Muscular Performances at the Ankle Joint
in Young and Elderly Men

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The effect of aging on mechanical and electromyographic characteristics of ankle joint muscles was investigated in 11 young (mean age 24 years) and 12 elderly (mean age 77 years) males. Maximal and submaximal isometric voluntary torques were measured during ankle plantarflexion and dorsiflexion. Electromyographic activities of triceps surae and tibialis anterior muscles were recorded. The elderly group developed equal maximal dorsiflexion torques (42 vs 45 N.m, \( p > .05 \)), but in plantarflexion, the elderly group was weaker (80 vs 132 N.m, \( p < .001 \)) and presented a decreased twitch amplitude (11 vs 16 N.m) and lower coactivation (8% vs 15%) than that of the young adults. We established a linear relationship between the percentage of coactivation and developed resultant torque. Our results showed that dorsiflexor muscles were not affected by aging, contrary to plantarflexors, in which the decline in torque was partly explained by changes intervening at the peripheral level.

Aging is commonly characterized by a progressive decline in neuromuscular function and performance. This deterioration can be estimated by measuring maximal voluntary torque, which decreases essentially after the sixth decade (1) at about 1.5%–2.5% per year (2). This weakness in elderly persons can diminish quality of life by making routine activities and recreational exercise more difficult. Whatever the muscles involved, this decline in strength in older persons (≥60 years or older) could involve values from −14% to −45% compared to those in younger persons (≥20 years) for upper and lower extremities (1,3–7). Explanations as to why the magnitude of the decline in strength varies among muscles with aging could be that they do not fulfill the same functions (tonic vs phasic muscles), and that their constitutions, in terms of fiber types, are different (slow vs fast fibers). Nevertheless, it has appeared that the highest (−45%) and lowest (−14%) reported strength declines correspond to decreases noted in the muscles of the ankle joint (1,5). The function of these muscles is of course highly relevant to locomotion, posture, and balance (8), and to the prevention of falls in older adults (9). Results reported in the literature about the ankle joint muscles with aging are conflicting, and studies examining both ankle plantar- and dorsiflexor muscles are rare. A major decline in plantarflexion (PF) torque with aging has been reported (1,10), but results are contradictory concerning age-related effects on dorsiflexion (DF) strength (1,5,10–12). It seems that plantarflexor muscles are the muscles most affected by aging, which must have an effect on postural stability (13,14). Furthermore, these plantarflexor muscles, especially soleus muscle, have essentially an antigravity function, whereas the tibialis anterior has a phasic role. Fiber type analysis revealed that the tibialis anterior muscle contains a rather high proportion of type I fibers [76% in younger, 84% in older persons (15)]. In terms of the triceps surae muscle group, the soleus muscle is composed almost entirely of type I fibers (89%) (16,17), and gastrocnemii muscles have approximately the same proportion of type I and II fibers (16) in healthy young adults. These different functions and compositions of the ankle muscles could be sources of dissimilar alterations in the muscular performances in PF and DF with aging. In addition, it is well established that there is a loss of muscle mass that comes about with aging (6,18,19); this loss leads to a certain decline in muscle strength and endurance. It was then hypothesized that the decline in strength with aging was different in plantarflexors and dorsiflexor muscles, consistent with existing data reporting that maximal muscle cross-sectional area decreased by =16% in the ankle dorsiflexor muscles in healthy older adults (20). In contrast, the muscle cross-sectional area of the plantarflexor compartment was smaller in elderly muscles (28%–36%) with greater amounts of nonmuscle tissue, that is, fat and connective tissue (81% more than in the young) (21). The reduction in maximal voluntary torques with aging could be linked to structural and/or nervous phenomena.

The decline in strength could first be explained by an alteration of capacity to maximally activate one’s muscles with aging. Indeed, the weakness associated with aging may be related to neural adaptations: The generation of strength relies on the ability of the nervous system to activate muscles effectively (22). Maximal muscular activation implies that all available motor units are recruited and that they are driven to their maximal firing rates (23). In a recent study (24), elderly males displayed a reduced activation capacity of the PF compared with young males. Conversely, other authors have concluded that most older individuals were able to use their descending motor pathways for optimal activation of the muscles of the ankle joint (1,5,12,25,26). Incomplete activation can be the result of a lowered motor unit discharge rate or a relatively larger number of motor units that cannot be voluntarily activated in the muscle (22). Some reports have suggested that discharge rates in motor units in elderly
individuals are smaller than those in younger individuals (12,27,28).

Second, a reduction in muscular torque could also be caused by an increase in antagonist activity, inducing a co-contraction. Indeed, measured muscular torque is a resultant one: it corresponds to the sum of the participation of agonist and antagonist muscles. This coactivity, usually called coactivation, is especially involved in stabilizing the joint and preserving its physical integrity (29–31). To our knowledge, there are very few studies relating to coactivation at the ankle joint muscles during maximal voluntary contractions (MVC). According to previous studies on other muscles, we could hypothesize that this coactivation increases (3,4,6) or remains stable (6,32) with aging. This coactivation could also evolve in different ways during PF and DF with aging, thus explaining a potential difference in the evolution of maximal voluntary torques in PF and DF.

Finally, this age-related decrease in muscular torque could also be due to adaptations occurring at the peripheral level. Some studies found that the amplitude of the twitch torque was affected (1,33), whereas others found that it was stable with aging (34–36). Although the twitch response could be affected by the quantity of contractile protein in a muscle and changes in series elasticity (7,37), the M-wave and corresponding twitch can give information on changes in the processes involved in the excitation–contraction coupling.

In light of these considerations, our hypothesis was that the age-related alterations of PF and DF torques might be different, the former probably being more affected. Thus, the present study was designed to compare maximal voluntary torque, neural drive, muscular coactivation, and peripheral parameters of dorsiflexor and plantarflexor muscles in young and elderly men, to examine the different mechanisms leading to the loss of force.

**Method**

**Participants**

Two groups of healthy male volunteers took part in this investigation: the first consisted of 11 young adults (age 23.91 ± 1.70 years; height 1.76 ± 0.04 m; weight 74.00 ± 8.53 kg), and the second of 12 elderly men (age 77.09 ± 1.81 years; height 1.68 ± 0.03 m; weight 74.45 ± 10.95 kg). As neurogenic muscle changes and atrophy seem to accelerate after age 70, participants in the second group were over 75 years old. The older adults underwent a complete medical examination, and only individuals free from muscular, neurological, cardiovascular, metabolic, and inflammatory diseases took part in the present investigation. To select moderately active individuals, an activity pattern profile was determined for each participant, taking into account the type and number of hours of exercise per week. Only those individuals taking part in recreational, noncompetitive, physical activities at a frequency of no more than twice a week were admitted to the study. None of the older participants had suffered a fall during the 2 years preceding the study. Written informed consent was obtained, and all experimental procedures conformed to the standards set by the Declaration of Helsinki, and were approved by the local Committee on Human Research.

**Torque Measurement**

In the aim of standardizing the tests, they were undergone between 2:00 PM and 6:00 PM for all participants. Torque and electromyographic (EMG) data were recorded concurrently during maximal and submaximal (30% and 60%) PF and DF efforts. Strength was measured by using an isokinetic ergometer (System 3; Biodex Medical Systems, Shirley, NY). Participants were supine with a knee angle of near 180° (full extension) and an ankle joint of 90°. The right leg was placed horizontally, and the axis of the ankle joint was aligned with the center of rotation of the dynamometer shaft. So as to minimize trunk and hip movement during contraction, the waist was stabilized by means of a belt; arms were positioned across the chest. The right foot was also attached to the dynamometer by means of a system with a shoe and straps. All torque measurements were gravity corrected by using Biodex software after calibration to determine the maximal effect of gravity on static torque (38). Strength signals obtained through the isokinetic ergometer were directly administered by Tida 4.11 software (Tida; Heka Electronik, Lambrecht/Pfalz, Germany), gathered with a sample frequency of 2 kHz.

**Electrical Stimulation**

Participants were stimulated at rest to evoke an EMG response (M-wave, see EMG recording below) associated with a mechanical response. The posterior tibial nerve was stimulated using a cathode ball electrode (0.5 cm diameter) pressed into the poplitea fossa. The anode was a large rectangular electrode (10 cm × 5 cm; Medicompex SA, Ecublens, Switzerland) placed on the anterior surface of the knee. The peroneal nerve was stimulated using an electrode, cathode, and anode side by side, placed just below the head of the fibula. The percutaneous electrical stimulus was a rectangular pulse of 1-millisecond duration for the tibial nerve and 0.5-millisecond width for the peroneal nerve, with a constant tension of 400 V and an adjustable intensity delivered by a Digitimer stimulator (model D57; Hertfordshire, England). Each participant was initially familiarized with several submaximal electrical stimuli. The current intensity was then progressively increased until maximal twitch torque in PF and in DF and, respectively, maximal soleus and tibialis anterior M-wave amplitude was achieved. When this optimal intensity was determined, four single stimuli were delivered, each separated by a 5-second interval. The individual intensity was then maintained for the entire session of electrical stimulation. The positioning and the upholding of the electrode were under the control of the experimenter.

Subsequently, six MVC in each condition were maintained for 5–6 seconds. Three paired stimuli (3 × 2 consecutive pulses at a frequency of 100 Hz) were applied to assess activation capacity [twitch interpolation technique (39)]. Superimposed doublets rather than twitches were used to increase the signal-to-noise ratio. The first doublet was applied 2 seconds before MVC as a control doublet, the second during MVC, and a further potentiation doublet applied after the MVC. We evaluated maximal muscular torques and levels of activation, activity, and coactivation.
from the highest contractile force records. During each maximal contraction, the participant was strongly encouraged by the experimenters.

Next, participants were asked for efforts corresponding to 30% and 60% of the maximal torque previously developed in PF and DF. The target effort and the exerted torque were displayed, in real time, on the computer monitor as visual feedback to the participant. Resting time of at least 30 seconds was taken between each trial to avoid any effect of fatigue on the measurements.

For one elderly participant, we did not manage to obtain correct measures for the tibial nerve stimulation. For two other participants, we were not able to calculate the level of maximal activation: For one, the stimulation intensity was too high and did not allow us to send a doublet, and, for the second, the maximal level of strength was not reached when the doublet was imposed.

**EMG Recording**

EMG activity from the soleus, gastrocnemius medialis, gastrocnemius lateralis, and tibialis anterior muscles was recorded by means of two silver-chloride surface electrodes of 10-mm diameter (Controle Graphique Medical, Brie-Comte-Robert, France), with an interelectrode (center-to-center) distance of 25 mm. For the soleus, the recording electrodes were placed along the mid-dorsal line of the leg, about 5 cm distal from where the two heads of the gastrocnemius join the Achilles tendon. For the gastrocnemius medialis, gastrocnemius lateralis, and tibialis anterior muscles, electrodes were fixed lengthwise over the middle of the muscle belly (40). Then, following the European Recommendations for Surface ElectroMyoGraphy (41), for the gastrocnemius medialis, electrodes were placed on the most prominent bulge of the muscle; for the gastrocnemius lateralis, electrodes were placed at one third of the line between the head of the fibula and the heel; and for tibialis anterior muscles, electrodes were fixed at one third of the line between the tip of the fibula and the tip of the medial malleolus. These sites were determined by eliciting the greatest M-wave amplitude that could be evoked for each muscle via nerve stimulation, for a given stimulus intensity. These procedures were performed so as to avoid the innervation zone and, therefore, obtain the optimal amplitude of the EMG response (42–47). The reference electrode was attached in a central position on the same leg, between gastrocnemii muscles. Low impedance (<5kΩ) at the skin–electrode interface was obtained by shaving, abrading, and cleaning the skin with an alcohol–ether–acetone mixture. EMG signals were amplified by a bandwidth frequency ranging from 15 Hz to 2 kHz and subsequently stored for offline analysis.

**Data Analysis**

For each participant, the highest MVC and the more accurate and steady recordings of submaximal efforts were analyzed. Indeed, when maximal ability between age groups is compared, an average score is not appropriate. For instance, to determine whether maximal voluntary activation differs between young and old men, the highest value attained, not the mean of all attempts, should be used in the assessment, and a sufficient number of attempts need to be made to attain the best MVC in older adults (48).

For voluntary torque testing, the root mean square (RMS) myoelectric activity of the soleus, gastrocnemius medialis, gastrocnemius lateralis, and tibialis anterior muscles was calculated. During isometric activity, the RMS EMG was measured over a 0.5-second period after the torque had reached a plateau.

As for the electrically evoked twitches, the average of four EMG signals and mechanical responses was considered and peak-to-peak amplitude and duration of soleus, gastrocnemius medialis, gastrocnemius lateralis, and tibialis anterior muscle M-wave were calculated. The following twitch contractile properties were computed: a) peak twitch (Pt), the highest value of twitch tension production; b) time to peak torque (TPT), the time to twitch maximal torque, calculated from the onset of the mechanical signal; and c) half-relaxation time (HRT), the time needed to obtain half of the decline in twitch maximal torque.

To reduce the differences in the EMG signals due to changes at the skin level, the RMS values were normalized to the amplitude of the M-wave of the session (49–53). For the PF, as the RMS of the agonist muscle group corresponded to the sum of the RMS of gastrocnemius and soleus muscles, the M-wave amplitude corresponded to the sum of the amplitudes of the M-waves obtained from the three muscles of the triceps surae. As for the duration, the mean of the M-waves of these three muscles was used.

To estimate the individual level of voluntary activation, the ratio of the amplitude of the superimposed doublet over the size of the postcontraction doublet was calculated as follows (4,12,24,39,48):

Voluntary activation (%) = \( \frac{(1 - \text{superimposed doublet torque}) \times 100}{\text{post-MVC doublet torque}} \)

To evaluate the level of coactivation, RMS values of the antagonist EMGs were calculated, normalized, and expressed as a percentage of the RMS value corresponding to the maximal isometric contraction when the muscle acted as an agonist (6,29,54). To establish the relationship between the coactivation and the corresponding torque developed, data from all participants, with the percentage of coactivation on the y-axis plotted against maximal and submaximal voluntary torques (100%, 60%, and 30%), were used.

Cross-talk was assessed in 6 of the participants (3 young and 3 elderly men). We examined amplitudes of soleus M-waves when stimulating the tibial or peroneal nerve at the optimal intensity, during rest, and at maximal (100%) and submaximal (60% and 30%) PF contractions. Under these conditions, cross-talk was found to be less than 9% of the maximal M-wave. The extent of cross-talk was thus considered negligible, both in our experiments (between the synergist or antagonist muscles) and in other studies (55,56).

**Statistical Analysis**

All statistical tests were performed with Statistica software (version 6.1; StatSoft, Tulsa, OK). Ordinary statistical
methods, including means and their standard deviations (SDs), standard errors (SEs), and the coefficient of variation (CV) were calculated for each parameter. The data are presented as means ± SD in the text and table, and as means ± SE in the figures. Normality of the data was checked using the Kolmogorov–Smirnov test, and equality of variances was verified by the Levene test. An unpaired Student t test was then executed to compare the results obtained from young and elderly men (maximal voluntary torques, activation levels, RMS/M_max during PF, parameters of mechanical twitches, and parameters of M-waves). However, a Mann–Whitney U test was performed on data that did not follow a normal law or that did not fulfill the equality of variances condition (RMS/M_max during DF). To determine whether there was a significant difference between the activation level of each group and the complete activation, the significance was tested by comparison with a standard value (100) using a t test. A three-way analysis of variance (2 × 2 × 3) with repeated measures on the intensity of the effort (30%, 60%, and 100%) and on the action type (DF and PF) was performed on the coactivation percentages, followed by a post hoc least significant differences (LSD) test. A Wilcoxon signed rank test was used to compare the slopes of the relationships between the percentage of coactivation and the corresponding developed torque, as the average of the three coactivation-to-torque ratios at 100%, 60%, and 30% in each participant, between DF and PF. To test the significance of the correlation coefficients of the coactivation (%)–torque (N.m) relationships, the Bravais–Pearson table was used. For all analyses, the level of significance was established at p < .05.

**RESULTS**

**Maximal Voluntary Torque**

Figure 1 shows that the elderly participants developed a maximal voluntary PF torque that was significantly inferior to that of young participants (p < .001): 79.6 ± 22.1 N.m versus 132.4 ± 19.1 N.m. In contrast, there was only a slight decreasing trend in DF with aging (42.0 ± 5.7 N.m vs 22.1 N.m; p = .15). However, CV were relatively high, showing values from 18.4% to 27.8% according to the group and the effort. These different muscular torque alterations resulted in an imbalance in the DF-to-PF maximal torque ratio developed at the ankle joint; this ratio had a value of 0.35 in young and 0.57 in elderly adults (p < .05).

**Central, Nervous Level**

**Activation level.**—In PF, contrary to DF, young and elderly men were not able to fully activate their agonist muscles, their maximal voluntary activation levels being significantly different from 100%. Older men were able to activate their muscles to the same extent as younger men, as statistical analysis did not reveal any significant difference in the maximal level of activation between younger and older men for a DF effort (100 ± 0% and 99.05 ± 2.99%) or for a PF effort (94.69 ± 6.39% and 91.04 ± 12.46%; p = .15) (Figure 2). However, there was a tendency toward a lower level of activation in the older individuals during PF, as shown by the difference of nearly 5% between the two groups. Moreover, a greater variability in the elderly men could mask a difference; indeed, the CV was 7.2% in the younger men and 13.8% in the older men.

**EMG activity.**—For a maximal contraction, there was no significant difference between young and old adults in RMS/M_max of the agonist muscle, either during a PF or a DF effort.

**Coactivation.**—Statistical analysis showed that there were significant interactions between age and intensity of the effort (p < .001) and age and action type (p < .01). Regardless of the intensity of the effort (30%, 60%, 100%), young adults had a greater level of coactivation than did elderly adults in PF (p < .001), but this was not the case in DF where there was no significant difference between the
two age groups. In the young participants, the percentages of coactivation were the same, whatever the ankle action type (DF or PF), but this was no longer true with aging, where the coactivation during PF was lower compared to coactivation during DF. Regardless of the action, there is an intensity effect: the greater the developed torque, the greater the level of coactivation. Besides, there was a linear relationship between the coactivation percentages and the developed submaximal (30% and 60%) and maximal torques (correlation coefficients: \( r = .83 \) in PF and \( r = .66 \) in DF, \( p < .01 \)) (Figure 3). Coactivation-to-torque ratios of DF and PF were significantly different (\( p < .001 \)); thus, the relationship slope in DF was higher than that in PF.

Peripheral Level

We noticed that the amplitude of the mechanical twitch at rest was significantly smaller (\( p < .001 \)) and that the kinetics (TPT and HRT) were significantly slower in PF with aging (\( p < .001 \)) (Table 1). However, in DF, only the HRT was significantly longer (\( p < .05 \)). No significant difference was established between the older and younger participants when maximal DF torques and PF torques were normalized by the amplitude of the agonist muscle’s mechanical twitch at rest. For the triceps surae muscle only, there was a decrease in the M-wave amplitude (18.84 ± 6.29 mV vs 28.66 ± 7.93 mV, \( p < .05 \)) and an increase in the M-wave duration (3.25 ± 0.58 ms vs 2.77 ± 0.48 ms, \( p < .05 \)) in elderly participants.

### Maximal Voluntary Torques

Elderly participants developed a maximal voluntary PF torque 40% inferior to that of the young participants: muscular strength is thus affected by aging. Davies and colleagues (57) had obtained similar results, finding a loss of 38% in men (70 years old) for torque exerted by PF muscles. Our results were congruent with studies on elderly adults (from 72 to 84 years) over a 12-year period (10) and on people aged 20–100 years (1): Authors found a clear decline in PF of –30% and –45%, respectively.

Surprisingly, for DF, there was a nonsignificant decrease between the two age groups (–7%). This slight decline in DF was nevertheless in accordance with results from other studies (5,10). Conversely, other authors (11) noted that the older (72-year-old) group produced an average of 21% less isometric torque in DF compared with the young (26-year-old) group; Vandervoort and McComas (1) reported findings of ~45%. It is possible that this biggest decline was due to the fact that the participants in this study were very old. Our participants (mean age 77 years) might be entering a phase in which they will lose more and more strength. Moreover, it has been found that ankle DF forces are much lower in participants reporting falls (58,59). This parameter seemed to be a good predictor of fall status. In our study, the elderly men had not undergone a fall during the previous 2 years; this could partly explain our slight decline in DF torque. The performance of triceps surae and tibialis anterior, involved in locomotion and important in the prevention of falls in elderly persons, were not altered by age in the same way. As a consequence, these different parameters in PF and DF were not altered by aging.

### DISCUSSION

The purpose of this study was to evaluate muscular performance at the ankle joint in young and elderly males. The main findings are: a) a decrease in the maximal PF torques with age, whereas the DF torques remain similar and, as a consequence, an imbalance in the ratio of maximal developed torques at the ankle joint; b) a high and similar level of activation of agonist muscles in young and elderly males in PF and DF; c) a linear relationship between the percentage of coactivation and the developed torque; and d) an alteration occurring at the peripheral level of PF muscles.

#### Table 1. Parameters of the Mechanical Twitch in Young and Old Adults (Mean ± SD)

<table>
<thead>
<tr>
<th>Action</th>
<th>Group</th>
<th>Pt (N.m)</th>
<th>TPT (ms)</th>
<th>HRT (ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>PF</td>
<td>Young</td>
<td>16.37 ± 2.73</td>
<td>105.20 ± 10.86</td>
<td>86.04 ± 10.73</td>
</tr>
<tr>
<td></td>
<td>Elderly</td>
<td>10.89 ± 3.43*</td>
<td>134.14 ± 24.09*</td>
<td>116.46 ± 23.77*</td>
</tr>
<tr>
<td>DF</td>
<td>Young</td>
<td>5.7 ± 1.41</td>
<td>96.09 ± 17.74</td>
<td>76.42 ± 23.32</td>
</tr>
<tr>
<td></td>
<td>Elderly</td>
<td>2.72 ± 1.59</td>
<td>102.28 ± 15.45</td>
<td>93.57 ± 10.39</td>
</tr>
</tbody>
</table>

Notes: PF = plantarflexion; DF = dorsiflexion; Pt = peak twitch; TPT = time-to-peak torque; HRT = half-relaxation time.

*Significant difference (\( p < .05 \)) between the values of the two age groups.

Figure 3. Relationships between coactivation percentages and torques developed in PF and DF in young and elderly males. To establish the relationship between the coactivation and the corresponding torque developed, data from all subjects were used, with the percentage of coactivation on the y-axis plotted against maximal and submaximal voluntary torques (100%, 60%, and 30%). □, Individual values for the 11 young participants; ■, individual values for the 12 elderly participants. A linear relationship (\( p < .01 \)) was found between the two variables in PF (A) and in DF (B).
alterations in the performance of muscular groups entailed an imbalance in the maximal DF-to-PF torque ratio, which had a value of 0.35 for the young and 0.57 for the elderly adults. This deficit of torque in plantarflexor antigravity muscles might explain a standing posture rather tilted backward (60) in some elderly people.

Supporting our hypothesis, dorsiflexor and plantarflexor muscles are differently altered by aging. It remains to be determined whether this difference was due to alterations in the pattern of muscle use or to intrinsic changes in muscle properties.

Motor Performance Alteration at the Nervous Level

Muscular activation is one neural factor that is critical to maximal torque production. A reduced ability in muscle activation could be a result of a lowered motor unit discharge rate or a relatively larger number of motor units that cannot be voluntarily activated in the muscle (22,23,27,61). The twitch interpolation technique is commonly used to test the ability to achieve complete activation of the muscles and, therefore, provides information concerning central motor drive (62). However, activation is a global parameter that does not give direct information on motor unit discharge rates and recruitment. Furthermore, this activation can be maximal even if the motor unit firing rate is reduced in the tibialis anterior muscle of elderly adults (12). The superimposition of evoked stimulation on a voluntary contraction is purported to activate those muscle fibers not voluntarily activated by the central nervous system. However, the precision of this technique has been questioned (63). The suggestion that tetanic stimulation during the twitch interpolation technique is a more sensitive tool is supported by studies (61,62) indicating that multiple evoked stimuli improve the signal-to-noise ratio. Single stimuli are much more tolerable to the participant than are trains of stimuli, which makes the use of superimposed single or double stimuli attractive. Twitch interpolation technique is based on an assumption that the evoked force decreases linearly, whereas the voluntary force increases; this relationship is in fact nonlinear (22), and the activation could be substantially lower than that estimated by this technique. The consistent finding that muscular activation does not reach a full 100% might be partially explained by limitations of the twitch interpolation technique rather than by physiological impairment. One might speculate that part of the failure to observe “full activation” is due to the stimulation procedure (23).

In the present study, maximal activation level did not seem to alter with aging, either for maximal DF effort or for maximal PF effort—that is, say, 99% and 90%, respectively—in elderly people. It appeared that the participants, whatever their age, were not always able to reach an activation level of 100% in PF. This finding of incomplete activation was in accordance with previous studies on plantarflexor muscles in young (49,64) and in elderly participants (62). Nevertheless, we could note that there was a slight decreasing trend in PF with aging. During maximal effort, complete activation of motor units was easily reached for the tibialis anterior, and with difficulty for the triceps surae, whatever the age of the participant, confirming results reported in young participants (65). In our study, we found that activation, not modified with aging despite a slight decreasing trend, showed large interindividual variability in PF in elderly people, as had already been observed (62). However, this finding was recently discussed by authors who found a reduction in activation capacity in the plantarflexors of elderly males (24). It might be that our elderly group had entered a phase of life in which PF activation begins to decrease. The EMG (RMS) normalized by the amplitude of the corresponding Mmax response, which did not seem affected by age in DF or PF, confirmed that the neural drive seemed not to be altered. Normalization of DF and PF maximal torques by the amplitude of electroinduced mechanical twitch at rest allowed us to rule out a possible intrinsic modification in the muscle. Under these conditions, there was no longer a significant difference between the two groups. Motor performance alteration was not primarily attributable to a change at the level of the nervous command of the agonist muscles, but indeed to modifications at the peripheral level and alterations of the command of the antagonist muscles. Indeed, torques developed at a joint depend on the interaction of multiple physiological, anatomic, and biomechanical factors, and on relative mechanical levels of agonist and antagonist muscles (66). In fact, as plantarflexor and dorsiflexor muscles did not behave in the same way with aging, we could suppose that when they acted as antagonist, the evolution of their coactivation would be different.

During MVC in PF, the tibialis anterior of the elderly participants showed a coactivation 48% lower than that of the young adults (7.88% vs 15.23%); this difference approximately corresponded to that observed in maximal voluntary muscular torque (40%) between the two age groups. The hypothesis that an increase in muscular coactivation could contribute to the decline in motor performance did not seem pertinent. It was interesting to observe that coactivation in PF in elderly participants during a maximal effort at 100% corresponded to that in the young adults during a submaximal effort at about 60%. In fact, PF torques in young men at 60% (77.81 ± 10.6 N.m) were not significantly different from maximal voluntary torques in elderly men (79.58 ± 22.10 N.m). The coactivation seems to be related to the intensity of the effort. In isometric contractions, coactivation of the antagonist may prevent extreme unilateral forces which could otherwise damage the joint. A study found that an increase in knee extension load was accompanied by an increase in RMS EMG activity of the coactive biceps femoris (67). In the present study, as young men developed higher maximal voluntary PF torques than did elderly men, their coactivation was greater. As the PF MVC torque is lower, the compensatory force contraction required, i.e., the torque developed by the coactive tibialis anterior, is also lower. These findings are thus the consequence of the positive relationship between coactivation and torque developed.

A greater level of coactivation with torque does not seem to be due to the greater cross-talk with increasing torque. In surface EMG recordings, the M-wave of the antagonist muscle is not observed during percutaneous stimulation of the agonist motor nerve. We could conclude that there is
negligible signal contamination (9% in our study) arising from adjacent muscles by virtue of their close proximity (67). The effect of cross-talk can be disregarded in most muscles of the extremities, spine, and upper trunk when, as recommended by Solomonow and colleagues (56), surface EMG is recorded using the appropriate size of electrodes, correct placement over the muscle, and short interelectrode distance.

In DF, in contrast, the coactivation of the triceps surae muscles remained stable with age, which was coherent with the stable level of maximal resultant torque and activation level. Thus, older men showed similar muscular coactivation when compared to younger males during PF and DF efforts, and, contrary to what we had expected, the decline in torque could not be linked to an increase of this coactivation.

Notably, a determining criterion in this coactivation was the level of developed torque. We have established a significant linear relationship between the percentage of coactivation and the developed resultant torque for all the participants, whatever their ages, and for each of the situations of DF and PF. For a given increase in torque, the coactivation was less in PF than in DF. Coactivation thus seemed to be linked to the absolute value from muscular torque and not to the relative value expressed according to maximal torque.

In brief, maximal muscular performance was different between the two studied populations in PF, but the origin of this phenomenon seemed to be linked neither to coactivation nor to levels of activation. The difference might therefore be due to alterations at the peripheral level.

**Motor Performance Alteration at the Peripheral Level**

The evoked twitch technique, shunting the central command, allowed us to examine intrinsic muscle events. Amplitudes of mechanical twitch and M$_{max}$ response at rest in elderly people were lower in PF, proving that changes intervened in the triceps surae muscles themselves, and thus explaining the decline in muscular torque in PF. This performance degradation could be explained by a slowing down of excitation and contraction processes (68), a modification in terms of myotypology (18), and an alteration of the stiffness (7). In DF, no significant difference between young and old men was found in the values of twitch amplitude of the tibialis anterior muscle, as had been observed in other studies (12,34,35).

In our study, there was a prolongation of the TPT and HRT for the plantarflexor muscles, which was coherent with the evolution of fiber typology in elderly people’s muscles (69). In DF, we observed an age-related slowing of relaxation only. Slowed contractile speeds in older adults during electrically stimulated isometric contractions had earlier been observed in the dorsiflexor muscles (5,26,70). This age-related slowing of contraction speed might be due to a number of changes in muscle morphology and muscle function with age, including a selective loss of type II muscle fiber area (19) and an increased proportion of type I fibers (71), especially for the gastrocnemii. Moreover, M-wave duration turned out to be higher in elderly people, the sign of a slower conduction velocity. In short, we noted modifications of the intrinsic properties of the plantarflexor muscles with a slowing down of the excitation–contraction coupling mechanisms. In contrast, the age effect in DF seemed slower in coming and at the moment, alterations most likely existed only at the intracellular level. Muscular torque decline was therefore linked to the atrophy phenomenon, to intrinsic muscle modifications; according to these results, it seemed that nervous factors might begin to intervene with aging.

**Conclusion**

The use of plantarflexor and dorsiflexor muscles, and thus the maximal torques to which they contribute, were not altered in the same way by aging, supporting our hypothesis. The aging of the plantarflexor muscles seemed to be of earlier onset compared with that of the dorsiflexor muscles, potentially leading to modifications in the standing posture. The decrease in the coactivation of the tibialis anterior during PF maximal effort would seem to compensate for this decrease in maximal torque, without, however, proving sufficient: The balance of DF-to-PF maximal torques ratio thus became very different with age. An original finding of the present study was the linear relationship between the percentage of coactivation and the developed torques. It would now be interesting to understand how the central nervous system controls agonist and antagonist activities of the muscular groups at the ankle to avoid falls. Do some elderly individuals adopt a standing posture which is rather tilted backward because of the decline in the performance of plantarflexor muscles, or is this particular position responsible for the deterioration of these muscles? Further investigations are also needed to determine if training could preserve the performance of triceps surae muscles in aging populations.

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