Age causes a redistribution of joint torques and powers during gait

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.

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

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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 the correct foot would contact the force platform in a normal stance and contribute to propelling the individual. Negative work was 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 \( \alpha \) level of <0.05 was used to indicate statistical significance.

Preliminary statistical analyses for gender (t-test, \( \alpha < 0.05 \)) 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 to a greater forward lean in the trunk. In contrast, elderly adults contacted the force platform in a more flexed knee position than 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 the 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 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 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.

### Table 1. Stride characteristics

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

Values are means ± SD.

### Table 2. Joint position variables

<table>
<thead>
<tr>
<th>Variable</th>
<th>Elderly Adults</th>
<th>Young Adults</th>
<th>P Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Avg stance pos ROM in stance</td>
<td>−11.9 ± 4.7</td>
<td>−3.9 ± 4.1</td>
<td>0.000</td>
</tr>
<tr>
<td>Knee</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Heel contact</td>
<td>−8.3 ± 5.9</td>
<td>−4.4 ± 5.6</td>
<td>0.048</td>
</tr>
<tr>
<td>Flex after contact</td>
<td>−21.3 ± 5.5</td>
<td>−26.3 ± 4.7</td>
<td>0.009</td>
</tr>
<tr>
<td>Ankle</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Avg stance pos ROM in stance</td>
<td>−1.9 ± 3.8</td>
<td>−4.2 ± 3.4</td>
<td>0.045</td>
</tr>
<tr>
<td>Max plantar flex</td>
<td>13.8 ± 4.0</td>
<td>17.4 ± 5.4</td>
<td>0.035</td>
</tr>
</tbody>
</table>

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.
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

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

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 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.

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 the 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% decline.
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 produced 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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