values in the older men as expected from the well-documented loss of muscle power with aging. Older men exhibited smaller peak muscle power at the knee and larger peak muscle power at the ankle and hip compared to young men. The increases in muscle power at the ankle and hip may result from a neuromuscular adaptation aimed at improving balance recovery ability by compensating for the age-related loss of muscle function.

Key Words: biomechanics, accidental falls, aging

An estimated 30–40% of community-dwelling adults over age 65 fall each year (Blake et al., 1988; Horak, Shupert, & Mirka, 1989). Tripping is responsible for up to 53% of these falls (Blake et al., 1988). In an effort to identify intrinsic factors in older adults that directly contribute to falls from trips, researchers have developed surrogate balance recovery tasks that are thought to involve biomechanical requisites similar to those of actual trip recovery. The advantage of these surrogate tasks is a higher level of experimental control for parameters such as initial body position, perturbation magnitude, and stepping characteristics. One of these surrogate tasks involves releasing participants from a static forward lean and allowing a single step for balance recovery. Studies employing this task have identified several performance-related differences between young and older adults that may contribute not only to the reduced ability of older adults to recover their balance after release from a forward lean, but also to the high rate of falls among older adults. Some of these factors include slower stepping velocities (Thelen, Wojcik, Schultz, Ashton-Miller, & Alexander, 1997; Wojcik, Thelen, Schultz, Ashton-Miller, & Alexander, 1999) and peak joint velocities (Madigan & Lloyd, 2005b) in older adults, and differing joint torques during the recovery step (Wojcik, Thelen, Schultz, Ashton-Miller, & Alexander, 2001) and after the recovery step (Madigan & Lloyd, 2005a) between young and older adults.

Muscle power, commonly calculated as the product of joint torque and joint angular velocity, is the net rate at which mechanical energy is generated or absorbed by the musculature at a joint. Any loss of muscle power production with aging

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may contribute to the difficulty that older adults have in quickly arresting the kinetic energy of the body after a postural perturbation (Cao, Schultz, Ashton-Miller, & Alexander, 1998; Hall & Jensen, 2002; Jensen, Brown, & Woollacott, 2001). In fact, muscle power has been proposed as a better indicator of functional status than muscle strength or muscle contraction velocity (Foldvari et al., 2000). Aging is associated with a loss of peak muscle power during constrained lower limb movements (De Vito et al., 1998; Dean, Kuo, & Alexander, 2004; Larsson, Grimby, & Karlsson, 1979; Martin, Farrar, Wagner, & Spirduso, 2000). These losses in muscle power may contribute to the high rate of falls among older adults. However, it is unknown whether similar losses are apparent during unconstrained, functional movements directly involved in fall prevention, such as trip recovery.

This study had two objectives. The first objective was to determine the main role of the hip, knee, and ankle in terms of mechanical energy generation and absorption during a surrogate task of trip recovery. This information could contribute to the development of exercise regimens aimed at fall prevention by determining the types of muscle contractions (concentric or eccentric) involved in trip recovery. The second objective was to investigate age-related differences in muscle power during a surrogate task of trip recovery. This information could help identify factors that contribute to high rate of falls in older adults. The age-related loss of muscle power during constrained tasks may lead to the expectation that peak muscle power at the hip, knee, and ankle would be lower in older adults compared to young adults. However, age-related differences in joint torques during a surrogate task of trip recovery (Madigan & Lloyd, 2005a; Wojcik, Thelen, Schultz, Ashton-Miller, & Alexander, 2001) and gait (DeVita & Hortobagyi, 2000) suggest otherwise. Smaller peak knee torques in older adults were attributed to the age-related loss of muscle strength, but larger peak hip torques in older adults were attributed to a neuromuscular adaptation that was thought to maximize functional capability by offsetting the loss of muscle strength, velocity, and power with aging. Based on these findings, it was hypothesized that older adults would exhibit smaller peak muscle power at the knee and larger peak muscle power at the hip and ankle compared to young adults.

Methods

Participants included ten young men (age: 19–23 years old; height: 1.76 ± 0.07 m; mass: 71.3 ± 14.0 kg) and ten older men (age: 65–83 years old; height: 1.71 ± 0.06 m; mass: 77.0 ± 14.0 kg). Inclusion criteria required all participants to be free of musculoskeletal injury for six months, and for older participants to pass a medical screening. This medical screening was performed by an internist to rule out individuals with any cardiac, respiratory, neurological, otological, or musculoskeletal disorders, or a history of repeated falls. All participants identified themselves as being right-hand and right-foot dominant (foot used to kick a ball). The study was approved by the Institutional Review Board at Virginia Polytechnic Institute and State University, and all participants signed informed consent forms prior to participation.

Forward falls were induced by releasing participants from a forward-leaning posture (Madigan & Lloyd, 2005a, 2005b). Upon release, participants attempted to recover their balance using a single step of the right foot. Each successful recovery was followed by another trial at a larger lean, and each failed recovery was followed by a second trial at the same lean. This process was repeated until participants failed to recover their balance with a single step for two consecutive trials at the same lean.

To start each trial, participants stood with their feet shoulders'-width apart and were leaned forward. Participants were held in this forward-leaning posture using a lean support rope spanning from the back of a belt worn by the participants to a releasable clasp affixed to a stable wooden structure. In this position, participants were asked to equally distribute their weight across both feet while maintaining heel contact with the ground. Equal (within 10%) weight distribution across both feet was verified post hoc using data from a force plate under each foot. Participants were asked to keep their arms folded across their chest throughout each trial. Once the participants were in position at the correct lean, they were verbally reminded to only take a single step with their right foot when released. Participants were released without warning 0–10 s after this verbal reminder. The initial lean corresponded to 12% body weight (BW) in the lean support rope, and lean was increased by 4% BW after each successful recovery.

In the event of an unsuccessful recovery, falls to the ground were prevented using a full-torso harness tethered to a ceiling-mounted support track with a fall-prevention lanyard. Three criteria were used to define a failed recovery: 1) when more than one step was taken with the right foot, 2) when more than 30% BW force was applied to the harness at any point during trip recovery, or 3) when the left foot took a step longer than 30% of the participant's body height. All participants practiced the single-step balance recovery before the start of the experiment.

Body segment positions were sampled at 200 Hz using an Optotrak optoelectronic motion analysis system (NDI, Waterloo, Ontario). Infrared markers were placed on the right side of the body at the fifth metatarsal head, heel, lateral malleolus, lateral femoral epicondyle, greater trochanter, and acromion. Equipment limitations precluded marker data from being collected from the left (nonstepping) lower limb. Thus, the muscle power analysis was performed on only the stepping limb. However, because muscle power is the product of joint angular velocity and joint torque and because joint angular velocity in the nonstepping limb are low, age-related differences in muscle power were thought to most likely occur in the stepping limb. Marker data were low-pass filtered at 7 Hz (fourth-order, zero-phase-shift Butterworth filter). In the initial forward-leaning posture, a force plate was under each foot (AMTI, Watertown, MA). Participants stepped onto a third force plate (90 cm × 90 cm, Bertec Corp., Columbus, OH) during recoveries to provide ground reaction forces during the support phase of balance recovery (i.e., after stepping). Force plate and harness load cell data were sampled at 1,000 Hz, and load cell data were subsequently low-pass filtered at 10 Hz (fourth-order, zero-phase-shift Butterworth filter). To estimate joint torques in the stepping leg, the body was modeled as a 2-D system of four rigid body segments, connected by frictionless pin joints, that included the right foot, right shank, right thigh, and a head/arms/trunk (HAT) segment. The mass and inertial characteristics of the body segments were defined using an anthropometric model (Hanavan, 1964). Sagittal plane joint torques were estimated for the right ankle, knee, and hip using the governing Newton-Euler equations as described by Winter (Winter, 1990). Joint angular velocities in the stepping leg were calculated from the marker data. Muscle powers were calculated as the product of joint angular velocity and joint torque (Winter, 1983).

Age-related differences in the maximum achieved lean ($lean_{max}$) were investigated using an independent t test. Results were significant when $p \leq 0.05$ here and throughout this article. Age-related differences in peak muscle power of the stepping limb were investigated using a repeated-measures ANOVA for each dependent variable. In this analysis, independent variables were age and lean magnitude, potential covariates were participant height and weight, and the correlation structure was autoregressive of order 1. A significant age effect indicated a significant difference in peak muscle power between age groups while holding lean magnitude (and all other independent variables) constant. Age-related differences in peak muscle power used during recovery from $lean_{max}$ were also investigated using an independent t test. Statistical analyses were conducted using SAS v. 9 (SAS Institute Inc., Cary, NC).

Results

Older participants achieved a smaller $lean_{max}$ compared to young participants (20.7 ± 6.2% BW [M ± SD] vs. 33.4 ± 5.4% BW, $p < 0.001$). One older participant was unable to recover from the smallest lean angle and was not included in the analysis. As a result, the final sample size included 10 young participants and 9 older participants.

Muscle powers during balance recovery followed a relatively consistent pattern in all participants (Figure 1). This pattern of muscle powers consisted of nine major phases: three at the hip (H1–H3), four at the knee (K1–K4), and two at the ankle (A1–A2). The area under each positive phase of the power curves was equal to the net energy generation by the joint musculature over the corresponding time interval, and the area under each negative phase of the power curves was equal to the net energy absorption by the joint musculature over the corresponding time interval.

Summing the area under the nine major phases of muscle power revealed the following: 1) the hip musculature was the main source of mechanical energy generation during balance recovery (78.1 J; Figure 1), 2) the knee musculature predominantly absorbed mechanical energy (–39.0 J), and 3) the ankle musculature initially generated mechanical

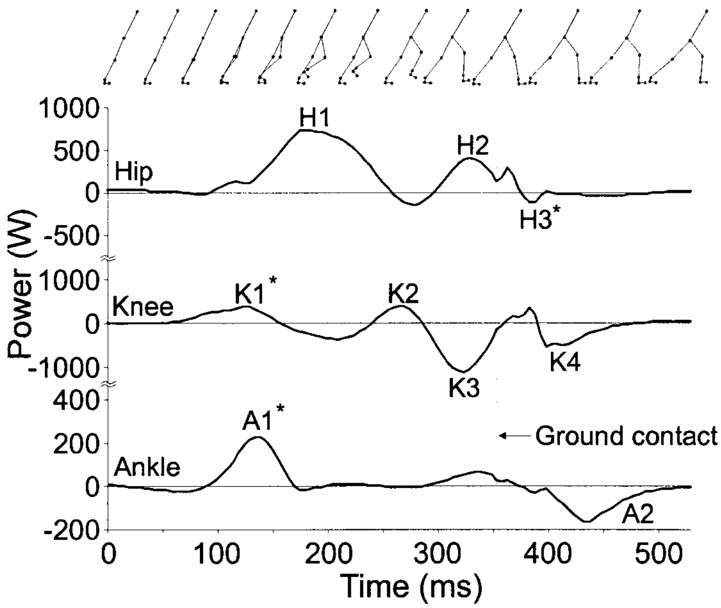

Figure 1 — Representative muscle powers at the hip, knee, and ankle after a young participant was released from a 32% body weight forward lean. Stick figures at the top illustrate body positions over the same time scale as the power curves to aid in interpretation. Positive power values indicate concentric activity of the dominant muscle group, and negative values indicate eccentric activity of the dominant muscle group. Dominant muscle group is determined by the joint torque. * = significant effect of age on peak muscle power when comparing across equivalent leans. Peaks not labeled did not consistently appear across all participants.

energy at the onset of the swing phase of recovery (9.7 J) and absorbed mechanical energy during the support phase of balance recovery (–14.8 J). This dual role led to a relatively small net change in energy at the ankle (–5.1 J) over the entire balance recovery task.

Peak muscle powers showed effects of both lean and age (Figure 2). All but one (K1) showed an increase in absolute value with increasing lean. The A1 peak was higher in older adults ($p = 0.031$), the K1 peak was lower in older adults ($p = 0.015$), and the H3 peak was higher (in absolute value) in older adults ($p = 0.049$). If peak muscle powers during recovery from lean$_{max}$ were compared across age groups, older adults generated smaller peak powers during six of the nine muscle power phases (Table 1).

Table 1 Peak Muscle Power During Recovery From Lean$_{max}$

	Age		
Phase	**Young**	**Older**	
A1	317.2 (106.7)	327.3 (85.5)	
A2	–173.8 (88.9)	–105.7 (16.5)	*
K1	263.5 (97.1)	178.1 (99.9)	*
K2	254.8 (89.4)	128.0 (64.6)	†
K3	–847.4 (293.1)	–422.6 (241.1)	†
K4	–541.5 (148.6)	–423.3 (261.6)	*
H1	533.7 (186.0)	214.8 (129.4)	†
H2	204.0 (167.3)	147.0 (101.4)	
H3	–211.7 (146.6)	–128.0 (80.5)	

Note. Values are means (*SD*). Statistical test was performed on peak muscle power normalized by (weight × height), but values presented here are not normalized to facilitate comparison with other studies.

*$p < 0.05$ for age effects.

†$p < 0.01$ for age effects.

Discussion

Visual inspection of muscle powers (Figure 1) and joint angular velocities and torques (Figure 3) allowed the types of muscle contractions and functional purpose of each contraction to be recognized. Upon release of the lean support rope, phase A1 represented a concentric contraction of the ankle plantar flexors to initiate the balance recovery step. This action also indirectly initiated hip flexion, which preceded significant hip flexor torque, owing to the dynamic coupling of body segments (Zajac & Gordon, 1989). Simultaneously, K1 represented a concentric contraction of the knee flexors, perhaps secondary to gastrocnemius muscle activation in A1 because this muscle also contributes to knee flexion. Flexing the knee not only assists in ground clearance of the stepping limb, but also decreases the moment of inertia of the limb about the hip joint to aid in the forward rotation of the entire lower limb during stepping. Next, H1 represented a concentric contraction of the hip flexors and was consistently

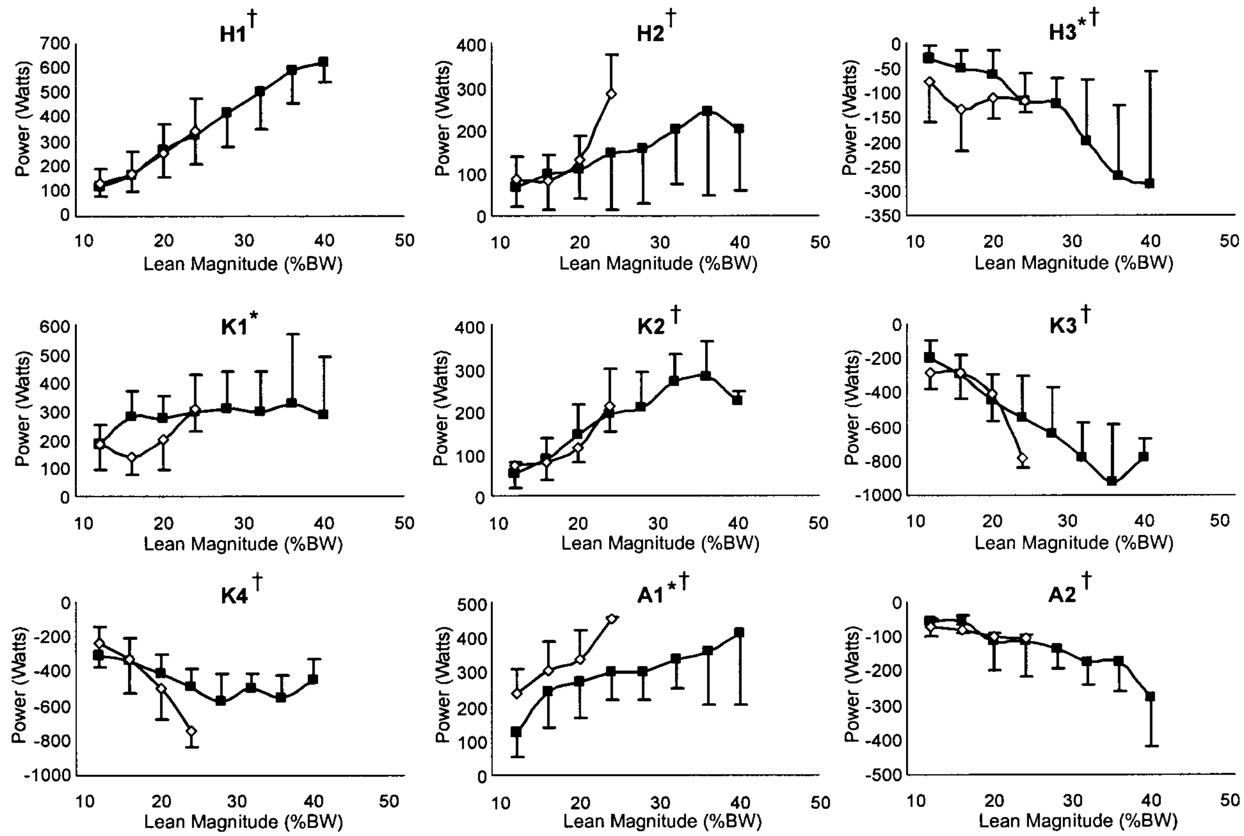

Figure 2 — Peak muscle powers during successful recovery at each lean. Symbols show mean values, and error bars show standard deviations. Solid symbols represent young participants, and hollow symbols represent older participants. Individual plot titles correspond to power phases defined in Figure 1. * = significant effect of age on peak muscle power. † = significant effect of lean on peak muscle power.

the predominant positive power phase during balance recovery. Phase K2 represented a concentric contraction of the knee extensors to extend the knee toward the end of the swing phase of the step, and K3 represented an eccentric contraction of the knee flexors to decelerate knee extension in preparation for ground contact. Phase H2 represented a concentric contraction of the hip extensors to extend the hip for ground contact. During the support phase of balance recovery (i.e., after ground contact), the major power phases occurred in a proximal-to-distal order. Phase H3 represented an eccentric contraction of the hip extensors, K4 represented an eccentric contraction of the knee extensors, and A2 represented an eccentric contraction of the plantar flexors. Most major power phases during the swing phase of balance recovery were positive (indicating generation of mechanical energy by the muscles), and all power phases during the support phase of balance recovery were negative (indicating absorption of mechanical energy by the muscles).

Results from the repeated-measures ANOVA revealed age-related differences in A1, K1, and H3 (Figure 3) while holding lean magnitude constant (unlike comparing peak powers at $lean_{max}$ because the two age groups exhibited different mean $lean_{max}$). As hypothesized, these differences did not always involve smaller peak power values in the older men, who exhibited larger peak powers at A1 and H3 (in absolute value) and smaller peak power at K1. This, in a sense, contradicts reports that peak muscle power decreases with aging. However, the task investigated in the present study was highly functional, involved movements of all three lower extremity joints, and could have conceivably been achieved with different movement strategies. Most other studies of peak muscle power involved only a single joint (Dean, Kuo, & Alexander, 2004; Izquierdo et al., 1999) or considered only the power generated by the entire lower limb (Bassey

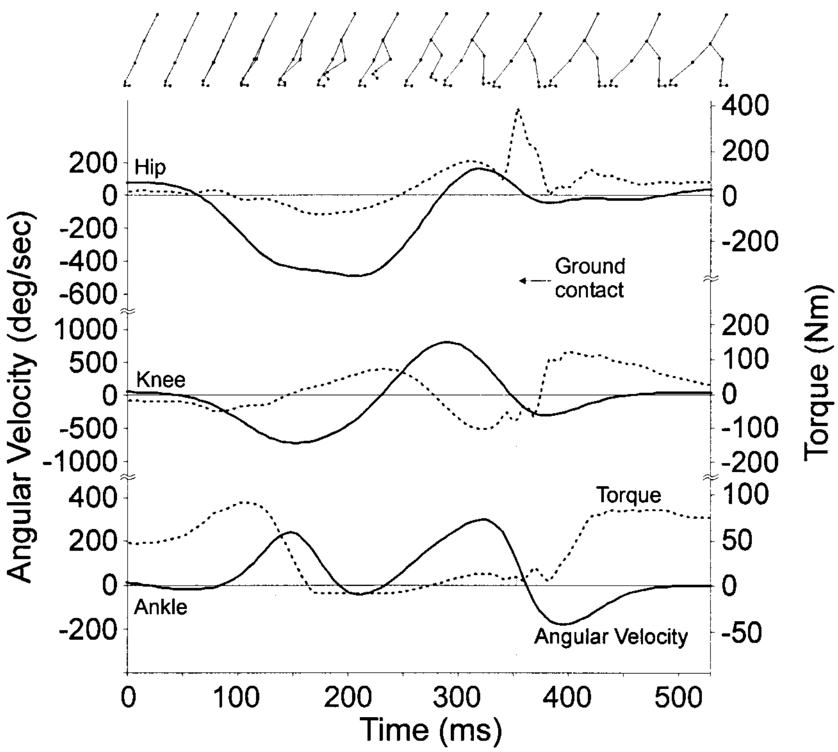

Figure 3 — Representative joint angular velocities and joint torques at the hip, knee, and ankle after a young participant was released from a 32% body weight forward lean (same trial as shown in Figure 1). Stick figures at the top illustrate body positions over the same time scale as the velocity and torque curves to aid in interpretation. Positive values indicate extension and extensor (plantar flexor) torque. Negative values indicate flexion and flexor (dorsiflexor) torque.

& Short, 1990). The increases in peak power at A1 and H3 may have resulted from an age-related neuromuscular adaptation to mitigate the deleterious effects of aging on muscle strength (Porter, Vandervoort, & Lexell, 1995), muscle power (Izquierdo et al., 1999), rate of torque development (Thelen, Schultz, Alexander, & Ashton-Miller, 1996), and movement speed (Dean, Kuo, & Alexander, 2004). This hypothesized adaptation is predicated on the dynamic coupling between body segments, which can lead to movements at a joint not spanned by the muscle group activated (Zajac & Gordon, 1989). For example, at the initiation of stepping (Figure 1), a concentric contraction of the ankle plantar flexors (A1) can contribute to knee flexion and hip flexion by moving the shank superiorly and anteriorly. In addition, after stepping to recover balance (Figure 1) a concentric contraction of the hip extensors (H3) can contribute to knee extension and ankle plantar flexion by moving the thigh posteriorly. This coupling could be beneficial to trip recovery if a muscle group weakened by aging is augmented by other muscle groups that contribute to the same movement.

Other studies have reported age-related differences in joint torques and/or muscle powers during various functional tasks that may also represent a neuromuscular adaptation, including gait (DeVita & Hortobagyi, 2000), compensatory stepping after support surface perturbations (Hall & Jensen, 2002), and the support phase of single-step balance recovery (Madigan & Lloyd, 2005a). Although the nature of these adaptations may vary among tasks, they generally involved larger joint torques or muscle powers at the hip in older adults (DeVita & Hortobagyi, 2000; Hall & Jensen, 2002; Madigan & Lloyd, 2005a). Larger ankle plantar flexor torques in older adults have also been reported during single-step recovery from a forward fall (Madigan & Lloyd, 2005a; Wojcik, Thelen, Schultz, Ashton-Miller, & Alexander, 2001). This increase is particularly interesting because the ankle plantar flexors have been reported to experience a larger age-related loss in strength compared to other muscle groups (Christ et al., 1992; Winegard, Hicks, Sale, & Vandervoort, 1996).

The relatively consistent pattern of muscle power (Figure 1) may be used to improve the beneficial effects of exercise interventions for fall prevention by mimicking the type of contractions during trip recovery. For example, the hip flexors only contract concentrically during trip recovery (H1), suggesting that this is how they should be trained to help prevent falls from trips. In addition, hip extensors contract concentrically (H2) and eccentrically (H3), knee flexors contract concentrically (K1) and eccentrically (K3), knee extensors contract concentrically (K2) and eccentrically (K4), and ankle plantar flexors contract concentrically (A1) and eccentrically (A2). The ankle dorsiflexors do not appear to have a large role in this surrogate task of trip recovery.

The smaller peak muscle powers in older men during recovery from $lean_{max}$ (Table 1) may be causally related to the smaller $lean_{max}$ in the older men. This would imply that an increase in these peak muscle powers would lead to an increase in $lean_{max}$ (and an increase in trip recovery capability). It seems reasonable to hypothesize that an improvement in any of these peak muscle powers would be beneficial, but it would be helpful to therapists to determine which muscle power would provide the most benefit in terms of fall prevention. This determination, however, is confounded by the neuromuscular adaptation reported here and elsewhere. Should exercise regimens train the knee musculature to improve power based on the decrease in muscle power? Or should training focus on the hip and ankle musculature based on the increase at these joints resulting from the neuromuscular adaptation? Answering these questions would likely require a forward dynamic model of trip recovery to isolate the beneficial effect of increasing power at each joint on trip recovery capability.

The surrogate task used here to mimic trip recovery has distinct differences from an actual trip recovery, which may compromise, to some extent, its external validity. These differences include using a static initial posture and limiting recovery to a single step. However, these differences improve internal validity to the extent that using a static initial posture allows the initial conditions to be controlled relatively easily, and limiting recovery to a single step for all participants facilitates comparison between groups that might not otherwise use the same number of steps. The surrogate task also does not require participants to step over an obstacle during recovery, which can affect stepping mechanics (Troy & Grabiner, 2005). Additional studies are needed to ascertain whether age-related differences found using this surrogate task reveal similar differences during actual trip recovery.

In conclusion, differences in peak muscle powers at equivalent leans suggest an age-related neuromuscular adaptation that may attempt to mitigate the effects of age-related muscle function loss on physical performance capabilities. Growing evidence suggests that it should be considered when identifying factors that may contribute to the high rate of falls in older adults.

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References

Bassey, E.J., & Short, A.H. (1990). A new method for measuring power output in a single leg extension: feasibility, reliability and validity. *European Journal of Applied Physiology and Occupational Physiology*, **60**(5), 385-390.

Blake, A.J., Morgan, K., Bendall, M.J., Dallosso, H., Ebrahim, S.B., Arie, T.H., et al. (1988). Falls by elderly people at home: Prevalence and associated factors. *Age and Ageing*, **17**(6), 365-372.

Cao, C., Schultz, A.B., Ashton-Miller, J.A., & Alexander, N.B. (1998). Sudden turns and stops while walking: Kinematic sources of age and gender differences. *Gait & Posture*, **7**(1), 45-52.

Christ, C.B., Boileau, R.A., Slaughter, M.H., Stillman, R.J., Cameron, J.A., & Massey, B.H. (1992). Maximal voluntary isometric force production characteristics of six muscle groups in women aged 25 to 74 years. *American Journal of Human Biology*, **4**, 537-545.

De Vito, G., Bernardi, M., Forte, R., Pulejo, C., Macaluso, A., & Figura, F. (1998). Determinants of maximal instantaneous muscle power in women aged 50-75 years. *European Journal of Applied Physiology and Occupational Physiology*, **78**(1), 59-64.

Dean, J.C., Kuo, A.D., & Alexander, N.B. (2004). Age-related changes in maximal hip strength and movement speed. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, **59**(3), 286-292.

DeVita, P., & Hortobagyi, T. (2000). Age causes a redistribution of joint torques and powers during gait. *Journal of Applied Physiology*, **88**(5), 1804-1811.

Foldvari, M., Clark, M., Laviolette, L.C., Bernstein, M.A., Kaliton, D., Castaneda, C., et al. (2000). Association of muscle power with functional status in community-dwelling elderly women. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, **55**(4), M192-199.

Hall, C.D., & Jensen, J.L. (2002). Age-related differences in lower extremity power after support surface perturbations. *Journal of the American Geriatrics Society*, **50**(11), 1782-1788.

Hanavan, E.P. (1964). *A mathematical model of the human body*. Dayton, OH: Wright Patterson Air Force Base, AMRL.

Horak, F.B., Shupert, C.L., & Mirka, A. (1989). Components of postural dyscontrol in the elderly: A review. *Neurobiology of Aging*, **10**(6), 727-738.

Izquierdo, M., Ibanez, J., Gorostiaga, E., Garrues, M., Zuniga, A., Anton, A., et al. (1999). Maximal strength and power characteristics in isometric and dynamic actions of the upper and lower extremities in middle-aged and older men. *Acta Physiologica Scandinavica*, **167**(1), 57-68.

Jensen, J.L., Brown, L.A., & Woollacott, M.H. (2001). Compensatory stepping: the biomechanics of a preferred response among older adults. *Experimental Aging Research*, **27**(4), 361-376.

Larsson, L., Grimby, G., & Karlsson, J. (1979). Muscle strength and speed of movement in relation to age and muscle morphology. *Journal of Applied Physiology*, **46**(3), 451-456.

Madigan, M.L., & Lloyd, E.M. (2005a). Age-related differences in peak joint torques during the support phase of single-step recovery from a forward fall. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, **60**(7), 910-914.

Madigan, M.L., & Lloyd, E.M. (2005b). Age and stepping limb performance differences during a single-step recovery from a forward fall. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, **60**(4), 481-485.

Martin, J.C., Farrar, R.P., Wagner, B.M., & Spirduso, W.W. (2000). Maximal power across the lifespan. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, **55**(6), M311-316.

Porter, M.M., Vandervoort, A.A., & Lexell, J. (1995). Aging of human muscle: structure, function and adaptability. *Scandinvaian Journal of Medicine & Science in Sports*, **5**(3), 129-142.

Thelen, D.G., Schultz, A.B., Alexander, N.B., & Ashton-Miller, J.A. (1996). Effects of age on rapid ankle torque development. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, **51**(5), M226-232.

Thelen, D.G., Wojcik, L.A., Schultz, A.B., Ashton-Miller, J.A., & Alexander, N.B. (1997). Age differences in using a rapid step to regain balance during a forward fall. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, **52**(1), M8-13.

Troy, K.L., & Grabiner, M.D. (2005). The presence of an obstacle influences the stepping response during induced trips and surrogate tasks. *Experimental Brain Research*, **161**(3), 343-350.

Winegard, K.J., Hicks, A.L., Sale, D.G., & Vandervoort, A.A. (1996). A 12-year follow-up study of ankle muscle function in older adults. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, **51**(3), B202-207.

Winter, D.A. (1983). Moments of force and mechanical power in jogging. *Journal of Biomechanics*, **16**(1), 91-97.

Winter, D.A. (1990). *Biomechanics and motor control of human movement* (2nd ed.). New York: Wiley-Interscience.

Wojcik, L.A., Thelen, D.G., Schultz, A.B., Ashton-Miller, J.A., & Alexander, N.B. (1999). Age and gender differences in single-step recovery from a forward fall. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, **54**(1), M44-50.

Wojcik, L.A., Thelen, D.G., Schultz, A.B., Ashton-Miller, J.A., & Alexander, N.B. (2001). Age and gender differences in peak lower extremity joint torques and ranges of motion used during single-step balance recovery from a forward fall. *Journal of Biomechanics*, **34**(1), 67-73.

Zajac, F.E., & Gordon, M.E. (1989). Determining muscle's force and action in multi-articular movement. *Exercise and Sport Sciences Reviews*, **17**, 187-230.

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