Age-Related Changes in the Initiation of Gait: Degradation of Central Mechanisms for Momentum Generation

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Objective: To investigate cross-sectionally age-related changes in the expression and biomechanical efficiency of the gait-initiation motor program.

Design: Case-control study.

Participants and Setting: Twenty healthy young research subjects and 20 healthy elderly subjects who volunteered from the community participated in this study at a university research laboratory.

Main Outcome Measures: Participants performed gait-initiation trials at three speeds from a starting position on a force platform while ground reaction force data, 3-D motion analysis data, and electromyographic data were collected. Measures included: latency of tibialis anterior (TA) activation and soleus (SOL) and gastrocnemius (GA) inhibition, magnitude of center of pressure (COP) displacement, magnitude of momentum generated, and final walking velocity.

Results: The expression of the central motor program governing gait initiation, as evidenced by the invariant timing between TA activation and SOL/GA inhibition, was seen in both the young and elderly populations, but the frequency was diminished in the latter group. The momentum-generating capacity of the COP shift mechanism was present but significantly diminished in the elderly population.

Conclusions: These findings suggest that the central nervous system uses stable, efficient mechanisms for dealing with the inherent instability of upright bipedalism and that the integrity of these mechanisms degrades with aging.

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The common act of taking a step involves complex interactions between neural and biomechanical factors that serve to move the body from a quasi-static state (quiet standing) to a dynamic state (walking). The initiation of gait is governed by a motor program.1-6 Stereotyped patterns of activity within an overall movement and invariant relative timing among the component parts of these patterns are common criteria used to determine whether a particular movement is governed by a motor program.1-3 In gait initiation, there is evidence of several patterns of invariance among the component parts of the movement, including a stereotyped pattern of muscle activity early in gait initiation, ie, inhibition of soleus (SOL) followed by activation of tibialis anterior (TA) in both the swing and stance limbs.3,5 A In addition, Crenna and Frigo1 found that during gait initiation the relative timing between SOL inhibition and TA activation is invariant.

The two biomechanical requirements for successful gait initiation are the generation of momentum (in the forward direction and in the direction of the stance limb) and the maintenance of balance. In bipedal animals, these requirements are usually in conflict, as the generation of significant amounts of momentum generally involves moving the center of mass (COM) beyond the base of support (as defined by the feet), thereby leading to instability. The gait-initiation motor program deals with this conflict as follows. Before any appreciable movement of the body's center of mass (COM) during gait initiation, the center of pressure (COP) under the feet moves backward and toward the foot that is to be lifted first (fig 1A).1,6,9-12 This shift of the COP posteriorly and toward the swing limb increases the components of the ground reaction force (GRF) anteriorly and toward the stance limb, respectively, over a period of time, thereby generating momentum in those directions (fig 1B). In addition, by increasing the components of the GRF through a COP shift rather than through movement of the COM, the gait-initiation motor program generates the initial momentum necessary for taking a step before the COM moves forward of the base of support (as defined by the feet). Crenna and Frigo7 found that this counter-intuitive movement of the COP is controlled, at least in part, by the sequence of muscle activities involved in the gait-initiation motor program. Thus, the central nervous system (CNS) uses stable, efficient mechanisms for dealing with the inherent instability of upright bipedalism during gait initiation.

Many falls in the elderly occur during transitions such as initiating and terminating gait and changing direction. For instance, many elderly persons fall when they are walking only short distances (implying that the transition periods are a major portion of the activity)13 and/or during activities when the COM is mildly shifted from the base of support.14 Given that gait initiation is a risky activity for the elderly, its mechanisms warrant study. Furthermore, problems with gait initiation can be exaggerated in pathologies associated with the elderly such as Parkinson disease, progressive supranuclear palsy, multi-infarct parkinsonism, normal-pressure hydrocephalus, andBinswanger disease.15-16 In addition, difficulty initiating gait sometimes occurs in the elderly as a more isolated syndrome, which has been termed "isolated gait failure."17-18
progressive freezing gait. Although the causes of these gait-initiation problems are not well understood, they appear to be caused in part by problems in the CNS. In addition, some modifications to the motor control system accompany even the normal aging process and such changes predispose the elderly to falls, which are a leading cause of morbidity and mortality among this group. An increased understanding of the functioning of the gait-initiation motor program and how it changes with aging may lead to improved rehabilitation treatment strategies to reduce the risk of falling for many elderly individuals.

The purposes of this study were (1) to determine whether the gait-initiation program (e.g., SOL inhibition followed by TA activation) is expressed in healthy older adults, and (2) to determine whether the functional effectiveness of the gait-initiation motor program in generating momentum is diminished in healthy older adults.

METHODS

Subjects
Twenty healthy young adults (10 women and 10 men: age, 18 to 29 yrs [mean, 25 yrs]; height, 1.52 to 1.93 cm [mean, 1.73 cm]; body weight, 51 to 88 kg [mean 69 kg]) and 20 healthy older adults (11 women and 9 men: age, 64 to 80 yrs [mean, 72 yrs]; height, 1.55 to 1.83 cm [mean, 1.68 cm]; body weight, 51 to 98 kg [mean 70 kg]) participated in the study. All subjects were free from any detectable gait, postural, musculoskeletal, or neurologic disorders. None of the subjects took medication that would affect gait or balance. The procedures for this study were approved by a university's Institutional Review Board, and each subject gave informed consent before participating in the study.

Equipment
The experimental set-up (fig 2) included a custom-designed electromyographic (EMG) system, an ELITE motion analysis system, and a Kistler 9287 multicomponent force platform. The EMG system consisted of low-noise, double-bar differential electrodes, flexible electrode-skin interfaces with a conductive gel, an isolated portable preamplifier, a signal-conditioning card, and a data-acquisition card in a 486 personal computer. The EMG system was used to record muscle activity from TA, SOL, and gastrocnemius (GA) of both lower limbs of each subject. The motion analysis system was used to collect three-dimensional time-displacement data from reflective markers on the shoulder, iliac crest, greater trochanter, distal end of the femur, proximal end of the tibia, lateral malleolus, heel, and fifth metatarsal of one side of the body of each subject. The force platform was used to measure the GRF and the displacements of the COP under each subject's feet.

Procedures
The subjects were asked to stand quietly on the force platform until an auditory signal was given to initiate walking. Data were recorded for 3 sec before the auditory signal and for 3 sec after the signal; thus, data were recorded for 6 sec in each trial. The subjects continued walking for several steps beyond the last recorded step. Subjects were asked to initiate gait at three different speeds: slow, normal, and fast. Speeds were...
subjectively determined by each subject. Ten trials were conducted at each speed. The presentation order of the speeds was varied among the subjects. Before testing, the subject’s dominant foot was determined, and the subject was instructed to lift that foot first in each trial. Also before testing, the subject was asked to stand comfortably on the platform and the position of the feet was marked so that initial foot position remained constant across all trials.

**Data Analysis**

EMG data were collected at 1024 Hz. The data were rectified and smoothed by computing the root mean square of the signal using a sliding window of 41 samples. Kinematic and kinetic data were collected at 100 Hz. The data were low-pass filtered, and the displacement data were time-differentiated to determine the velocity of each point. This processing was conducted using an algorithm based on the LAMBDA algorithm.25

The following parameters were extracted from the data: muscle inhibition/activation times, the maximum forward walking speed reached by the end of the first step (which will henceforth be referred to simply as walking speed), time of swing-limb toe-off, time of stance-limb toe-off, the magnitude of the COP shift in a given direction, and the amount of momentum generated in a given direction. The time at which each muscle was inhibited or activated during the gait-initiation cycle was determined using the criterion that a muscle was active when the amplitude of its EMG signal was greater than the noise floor by at least 3% of the maximum signal recorded for that muscle during the testing. The muscle inhibition/activation times were converted to latencies from the time of movement onset (defined here as toe-off) in the corresponding limb.3 The maximum forward speed of the iliac-crest marker during the gait-initiation cycle was used to estimate the walking speed for a given trial. (We assumed that the COM lies in the pelvic area during stance and gaitz6sz7) The time of swing-limb toe-off was determined from the metatarsal marker on the swing limb.3 The maximum forward speed of the iliac-crest marker during the gait-initiation cycle was used to estimate the walking speed for a given trial. The muscle inhibition/activation times were converted to latencies from the time of movement onset (defined here as toe-off) in the corresponding limb.3 The maximum forward speed of the iliac-crest marker during the gait-initiation cycle was used to estimate the walking speed for a given trial. (We assumed that the COM lies in the pelvic area during stance and gaitz6sz7) The time of swing-limb toe-off was determined from the metatarsal marker on the swing limb, and the time of stance limb toe off was determined from the force-platform data.

Because the time integral of the COP movement and the forward momentum of the COM are not commonly measured parameters, the methods used to calculate these quantities are discussed here in some detail. The movement of the COP over time, rather than the maximum COP displacement, is the relevant variable for investigating momentum generation. Therefore, the magnitude of the COP shift in a given direction (ie, anteroposterior [AP] or mediolateral [ML]) during gait initiation was parameterized by computing the time integral of the COP displacement in that direction from the time of the start signal to the time when the integral was maximized. (In the AP and ML directions, respectively, the integral was maximized when the COP moved in front and to the stance side of its initial position. Note that the COP time series was normalized such that at the time of the start signal, the AP and ML components of the COP were each equal to zero.)

Three sets of parameters—walking speed, the COP displacements, and the amounts of momentum generated—were normalized to allow intersubject comparisons. The first two sets of parameters were normalized by subject lower-limb length, and the momentum measures were normalized by subject lower-limb length and body mass. The resultant parameters had units independent of subject morphology (ie, time units only).

Linear regression analysis was used to examine the relationships between the latencies of muscle activation and inhibition and the relationships between various kinetic and kinematic parameters. Single-factor ANOVA analysis was used to compare the regression-analysis results for the young and elderly populations. The Mann-Whitney rank-sum test was used to compare the occurrence frequency of muscle inhibition/activation patterns in the young and elderly populations.

**RESULTS**

**Presence of the Gait-Initiation Motor Program**

To test for the existence of the gait-initiation motor program in both the young and elderly populations, we examined the muscle activity patterns during gait initiation. We found that most subjects in both populations exhibited the previously reported pattern of SOL inhibition followed by TA activation (eg, fig 3). We also observed a similar pattern wherein GA is inhibited before TA activation. These muscle-activity patterns occurred in both the swing and stance limbs before toe-off of each limb, respectively. However, in the older population, we observed a significant decrease in the frequency with which the gait-initiation program was expressed, as indicated by the number of trials in which the pattern of SOL inhibition/TA activation was observed (fig 4, tables 1 and 2). In addition, the...
central mechanisms for momentum generation, Poleyn

percentage of trials in which the pattern of GA inhibition/TA activation was observed was lower in the elderly population, but this difference did not reach statistical significance. The percentages of trials in which these patterns were found in both populations are presented in tables 1 and 2.

We also examined the invariance of the relative timing between these sequences of muscle activity. We calculated the linear correlation (using linear regression analysis) between the latency of TA activation and the latency of SOL inhibition and GA inhibition, respectively. We found that for the trials in which the muscle pattern indicative of the gait-initiation motor program was expressed, the relative times of SOL inhibition and TA activation were highly correlated in both the young and elderly populations (eg, figs 5A and 6A). We also found a similar relationship between the relative times of GA inhibition and TA activation. Summary statistics on these relationships are given in tables 1 and 2.

Table 1: Expression of the Gait-Initiation Motor Program in the Stance Limb in the Young and Elderly Populations

<table>
<thead>
<tr>
<th>Measures</th>
<th>SOL vs TA</th>
<th>GA vs TA</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of trials included in analysis*</td>
<td>293 (of 452)</td>
<td>427 (of 504)</td>
</tr>
<tr>
<td>Percent of trials exhibiting pattern</td>
<td>76%</td>
<td>49%</td>
</tr>
<tr>
<td>Regression line correlation coefficient (+SD)</td>
<td>.94 ± .07†</td>
<td>.95 ± .05†</td>
</tr>
</tbody>
</table>

* A trial was included in the analysis only if SOL or GA was active and TA was inactive at the time of the start signal. (SOL/GA activity was difficult to detect consistently in some subjects.)
† Correlation coefficient significant at a level of $p < .01$.

Table 2: Expression of the Gait-Initiation Motor Program in the Swing Limb in the Young and Elderly Populations

<table>
<thead>
<tr>
<th>Measures</th>
<th>SOL vs TA</th>
<th>GA vs TA</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of trials included in analysis*</td>
<td>401 (of 541)</td>
<td>416 (of 505)</td>
</tr>
<tr>
<td>Percent of trials exhibiting pattern</td>
<td>68%</td>
<td>24%</td>
</tr>
<tr>
<td>Regression line correlation coefficient (+SD)</td>
<td>.88 ± .14†</td>
<td>.91 ± .08†</td>
</tr>
</tbody>
</table>

* A trial was included in the analysis only if SOL or GA was active and TA was inactive at the time of the start signal. (SOL/GA activity was difficult to detect consistently in some subjects.)
† Correlation coefficient significant at a level of $p < .01$.

To evaluate the functional effectiveness of the gait-initiation motor program in generating momentum we examined the relationship between the time integral of the COP displacement and the amount of momentum generated in the AP and ML directions, respectively. We found that the time integral of the backward COP shift was highly correlated with the amount of forward momentum generated in both the young and elderly populations (eg, figs 5C and 6C). Likewise, we found the time integral of the swing-side COP shift and the amount of momentum generated in the direction of the stance limb to be highly correlated (eg, figs 5D and 6D) in both populations. Summary statistics on these relationships are given in table 3.

Table 3: Summary Statistics on the Relationships Between the Time Integral of the COP Shift and the Amount of Momentum Generated

<table>
<thead>
<tr>
<th>Measures</th>
<th>AP COP</th>
<th>ML COP</th>
</tr>
</thead>
<tbody>
<tr>
<td>Regression line correlation coefficient (+SD)</td>
<td>.89 ± .05†</td>
<td>.91 ± .07†</td>
</tr>
</tbody>
</table>
| * A trial was included in the analysis only if SOL or GA was active and TA was inactive at the time of the start signal. (SOL/GA activity was difficult to detect consistently in some subjects.)
† Correlation coefficient significant at a level of $p < .01$.

As an additional measure of the functional effectiveness of the gait-initiation motor program, we examined the relationship between the time integral of the COP displacement and walking speed. In both the young and elderly populations, the time integral of the AP COP displacement was highly correlated with walking speed, but the correlation coefficients were significantly higher in the young population than in the elderly population ($p < .005$). The summary statistics for this relationship are presented in table 3. The time integral of the swing-side COP shift and walking speed were not correlated in either population. These results were expected given that as walking speed increases, more forward momentum is needed, but the demands for stance-side momentum remain unchanged (ie, the task of shifting the body’s weight to the stance limb does not change with walking speed).

From the regression relationships between the time integral of the COP shift and the amount of momentum generated, we also computed (1) the slope of the regression line, which indicates the amount of momentum generated for each incremental change in the magnitude of the COP shift, and (2) the
Fig 5. Evidence for and functioning of the gait-initiation motor program in a representative young subject: (A) latency of SOL inhibition versus latency of TA activation (subjects were included in the EMG analyses only if SOL was active before gait initiation in at least five trials); (B) AP COP displacement time integral versus the maximum amplitude of the initial burst of TA activity (averaged across the swing and stance limbs); (C) forward momentum versus the AP COP displacement time integral; and (D) stance-side momentum versus the ML COP displacement time integral. In (B), (C), and (D), the quantities were normalized. ○, Slow target speed; ×, normal target speed; ▲, fast target speed.

The intercept of the regression line, which indicates the amount of momentum generated by mechanisms other than the COP shift mechanism. We found that the slope of the relationship between the time integral of the backward COP shift and the amount of forward momentum generated was significantly lower ($p < .001$) in the elderly population (fig 7). This decrease indicates that in older subjects less momentum is generated for each incremental change in COP displacement. We found, however, that the maximum amount of forward momentum generated by the elderly subjects during gait initiation was not significantly different from that generated by the young subjects. This apparent discrepancy can be accounted for by the
Fig 6. Evidence for and functioning of the gait-initiation motor program in a representative elderly subject: (A) latency of SOL inhibition versus latency of TA activation; (B) AP COP displacement time integral versus the maximum amplitude of the initial burst of TA activity (averaged across the swing and stance limbs); (C) forward momentum versus the AP COP displacement time integral; and (D) stance-side momentum versus the ML COP displacement time integral. In (B), (C), and (D), the quantities were normalized. ○, Slow target speed; ×, normal target speed; ▲, fast target speed.

Finally, we found that the ranges of the respective functional parameters (ie, walking speed, the COP displacement time integrals, and the amounts of momentum generated) for the young and elderly populations exhibited much overlap.

DISCUSSION

We found that healthy older adults use the same gait-initiation motor program that healthy young adults use. However, some age-related differences in the expression and func-
From a biomechanical standpoint, we demonstrated that the momentum-generating capacity of the COP shift mechanism is significantly diminished in older adults. Specifically, we showed that the slope of the relationship between the time integral of the COP displacement and forward momentum for gait initiation is not significantly different from that generated by young individuals. This apparent discrepancy can be accounted for by the fact that although the slopes for the regression lines for the AP COP shift–momentum relationship were significantly lower in the elderly subjects, the intercepts were significantly higher. For a given subject, the regression-line intercept for the AP COP shift–momentum relationship represents momentum that is generated in addition to that produced by the COP shift. In the young subjects, the intercept was close to zero, whereas in the elderly subjects, it was significantly greater than zero. This result implies that older adults use other strategies, in addition to the backward COP shift, to generate the forward momentum needed to initiate gait.

Clearly, strategies other than the COP shift mechanism could be used to generate the momentum needed for gait initiation; to do so from a quiet-standing position, the resultant GRF simply must be displaced from its essentially vertical position. This could be accomplished, for instance, by shifting the upper body forward and toward the stance limb. A strategy of this sort, however, would lead to an unstable body configuration because it would result in the trunk being moved beyond the base of support. The COP-shift mechanism (fig 1B) is a more stable (and potentially optimal) strategy because it generates the initial momentum needed for gait initiation without requiring the upper body to be moved from over the base of support. In this manner, balance is maintained during the initial transition period. It is unclear from the study what additional mechanisms are being used by the older subjects to generate the necessary momentum for gait initiation, and this question merits further study.

The subtle age-related changes in the motor control systems of healthy older adults may be exaggerated in the diseases of the CNS associated with aging. Thus, these results may be relevant to the gait-initiation difficulties presented by the elderly patients with Parkinson disease, frontal lobe disease, and other degenerative disorders such as Alzheimer disease.15,17,18 For example,
the festinating gait seen in Parkinson disease is characterized by forward flexion and progressively increasing walking speed, often until the patient falls to the ground. Furthermore, the reduced inhibition of SOL and/or GA muscle activity in such patients may result in increased lower extremity rigidity and postural instability.

CONCLUSIONS

In general, our findings suggest that the CNS uses stable, efficient mechanisms for dealing with the inherent instability of upright bipedalism and that the integrity of these mechanisms degrades with aging. Specifically, the CNS uses a motor program to control gait initiation. This program is not expressed postural instability.

Furthermore, the CNS uses a motor program to control gait initiation. This program is not expressed until the patient falls to the ground. Furthermore, the CNS uses alternative momentum-generating strategies to compensate for this inefficiency, and these alternative strategies are potentially less stable than those implemented by the standard gait-initiation motor program. Rehabilitation strategies specifically aimed at improving the COP shift mechanism would be potentially beneficial to elderly adults, especially those who present gait-initiation difficulties or who are at risk for falling. For instance, elderly people might be able to learn to inhibit the SOL and GA muscles during gait initiation through biofeedback techniques or relaxation exercises. In addition, these results could have important implications for a general understanding of the pathophysiology of falls in the elderly, which are a major cause of morbidity and mortality.

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References


Suppliers

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b. Kistler Instrumentation Corp., 75 John Glenn Drive, Amherst, NY 14228.