Gait Pattern Classification of Healthy Elderly Men Based on Biomechanical Data

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Objectives: To distinguish the gait patterns of young subjects from those of elderly men using three-dimensional (3D) gait data, to determine if elderly subjects displayed other than a typical gait pattern, and to identify which parameters best describe them.

Design: Nonrandomized study in which video and force plate data were collected at the subject's own free walking speed and used in a 3D inverse dynamic model. Cluster analysis was chosen to identify the gait families, and analyses of variance were performed to determine which parameters were different.

Setting: A gait laboratory.

Participants: The sample of convenience involved a single but mixed group consisting of 16 able-bodied elderly subjects (mean age, 62yrs) and 16 able-bodied young subjects aged between 20 and 35 years.

Main Outcome Measures: Phasic and temporal gait parameters, as well as the 3D muscle powers developed in the joints of the right lower limb during the gait cycle.

Results: The walking patterns in elderly subjects were found to be different from those of the young adults. Three elderly gait families or groups forming a specific gait pattern were identified, and differences were found in the phasic and temporal parameters as well as in 6 peak muscle powers. Four of the peak powers occurred in the sagittal plane, and half of them were related to the hip.

Conclusions: Biomechanical parameters can be used to classify the gait patterns of young and elderly men using cluster analysis rather than age alone. The muscle powers in elderly subjects are perturbed throughout the gait cycle and not only at push-off. It appears that the plane in which the peak powers occurred was related to their occurrence in the gait cycle.

Variability in the gait patterns of elderly subjects could reflect natural adaptations or compensations. These should not be indicative of a deficient gait or be misconstrued as some age-related pathology.

Key Words: Gait families; Healthy elderly men; Three-dimensional analysis; Kinetic parameters; Rehabilitation.

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MAINTAINING WALKING abilities is important to elderly people, because it is instrumental in activities of daily living and required in many tasks for independent living. Because locomotion is recognized as a risk factor associated with falls, gait patterns in elderly, able-bodied subjects have been documented to establish relationships with walking speeds, to compare them with those obtained from young adults or with those of known fallers. The recruitment strategy in all these studies and in many others was to divide the population based on age alone, usually above 60 years.

Documented changes in some gait parameters, such as shorter stride length, reduced walking speed, and lower ankle push-off muscle power, may be more indicative of gait adaptations selected by elderly men rather than the results of age specific impairments. Grouping populations by age has the inconvenience of masking these gait-related adaptations attributed to aging. We hypothesize that the walking patterns in elderly subjects are different from those of the young adults, and that they can be distinguished according to the biomechanical gait parameters of each individual rather than using age as a grouping factor.

An activity such as walking can be an overall result of several movement parameters, which can vary within the gait of the individual, while the activity itself can be fairly representative of the person's performance. Classifying gait patterns has the advantage of taking into account several parameters at the same time rather than a single one for each individual.

Using peak muscle powers developed at the hip, knee, and ankle, Vardaxis and colleagues identified 5 gait families in 19 young adults using cluster analysis. A gait family was formed by subjects that displayed a strong affinity based on several parameters obtained from each individual gait trial and that were significantly different from the other clusters of subjects having their own gait similarities. The subjects in the first family displayed a strong hip pull and ankle push to propel themselves forward. For families 2 to 5, forward progression was ensured by an increasing action of the sagittal hip power shortly after heel-strike. These results highlight the multiple normal dynamic strategies selected by able-bodied subjects in walking. We further speculate that gait patterns for elderly subjects differ even within that age category.

Muscle power that is the product of the net muscle moment and the joint angular velocity has been recognized as a valuable gait descriptor, because it combines both kinematic and kinetic information. It is widely used to characterize able-bodied gait, cerebral palsy locomotion, and the gait of subjects with various foot prostheses or total hip implants. In
healthy elderly gait, phasic and temporal-distance parameters must also be considered, because walking speed has been found to influence the mechanical work developed in the lower limb, and reduced walking speed has been reported in this population.

Most gait studies involving elderly subjects have been performed using a control group consisting of healthy young adults. These studies have examined one gait parameter (eg, locomotion speed, stance time, joint peak muscle power) at a time; to our knowledge, no studies have grouped several parameters together to describe the global locomotion pattern, as has been done for healthy young adults. Patients with an anterior cruciate ligament deficiency, or patients with hemiplegia, while muscle powers have been used to characterize gait in elderly subjects, these studies either were limited to a planar analysis or, when a three-dimensional (3D) analysis was reported, only sagittal plane information was given. The primary objectives of this study were twofold. Using a single but mixed group of able-bodied young and elderly men, we hoped (1) to distinguish the gait patterns of these populations based on phasic and temporal parameters as well as 3D peak muscle powers rather than age alone, and (2) to determine whether elderly subjects displayed different gait patterns. We also hoped to identify from the parameters determined above those that best describe elderly gaits.

METHODS

A single mixed group of 16 able-bodied young and 16 elderly men (32 subjects) participated in the study, which was reviewed and approved by the local institutional review ethics board. Demographic information is given in table 1. The subjects were later classified into different groups according to their gait characteristics.

The 16 young male volunteers who participated had an average age of 28.0 years, and none were over 35 years old. None had a limb length discrepancy larger than 1.5 cm, and all were in good health. The exclusion criteria were musculoskeletal ailments, scoliosis, joint replacement, recent surgery, use of medication, or a history of neurologic, pulmonary, cardiac, or locomotor disorders. Half of the young men participated in some sport-related activities, but they trained not on a regular basis, while the others were essentially sedentary.

Sixteen healthy elderly men also took part. Their mean age was 61.7 years. The youngest subject was 56 years old; the eldest was 71 years. The inclusion criteria were the same as for the young men, but with the following additional requirements: the subjects had to be capable of independent walking and be over 55 years old (due to recruitment difficulties). The exclusion criteria for this group added the following conditions to those of the young-men group: vertigo, stroke or other heart problems, diabetes, eyesight problems, and pulmonary diseases. Seven of the elderly subjects were sedentary and 5 were active but did not participate in any organized sport-related activities, while the remaining 4 were active in occasional sport activities (tennis). This demographic information concerning both the healthy young adults and the elderly subjects was not included in the subsequent analysis.

The gait analyses were performed in the Human Movement Laboratory of the Service d'Exploration Neurophysiologique at the Centre Hospitalier Regional Universitaire of Lille. The kinematic data were collected by a Vicon 360 system. Two 50 Hz cameras were located at 3.5 m from the center of a 10 m walkway to cover the volume required for at least one complete gait cycle of the right lower limb. A third camera was placed in front of the subject and aligned along the axis of progression. Reflective markers were located on the subject before the gait analysis according the Vicon Clinical Manager protocol. An AMTI force plate (250 Hz) located in the middle of the walkway was synchronized to the cameras of the Vicon system.

Before data collection, the cameras were calibrated using 20 markers located within a calibration volume of 2.8 m long by 1.0 m wide and 1.80 m high. The subject was asked to walk at his natural speed and step with his right leg on the force plate before coming to a full stop 5 m farther on. A few practice trials were permitted before the gait data were collected. The subject performed at least 5 successful trials for which both kinematic and kinetic informations were available. Because there were some variations in the walking speed of the elderly subjects, two trials were selected for further analysis. Including the trials of two populations (young and elderly), there was a total of 64 gait trials.

The 3D coordinates were calculated and then filtered using the routines provided in the Vicon system software. The phasic and temporal gait parameters were obtained from video and force-plate data. Speed, stride length, cadence, and stance phase relative duration were selected because these parameters were reported to be modified in the gait of elderly men. The 3D joint reaction forces and net muscle moments were calculated by the Vicon Clinical Manager software. All the kinematic and kinetic data were normalized with respect to 100% of the gait cycle (GC) duration defined by 2 consecutive heel strikes of the right lower limb.

The 3D muscle powers were estimated by the product of the net muscle moment developed at each joint of the lower limb and the corresponding angular velocity calculated about each joint axis. When the muscle moments and the corresponding angular velocities have the same polarity, the muscle power is positive, and it is assumed that energy is being expended during a concentric contraction. When the polarities are different, then the muscle power is negative, and it is assumed that energy is being absorbed in an eccentric contraction. The muscle powers were normalized with respect to the

Table 1: Mean Values of the Demographic Information and of the Phasic and Temporal Gait Parameters of the Young Able-bodied Subjects and Healthy Elderly Men

<table>
<thead>
<tr>
<th></th>
<th>Young Subjects (Y1)</th>
<th>Elderly Subjects (all)</th>
<th>Family E1</th>
<th>Family E2</th>
<th>Family E3</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Number of subjects</strong></td>
<td>16</td>
<td>16</td>
<td>16</td>
<td>16</td>
<td>16</td>
</tr>
<tr>
<td><strong>Number of trials</strong></td>
<td>32</td>
<td>32</td>
<td>10</td>
<td>6</td>
<td>16</td>
</tr>
<tr>
<td><strong>Age (yrs)</strong></td>
<td>28.04±8</td>
<td>61.73±9</td>
<td>61.14±2</td>
<td>64.33±4</td>
<td>61.12±1</td>
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<tr>
<td></td>
<td>(3.60)</td>
<td>(3.81)</td>
<td>(5.21)</td>
<td>(1.88)</td>
<td>(3.03)</td>
</tr>
<tr>
<td><strong>Height (m)</strong></td>
<td>1.81±6</td>
<td>1.74±4</td>
<td>1.70±1</td>
<td>1.72±1</td>
<td>1.75±1</td>
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<tr>
<td></td>
<td>(0.045)</td>
<td>(0.046)</td>
<td>(0.019)</td>
<td>(0.067)</td>
<td>(0.038)</td>
</tr>
<tr>
<td><strong>Weight (kg)</strong></td>
<td>80.22</td>
<td>76.41</td>
<td>76.42</td>
<td>76.17</td>
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<td></td>
<td>(6.25)</td>
<td>(8.36)</td>
<td>(11.13)</td>
<td>(3.17)</td>
<td>(7.92)</td>
</tr>
<tr>
<td><strong>Speed (m/sec)</strong></td>
<td>1.348±8</td>
<td>1.22±2</td>
<td>1.01±7</td>
<td>1.26±2</td>
<td>1.33±2</td>
</tr>
<tr>
<td></td>
<td>(0.097)</td>
<td>(0.189)</td>
<td>(0.100)</td>
<td>(0.096)</td>
<td>(0.132)</td>
</tr>
<tr>
<td><strong>Cadence (step/min)</strong></td>
<td>110.70±1</td>
<td>112.92±1</td>
<td>110.41±1</td>
<td>103.74±1</td>
<td>117.93±6</td>
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<tr>
<td></td>
<td>(1.83)</td>
<td>(5.87)</td>
<td>(1.33)</td>
<td>(1.77)</td>
<td>(2.09)</td>
</tr>
<tr>
<td><strong>Stride (m)</strong></td>
<td>1.45±8</td>
<td>1.34±4</td>
<td>1.21±4</td>
<td>1.45±3</td>
<td>1.36±4</td>
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<tr>
<td></td>
<td>(0.098)</td>
<td>(0.131)</td>
<td>(0.037)</td>
<td>(0.054)</td>
<td>(0.133)</td>
</tr>
<tr>
<td><strong>Stance (m)</strong></td>
<td>59.00±8</td>
<td>63.03±8</td>
<td>64.16±8</td>
<td>62.43±8</td>
<td>62.65±8</td>
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<tr>
<td></td>
<td>(0.87)</td>
<td>(1.84)</td>
<td>(0.90)</td>
<td>(1.65)</td>
<td>(2.09)</td>
</tr>
</tbody>
</table>

Standard deviation in parentheses.

* p < .05, between Y1 and all elderly subjects; t p < .05, between Y1 and E1, E2, and E3; 7 p < .05, between Y1 and E1 and E2, and E3; 1 p < .05, between E1 and Y1, E2 and E3; between Y1 and E2; between E2 and E3.
individual body mass and labeled according to Eng and Winter. For example, K3F corresponds to the third peak knee power developed in the frontal plane. In all, 27 3D peak power values were identified for each subject and for each trial.

The principal objectives of this study were to determine if gait data are sufficient to distinguish the patterns observed in the young able-bodied subjects from those obtained in elderly healthy men, and if there was more than a single pattern in the elderly gait, although neither the number nor the members of the family groups were known. To answer these objectives, a cluster analysis was performed using gait parameters only. Cluster analysis was successfully used by Vardaxis and coworkers to classify gait patterns in healthy young adults and by Winters and colleagues to categorize the diagnosis of patients with shoulder complaints.

A gait trial was defined by 31 discrete parameters consisting of the 4 phasic and temporal parameters and the 27 3D peak power values. The hierarchical cluster method proposed by Ward assigns the gait trials to clusters (families) in a stepwise fashion. It begins with as many clusters as there are trials, 64 in this study. Afterward, these individual clusters are joined together to form new clusters, and so on. The procedure ends by grouping all the trials in a single cluster, which forms a hierarchical tree called a dendrogram. The trials are first joined into clusters according to the degree of similarity among the 31 gait parameters of every trial. This is shown in figure 1, in which trials are grouped by pairs at different linkage distances. Though there are other methods available, the measurement of similarity was based on the Euclidean distance between the trials, which is the most commonly chosen type of distance calculation.

Clusters having different characteristics are distanced from each other. This distance was determined by the complete linkage method and is calculated using the greatest distance between any 2 trials in the different clusters. This method performs well in cases when the trials naturally form distinct families, as was expected in this study. In figure 1, trials 1 and 2 (right side) are much different from trials 59 and 60 (left side), in terms of the effect of the combined 31 gait parameters. The optimal number of clusters can be determined in several ways. Winters and coworkers calculated the criterion of Hartigan and the maximum value of the cubic clustering criterion, while Vardaxis and colleagues used the R ratio, which is a measure of the reduction of the inner cluster variability. The R ratio was applied in this study to determine the number of meaningful gait families.

Once the number of optimal clusters or gait families was identified, univariate analysis of variance tests were performed on each of the demographic parameters, which were not considered in the cluster analysis and on the 31 dependent variables to determine which parameters best describe the gait patterns of the young and elderly subjects. This was followed by a post-hoc Tukey test for unequal N multiple comparison tests. The statistically significant differences found were used to describe the patterns of the young and elderly gait families.

RESULTS

The ascending hierarchical cluster method formed a tree-like structure (fig 1) outlining 4 main branches from bottom to top. The first division occurred at a Euclidean distance of 7.3 where 32 trials (33 to 64) formed a cluster. This first division separates those trials, which were much different from the single group. They represent a family of 16 young subjects, Y. This group could be subdivided into 3 groups, but the divisions within the family Y occurred too closely together to be able to distinguish them using the R ratio. Both trials of the same individual were grouped together, with the exception of 3 subjects (trials 49-50, 51-52, 53-54). Vardaxis and colleagues reported a similar observation in their group of able-bodied subjects.

The second and third divisions occurred at a Euclidean
Fig 2. Ankle, knee, and hip muscle power curves developed in the (A) sagittal, (B) frontal, and (C) transverse planes by the Y1, E1, E2, and E3 families, over which the standard deviation of the Y1 family is overlaid in bold.

distance of 8.7 and 13.7, respectively. They form one family of 10 trials, E1, and a smaller one, E2, of 6, all consisting of elderly men. Though all the trials of the E1 cluster were paired according to the subjects, this was not always observed in the E2 group. This reflects a variable pattern between the 2 trials of the same individual. The fourth family of 16 gait trials was made up exclusively of the 8 elderly subjects, E3. All but 2 subjects did not have their trials grouped together (15-16, 21-22).

The phasic and temporal gait parameters for the young and elderly groups and each family are given in Table 1. The data for the group of young able-bodied subjects fall within previously reported values. Only the significant differences, p < .05, were reported. Generally, elderly subjects were older (F = 401.74) and slightly shorter (4.6%; F = 15.82) than the young men. Their walking speed was about 9.3% (F = 27.69) slower due to a decrease in the stride length (F = 16.41). The stance phase's relative duration was approximately 5% longer (F = 31.83) in the healthy elderly men. The variability expressed by the standard deviation was always larger in the elderly group than in the young men. Similar observations have previously been reported elsewhere.

The mean 3D muscle power curves calculated for each joint of the healthy young men with their respective standard deviation are given in figure 2. The mean power curves for the 3 families of elderly subjects are also presented with the 27 peak powers used in this study. Sagittal plane data were within the values previously reported for able-bodied young and elderly subjects. Of 27 3D peak muscle powers, only 8 were found to display significant differences between families. Though it was shown that limb length influenced muscle powers, normalizing the peak power did not modify our findings. These findings are presented in figures 3 and 4. There are some slight discrepancies between the mean muscle power curves (fig 2) and the mean peak powers (figs 3, 4). This stems from the fact that the individual peak values do not necessarily occur exactly at the same time in each individual trial and for each individual, but rather within 1% to 3% time range.
Thus, the values reported in figures 3 and 4 represent the peak power bursts taken from the individual trials, and not from the mean curve. Half of the statistically significant peak muscle powers were related to the hip. Three peak powers were associated with the frontal plane and occurred at heel-strike (20% GC). In the transverse plane, the peak power (H3T) manifested itself in the midstance period (40% GC). The remaining 4 peak powers were related to the sagittal plane and happened between the push-off and the terminal swing periods.

Three gait patterns were found in the healthy elderly subjects. The first group of elderly subjects, E1, had the slowest walking speed (1.017 m/sec), which was significantly different from all the other groups. This was attributed to a 16.7% decrease in stride length and a small but not significant decrease in cadence. This family had the most significant peak power differences with the other groups. This group of elderly subjects was the first to cluster itself with the healthy young men and displayed 7 of 8 significant differences in the peak muscle powers as compared with the other families. The E1 family usually had the lowest peak muscle power values. These lower values could be the result of the slower walking speed, as reported by Chen and colleagues, who noted that speed influenced the sagittal plane hip and ankle muscle powers.

The second family of elderly men, E2, had a slow walking speed (1.260 m/sec), but was not statistically different from that of the young men, due to a 6% reduction in cadence coupled with a slightly smaller stride length. The cluster analysis was able to differentiate this family from the others because of the temporal and phasic parameters, as well as for 3 peak powers higher than those of the E1 family.

The E3 was characterized by a walking speed similar to the Y1 group. This was mainly achieved by a significant increase in cadence (6.5%) and a corresponding but not significant decrease in stride length. The E3 subjects usually had muscle power values close to those of the Y1 subjects. These differences were not found to be significant with the exception of the H2F peak power. However, the E3 family displayed 6 higher peak powers than those of the E1 family. In the dendrogram, this group was located at the opposite end from the young men’s group and E1 family, and was the last to join the other clusters.

**DISCUSSION**

The aim of this work was to distinguish the gait patterns of 16 able-bodied young and 16 elderly men from a single group of 32 subjects. Phasic and temporal parameters and 3D peak muscle powers were used in the classification process, rather than age alone. The hierarchical cluster method proposed by Ward was used successfully for this purpose. The young subjects were the first to form a cluster and dissociate themselves from the 3 elderly subjects’ gait trials (fig 1).

The cluster analysis method is sensitive to variations in the gait trials, especially where reproducibility may be difficult to attain, particularly in the elderly subjects. The trials of a new subject may not be grouped together, but rather may be located anywhere in the dendrogram, and at the same time, adding new subjects can perturb the tree architecture, making it totally useless. It is also possible that the trials of a young subject may be located in one of the elderly subject’s clusters. Inversely, the trials of an elderly individual could end up within the cluster of the young, able-bodied subjects. This study was susceptible to this type of perturbation, even though our elderly subjects were relatively young and physically fit. Two additional subjects, one in each group, were not included in this study, because they were unable to develop reproducible and consistent gait data between trials. This was mostly observed in the temporal gait parameters and in the sagittal peak muscle powers. The lack of reproducibility between these individual gait data was deter-
mined by the cluster analysis. This was based on the Euclidean distances between these subjects' own two trials. These trials were located far from each other. Including those 2 subjects with the present 32 led to a confused dendrogram in which several young and elderly subjects were grouped indiscriminately.

The second purpose of this study was to determine if elderly subjects displayed different gait patterns and to identify among these gait parameters those that best describe the gait of the young and elderly subjects. It was the combined effect of the 27 peak muscle powers and 4 phasic and temporal parameters that enabled the classification of the elderly gait families and allowed us to distinguish them from the young men. Three families of elderly men were identified by means of the R ratios, following cluster analysis.

The phasic and temporal gait parameters were recognized as important in this study as well as in others. The variability expressed by the standard deviation in the temporal and phasic parameters was always larger in the elderly group than in the group of young men. This could emphasize the possibilities of different families of elderly gaits. Taken alone, these parameters can be misleading due to their interdependence. This explains, in part, the conflicting results reported in the literature on the walking patterns of the elderly. One of the major findings of this study was to explain the reported contradictory observations related to the phasic and temporal gait parameters according to the 3 families of elderly men.

The E1 family had the slowest speed because of a reduction in stride length. In fact, our whole group of 16 elderly subjects can be classified as having slow speed due to reduced stride length. Judge and Kerrigan and colleagues previously described elderly gait as such. Nonetheless, others have attributed reduced walking speed to a decrease in cadence. The cluster analysis revealed that our E2 family displayed this characteristic. Finally, the E3 family developed the same walking speed as the young men by increasing their cadence. Blanken and Hageman found a 6% increase in walking speed in a group of 12 elderly men aged between 60 and 74 years who had a short stride length. From these observations, one can only conclude that the cadence of the elderly men was much higher than that of the young-men's group, as was the case in this study. Thus, we cannot generalize that elderly subjects aged between 60 and 70 years have a slower walking speed and a shorter stride length. We must associate the temporal and phasic gait parameters to their different able-bodied walking patterns.

In previous studies, muscle powers have been used to describe the dynamic walking pattern in healthy elderly men. Some of these performed a planar (2D) analysis, while others followed a 3D gait analysis protocol. Nevertheless, only sagittal plane muscle powers were reported. In this study, 4 of the 8 significant peak powers occurred in the sagittal plane, and half of them at the hip, which supports, in part, the previous studies. These peak powers were distributed throughout the whole gait cycle. It appears that the plane in which these statistically significant powers occurred was related to their occurrence in the gait cycle. Shortly after heel-strike, the frontal-plane peak powers were associated with the control needed for body-weight transfer. During midstance, a transverse-plane peak power occurred that stabilized the support during single-limb balance. From push-off to terminal swing, the sagittal-plane peak powers were required for progression and control in the forward direction.

Generally, the A1F, K2F, and H2F frontal peak powers reflect the energy generated between the 10% and 25% mark on the gait cycle. These 3 peak powers are respectively associated with a hip and knee abduction moment and an ankle eversion moment. They counterbalance the frontal-plane displacement of the center of mass toward the supporting limb. These peak powers have not been well documented in the literature.

In this study, at the hip, the H2F was higher for the elderly families, with the E3 group having the highest value (.585 ± .432W/kg). According to Eng and Winter and MacKinnon and Winter, H2F activity was attributed to the raising of the pelvis and trunk to its neutral position. We hypothesize that the elderly subjects were mostly concerned with achieving a stable base of support by controlling hip action. This was less marked in the slowest E1 group (.231 ± .202W/kg). The knee K2F values were less than that of the young adults, with the exception of the E2 group (.581 ± .565W/kg). The K2F peak power generation was associated with a knee abduction moment that acted with the pelvis H2F abduction. The highest K2F value could reflect a greater frontal stability in developing the largest stride length of the elderly groups. Only the E1 family displayed a higher absorption of the A1F (−.107 ± .109W/kg) due to an adduction ankle moment. The A1F power generation can be considered as a reaction to the knee abduction. It appeared that the subjects of E1 family were passively controlling the body-weight transfer, while the subjects in the E2 and E3 families were actively contributing by generating the A1F. The control subject also had a low A1F peak power absorption (−.005 ± .032W/kg).

During midstance, both young and elderly subjects developed a positive H3T peak power associated with an active thigh internal rotation moment that occurred late in midstance. The H3T was assumed to be related to body-weight transfer on the contralateral limb in preparation for the subsequent heel-strike by propelling the body forward. The E1 family had the highest, significantly different H3T (.320 ± .372W/kg). As a result, normal cadence could be maintained. The E3 family, which had the highest cadence, did not increase its H3T, because other power strategies were available.

In late midstance, the H2S muscle power absorption peak is responsible for the deceleration of the thigh extension. The H2S peak powers in the E1 (−.525 ± .519W/kg) and E2 (−.818 ± .278W/kg) families were less than that of the Y1 group. The subjects in the E3 family developed a higher (−1.288 ± .700W/kg), but not significantly different, H2S from that of the Y1 family (−1.01 ± .308W/kg). Only the H3S of the E3 family was statistically different from that of the E1 group. We postulate that a reduction in the H2S can be associated with a shorter stride length, resulting in a slower walking speed as observed in the E1 and E2 families. A
reduction in the H2S could also be attributed to trunk forward inclination as commonly observed in the elderly population. At push-off, the H3S activity was associated with propulsion in healthy young men. It has been shown that the hip-pulling action (H3S) contributes actively to push-off in the gait of elderly subjects. We speculate that the subjects of the E3 family developed the highest H3S muscle power (1.930 ± .811W/kg) to increase their cadence and maintain a walking speed similar to that of the young-men's group. Though lower H3S values were observed in other elderly subjects, there was a significant difference only between the E1 (.962 ± .598W/kg) family, which had the slowest walking speed, and the E3 group.

A lower ankle peak power has been observed in the gait of healthy elderly subjects during push-off. The ankle peak power (A2S) responsible for push-off has been found to explain more than 52% of the step-length variance. A comparable value (64%) was found in this study. Interestingly, the E3 family compensated for a reduction in the A2S peak power (2.810 ± .589W/kg) by an increase in the hip-pulling action (H3S). The members of the other 2 elderly families were not able to compensate for a low A2S, which resulted in a slower walking speed. This was more pronounced in the E1 family, in which the slower values were observed for the walking speed as well as for the A2S (1.869 ± 867W/kg) and the H3S peak powers. Though the E2 family developed a slightly higher A2S (2.914 ± .544W/kg) among the elderly families, it was insufficient to compensate for their low H3S value.

The last peak power that was observed to be lower in the gait of healthy elderly subjects was related to the knee (K4S). The K4S peak power controls the lower limb's forward projection in preparation for the following heel-strike. A higher knee generation at push-off (K3S) and a decrease in the knee absorption (K4S) has been reported during late swing in a group of elderly men. A reduction in the K4S, also observed in this study, may help increase the stride length, but at the cost of a higher horizontal speed of the foot, which causes stumbling. Though this could be applicable to all 3 elderly families, the E1 (−.573 ± .386W/kg) had the largest reduction (54%) compared with the young-men's group (−1.250 ± .432W/kg).

Considering the major involvement of the hip-muscle powers in the 3 families of the healthy elderly subjects, our findings support Kerrigan and coworkers' proposition that a stretching-exercise program would improve walking performance. They also proposed that reduced plantar flexion may not be representative of an impairment, but rather of a strategy to preserve balance during walking. We postulate that this observation could be extended to the muscle power patterns in the elderly. This may explain why the elderly subjects, particularly the E1 family, had as a gait strategy a slow walking speed, a short stride length, and a long stance phase duration. Low peak ankle power (A2S) and hip-pulling action (H3S) at push-off may lessen forward propulsion and result in a short stride length and a slow walking speed. Having low muscle powers at push-off may require less knee control in the limb's forward projection as shown by the K4S power burst, which can lead to a faster foot speed and an increase in the risk of falling.

In summary, the muscle powers in elderly subjects are perturbed throughout the gait cycle, and not only at push-off. These gait adaptations are important to recognize in assessing elderly subjects, because they could falsely be attributed to age-related pathologies, and not to age alone. This article highlights different healthy elderly gait families classified according to their own phasic and temporal parameters and muscle power characteristics. The E1 family was characterized by a slow walking speed due, in greater part, to a short stride length. These subjects needed more internal rotation (H3T), less hip extension deceleration (H2S) during midstance, and less knee absorption (K4S) at the end of the swing phase to compensate for their small stride length. The E2 family also had a slow walking speed, though it resulted from a decrease in cadence. Stride length was possibly maintained by a large abduction activity (K2F). Though the strongest push-off (A2S) was reported in the elderly families, it was insufficient to compensate for the weak hip-pulling action (H3S). The E3 family was able to maintain a walking speed similar to the young subjects by an exaggerated cadence. This led to a greater need for frontal hip stability (H2F) shortly after heel-strike and a high ankle push-off (A2S) combined with an active hip-pulling action (H3S). These families must not be confused or interpreted as part of some age-related pathology, but rather as natural adaptations or compensations in the diversified evolution of able-bodied gait.

The walking patterns in elderly subjects were found to be different from those of the young adults by means of hierarchical cluster analysis based on biomechanical parameters, rather than using age alone as a grouping factor. Four phasic and temporal gait parameters as well as 27 3D peak powers for each individual trial were used in the classification process, rather than taking a single gait parameter at a time to determine the 3 families of elderly subjects. Cluster analysis enabled group comparison of young men and elderly subjects, while allowing the characterization of individual performances. Significant differences between the 3 elderly families were found in the phasic and temporal gait parameters, as well as in 8 peak muscle powers. Five peak powers, namely H2S, and 4 frontal- and transverse-plane powers were identified for the first time due to the clustering of gait patterns, and partially complete the dynamic description of elderly gait. The muscle powers in elderly subjects were perturbed throughout the gait cycle, and not only at push-off. It appears that the plane in which they occurred was related their occurrence in the gait cycle.

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References

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