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[Abstract] [Full Text] [PDF]

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K. S Rudolph, L. C Schmitt and M. D Lewek
Physical Therapy, November 1, 2007; 87 (11): 1422-1432.
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P. DeVita, J. Helseth and T. Hortobagyi
J. Exp. Biol., October 1, 2007; 210 (19): 3361-3373.
[Abstract] [Full Text] [PDF]

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J. D. Ortega and C. T. Farley
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Age causes a redistribution of joint torques and powers during gait

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DeVita, Paul, and Tibor Hortobagyi. Age causes a redistribution of joint torques and powers during gait. *J Appl Physiol* 88: 1804–1811, 2000.—At self-selected walking speeds, elderly compared with young adults generate decreased joint torques and powers in the lower extremity. These differences may be actual gait-limiting factors and neuromuscular adaptations with age or simply a consciously selected motor pattern to produce a slower gait. The purpose of the study was to compare joint torques and powers of young and elderly adults walking at the same speed. Twelve elderly and fourteen young adults (ages 69 and 21 yr) walked at 1.48 m/s over a force platform while being videotaped. Hip, knee, and ankle torques and powers were calculated from the reaction force and kinematic data. A support torque was calculated as the sum of the three joint torques. Extensor angular impulse during stance and positive work at each joint were derived from the torques and powers. Step length was 4% shorter and cadence was 4% higher in elderly adults (both $P < 0.05$) compared with young adults. Support angular impulse was nearly identical between groups, but elderly adults had 58% greater angular impulse and 279% more work at the hip, 50% less angular impulse and 39% less work at the knee, and 23% less angular impulse and 29% less work at the ankle compared with young adults (*t*-test, all $P < 0.05$). Age caused a redistribution of joint torques and powers, with the elderly using their hip extensors more and their knee extensors and ankle plantar flexors less than young adults when walking at the same speed. Along with a reduction in motor and sensory functions, the natural history of aging causes a shift in the locus of function in motor performance.

aging; locomotion; neuromuscular function; work; walking

GAIT KINEMATICS ARE DIFFERENT in healthy, elderly adults compared with healthy, young adults when walking at freely chosen speeds. Elderly adults select a slower gait velocity with a shorter step length, shorter relative swing phase time, and less range of motion (ROM) at the hip, knee, and ankle joints compared with young adults (5, 10, 12, 18, 22–24, 33, 35, 36, 48). The underlying joint kinetics associated with these kinematic reductions with age appear to be primarily lower plantar flexor torque and power at the ankle joint (22, 24, 48) and possibly reduced torque at the knee and hip joints (5, 24). It is not clear, however, whether the reduced joint torques were gait-limiting factors and actual neuromuscular adaptations in elderly adults or

whether elderly adults simply chose to walk slower and, therefore, consciously reduced their ankle plantar flexor and knee and hip extensor functions.

Many gait characteristics are dependent on gait velocity (1, 5, 25, 32, 41, 44, 45). Kirtley et al. (25), for example, showed the strong direct relationship between velocity and stride length ($r = 0.95$) and the inverse relationship between velocity and stance phase duration ($r = -0.71$). Joint torque and joint power at the ankle are also directly related to walking velocity (2, 44, 47). Indeed, a group of young adults walking at fast and slow velocities showed many of the same differences as observed between young and elderly adults walking at freely chosen speeds (44, 45). It should be noted that both Larish et al. (27) and Elble et al. (10) have already indicated that gait differences attributed to age may actually be due to differences in gait velocity. Therefore, a comparison of young and elderly adults walking at different speeds may not provide a valid analysis because the differences between young and elderly adults in gait would be due to age as well as velocity.

In the few gait comparisons made at identical velocities, stride length was reduced in elderly compared with young adults but only at faster velocities (5, 27). Concomitant with the shorter stride length was a higher stride frequency in the elderly subjects. Winter (44, 45) has shown that increased cadence is directly associated with increased joint torque and power. In addition, aging causes larger deficits in torque production (3, 43) and mitochondrial enzyme activity (21) in the plantar flexors compared with other muscle groups. These observations lead to the hypothesis that, in compensation for the nonuniform reductions in muscle function at the three lower extremity joints, elderly adults would actually reveal increases in joint torques at some joints and decreases at other joints to walk at the same speed as young adults. The purpose of the present study was to compare lower extremity joint torques and powers between healthy young and elderly adults walking at the same speed.

METHODS

Subjects. Twelve healthy elderly and fourteen healthy young adults volunteered. Elderly adults were, on average, 47.4 yr older than young adults (elderly: 69.0 ± 6.5 yr; young: 21.6 ± 2.7 yr). Eight elderly and five young adults were women. Mass and height were similar between groups. Elderly adults averaged 72.4 ± 13.7 kg and 1.61 ± 0.09 m and young adults averaged 69.8 ± 11.1 kg and 1.66 ± 0.07 m in mass and height, respectively. All subjects were healthy, and

elderly subjects provided a physician's approval to participate in the study and completed a medical questionnaire to determine their eligibility. Subjects were excluded if they had more than two risk factors for coronary artery disease, osteoporosis, osteoarthritis, orthopedic or neurological conditions (i.e., stroke), a body mass-to-body height ratio >28, high blood pressure (140/90 mmHg), a heart condition, medication that causes dizziness, or a history of falls or smoking. The testing protocol was readily performed by all subjects, and no subjects withdrew from the study because of an inability to perform the walking test. All subjects provided written informed consent before testing.

Experimental setup. A force platform (model LG6-4-1, AMTI, Newton, MA) located in the center of a 20-m, level walkway was used to measure vertical and anteroposterior ground reaction forces and the mediolateral moment at 1,000 Hz. A video camera (model SSC M350, SONY) and videocassette recorder (model HRS5100U, JVC) were used to videotape the subjects in the sagittal plane at 60 Hz as they walked over the force platform. An infrared system (model 63520, Lafayette) was used to measure the time required for each subject to walk a 6-m interval over the platform. The field of view for the video image was ~2.5 m wide and 2.0 m high, which maximized the image size.

The vertical force from the platform was calibrated with known weights ranging from 0 to 2,100 N. The voltage output was highly linear throughout the tested range, and the coefficient of determination between force and voltage was $R^2 = 0.999$.

Testing protocol. Subjects wore black spandex bike shorts, a tight-fitting T-shirt, and athletic shoes. Standing height and mass were measured along with circumferences of the proximal thigh, knee, ankle, and metatarsals. Reflective markers were placed on the subjects' right side on the lateral border of their fifth metatarsal head, the lateral heel of the shoe, the lateral malleolus, the lateral femoral condyle, the greater trochanter, and the shoulder. Subjects walked through the experimental area for several minutes until they were relaxed and comfortable. A starting point was selected so that the correct foot would contact the force platform in a normal stride at the required pace. The criterion walking speed was 1.5 ± 0.1 m/s for all subjects. The calculated, mean speeds were nearly identical for the two groups and were 1.484 ± 0.110 and 1.481 ± 0.089 m/s for the elderly and young subjects, respectively. The speed for the elderly group was faster than values generally reported for this population (e.g., Refs. 16, 20, 22, 31). Elderly subjects performed at a freely chosen walking speed before testing at the nominal speed. These data were also analyzed but are not presented here. Their average, freely chosen speed was 1.43 m/s, which was only 3.5% slower than the test speed. All elderly adults were comfortable at both freely chosen and test speeds, and all were able to walk normally at the test speed. The relatively fast, freely chosen speed used by the elderly subjects indicated that these individuals were relatively fit and had good neuromuscular function. Trials were discarded if the subject's velocity was outside the acceptable range, the foot was not completely on the force platform, or the subject made visually obvious stride alterations to contact the force platform. A minimum of five successful trials were collected for each subject. No subjects reported fatigue or required rest during the test session.

Data reduction. Cartesian coordinates of the reflective markers were derived from the video records during the swing phase before contact with the force platform and the subsequent stance phase on the platform (7) by using the Peak5 system (Peak Performance Technologies, Englewood, CO). The accuracy and repeatability of the digitizing process were evaluated by redigitizing 10 frames of videotape 10 times. The mean position of each marker within the frame was used as the true position, and the differences between this position and the repeated positions were calculated and averaged across markers and frames. The mean errors in the horizontal and vertical directions were 0.005 and 0.002 m, respectively, and were considered small. High-frequency error was removed from the digitized coordinates with a low-pass digital filter that automatically selected the cutoff frequency on the basis of Winter's method (46). The mean cutoff frequency was ~5 Hz, and the range of frequencies was 2.9–8.5 Hz. Linear position data were interpolated to 200 Hz by using a cubic spline routine without further smoothing. The interpolation was done to set the frequency of the kinematic data to a multiplicative factor of the kinetic data so that the data sets could be synchronized. Linear velocity and acceleration were calculated for each point during the walking cycle. Joint angular position and velocity were calculated at the hip, knee, and ankle, and the joint-position curves were evaluated with a set of seven variables describing stance-phase kinematics (see Table 2).

The lower extremity was modeled as a rigid, linked segment system. Magnitude and location of the segmental masses and mass centers in the lower extremity along with their moments of inertia were estimated from the position data by using a mathematical model (19), segmental masses reported by Dempster (6), and the individual subject's anthropometric data. Center of pressure in the anteroposterior direction was calculated from the ground reaction forces and the mediolateral moment on the platform. It was then converted from a force platform-based system to the kinematic reference frame on the basis of the digitized location of the force platform. The accuracy of the prediction of the center of pressure on the force platform was tested by applying a force of 550 N at seven known locations on the platform. The average error in the anteroposterior direction was 0.003 m. Error of this magnitude will produce ~4% error in the prediction of lower extremity joint torques (30).

Inverse dynamics using linear and angular Newtonian equations of motion were used to calculate the joint reaction forces and torques at each joint throughout the gait cycle. A support torque was calculated as the sum of the three joint torques (48). The support torque quantifies the total torque output from the extremity, and it provides a quantitative assessment of the support and propulsive effort of the musculature in the entire limb. It also provides a total torque value against which each of the individual joint torques can be compared with to assess the relative contribution of the individual joints to the performance. Angular impulse by the extensor or plantar flexor muscles was calculated throughout stance from the support and ankle torques and during the initial half of stance from the knee and hip torques. Flexor angular impulse at the hip in late stance and the maximum plantar flexor torque at the ankle were also assessed.

Joint powers were calculated as the product of the joint moments and angular velocities. Work at each joint was calculated as the area under portions of the power curves during extensor torque generation. Positive work indicates that the observed joint torque generated mechanical energy and contributed to propelling the individual. Negative work indicates that the observed joint torque absorbed mechanical energy. Positive work was calculated at the hip during the first half of stance, at the knee during midstance, and at the ankle during late stance. Negative work was calculated at the knee joint during early stance.

Statistical analysis. Each variable was averaged over the five trials for each subject. These mean values were considered to be the best estimate of the particular variable for each subject, and they were entered into the statistical tests. All variables were analyzed with a *t*-test comparing young and elderly subjects. An α level of <0.05 was used to indicate statistical significance.

Preliminary statistical analyses for gender (*t*-test, α < 0.05) were conducted to verify that the observed age differences were not confounded by gender. Five of the twenty-one variables, including two torque and two power variables, were significantly different between genders. The men, however, were 18% more massive than the women, and, therefore, the kinetic data were subsequently normalized for mass. The analysis on these data showed no gender differences. We concluded that gender had virtually no effect on the present results because only 1 of 21 variables, knee position at heel contact, was significantly different when mass was controlled.

RESULTS

Although the two groups walked at nearly identical speeds, stride characteristics were significantly different between the groups (Table 1). Elderly adults spent less relative time in swing and walked with a shorter stride length and a higher stride frequency compared with young adults. The magnitude of the differences in stride characteristics between groups was not large, averaging ~5%, but the consistency in performance between subjects in each group was high, as shown by the low SD values and the mean coefficient of variation for these four variables, which was 4%.

The joint angular kinematics were significantly different between the age groups (Table 2, Fig. 1). Elderly adults had a larger ROM and were flexed more at the hip joint throughout the gait cycle compared with young adults. This second adaptation was due a greater forward lean in the trunk. In contrast, elderly adults contacted the force platform in a more flexed knee position but flexed less at this joint compared with young adults. Ankle kinematics were generally minimized in the elderly compared with the young subjects because they had less ROM and less plantar flexion. In short, they maintained a more neutral ankle position through the stance phase.

The joint torques were significantly different between groups at each joint, but they combined to produce nearly identical support torque curves (Table 3, Fig. 2). Elderly adults increased the extensor output from their hip musculature by exerting a net extensor torque longer into the stance phase compared with young adults. Along with this adaptation, elderly adults also used their hip flexors 37% less than did young adults. The knee joint torque curves were similar

Table 1. *Stride characteristics*

Variable	Elderly Adults	Young Adults	P Value
Swing time, %	35.8 ± 1.5	38.8 ± 2.8	0.002
Stance time, %	64.2 ± 1.5	61.2 ± 2.8	0.002
Step length, m	0.72 ± 0.04	0.75 ± 0.04	0.018
Cadence, steps/min	124 ± 5	119 ± 6	0.030

Values are means ± SD.

Table 2. *Joint position variables*

Variable	Elderly Adults	Young Adults	P Value
Hip			
Avg stance pos	−11.9 ± 4.7	−3.9 ± 4.1	0.000
ROM in stance	39.5 ± 3.6	36.2 ± 3.9	0.017
Knee			
Heel contact	−8.3 ± 5.9	−4.4 ± 5.6	0.048
Flex after contact	−21.3 ± 5.5	−26.3 ± 4.7	0.009
Ankle			
Avg stance pos	−1.9 ± 3.8	−4.2 ± 3.4	0.045
ROM in stance	26.7 ± 2.4	29.4 ± 3.9	0.023
Max plantar flex	13.8 ± 4.0	17.4 ± 5.4	0.035

Values are means ± SD in degrees; negative values are flexed or dorsiflexed positions. ROM, range of motion; Avg stance pos, average angular position during stance phase; Flex after contact, maximum knee flexion in early stance; Max plantar flex, maximum ankle plantar flexion at end of stance.

between age groups through most of the gait cycle, but elderly adults walked with less extensor torque during the crucial extensor period after heel strike. Plantar flexor ankle function was also reduced in the elderly subjects during stance by ~25% compared with the young adults. The differences in torque caused by age seem to be reliable because the average difference in joint torque variables was 36% and the error in torque predictions due to center of pressure error was only 4%. In summary, elderly adults walked at the same speed as young adults by producing the same support torque, but they produced this total output with a larger contribution from the hip extensors and smaller contributions from knee extensors and ankle plantar flexors.

Joint power results showed significant differences between groups at all joints during the stance phase (Table 4, Fig. 3). The positive power at the hip during the first half of stance was larger in the elderly adults, whereas the negative and positive power phases at the knee and ankle were lower compared with young adults. Elderly adults produced more work with their hip extensors and less with their knee extensors and ankle plantar flexors compared with young adults.

DISCUSSION

The operative concept in the biomechanics and physiology research on aging is that elderly adults have less than young adults. Healthy, elderly adults have less muscle mass, strength, power, and rate of force production compared with healthy, young adults (3, 15, 16, 26, 28, 29, 31, 38, 39, 42). These declines are associated with a slower, freely chosen speed, a shorter step length, and a shorter swing phase during walking (10, 12, 18, 22, 24, 33, 35, 36, 48). Elderly adults have impaired balance and proprioceptive abilities, have less joint range of motion, and are less able to exert steady or constant muscle forces (11, 14, 17, 34, 37). Nearly all aging-related studies are designed to identify and quantify deficits in neuromuscular function, although it is not always clear whether aging per se, or other factors, such as inactivity (20) or volitional choice, causes these deficits.
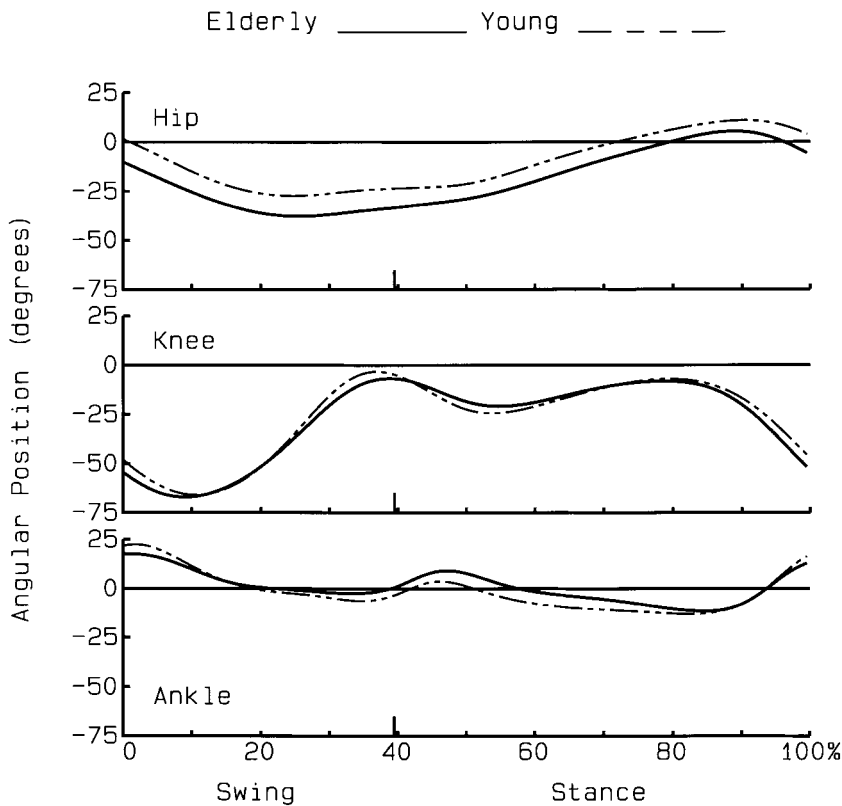
Elderly _____ Young _ _ _ _

Fig. 1. Mean joint position curves. Zero is an erect, standing position (foot at 90° to leg), and negative values indicate flexion or ankle dorsiflexion. Elderly adults ($n = 12$) were flexed more at the hip throughout gait cycle and had more hip range of motion than young adults. Young adults ($n = 14$) contacted the force platform with more extension at the knee joint, and they flexed more during early stance. Ankle kinematics were generally minimized in elderly subjects compared with younger subjects because they maintained a more neutral ankle position through stance phase and their final position was less plantar flexed.

We propose an alternative neuromuscular adaptation with aging: aging causes a shift in the locus of function in motor performance. Along with a reduction in motor and sensory functions, the natural history of aging includes a reformation of the motor pattern used to perform a movement skill compared with the pattern used by young adults. The altered motor pattern is manifested as a redistribution of joint torques and powers, which alters the relative contributions from the various muscle groups to the total performance. The functional outcome of such a process is that elderly adults might exert greater torques and powers with one or more muscle groups when performing a movement skill compared with young adults performing the same skill. A critical concept in the present study is that the movement skills performed by young and elderly adults were identical. Thus the total effort required to perform the movements would be identical.

The redistribution of joint torques in the elderly adults changed the relative contribution of the individual muscle groups to the total output of the limb, the support torque, particularly during the first half of stance. The angular impulses from the support torques during this period were 19.2 and 20.3 Nms in elderly and young adults, respectively. The relative contributions of the extensor torques at the hip and knee and the plantar flexor torque at the ankle to these impulses were 74, 13, and 12% in elderly adults and 37, 35, and 28% in young adults. The joint power and work results showed a similar shift with age. The total positive work performed at the three joints was 44.8 and 43.9 J in elderly and young adults. Hip, knee, and ankle joint muscles produced 44, 5, and 51% of this work in elderly adults and 16, 11, and 73% of the work in young adults, respectively. Elderly adults used their proximal hip extensor musculature much more than their other muscles for support, and they had similar amounts of work done at the hip and ankle for propulsion. In contrast, young adults had similar torque contributions from their hip and knee extensors and only slightly less contribution from their ankle plantar flexors, with the majority of work done by the ankle plantar flexors. We identify this neuromuscular adaptation toward the proximal hip musculature in the elderly adults as a shift in the locus of function.

A similar, although not identical, response in hip joint torque was reported by Kerrigan et al. (24).

Table 3. *Joint torque variables*

Variable	Elderly Adults	Young Adults	P Value
Support			
Ext ang imp, Nms	33.9 ± 13.7	34.3 ± 11.2	0.469
Hip			
Ext ang imp, Nms	17.1 ± 9.3	10.8 ± 4.1	0.016
Flx ang imp, Nms	−8.4 ± 7.2	−13.4 ± 6.9	0.046
Knee			
Ext ang imp, Nms	5.1 ± 3.3	8.4 ± 4.6	0.029
Ankle			
Plnt flx ang imp, Nms	24.6 ± 7.5	32.0 ± 9.8	0.025
Max plnt flx torque, Nm	102 ± 29	136 ± 27	0.003

Values are means ± SD. All variables applied to stance phase. Ext, flx, plnt flx ang imp: extensor, flexor, plantar flexor angular impulse, respectively. Impulses are areas under torque curves; support, ~40 to 95% of cycle; hip, ~40 to 65% of cycle; knee, ~45 to 65% of cycle; ankle, ~50 to 100% of cycle. max, Maximum.

Elderly ——————— Young — — — —

Fig. 2. Mean joint torque curves. Positive values indicate extensor or plantar flexor torque. Joint torques were significantly different between groups at each joint, but they combined to produce nearly identical support torque curves. Elderly adults ($n =$ 12) increased extensor output from their hip musculature such that they exerted a net extensor torque longer into stance phase compared with young adults ($n = 14$). Extensor angular impulse at the hip was larger in elderly subjects. Elderly adults also used their hip flexors less than young adults. Elderly adults walked with a low knee extensor torque in early stance and exerted less knee extensor angular impulse. Plantar flexor ankle function was also reduced in elderly subjects.
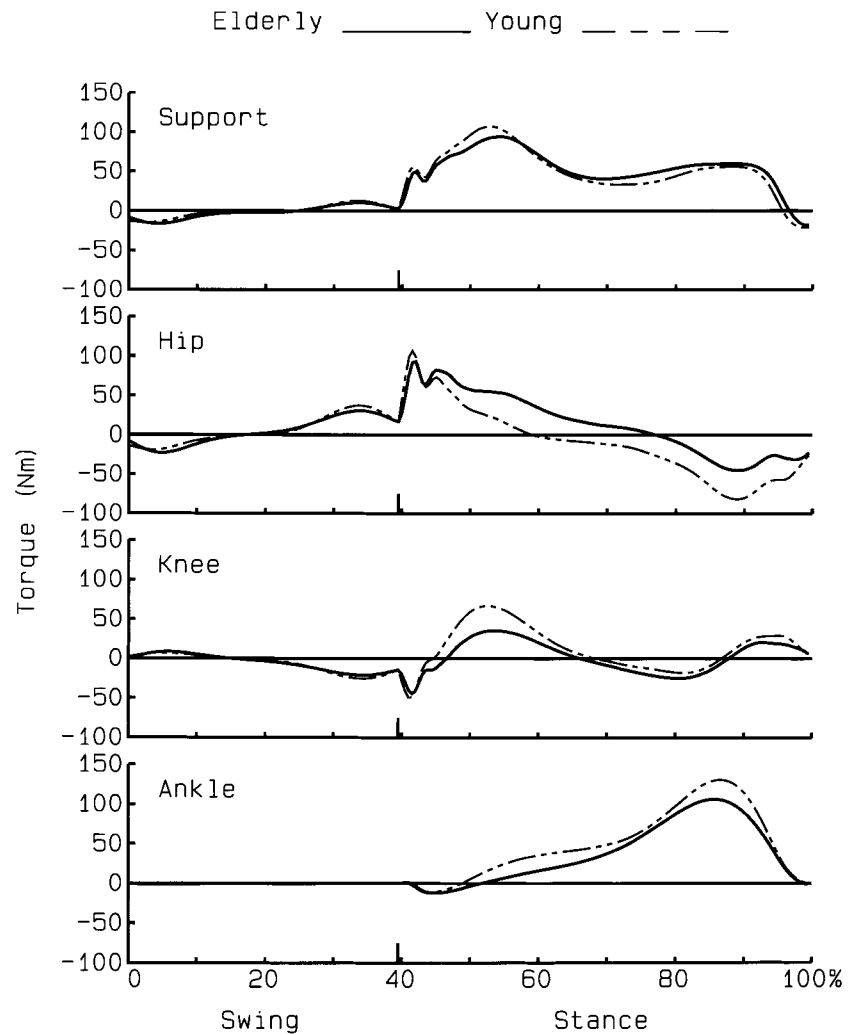
Elderly adults walked at comfortable and fast speeds, and the faster speed was produced by increasing the extensor torques produced by the hip and knee muscles. The hip joint torque, but not joint power, was greater in the elderly adults than in young adults, who walked at a slower velocity. Elderly and young subjects were not actually performing the same motor task because of the difference in speeds. The results of Kerrigan and colleagues do not, therefore, show a redistribution of joint torques, but they do support the present results that identify increased output from the hip extensor muscles in elderly compared with young adults.

Table 4. *Work variables from joint power curves*

Variable	Elderly Adults	Young Adults	P Value
Hip			
Positive (~40–65%)	19.5 ± 13.5	7.0 ± 3.9	0.002
Knee			
Negative (~45–55%)	−2.8 ± 2.3	−5.6 ± 2.7	0.006
Positive (~55–65%)	2.3 ± 1.8	4.7 ± 3.2	0.019
Ankle			
Positive (~85–100%)	23.0 ± 4.5	32.2 ± 6.5	0.025

Values are means ± SD in J. %Ranges apply to entire gait cycle.

The reduced torque and power at the ankle joint in the elderly subjects was in agreement with previous analyses on this population. Although the elderly and young adults in the present study walked at the same speed, elderly subjects used ~27% less plantar flexor angular impulse and work at the ankle. Other elderly subjects had between 17 and 35% less plantar flexor torque and power while walking at slower speeds than did young adults (23, 24, 48), and they still had 12% less plantar flexor work while walking at a 13% faster speed (24). Ankle plantar flexors have larger biomechanical and physiological deficits compared with other muscle groups in elderly compared with young adults. Christ et al. (3) reported the rate of decline in maximal voluntary isometric force between the ages of 25 and 74 yr was largest in the plantar flexors compared with five other muscle groups, and Winegard et al. (43) showed a fourfold larger decline in plantar flexor strength compared with dorsiflexor strength in a longitudinal study of 73- to 97-yr-old people. Houmard et al. (21) reported a 25% decline in citrate synthase activity in the gastrocnemius of subjects between the ages of 18 and 80 yr, whereas no decline was observed in the vastus lateralis. In addition, Coggan et al. (4) reported 13–31%
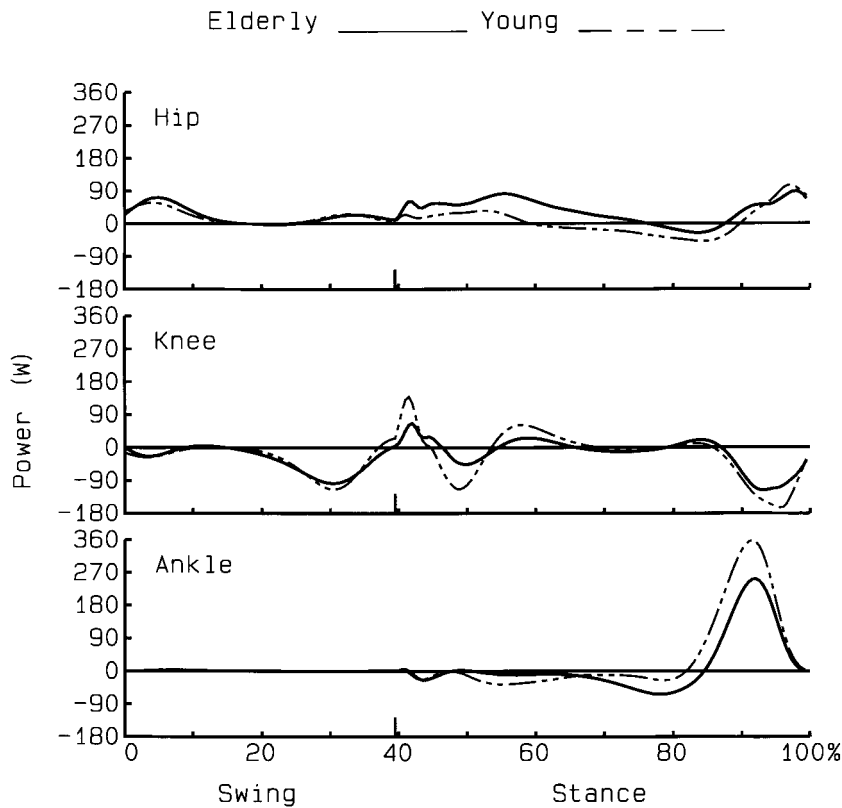
Elderly _____ Young __ _ _ __

Fig. 3. Mean joint power curves. Positive work indicates that observed joint torque generated mechanical energy and contributed to propelling the individual. Negative work indicates that observed joint torque absorbed mechanical energy. Positive power at the hip during the first half of stance was larger in elderly adults ($n = 12$), and it produced more work. Knee and ankle joint powers were lower in elderly adults. Work produced during negative and positive knee power bursts in early to midstance was ~50 J less than corresponding work produced by young adults ($n = 14$). Work from ankle plantar flexors during late stance was 29% lower in elderly adults.

reductions in type IIa and IIb fiber areas and 25% lower mitochondrial enzyme activity in gastrocnemius of elderly adults compared with young adults.

Although it is possible, we do not think the elderly subjects had lower plantar flexor function than the young subjects because they had reached their physiological limits in torque and power production. Young adults had a maximum plantar flexor torque of ~325 Nm while sprinting (40), which indicates the present young adults used ~40% of their maximum plantar flexor torque while walking. Although functional decline with age is larger in the plantar flexors than in other muscle groups, elderly retain ~60% of their maximum isometric and concentric torque at age 80 yr (3, 13). The elderly adults in the present study, therefore, may have had some remaining torque and power capabilities in their plantar flexors, which they did not use. In addition, elderly adults of the present study were only 69 yr old, walked very comfortably at the test speed, and clearly could have walked faster with presumably greater ankle plantar flexor torque and power. We conclude that reduced physiological and biomechanical properties of human plantar flexors with age cause elderly adults to use less torque and power from these muscles compared with young adults while walking, even though elderly adults still have the capability to generate the amounts of torque and power used by young adults in walking. We also acknowledge the limitation of the single walking speed used in this study and the possibility that the ankle deficit and torque redistribution may not occur at slower speeds. Further investigations may be required to fully understand the

nature of the change in neuromuscular locus of function with age.

Redistribution of joint torques and powers is not an entirely novel observation. Joint torque and power redistribution has been observed during walking and running in individuals with anterior cruciate ligament (ACL) injury and reconstruction surgery (8, 9). The total output of the lower extremity during the stance phases of walking and running, as measured by the support torque, was identical between healthy and rehabilitated ACL-injured individuals, but the ACL subjects had greater torque and power at the hip and ankle and less torque and power at the knee compared with the healthy subjects. The present results from healthy but aged individuals and those from injured individuals identify the flexibility inherent in the human neuromuscular system and our ability to shift function from weakened or disabled muscle groups to those with better neuromuscular function.

The effect of age on the general stride characteristics and the angular kinematics at the knee and ankle were in general agreement with previous reports. Step length was 4% shorter in elderly compared with young adults in the present study and 10% shorter, on average, than in other studies (23, 24, 27, 48). Relative stance time was 5% longer in the present elderly group and 6% longer in other elderly adults (12, 33, 48). The differences in present relative times were actually due to a shorter absolute swing time (elderly: 0.35 s, young: 0.39 s), but there was virtually no difference in absolute stance time (elderly: 0.62 s, young: 0.62 s). The reduced step length in elderly adults was related to a shorter

time used for swinging the limb forward. The shift toward a longer relative stance time with age probably increased the subjects' stability by increasing the amount of time with both feet in contact with the floor.

These basic differences in stride characteristics seem to occur when elderly adults walk at either slower or identical speeds as young adults. The elderly subjects in the present study had a 4% higher cadence than did young subjects, and they matched the gait speed of the young adults by increasing their stride frequency to compensate for the shorter step length. Elderly subjects had higher cadences while walking at the same but relatively fast velocity (5) and at a faster velocity (24) as young adults in previous investigations. At the slower, yet freely chosen speeds, however, cadence was unchanged with age (10, 24, 33, 36, 48). It seems clear that, whereas elderly and young adults have identical cadences at freely chosen speeds, elderly adults walk at relatively fast speeds with a larger emphasis on frequency (i.e., higher cadence) and a lower emphasis on amplitude (i.e., shorter step length) compared with the cycle used by the younger sample.

The increased hip flexion and reduced knee flexion, ankle ROM, and ankle plantar flexion in late stance in the elderly compared with young subjects have been observed in other studies (10, 18, 22, 24, 33, 35, 36). Increased hip flexion may be a postural adjustment to further stretch the hip extensor muscles and enable them to produce larger amounts of torque and power during the stance phase compared with the young adults. Elderly subjects in the present study had a greater ROM at the hip compared with the young adults and therefore compensated for reduced ankle plantar flexion by increasing hip extension, and this result is in contrast to the smaller ROM reported for elderly adults while walking at the same speed as young adults (5). The relatively fast freely chosen speed in the present elderly subjects (see METHODS) and the ease they showed while walking at 1.48 m/s suggest that they may be healthier or fitter than those measured previously and, therefore, have the ability to increase hip ROM when necessary. The reduced ROM at the ankle used by the elderly adults may have contributed to the reduced torque and power produced by the ankle plantar flexors. In general, while walking at identical speeds, elderly adults had more forward trunk lean, creating more hip flexion, and they used a larger ROM at the hip compared with young adults. However, elderly adults also had a more erect lower extremity, with less knee flexion in early stance and less ankle plantar flexion in late stance.

In summary, motor performance in elderly and young adults was compared while these groups walked at the same speed. The total motor output, represented by the support torque, was identical between groups, but the individual contributions of the hip, knee, and ankle joint muscle groups varied between ages. Elderly adults showed a redistribution of joint torques and powers, which emphasized the hip extensors and deemphasized the knee extensors and ankle plantar flexors compared with the young adults. In fact, elderly subjects pro-

duced a larger extensor torque and more work at the hip during the first half of stance than the young subjects. This redistribution of joint torques and powers was interpreted as a change in the locus of function in motor performance, and it represents an alteration in the motor pattern used to perform the task. The present results also support the concept that the biomechanical and physiological consequences of aging are not solely a reduction in motor abilities but are a qualitative change in the underlying neuromuscular components of a motor performance.

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REFERENCES

1. **Andriacchi TP, Ogle JA, and Galante JO.** Walking speed as a basis for normal and abnormal gait measurements. *J Biomech* 16: 261–268, 1977.
2. **Chen IH, Kuo KN, and Andriacchi TP.** The influence of walking speed on mechanical joint power during gait. *Gait Post* 6: 171–176, 1997.
3. **Christ CB, Boileau RA, Slaughter MH, Stillman RJ, Cameron JA, and Massey BH.** Maximal voluntary isometric force production characteristics of six muscle groups in women aged 25 to 74 years. *Am J Hum Biol* 4: 537–545, 1992.
4. **Coggan AR, Spina RJ, King DS, Rogers MA, Brown M, Nemeth PM, and Holloszy JO.** Histochemical and enzymatic comparison of the gastrocnemius muscle of young and elderly men and women. *J Gerontol* 47: B71–B76, 1992.
5. **Crowinshield RD, Brand RA, and Johnston RC.** The effects of walking velocity and age on hip kinematics and kinetics. *Clin Orthop* 132: 140–144, 1978.
6. **Dempster W.** *Space Requirements of the Seated Operator.* Dayton, OH: Wright Patterson Air Force Base, WADC, 1955. (Wright Air Development Center Technical Rep. 55-159)
7. **DeVita P.** The selection of a standard convention for analyzing gait data based on the analysis of relevant biomechanical factors. *J Biomech* 27: 501–508, 1994.
8. **DeVita P, Blankenship-Hunter P, and Skelly WA.** Effects of a functional knee brace on the biomechanics of running. *Med Sci Sports Exerc* 24: 797–806, 1992.
9. **DeVita P, Hortobagyi T, and Barrier J.** Gait biomechanics are not normal after anterior cruciate ligament reconstruction and accelerated rehabilitation. *Med Sci Sports Exerc* 30: 1481–1488, 1998.
10. **Elble RJ, Thomas SS, Higgins C, and Colliver J.** Stride-dependent changes in gait of older people. *J Neurol* 238: 1–5, 1991.
11. **Enoka RM.** Neural strategies in the control of muscle force. *Muscle Nerve Suppl* 5: S66–S69, 1997.
12. **Finley FR, Cody KA, and Finizie RV.** Locomotion patterns in elderly women. *Arch Phys Med Rehabil* 50: 140–146, 1969.
13. **Gajdosik RL, Vander Linden DW, and Williams AK.** Concentric isokinetic torque characteristics of the calf muscles of active women aged 20 to 84 years. *J Orthop Sports Phys Ther* 29: 181–190, 1999.
14. **Galganski ME, Fuglevand AJ, and Enoka RM.** Reduced control of motor output in a human hand muscle of elderly subjects during submaximal contractions. *J Neurophysiol* 69: 2108–2115, 1993.
15. **Gallagher D, Visser M, De Meersman RE, Sepulveda D, Baumgartner RN, Pierson RN, Harris T, and Heymsfield**

SB. Appendicular skeletal muscle mass: effects of age, gender, and ethnicity. *J Appl Physiol* 83: 229–239, 1997.

16. **Grimby G.** Muscle performance and structure in the elderly as studied cross-sectionally and longitudinally. *J Gerontol A Biol Sci Med Sci* 50: 17–22, 1995.

17. **Gu MJ, Schultz AB, Shepard NT, and Alexander NB.** Postural control in young and elderly adults when stance is perturbed: dynamics. *J Biomech* 29: 319–330, 1996.

18. **Hageman PA and Blanke DJ.** Comparison of gait of young women and elderly women. *Phys Ther* 66: 1382–1387, 1986.

19. **Hanavan EP.** *A Mathematical Model of the Human Body.* Dayton, OH: Wright Patterson Air Force Base, AMRL, 1964. (Aerospace Medical Research Laboratories Tech. Rep. 64-102)

20. **Holloszy JO, and Kohrt WM.** *Exercise: Aging.* New York: Oxford Univ. Press, 1995.

21. **Houmard JA, Weidner ML, Gavigan KE, Tyndall GL, Hickey MS, and Alshami A.** Fiber type and citrate synthase activity in the human gastrocnemius and vastus lateralis with aging. *J Appl Physiol* 85: 1337–1341, 1998.

22. **Judge JO, Davis RB III, and Ounpuu S.** Step length reductions in advanced age: the role of ankle and hip kinetics. *J Gerontol A Biol Sci Med Sci* 51: M303–M312, 1996.

23. **Judge JO, Underwood M, and Gennosa T.** Exercise to improve gait velocity in older persons. *Arch Phys Med Rehabil* 74: 400–406, 1993.

24. **Kerrigan DC, Todd MK, Della Croce U, Lipsitz LA, and Collins JJ.** Biomechanical gait alterations independent of speed in the healthy elderly: evidence for specific limiting impairments. *Arch Phys Med Rehabil* 79: 317–322, 1998.

25. **Kirtley C, Whittle MW, and Jefferson RJ.** Influence of walking speed on gait parameters. *J Biomed Eng* 7: 282–288, 1985.

26. **Klitgaard H, Mantoni M, Schiaffino S, Ausoni S, Gorza L, Laurent-Winter C, Schnohr P, and Saltin B.** Function, morphology and protein expression of ageing skeletal muscle: a cross-sectional study of elderly men with different training backgrounds. *Acta Physiol Scand* 140: 41–54, 1990.

27. **Larish DD, Martin PE, and Mungiole M.** Characteristic patterns of gait in the healthy old. *Ann NY Acad Sci* 515: 18–32, 1988.

28. **Larsson L, Grimby G, and Karlsson J.** Muscle strength and speed of movement in relation to age and muscle morphology. *J Appl Physiol* 46: 451–456, 1979.

29. **Lynch NA, Metter EJ, Lindle RS, Fozard JL, Tobin JD, Roy TA, Fleg JL, and Hurley BF.** Muscle quality. I. Age-associated differences between arm and leg muscle groups. *J Appl Physiol* 86: 188–194, 1999.

30. **McCaw ST and DeVita P.** Errors in alignment of center of pressure and foot coordinates affect predicted lower extremity torques. *J Biomech* 28: 985–988, 1995.

31. **Metter EJ, Conwit R, Tobin J, and Fozard JL.** Age-associated loss of power and strength in the upper extremities in women and men. *J Gerontol A Biol Sci Med Sci* 52: B267–B276, 1997.

32. **Murray MP, Drought AB, and Kory RC.** Walking patterns of normal men. *J Bone Joint Surg* 46A: 335–360, 1964.

33. **Murray MP, Kory RC, and Clarkson BH.** Walking patterns in healthy old men. *J Gerontol* 24: 169–78, 1969.

34. **Nigg BM, Fisher V, Allinger TL, Ronsky JR, and Engsberg JR.** Range of motion of the foot as a function of age. *Foot Ankle* 13: 336–343, 1992.

35. **Nigg BM, Fisher V, and Ronsky JL.** Gait characteristics as a function of age and gender. *Gait Post* 2: 213–220, 1994.

36. **Ostrosky KM, VanSwearingen JM, Burdett RG, and Gee Z.** A comparison of gait characteristics in young and old subjects. *Phys Ther* 74: 637–644, 1994.

37. **Petrella RJ, Lattanzio PJ, and Nelson MG.** Effect of age and activity on knee joint proprioception. *Am J Phys Med Rehabil* 76: 235–241, 1997.

38. **Porter MM, Vandervoort AA, and Kramer JF.** Eccentric peak torque of the plantar and dorsiflexors is maintained in older women. *J Gerontol A Biol Sci Med Sci* 52: B125–B131, 1997.

39. **Poulin MJ, Vandervoort AA, Paterson DH, Kramer JF, and Cunningham DA.** Eccentric and concentric torques of knee and elbow extension in young and older men. *Can J Sport Sci* 17: 3–7, 1992.

40. **Stefanyshyn D and Nigg B.** Dynamic angular stiffness of the ankle joint during running and sprinting. *J Appl Biomech* 14: 292–299, 1998.

41. **Sutherland DH, Olshen R, Cooper L, and Woo SL.** The development of mature gait. *J Bone Joint Surg Am* 62: 336–353, 1980.

42. **Thelen DG, Schultz AB, Alexander NB, and Ashton-Miller JA.** Effects of age on rapid ankle torque development. *J Gerontol A Biol Sci Med Sci* 51: M226–M232, 1996.

43. **Winegard KJ, Hicks AL, Sale DG, and Vandervoort AA.** A 12-year follow-up study of ankle muscle function in older adults. *J Gerontol A Biol Sci Med Sci* 51: B202–B207, 1996.

44. **Winter DA.** Biomechanical motor patterns in normal walking. *J Motor Behav* 15: 302–330, 1983.

45. **Winter DA.** Energy generation and absorption at the ankle and knee during fast, natural, and slow cadences. *Clin Orthop* 175: 147–154, 1983.

46. **Winter DA.** *Biomechanics and Motor Control of Human Movement.* Waterloo, ON, Canada: Wiley, 1990.

47. **Winter DA.** *The Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological.* Waterloo, ON, Canada: Univ. of Waterloo Press, 1991.

48. **Winter DA, Patla AE, Frank JS, and Walt SE.** Biomechanical walking pattern changes in the fit and healthy elderly. *Phys Ther* 70: 340–347, 1990.