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Research Report

Biomechanical Walking Pattern Changes in the Fit and Healthy Elderly

A descriptive study of the biomechanical variables of the walking patterns of the fit and healthy elderly compared with those of young adults revealed several significant differences. The walking patterns of 15 elderly subjects, selected for their active life style and screened for any gait- or balance-related pathological conditions, were analyzed. Kinematic and kinetic data for a minimum of 10 repeat walking trials were collected using a video digitizing system and a force platform. Basic kinematic analyses and an inverse dynamics model yielded data based on the following variables: temporal and cadence measures, heel and toe trajectories, joint kinematics, joint moments of force, and joint mechanical power generation and absorption. Significant differences between these elderly subjects and a database of young adults revealed the following: the same cadence but a shorter step length, an increased double-support stance period, decreased push-off power, a more flat-footed landing, and a reduction in their “index of dynamic balance.” All of these differences, except reduction in index of dynamic balance, indicate adaptation by the elderly toward a safer, more stable gait pattern. The reduction in index of dynamic balance suggests deterioration in the efficiency of the balance control system during gait. Because of these significant differences attributable to age alone, it is apparent that a separate gait database is needed in order to pinpoint falling disorders of the elderly. [Winter DA, Patla AE, Frank JS, et al. Biomechanical walking pattern changes in the fit and healthy elderly. Phys Ther. 1990;70:340–347]

Key Words: Equilibrium, Geriatrics, Kinesiology/biomechanics, gait analysis, Posture, tests and measurements.

The reduction of frequency of falls among the elderly is the goal of many researchers addressing the resultant injuries, death, and loss of mobility.1 Research has focused on epidemiological studies to provide a better description and assessment of the extent of the problem and on characterizing the changes in the standing balance control system that occur with age. The epidemiological data have implicated some aspects of locomotion (ie, initiation of walking, turning, walking over uneven surfaces, stopping) in almost all incidences of falls.2-5

Despite this strong evidence linking locomotion to falls, studies of changes in the balance control system have been limited mainly to tests that probe the integrity of the system during quiet standing. Performance on these tests does not correlate with incidence of falls and is a poor pre-
During standing, the goal is to maintain the body's center of gravity (CG) within the base of support. The initiation of gait, however, is an unstabilizing event whereby the body's CG is made to fall forward and outside of the stance foot. By the time the selected cadence is achieved, the only stabilizing period is double-support stance; and even during that time period the one limb is pushing off with considerable force while the other limb is accepting the full weight of the body. During natural cadence, 80% of the stride period is single-support stance, when the CG of the body has been shown to be outside the foot; the closest it gets to the base of support is when it passes forward along the medial border of the foot. Even during the two 10% double-support stance periods, both feet are not flat on the ground. During the first half of double-support stance, or heel contact (HC), the weight-accepting foot is being lowered to the ground, during the latter half of double-support stance, the final stage of push-off has weight only under the toes. Thus, the body is in an inherent state of instability. Most of the findings from balance studies during standing, therefore, have very limited relevance to gait. The dynamic balance of the head, arms, and trunk (HAT) and the safe transit of the foot during the swing phase of gait (safe toe clearance and a gentle foot landing) present a challenge to the central nervous system during walking. The HAT constitutes two thirds of the body mass, and the HAT's center of mass (CM) is located about two thirds of the body height above ground level. The CM is the point where all the mass of the HAT can be considered to act in all three axes as compared with the CG, which is its location in the gravitational axis. In the sagittal plane, even in slow walking, the horizontal momentum of the HAT results in inherent instability. The role of the ankle muscles in standing balance is paramount, but in walking the role of the ankle plantar-flexor and dorsiflexor muscles for balance has not been seen to be important. The moment of inertia of the HAT about the ankle is about eight times what it is about the hip. Thus, during the first half of stance, for example, when a posterior acceleration at the hip is attempting to collapse the HAT in the forward direction, the ankle muscles do not act to intervene. If they did, they would require a plantar-flexor moment of about 300 N-m to control the huge inertial load. Instead, the ankle muscles produce a small dorsiflexor moment to lower the foot to the ground, followed by a small plantar-flexor moment to control the forward leg rotation. The hip extensor muscles, however, intervene to control the lesser inertial load in conjunction with a tight coupling with the knee muscles. The tight coupling of these two motor patterns has been labeled an "index of dynamic balance." This balance control of the large inertial load of the HAT acts primarily during single-support stance with a transfer of responsibility between limbs taking place during double-support stance.

The swing phase of gait has been shown to be executed with considerable precision with average toe clearances of about 1 cm, and this clearance occurs while the horizontal velocity is maximal (3.6-4.5 m/sec). The heel velocity is also reduced drastically in both horizontal and vertical directions immediately prior to HC. Thus, any degeneration in this fine motor control of the foot may result in problems of stumbling during swing and in rebalancing immediately after HC. Numerous studies have addressed the changes in the gait patterns of the elderly compared with those of the younger adult. The majority of these studies have concentrated on basic outcome measures (ie, stride length, cadence, velocity) and the variability of those measures. Several of these studies have related these gait changes to falls, mobility, and post-fall anxiety. All of these studies have made inferences about the reasons for the observed changes: lower cadence, shorter and more variable step length, increased head and torso flexion, and increased knee and elbow flexion. The suggested reasons imply a degeneration of balance control combined with a general loss of muscle strength. The measures reported, however, were outcome measures, which provide limited insight into the changes in the motor system for balance control and limit our ability to identify the mechanisms behind the observed changes.

With this background in mind, there is a need to document the motor pattern changes that occur in the gait of the elderly and to determine whether those changes are related to balance. Fit and healthy elderly individuals were chosen for this initial study to eliminate effects of a sedentary lifestyle or pathological conditions on walking patterns. Of interest was the normal biological degeneration that takes place with age prior to the advent of any identifiable neural, muscular, or skeletal disorder. All kinematic and kinetic patterns were examined in detail in order to pinpoint major or subtle changes that would point to the degeneration or to compensations that reduce the chance of stumbling or losing balance. Simultaneously, a second major goal was achieved, that of developing a full database of kinematic and kinetic profiles against which to compare individual elderly patients with known or suspected balance or tripping disorders.

Method

Subjects

Fifteen elderly subjects were screened based on a life-style and medical questionnaire and examined by a geriatrician to eliminate any volunteers who had any pathological condition related to the human locomotor system. Informed consent forms were
signed by each subject prior to the walking trials. These fit and healthy elderly individuals (10 men, 5 women) ranged in age from 62 to 78 years ($\bar{X} = 68$ years).

**Procedure**

The protocol for the biomechanical gait analyses was identical to that reported previously and is summarized as follows. Each subject was instrumented with reflective markers to define the following joint centers and segments: toe, fifth metatarsal, heel, lateral malleolus (ankle), head of the fibula, lateral epicondyle of the femur (knee), and greater trochanter (hip). Additional markers, not part of this link-segment analysis, were also attached to the trunk and head to define upper body kinematics: L4-L5, sternum, C1-C2, ear canal, and forehead. A standard link-segment model of the lower limb was developed for the foot, leg, and thigh segments in order to calculate the moments of force at the ankle, knee, and hip. Each subject walked at his or her natural cadence on a level walkway a minimum of 10 times; the walkway recorded the marker trajectories: L4-L5, sternum, C1-C2, ear canal, and forehead. A standard link-segment model of the lower limb was developed for the foot, leg, and thigh segments in order to calculate the moments of force at the ankle, knee, and hip.

Each subject walked over a force platform while a Charge-Coupled Device (CCD) video camera located 6 m to the side of the walkway recorded the marker trajectories over the stride period. The CCD camera was electronically shuttered at 1 msec with a field rate of 60 Hz. The video signal was stored on a Sony Motion Analyzer and subsequently digitized using a specially designed video interface into an IBM PC-AP® computer. The precision of the marker centroids was calculated to within 1 mm. The raw coordinate data were digitally filtered with a fourth-order zero-lag Butterworth filter with a cutoff at 6 Hz. The smoothed coordinates then became inputs to the standard link-segment model.

In addition to the joint moments of force, the mechanical power generated and absorbed at each joint was calculated and the area under each power burst was integrated to determine the mechanical work performed during each of the generating and absorbing phases. The support moment, as defined a decade ago, was calculated and is equal to the sum of the moments at the ankle, knee, and hip (extensor moments were set positive, and flexor moments were set negative). The support moment is the total motor pattern of the lower limb, which has been seen to be positive (extensor) during most of stance, negative (flexor) during late double-support and early swing, and positive (extensor) during late swing. The ensemble average of the moment-of-force patterns over all the strides yielded a mean variance measure for the ankle, knee, and hip powers, from which the hip-knee and knee-ankle covariances were readily calculated. The kinematics of toe markers over the stride period yielded the toe clearance during midswing. Toe clearance was defined as the difference in the vertical displacement of the toe marker at its lowest point in stance (just before toe-off) and its lowest point in mid-swing.

**Data Analysis**

Identical measures were taken from our database on 12 young adults (7 men, 5 women), ranging in age from 21 to 28 years ($\bar{X} = 24.6$ years). Because the population variances were not identical, a modified $t$ test was used to determine any significant differences between selected kinematic and kinetic variables that had potential impact on balance and falling during walking. These variables are presented in the Table.

**Results and Discussion**

The kinematic and kinetic patterns of one elderly subject are used in this section to illustrate the nature and format of the data. The mean cadence for this subject was 105 steps/min ($s = 1.8$), and the following ensemble-averaged waveforms were plotted at 2% intervals over the stride period ($HC = 0$, next $HC = 100$). The average toe-off for this subject was 65.7%, so it was set to the nearest 2% interval (66%). The following profiles are presented: ankle, knee, and hip angles (Fig 1); toe vertical displacement, vertical velocity, and horizontal velocity (Fig 2); ankle, knee, hip, and support moments (Fig 3); and ankle, knee, and hip powers (Fig 4). In all of these diagrams, the mean of the repeat trials is plotted as a solid line with one standard deviation plotted at each 2% interval over the stride period. The mean coefficient of variation (CV) is reported and represents the average variability over the stride period expressed as a percentage of the mean signal amplitude. The CV measure is a single score that allows comparison of the percentage of variability of any waveform over any group of repeat walking trials.

Figure 1 shows the variability of this subject's ankle, knee, and hip joint angles to be quite low. The CV for the ankle, knee, and hip joints was 21%, 8%, and 8%, respectively. Similar low variabilities have been reported for inasubject repeat trials performed across days as well as minutes apart on young adults. These consistent results caution against any inferences about similar invariance in the motor patterns. The indeterminacy of the human motor system during stance is such that many combinations of moments of force at the ankle, knee, and hip can still result in the same lower limb kinematics, especially at the hip and knee; and this finding is supported by the data for this subject.
The toe trajectory data (Fig 2) show the vertical displacement (upper trace), the vertical velocity (middle trace), and the horizontal velocity (lower trace). These trajectory plots all have low CVs, indicating a highly consistent control of the distal segment of the limb, the toe. The average toe clearance of 1.5 cm ($\pm$ 0.5) for this subject occurred at 80% of stride as the toe reached its peak horizontal velocity of 4.3 m/sec. The complex nature of this end-point control task needs to be recognized. The length of the link-segment chain is over 2 m, starting with the stance phase foot and continuing up to the hip, across the pelvis, and down the swing limb, and the chain involves at least 12 degrees of freedom at the joints and scores of muscles. The generation and execution of such a consistent toe trajectory is evidence of fine motor control.

The moment-of-force curves for this elderly subject are presented in Figure 3 with extensor moments plotted as positive, along with the support moment, which is the algebraic sum of the three joint moments. The interpretation of the support-moment pattern has been discussed in detail previously. In summary, the support moment quantifies the total limb synergy, which is extensor during most of stance, becomes flexor during late double-support and early swing, and returns to extensor during late swing. We have identified this support synergy in over 50 assessments on a wide variety of gait pathologies in healthy young ($n = 200$) and elderly ($n = 15$) subjects.

The variability of these moment patterns varies with the joint. This subject's CV was 9% at the ankle, 31% at the knee, 19% at the hip, and only 9% in the support moment. Because CV is a ratio of mean variance and mean signal, the low CV for support moment was partially due to increased mean signal as well as decreased mean variance. It has been shown that the variance in these motor patterns is not random, especially in the highly variable hip and knee patterns. There is a tight neural and anatomical coupling between the knee and hip motor patterns. The covariance between the hip and knee moments can reach 89% in repeat strides assessed days apart and ranges from 60% to 70% for repeat assessments performed minutes apart. This covariance is expressed as a percentage of the maximum possible and would reach 100% if the covariance were equal to the sum of the knee and hip variances. This coupling between the joint moments is revealed in the small CV for the summation of hip and knee moments, which was 14% for this set of repeat trials. The reason for these trade-offs between the hip and knee moments is related to a second limb synergy, that of dynamic balance. This balance synergy is described as follows: On a stride-to-stride basis, the anterior-posterior balance of the HAT is controlled by the hip flexors and extensors during stance (mainly single-support stance). Each stride is somewhat different, and the regulation of this large mass (two thirds of body mass) requires a modified hip motor pattern on each stride. Thus, the high variance in the hip moment during stance is directly due to a continuously changing balance control task. The hip moment, however, is also part of the support synergy. To keep the support pattern nearly constant, there must be an opposite change in the knee moment, which is almost as variable, but in the opposite direction. Such a trade-off between
The knee-hip covariance (% COV hip-knee) was marginally less for the elderly subjects (elderly subjects, 57.7%; young adults, 67.0%; p < .07). The interpretation of this score as an index of dynamic balance suggests that the elderly are less able to make the anterior-posterior shifts in the moment patterns on a stride-to-stride basis to dynamically control the balance of the HAT in the sagittal plane and at the same time maintain the extensor support moment. Currently, it is not possible to speculate whether the covariance reduction is functionally significant. Only after a large number of balance-impaired patients are analyzed will the safety threshold of this synergy be evident. Because of the somewhat higher variability in the hip-knee covariance score for the elderly subjects, these individual scores were examined and revealed that the elderly subjects had a bi-modal distribution, with 10 of them falling within the same range as the young adults and 5 of them with quite low covariances. Our cautious interpretation of this finding is that some of our healthy elderly subjects may have a balance impairment that has not yet been detected by the current simple clinical tests.
The last three significant differences were seen in the mechanical power profiles at the three joints. The work performed (absorbed or generated) during each of these concentric and eccentric bursts is illustrated by the power curves shown in Figure 4 and is described in the Table. Figure 4 shows the average power plots for the 11 repeat trials for the same subject discussed previously. The time integral of each of these power phases (in watts per kilogram) yields the mechanical work (in joules per kilogram) performed by the muscles. The push-off generation (A4 work) by the elderly subjects was considerably reduced (elderly subjects, 0.191; young adults, 0.296 J/kg; \( p < .01 \)) at the same time as the absorbed energy (K3 work) was increased (elderly subjects, -0.087; young adults, -0.047 J/kg; \( p < .01 \)). Thus, the vigor of push-off by the elderly individual is drastically reduced. As stated previously, push-off normally starts at about 40% to 44% of the walking cycle, when the push-off leg is about 30 degrees forward of vertical and the contralateral limb has not yet reached HC.22 Thus, a normal push-off is a "piston-like" thrust from the ankle, which acts upward and forward, and is destabilizing. The elderly subjects in this study appear to have recognized this fact and are reducing that potential for instability. Another possibility is that their plantar flexors may have reduced in strength, and, because of the overpowering gravitational load associated with push-off, a small reduction in strength resulted in a significant reduction in power generation. By-products of this weaker push-off were a shorter step length and the increased double-support time already discussed. Finally, because of the shorter step length, the angle of the foot relative to the ground at HC was reduced in the elderly subjects; thus, the need for absorption of energy by the dorsiflexors (A1 work) in lowering the foot to the ground would be reduced. This difference was borderline significant (elderly subjects, -0.0028; young adults, -0.0074 J/kg; \( p < .08 \)).

Note that all the remaining variables that showed a significant difference were related and reflect functional changes in the gait pattern of the elderly subjects, as represented in the "circular" interrelationships presented in Figure 5. Three possible causes could equally account for all of the observed changes. First, the elderly subjects may have increased their double-support time and reduced the foot angle at HC to improve their restabilizing time. This adaptation would be accomplished with a shorter step length, which could be achieved at the motor level by a less vigorous push-off. A second cause could be that they felt more stable with a shorter step length or a lower velocity, with the associated more flat-footed landing achieved by a weaker push-off and with the longer double-support stance time being a natural consequence. Finally, the primary adaptation may have been a reduced push-off, caused either by muscle weakness or the inherent instability involved in that task, the consequence being a shorter step length and increased double-support stance time. With these three equally acceptable explanations, the exact primary cause of the adaptations may never be known. However, these age-related adaptations by the healthy elderly are important to recognize when researchers and therapists assess elderly individuals with balance disorders. This recognition will enable researchers and therapists to pinpoint...
changes attributable to the disorder and not to age.

Based on previous findings with young adults where no gait-related sex differences were evidenced, this study assumed that the mix of sexes in our elderly group would not alter our findings. In future work, we plan to expand the elderly subject pool to determine whether that assumption was correct.

Summary and Conclusions

This biomechanical study of the gait of young adult and fit and healthy elderly subjects revealed the following:

1. The natural walking velocity of the elderly subjects was significantly reduced; this reduction was not due to a decrease in cadence, but rather to a reduction in stride length. Accompanying this decrease was an increased double-support stance time.

2. Toe clearance in the elderly subjects was not significantly different from that of the younger adults.

3. The covariance between the hip and knee moments of force patterns, which has been identified as an "index of dynamic balance," was reduced slightly in the elderly subjects.

4. Significant differences, which were related to a less vigorous push-off and a more flat-footed landing, were noted in the mechanical power patterns.

5. The significant differences noted above are all attributable to an adaptation related to a safer (less destabilizing) gait stride.

6. Because of the significant differences attributable to age alone, it appears that a separate database is necessary in order to pinpoint falling disorders of the elderly.

Acknowledgment

We acknowledge the technical research assistance of Paul Guy.

References

<table>
<thead>
<tr>
<th>Variable*</th>
<th>Young Adult (n = 12)</th>
<th>Elderly (n = 15)</th>
</tr>
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<tbody>
<tr>
<td>Age (yr)</td>
<td>24.6 2.2</td>
<td>68.0 3.9</td>
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<tr>
<td>Weight (kg)</td>
<td>69.2 10.4</td>
<td>77.2 13.4</td>
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<tr>
<td>Height (m)</td>
<td>1.73 0.10</td>
<td>1.72 0.09</td>
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<td>Cadence (steps/min)</td>
<td>111.0 8.7</td>
<td>110.5 7.3</td>
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<tr>
<td>Stride length (m)</td>
<td>1.55 0.103</td>
<td>1.39 0.14</td>
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<td>Stride length (statures)</td>
<td>0.895 0.047</td>
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<tr>
<td>Stance time (%)</td>
<td>62.3 1.55</td>
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<tr>
<td>Toe clearance (cm)</td>
<td>1.27 0.588</td>
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<td>Toe clearance variance (cm)</td>
<td>0.45 0.311</td>
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<td>% COV (hip-knee)</td>
<td>67.0 8.74</td>
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<td>% COV (knee-ankle)</td>
<td>50.9 17.1</td>
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<td>A1 work (J/kg)</td>
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*Work phase: A1 = absorption by dorsiflexors after heel contact; A2 = generation by dorsiflexors to pull the leg forward over foot; A3 = absorption by plantar flexors as leg rotates forward over foot; A4 = generation of energy by plantar flexors at push-off; K1 = energy absorbed at knee by quadriceps femoris muscle during weight acceptance; K2 = energy generated by quadriceps femoris muscle as knee extends during mid-stance; K3 = energy absorbed by quadriceps femoris muscle as knee flexes during late stance and early swing; K4 = energy absorbed by knee flexors (hamstring muscles) as knee extends late in swing; H1 = energy generated by hip extensors as hip extends (hip flexion reduces) during weight acceptance; H2 = energy absorbed by hip flexors in mid-stance as backward-rotating thigh is decelerated; H3 = energy generated by hip during late stance and early swing to accelerate to lower limb upward and forward.

% COV = percentage of covariance.
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