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Old Adults Perform Activities of Daily Living Near Their Maximal Capabilities

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Background. Old adults' ability to execute activities of daily living (ADLs) declines with age. One possible reason for this decline is that the execution of customary motor tasks requires a substantially greater effort in old compared with young adults relative to their available maximal capacity.

Methods. We tested the hypothesis that the relative effort (i.e., the percentage of joint moment relative to maximal joint moment) to execute ADLs is higher in old adults compared with young adults. Healthy young adults (n = 13; mean age, 22 years) and old adults (n = 14; mean age, 74 years) ascended and descended stairs and rose from a chair and performed maximal-effort isometric supine leg press. Using inverse dynamics analysis, we determined knee joint moments in ADLs and computed relative effort.

Results. Compared with young adults, old adults had 60% lower maximal leg press moments, 53% slower knee angular velocity at peak torque, and 27% lower knee joint moments in the ADLs (all $p < .05$). Relative effort in ascent was 54% ($SD \pm$ 16%) and 78% (±20%) in young and old adults, respectively; in descent, it was 42% (±20%) and 88% (±43%); and in chair rise, it was 42% (±19%) and 80% (±34%) (all $p < .05$). The relative electromyographic activity of the vastus lateralis and the coactivity of the biceps femoris associated with this relative effort were, respectively, 2- and 1.6-fold greater in old compared with young adults in the 3 ADLs ($p < .05$).

Conclusions. For healthy old adults, the difficulty that arises while performing ADLs may be due more to working at a higher level of effort relative to their maximum capability than to the absolute functional demands imposed by the task.

OLD adults' ability to execute activities of daily living (ADLs) declines with age (1–9). One possible reason for this decline is that the execution of customary motor tasks such as walking, ascending or descending stairs, and rising from a chair requires a substantially greater effort in old compared with young adults relative to their available maximal capacity. Indeed, there is a wealth of historical data on cardiovascular function indicating that, due to a reduction in peak oxygen uptake, impairment of the oxygen delivery system, and changes in muscle fiber type with aging, old adults walk at a significantly higher percentage of their peak oxygen uptake, about 50%, compared with young adults, who walk at about 30% (10–12). A significant increase with age in the physiologic relative effort (i.e., the level of effort needed to execute a task as a percentage of the available maximal capacity) forces old adults to operate at high effort levels, causes premature fatigue, and in some cases leads to motor accidents.

Knowledge of the cardiovascular relative effort for the executions of ADLs is important, especially for exercise prescription (13). Yet intuitively it appears that the quantification of relative effort in ADLs in terms of muscle strength or joint torques should bring us closer to better understanding the causes of mobility limitations with age. However, such data for the neuromuscular system of healthy old adults, though highly needed, are scant. Stepping responses to small postural perturbations require modest joint torques, far below the presumed maximal torque reserves of old adults (14), but in reactions to impending falls, joint torques were 70 N · m at the ankle, 82 N · m at the knee, and 73 N · m at the hip in young adults. Such torque requirements, especially if combined with a need for rapid torque production, may reach or even exceed old adults' maximal capabilities. Indeed, it was suggested that the ability to ascend and descend stairs may require joint moments that exceed the available maximal levels in some healthy old adults and particularly in frail individuals (5). The knee joint torque requirements to execute ADLs such as stair ascent, stair descent, and chair rise range between 50 and 100 N · m (15–19), but the relationship between these values and maximal torque capabilities within individual old adults is unknown.

Determination of the relative effort in stairway locomotion is especially important because falls on stairs are the leading cause of accidental deaths. In addition, 80% of stairway accidents occur in descent (20). Data from some studies suggest that the relative effort needed to rise from a chair is relatively low (21–23), but when quantified, relative effort to rise from chairs of 0.38 to 0.58 m in height ranged from 70% to nearly 100% of maximal available knee torque in frail adults (18). These initial data on relative effort by Hughes and colleagues (18) combined with the well-established reduction in the maximal torque-producing capacity of the old adults' neuromuscular system prompted us to hypothesize that the relative effort to execute ADLs is higher in old compared with young adults. This hypothesis implies that old adults' difficulty in performing ADLs is not due to the absolute task demands but to their performing ADLs with much less reserve capacity compared with young adults.

One of the many adjustments in the aging neuromuscular system is the increase in the antagonistic muscle coactivation in responses to sudden perturbations or in the execution of single-joint and multijoint voluntary muscle contractions (24–29). Specifically, we have observed that old adults perform downward stepping with heightened antagonistic muscle preactivation and coactivation (28). Here we expand on the stepping data by testing the hypothesis that elevated muscle coactivity is an omnipresent phenomenon when old adults perform ADLs (28). In total, the purpose of the present study was to determine the relative effort necessary for old adults to execute ADLs and to assess the magnitude of antagonistic muscle coactivity while ascending and descending stairs and rising from a chair.

Methods

Subjects

We recruited 13 young white adults (6 women) and 14 community-dwelling old white adults (7 women) by newspaper advertisement and word of mouth. Young subjects were aged 19 to 25 years (mean ± *SD*, 22 ± 2 years), and old subjects were aged 69 to 77 years (74 ± 3 years). Young and old adults were similar in height (young, 1.71 ± 0.01 m; old, 1.65 ± 0.01 m) and mass (young, 70 ± 10 kg; old, 73 ± 10 kg). All subjects were apparently healthy and had not exercised more than once a week during the year preceding the study. Old adults were required to provide a physician's approval to participate in the study. This approval report and a medical questionnaire were used to determine whether an old adult met the criteria of having fewer than two major risk factors for coronary artery disease as defined by the American College of Sports Medicine (13). We excluded subjects with more than 2 risk factors for coronary artery disease, a history of falls, osteoporosis, osteoarthritis, orthopedic or neurologic conditions (i.e., stroke, Parkinson disease), use of medication that causes dizziness, smoking, a body mass–height ratio greater than 28, high blood pressure (140/90 mm Hg), or a heart condition. Before testing, subjects read and signed a written informed consent document approved by the university and medical center institutional review board.

Subject Preparation

Subjects dressed in black spandex bicycle shorts and wore dark running shoes. Height and mass were measured. Reflective markers were placed on the left side of the body, including the lateral side of the shoe at the heel and over the fifth metatarsal head, and on surface locations above the lateral malleolus, the femoral condyle, and the greater trochanter, and on the front corner of the force platform.

The skin over the left fibula head, vastus lateralis, and biceps femoris was shaved and washed with alcohol. Two single-use diagnostic electrocardiographic electrodes (ConMed Inc., Utica, NY) were attached to each muscle belly with a 3.5-cm center-to-center interelectrode distance to detect surface electromyographic (EMG) activity. The ground electrode was placed on the left fibula head. To avoid interference with task execution, the electrode leads were placed inside the spandex bicycle shorts. After being prepared, as a general warm-up, subjects bicycled for 5 minutes on a cycle ergometer at a resistance of 1 to 1.5 kg.

Protocol

Data were collected in one 2-hour session with 5 to 10 minutes of rest between the tasks. Subjects ascended and descended on a custom-built four-step wooden stairway that had a removable force platform embedded in the center of the second step. The step height was 0.19 m, and the step depth was 0.27 m. Subjects finished ascent on a landing measuring 1.0 × 1.0 m on the top of the stairway. The stairway was equipped with railings, but the data reported here do not include trials during which the subjects touched the railings. During ascent and descent, subjects synchronized their foot contact with the beat of a metronome that was set at 80 beats per minute. Subjects stepped on the force platform with their left leg. For the ascending trials, subjects stood one step in front of the stairway (about 0.5 m). They took an initial step with the left leg on the floor surface and continued up the stairway. For the descending trials, subjects stood on the landing area one step in front of the edge of the top step (about 0.5 m); they then took an initial step with the left leg on the landing surface and continued walking down the stairway. As a warm-up, subjects walked up and down the four-step stairway for 1 minute.

Subjects performed the sit-to-stand task on a wooden bench without arm and back support. The bench seat was 1.0 m long and 0.23 m wide, and it was set at 25% of each subject's height. The bench straddled the force platform so that the subject's left foot was on the platform and the subject's right foot was off the platform. While rising from the bench, subjects kept their hands in their laps. As a warm-up, subjects sat down and rose from the bench several times.

Subjects performed five trials per movement. The order of stair ascent and descent was alternated between subjects, as was the order of stairway locomotion and chair rise. During all movements, movement of the left side of the body was recorded with a video camera.

Maximal leg strength, used here as a reference relative to the torque requirements for the ADLs, was measured in the supine position on a leg press machine (model 7412, Cybex, Inc., Owatonna, MN). A force platform was firmly attached to the footrest of the leg press machine. Subjects placed their left foot on the center of the platform and rested their right foot on a stool on the side. Subjects were carefully instructed to press the platform with the midsole of their foot to reliably produce extensor moments at the knee. To facilitate subjects' efforts to produce knee extensor moment, the pelvis was strapped to the sled with wide leather weightlifting belts. As a warm-up, subjects performed three efforts at 50% to 60% of maximum effort in the leg press machine at each of five joint positions. Subjects performed no more than three maximal isometric efforts at 15°, 30°, 45°, 60°, and 75° of knee flexion (0° = fully extended knee) with 1 minute of rest between efforts. From the video data, we also determined the knee joint position at the peak knee extensor moment.

Data Recording and Analysis

Subjects performing the ADLs and the leg press were videotaped in the sagittal plane at 60 Hz with a Sony CCD-Iris black and white video camera (model SSC-M350) and were recorded with a super VHS JVC videocassette recorder (model HR S5100U). The camera was placed about 6 m from the subject, and its optical axis was at 90° in relation to the subject's sagittal plane. The field of view for the video image was approximately 2.5 m wide and 2.0 m high, which maximized the image size.

A force platform (model OR6-6, AMTI, Newton, MA) was firmly affixed to the footrest of the Cybex leg press machine to assess the maximal isometric strength of the subjects' lower extremity. For the stair locomotion trials, the force platform was removed from the leg press machine and mounted in the center of the second step of the stairway. Subjects performed the chair rise task on a second force platform (model LG-6-4-1, AMTI) embedded in an elevated walkway.

We collected EMG data with the TeleMyo telemetric hardware system (Noraxon USA, Inc., Scottsdale, AZ). As detailed previously (28), this system contains a battery-powered transmitter tied to the subject's waist with a cloth belt so that the subject can freely move while performing the ADLs. The differential amplifiers have a gain of 2000, an input impedance of 10 MΩ, and a common mode rejection ratio of 130 dB. The EMG signals have a bandwidth of 3 dB at 16 to 500 Hz.

We used our standard methods to reduce the kinematic, kinetic, and EMG data (28,30). In brief, the vertical and anteroposterior ground reaction forces, the mediolateral moment from each force platform, and the two channels of EMG signals were digitized at 1 kHz by a 12-bit analog-to-digital converter (DAS 1402, Kethley Metrabyte, Taunton, MA) and stored by the Myosoft software (Noraxon USA, Inc.) on a Pentium personal computer. For ADLs and the leg press efforts, Cartesian coordinates of the reflective markers were derived from the video records using the Peak5 system (Peak Performance Technologies, Englewood, CO). High-frequency error was removed from the digitized coordinates with an automatic, low-pass digital filter using an average cutoff frequency of about 5 Hz. Linear position data were interpolated to 200 Hz using a cubic spline routine without further smoothing, and linear velocity and acceleration were calculated for each joint. The position data were then interpolated to 200 Hz; this higher frequency allowed the synchronization of kinematic and kinetic data by relating every fifth force data point to each interpolated video data point. Joint angular position and velocity were calculated at the hip, knee, and ankle, and the joint position curves were evaluated with six variables describing stance phase kinematics.

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 using a mathematical model (31), segmental masses reported by Dempster (32), and the individual subject's anthropometric data. Center of pressure 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 based on the digitized location of the force platform. Inverse dynamics using linear and angular Newtonian equations of motion were used to calculate the joint reaction forces and moments at the ankle, knee, and hip joints, but here we report only the knee moment data.

The root-mean-square (RMS) of the direct EMG signal was obtained by using a 20-millisecond smoothing window. Area under the RMS envelope (in millivolt-seconds) was determined between two cursors; the first cursor was set 200 milliseconds before foot contact with the force platform, and the second was set at the end of the stance phase for stair ascent and descent. For chair rise, the analysis started at the first deflection of the vertical ground reaction force and ended at the end of movement. For the leg press, the area under the RMS curve was determined for a 1-second window surrounding the peak knee extensor moment. EMG activity of the vastus lateralis recorded during ADLs was expressed as a percentage of the maximal EMG activity of the vastus lateralis determined during the leg press. Coactivity was computed as the quotient of biceps femoris RMS EMG activity divided by vastus lateralis RMS EMG activity. Figure 1 shows a typical example of EMG recordings during stair ascent and leg press in a young subject and in an old subject.

Statistical Analyses

For each subject, five trials for each ADL were analyzed, and the average of these trials was used in the data analysis. Of the leg press trials, the trial that produced the highest knee extensor moment at each joint position was included in the data analysis. We report means ± *SD*s. In a separate study conducted over 2 separate days, we assessed reliability of the dependent variables in 12 subjects, and the intraclass correlation coefficient (R) ranged from 0.63 to 0.88. Kinematic dependent variables for the three ADLs were knee joint position at contact or start of chair rise, position at peak torque, position at toe-off, range of motion, peak extension or flexion velocity, and velocity at peak torque. Kinetic dependent variables included peak knee joint moment during ADLs and leg press, normalized for body mass. The peak knee joint moments observed during ADLs were expressed relative to maximal isometric knee joint moments observed during the leg press, yielding the *relative effort* ratios for ADLs. The EMG dependent variables included the average RMS EMG of the vastus lateralis over the task period relative to the maximal EMG observed during the leg press task. EMG coactivity during ADLs was expressed as a ratio of biceps femoris RMS EMG relative to vastus lateralis RMS EMG activity. The leg press position–joint moment data were analyzed with an age group by position ANOVA with repeated measures, followed by a Tukey post hoc contrast test. Other comparisons between age groups were done by unpaired, two-tailed t tests at a significance level of $p < .05$.

Results

Table 1 shows the group results for six kinematic variables relative to the knee joint in the ADLs. Knee angular velocity at peak torque was 18% slower during ascent and chair rise, and 124% faster during descent in old compared with young subjects ($p < .05$). No other functionally meaningful differences were noted between the two age groups in the kinematics of the three ADL tasks. The

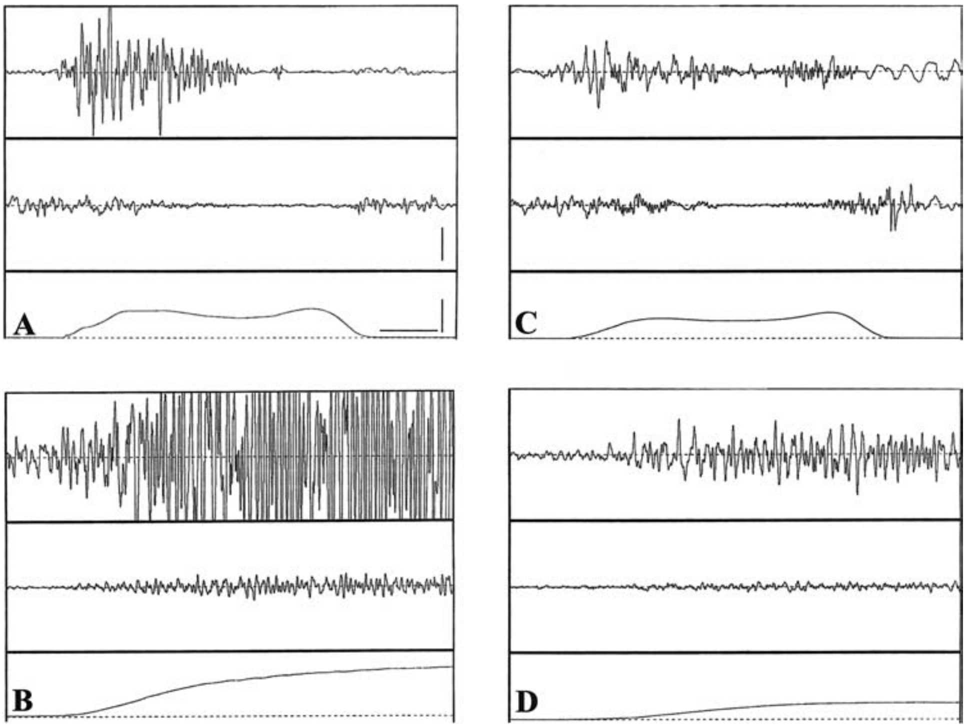

Figure 1. Typical example of direct surface electromyographic (EMG) recordings and vertical ground reaction force during stair ascent (**A** and **C**) and supine leg press (**B** and **D**) with the left leg in a young adult (age, 22 years; **A** and **B**) and an old adult (age, 69 years; **C** and **D**). In each panel, the top tracing is vastus lateralis EMG activity, the middle tracing is biceps femoris EMG activity, and the bottom tracing is vertical ground reaction force measured on a force platform. Note the large difference between the young (B) and old (D) subjects in maximal strength. In these particular individual trials, the biceps femoris coactivity relative to vastus lateralis EMG activity, based on root-mean-square (RMS) EMG envelope 200 milliseconds before foot contact to the end of the ascent movement (RMS EMG tracings not shown), was 31% in the young subject and 47% in the old subject. Time calibration is 125 milliseconds, EMG calibration is 0.125 mV, and force calibration is 1000 N.

Table 1. Knee Joint Kinematics During Stair Ascent, Stair Descent, and Rising From a Chair

Movement and Variable	Young Adults		Old Adults	
Ascent				
Position at contact, °	−68	5	−67	3
Position at peak torque, °	−54	6	−53	6
Position at toe-off, °	−29	6	−26	6
Range of motion, °	57	6	53	3*
Peak extension velocity, °/s	151	19	141	25
Velocity at peak torque, °/s	122	17	105	19*
Descent				
Position at contact, °	−13	4	−14	5
Position at peak torque peak, °	−27	4	−26	5
Position at toe-off, °	−95	4	−96	4
Range of motion, °	82	2	82	6
Peak flexion velocity, °/s	−124	26	−114	18
Velocity at peak torque, °/s	−25	26	−56	26*
Chair rise				
Position at start, °	−87	4	−90	7
Position at peak torque, °	−78	4	−80	10
Final position, °	−4	3	−8	2*
Range of motion, °	81	4	82	8
Peak extension velocity, °/s	141	36	138	25
Velocity at peak torque, °/s	51	14	41	12*

Note: All data are for the left knee. Negative joint position values indicate knee flexion, and positive values indicate knee extension.

* $p < .05$ for difference between age groups.

average knee joint position where peak torque occurred in ascent in young and old subjects was 54° (±6°), and the position at which we extracted each subject's torque from the leg press position-torque was 54° (±9°). In descent, these two joint position values were 27° (±5°) and 32° (±6°), respectively. In chair rise, the values were 78° (±7°) and 71° (±9°), respectively. On average, we were able to match joint positions with 9% difference ($p > .05$).

Figure 2 shows that old adults produced significantly lower knee joint moments normalized for body mass at each of the five positions during a supine leg press with the left leg. The reductions ranged from 55% to 69%, with a mean reduction of 60% (age main effect, F = 155.8, $p < .001$). The position main effect was also significant and, as expected, at extreme knee joint positions, the moments were lower than in the middle positions (F = 65.3, $p = .003$). The group by knee joint position interaction was not significant.

Table 2 and Figure 3 display the group data for required knee joint moments in the ADLs. Table 2 also shows the group data of the leg press knee joint moments that were selected for each subject from the position–moment curve (Figure 2) to match the knee joint position at which peak extension moment occurred in each ADL. While performing identical tasks with almost similar kinematics, old adults executed ADLs by producing 27% less knee joint moments than young adults ($p < .05$ for each task).

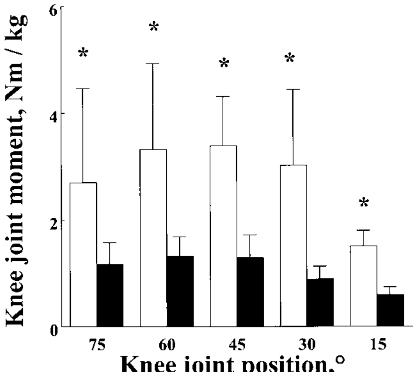

Figure 2. Knee joint moments, normalized for body mass, at five knee joint positions during supine leg press with the left leg. Young subjects (open columns) produced significantly greater knee joint moments than old subjects (filled columns) at each joint position. $^{*}p < .05$ for young compared with old subjects.

Figure 4 shows that the relative effort to execute the ADLs was 1.8-fold greater for old compared with young adults. Relative effort was expressed as the percentage of knee extensor moment produced during an ADL in relation to the maximal knee extensor moment produced at a similar knee joint position during a maximal-effort supine leg press with the left leg. In ascent this effort was 54% (±16%) and 78% (±20%) in young and old adults, respectively; in descent, it was 42% (±20%) and 88% (43%); and in chair rise, it was 42% (±19%) and 80% (±34%) (all $p < .05$). Similarly, the EMG activity associated with this torque demand was twofold greater in the old compared with the young adults in the three ADLs. Relative vastus lateralis EMG activity was expressed as a percentage of maximal vastus lateralis EMG activity measured during maximal-effort supine leg press with the left leg. In ascent, the relative EMG activity was 28% (±20%) in young subjects and 62% (±22%) in old subjects, respectively; in descent, it was 33% (±21%) and 79% (±25%); and in chair rise, it was 29% (±22%) and 77% (±23%) (all $p < .05$). We observed a moderately close association between relative effort and relative EMG activity. Relative EMG activity accounted for 44% of the variance in relative effort in the three ADLs in the two age groups combined (Figure 4). The linear regression of relative EMG activity on relative moment was described by the equation of $y = 0.53x + 33.4$, $r = .66$, $F = 119.7$, and $p < .001$.

Table 2. Knee Joint Moments During Stair Ascent, Stair Descent, Rising From a Chair, and Supine Leg Press

Movement and Variable	Young Adults		Old Adults	
Ascent				
Peak extension moment	1.55	0.24	1.00	0.22*
Leg press moment†	3.09	1.14	1.17	0.29*
Descent				
Peak extension moment	0.90	0.24	0.64	0.24*
Leg press moment†	2.27	1.22	0.89	0.38*
Chair rise				
Peak extension torque	0.84	0.18	0.69	0.13*
Leg press moment†	2.50	0.97	0.93	0.25*

Note: Values are the highest knee extension moments for the left leg normalized for body mass (N · m/kg) that occurred during the specific task.

† Knee joint extensor moments from each subject's joint position–moment curve in the leg press, nearest to the peak knee extension moment in the specific activity of daily living. Subjects performed the supine leg press with the left leg contacting the force platform.

* $p < .05$ for difference between the two age groups.

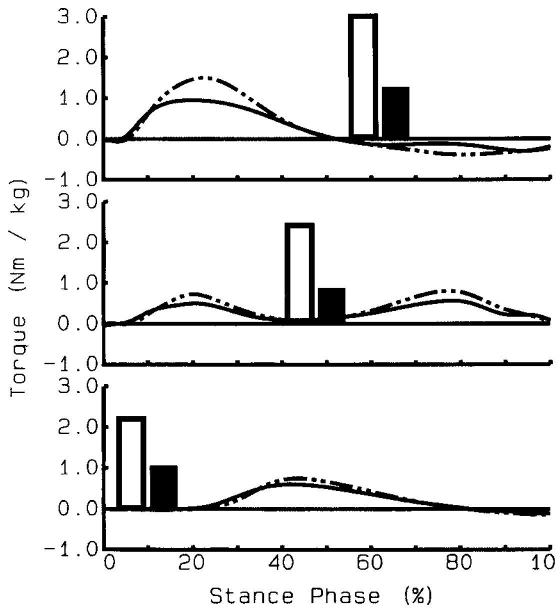

Figure 3. Group mean, body mass–normalized knee joint moments during stair ascent, stair descent, and sit-to-stand in young and old adults. For stair ascent and descent, 0% corresponds to initial foot contact with the stair, and 100% marks toe-off. For chair rise, 0% corresponds to lift-off, and 100% marks fully erect position at end of rise. The inset columns represent the group mean of the maximal knee joint moments measured at similar knee joint positions in a leg press task at which the peak torques occurred in the ADLs. Dashed lines and open columns denote young adults, and solid lines and filled columns indicate old adults. *SD*s are omitted for clarity.

The amount of hamstring EMG activity relative to the vastus lateralis EMG activity revealed that in the ADLs, the hamstring muscle coactivity was 1.6-fold greater in old adults than in young adults. In ascent, the coactivity ratio (a dimensionless value) was 26 (±21) in the young adults and 51 (±36) in the old adults ($p < .05$). In descent, it was 46 (±37) and 64 (±32), and in chair rise, the corresponding values were 24 (±22) and 34 (±20), respectively, for young and old adults (both $p < .05$). As relative effort increased, hamstring coactivity also increased, and it accounted for about 30% of the variance in relative effort. The linear regression of relative effort on hamstring muscle coactivity in stair ascent, stair descent, and chair rise performed by young and old adults was described by the equation of $y = 0.40x + 21.4$, $r = .54$, $F = 34.9$, and $p = .003$.

Discussion

According to the hypothesis, the relative effort necessary to execute ADLs was almost twice as great in old adults as in young adults. Old adults performed near their maximal strength capabilities while ascending and descending stairs and rising from a chair. The relative knee joint effort in the 3 tasks averaged 46% and 82% in young and old adults,

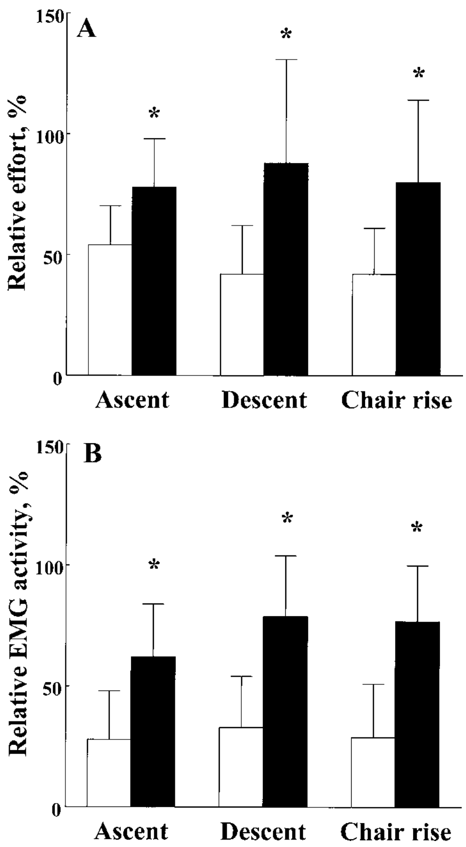

Figure 4. Relative effort at the knee joint (**A**) and the relative electromyographic (EMG) activity of the vastus lateralis associated with this moment during the execution of activities of daily living (ADLs) (**B**). Relative effort is expressed as the percent of left knee extensor moment produced during an ADL in relation to the maximal knee extensor moment produced at a similar knee joint position during maximal-effort supine leg press with the left leg. Relative vastus lateralis EMG activity is expressed as a percentage of maximal vastus lateralis EMG activity measured during maximal-effort supine leg press with the left leg. $^*p < .05$ for difference between age groups. Open symbols denote young adults, and filled symbols denote old adults.

respectively. The high relative effort values in old adults were caused by a combined effect from 60% lower maximal isometric lower extremity moments and 27% lower joint moments produced in the ADLs compared with those of young adults.

Our relative effort value of 82% and specifically the 80% value for chair rise from a chair height of 25% of body height (~0.43 m) is remarkably similar to the 78% value reported previously for chair rise in somewhat older subjects (age, 78 years) (18). The similarity of these relative effort values is somewhat surprising. By applying highly stringent inclusion criteria, our subjects were unusually fit and mobile, whereas five subjects in the previously noted study were unable to rise from a chair set at about the same relative height as in the present study, indicating a high level of frailty (18). At lower chair heights, which we did not include in testing, frail adults use 97% of their maximal leg strength (18). Even though the relative effort values are similar in the two studies, we suspect for two reasons that the similarities may be coincidental. One reason is methodologic. Hughes and colleagues (18) obtained joint torques in chair rise by the process of inverse dynamics, but referenced these values to maximum torques from isokinetic measurements. They also used only one knee joint position (60°) in the reference task, but used a range of different seat heights. In contrast, we derived ADL and maximal-effort joint torques through the identical process of inverse dynamics analysis. We used position-specific knee moments measured in the supine leg press, a task that also afforded similar muscle lengths at the knee joint as well as comparable alignment between the hip and ankle joints in the ADL and the leg strength test.

Another difference between the present and the previous study (18) is that although the required torque was similar in young and old subjects in the chair rise task (18), our old subjects executed the ADLs with ~30% lower knee joint moments compared with the young subjects (Table 2). These lower levels of required torques suggest that the fit old adults in our study used a different motor strategy to execute ADLs, including the chair rise, compared with the frail subjects in the study of Hughes and colleagues (18). As was the case here (Table 1) and in many previous studies (2,6,33), but not in the study of Hughes and colleagues (18), old adults executed ADLs more slowly than young adults. Even though we controlled the average velocity of ADLs with a metronome, and the gross movement kinematics of ADLs was similar in young and old adults, we also found that velocities at individual joints were different between the two age groups (34). A slower execution of ADLs allows old adults to seek and attain acceptable postural stability while performing an ADL (23). The lower required torques in old adults also suggest that referencing torque requirements of ADLs to the torque levels used by young people (1.5–2.0 N · m/kg) may be misleading (5) (Table 2). Old adults may simply adopt a strategy to execute an ADL with less torque at the critical joint and alter torque or velocity pattern at adjacent joints. Evidence for such a strategy is also apparent in level walking, during which the total support moments are similar in young and old adults, but the distribution of moments between the joints are different in the two age groups (30). Preliminary data on stairway locomotion also suggest that even though knee joint moments are critical for the successful execution of ADLs (18,23,30,35,36), old adults may adopt a strategy of redistributing the total available torque by using proportionally less knee torque and more hip torque (19). A switching in the velocity patterns between knee and hip joint is another manifestation of this strategy; functionally impaired old adults increase hip flexion and optimize knee joint velocities while rising from a chair (34,35). Thus, lower torque production at one joint in old compared with young adults while executing ADLs does not necessarily signify mobility impairment; instead, it may indeed indicate the adaptive mechanical plasticity of the aging neuromuscular system.

Because the high relative effort in old adults is due to significantly reduced capability to produce maximal leg strength, one would also expect that people execute more difficult ADLs at higher levels of relative effort. For example, it has been suggested that subjects need to produce knee moments during stair descent that are more than

1.5 times the knee moments required to rise from a chair (5), but the required torque values for stair descent and chair rise within each age group were very similar (Table 2). As expected, of the 3 ADLs, ascent still required the most absolute torque. However, the relative effort for the ADLs was within a narrow range in both young adults (42–54%) and old adults (78–88%). The similarity between the levels of relative effort used by individuals in the different ADLs was probably caused by a match between mechanical requirements of an ADL and muscle mechanics in the reference task (Table 1). We were able to determine the relative effort in an ADL from the knee joint position–knee joint moment curve with less than 10% difference, even though we were not able to match the type of muscle contraction in the ADL with the reference task. This latter limitation probably has little effect on the accuracy of the relative effort values. It is possible that we overestimated relative effort for stair descent because eccentric strength loss is less with increasing age (37,38). However, due to a relative preservation of eccentric strength with age, the difference between eccentric and isometric strength is very small on the quadriceps torque–velocity curve of old adults. It is thus not likely that the 88% relative effort for stair descent in old adults represents a gross overestimation. Isometric torques are higher than concentric torques in the torque–velocity relationship, and thus the use of isometric instead of concentric moments to estimate relative effort in chair rise and stair ascent could have been underestimated. However, knee joint velocity at peak moment, used in the relative effort computation, was only 40° to 50° per second, suggesting a near isometric state and very little, if any, underestimation of the relative effort (Table 1).

A final observation in this study was the association between increased relative effort at the knee joint during the execution of ADLs and increased muscle coactivation. The amount of hamstring EMG activity relative to the vastus lateralis EMG activity was 1.6-fold greater in old than in young adults. As relative effort in the ADLs increased, hamstring coactivity significantly increased. These findings extend on previous reports of increased, abnormal, or ill-timed coactivity in aging during single-joint motor tasks (24–26), multijoint voluntary tasks (27,28), and multijoint tasks performed under sudden perturbations (29). The increased coactivity in freely moving old adults executing ADLs complements our similar observation made during downward stepping (28).

The source of this increased coactivity could be an aberration in the central partitioning of neural drive to the target muscles (39,40), an alteration of segmentally mediated reflexes (41), or a combination of these two factors (28). A more mundane explanation for increased hamstring coactivity is that old adults execute the ADLs with greater hip flexion (i.e., leaning forward), requiring an increase in hip extensor moments that would require increased hamstring activity (19,34,35,42). At any rate, the functional interpretation of the increased coactivity around the knee joint is that old adults increase muscle coactivation to compensate for the reduced net torque production while executing ADLs (Table 2), concurrently contributing to the increased relative effort. It is also possible that the increased coactivity actually causes the reduction in net knee torque. Because coactivity increased in all three ADLs, we propose to revise the conventional interpretation that it is a maladaptation to aging. Instead, we suggest that increased coactivity is the evolving functional mechanism to increase limb and joint stability or limb and joint stiffness that compensates for neuromotor impairments, including loss of muscle strength.

In conclusion, we found that healthy old adults execute three ADLs near their maximal torque-producing capabilities of the knee musculature. This increased relative effort at the knee joint was associated with increased neural drive to the involved muscle as well as an enhanced coactivation of the opposing muscle. These results provide a conceptual justification for the many training studies designed to improve maximal performance capabilities of the aged neuromuscular system. For healthy old adults, the difficulty that arises while performing ADLs may be due more to working at a higher level of effort relative to their maximum capability than to the absolute functional demands imposed by the task.

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