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Whole Body Momentum During Gait: A Preliminary Study of Non-Fallers and Frequent Fallers

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Whole-Body Momentum During Gait: A Preliminary Study of Non-Fallers and Frequent Fallers

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The importance of momentum in compensating for elderly individuals' strength deficits to achieve activities of daily living, such as rising from a chair has been demonstrated in earlier studies. Here we present a case-control study of three healthy "non-fallers" and two "frequent fallers." All 5 elders were community-living and were tested in the gait laboratory. A four-camera Selspot system was used to obtain whole-body momentum from an 11-segment kinematic model. Ground reaction forces and kinematics were used to calculate lower extremity joint moments. With the exception of the whole-body's angular momentum about the vertical axis, linear and angular momenta during gait were minimum during mid-single limb support and maximum near heel contact. Whole-body momentum values for individuals with a history of falls were similar to those measured in non-fallers. However, subjects with a history of falls had between 17 and 37% smaller maximum ankle and knee torque values than the subjects without a history of falls during ambulation. A comprehensive description of whole-body linear and angular momenta during steady-state gait in older individuals is presented. While whole-body momentum characteristics and magnitude were similar between fallers and non-fallers, the consequences of the lesser torque values in the fallers' knees and ankles to generate and control this momentum warrant further investigation.

Key Words: falls, elderly, balance

While joint angular displacements, angular velocities, torques, and powers are frequently described and discussed in the gait literature, only rarely is momentum of the whole body or of individual segments mentioned (Brunt, Vander Linden, & Behrman, 1995; Kaya, Krebs, & Riley, 1998). Momentum is present during all human motion, including gait. Although whole-body momentum has been relatively overlooked as a parameter of gait, we suggest that it may provide helpful information to better understand the control of balance (and perhaps the failure to control balance) during ambulation.

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Both linear and angular momenta exist during most human movement. Using a three-dimensional orientation system, whole-body linear momenta can be described in the anteroposterior direction (corresponding to the plane of progression of a person ambulating along a straight line), a mediolateral direction and a vertical direction. Whole-body angular momenta can be measured about each of the three orthogonal axes used to describe linear momenta (Figure 1).

Most of the studies investigating whole-body momentum have been performed in order to describe movements taking place during sports such as gymnastics (Gervais, 1994), diving (Yeadon, 1993), and pole-vaulting (Morlier & Cid, 1996). In addition, and of particular interest to our study, a few authors have investigated the role of whole-body momentum in the sit-to-stand maneuver (Hanke, Pai, & Rogers, 1995; Pai, Naughton, Chang, & Rogers, 1994; Pai & Rogers, 1991; Schenkman, Berger, Riley, Mann, & Hodge, 1990). In these particular studies, momentum generation was investigated to establish its usefulness in compensating for lower extremity strength deficits when rising from a chair.

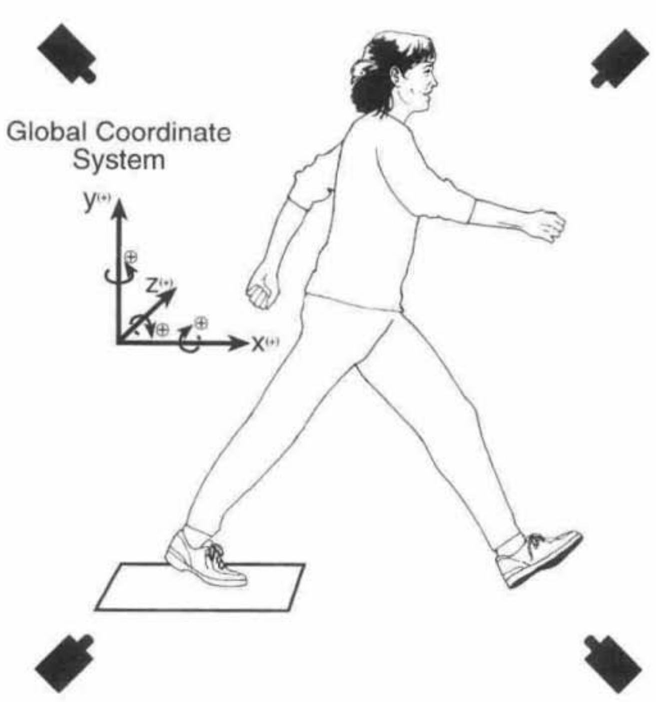

Figure 1 — Graphical representation of the coordinate system. The subject is viewed from the right side with the right foot making contact with the force platform. Arrows indicating the positive direction for each axis and moments about these axes illustrate the global coordinate system. Data are reported according to the right hand rule except that the X-axis valence is reversed in the figure and all data presentations conform to common clinical use showing forward walking as being positive. Since the X valence is reversed only in the final presentation stage, the calculations are not affected.

While momentum may be found to be useful in accomplishing some tasks, such as rising from a chair, excessive or uncontrolled whole-body momentum may also result in detrimental consequences such as falling. This preliminary research provides an initial insight in the role that whole-body momentum may play in fall events experienced by frequent fallers. Therefore, we describe whole-body linear and angular momenta during ambulation for three healthy (without a history of falls) elderly subjects and for two elderly subjects with a history of falls. We hypothesized that the generation and control of whole-body momentum may be altered in individuals with unsteady gait and that it may be a contributing factor to the frequent falls seen in this population.

Methods

Subjects

The control group consisted of three elderly subjects (two females and one male, ages 68, 71, and 73 years, respectively) without self-reported balance disorders (Table 1). These three subjects were selected from a large database of elderly subjects who had been tested in the laboratory. Their selection was based on age and on the horizontal velocity of gait, which closely matched the gait velocity of our subjects with a history of falls when walking at a cadence of 120 steps per minute. The large original pool of healthy elderly subjects was randomly gathered from Medicare lists and had been screened for any impairment through a telephone interview. At their visit to the laboratory, volunteers were examined to rule out any sensory (including vestibular, peripheral nerve, and vision) or motor deficits. They also completed a questionnaire on the occurrence of falls over the previous year.

Two individuals with balance disorders were also tested: an 84-year-old female with a history of falling (more than one fall per month over the past year, no identifiable pathology) and a 61-year-old male diagnosed with bilateral vestibular hypofunction (two fall events over the past year). These two subjects were also examined to rule out range-

Table 1 Subjects' Profiles

Code	Subject	Gender	Age (yrs)	Height (m)	Mass (kg)	Speed (m/s)
NF1	No history of falls	Female	69	1.70	68.0	1.30
NF2	No history of falls	Male	73	1.80	83.9	1.20
NF3	No history of falls	Female	71	1.55	62.1	1.37
F1	Undiagnosed faller	Female	83	1.57	57.6	1.20
F2	Vestibular hypofunction	Male	61	1.78	63.5	1.15

NF1: No history of neurological or vascular disorders; reported one fall due to tripping during the past year; at the time of testing, subject was taking anti-inflammatory drugs. NF2: Walks a mile to work daily; no history of hospitalizations and no history of neurological or vascular disorders. NF3: High blood pressure for 3 years (takes Vasotec); no history of neurological disorders or hospitalization. F1: Status post right total hip replacement (6 years previously) and paroxysmal tachycardia; came to the Biomotion Laboratory because of increased number of falls over the last year; a year prior to testing, she suffered unilateral VIII nerve inflammation. F2: Bilateral vestibular hypofunction; history of sinus tumor removed 12 years ago; in same year, started to experience dizziness, which worsened 6 years ago.

of-motion, motor, and other sensory or balance deficits. With the exception of vestibular deficits in the 61-year-old individual, all other findings were considered normal.

All subjects were tested as part of a larger study investigating falls in the elderly. In this study, all participants (which included individuals with and without a history of falls) were asked to walk at a cadence of 120 steps per minute. The 5 subjects for whom we report momentum data had a walking speed between 1.15 and 1.37 m/s. Only walking cadence was monitored during testing, the limited range of walking velocity ensured that step length was similar for all subjects. All subjects signed an informed consent statement prior to participating in this study.

Instrumentation

The instrumentation used in this study has been described in detail previously (Jevsevar, Riley, Hodge, & Krebs, 1993; Krebs, Wong, Jevsevar, Riley, & Hodge, 1992). Kinematic data were collected by four Selspot II optoelectronic cameras (Selective Electronics, Partille, Sweden) with a viewing volume of $2 \times 2 \times 2$ m. The cameras contain detectors that track the Selspot system's infrared light-emitting diodes (LEDs) embedded in rigid arrays. Arrays were attached to 11 body segments: head, trunk, pelvis, arms (2), thighs (2), lower legs (2), and feet (2). Each segment was modeled as a rigid body with six degrees of freedom (three translations and three rotations). Telemetered Rapid Automatic Computerized Kinematic (TRACK; developed at the Massachusetts Institute of Technology, Cambridge, MA) software was used to acquire and analyze each segment's kinematic data according to techniques described by Riley et al. (1990). Accuracy error in determining the LED position is less than ±1 mm for linear displacement and less than ±1° orientation for angular displacement (Antonsson & Mann, 1989). Kinetic data were collected with two Kistler force plates (type 9281A, Kistler Instruments AG, Winterthur, Switzerland), which have an accuracy equal to ±1% of full scale. Body segment mass, mass centers, and inertial parameters were estimated using regression equations (McConville, Clauser, Churchill, Cuzzi, & Kaleps, 1980; Young et al., 1983). Torques and segmental and whole-body linear and angular momenta were calculated using Newton-Euler inverse dynamics (Hutchinson, Riley, & Krebs, 1994). Both kinematic and kinetic data were sampled at a rate of 150 Hz. The kinematic data were low-passed filtered at 6 Hz (Angeloni, Riley, & Krebs, 1994).

Testing Procedures

The subjects were tested while walking barefoot and wearing shorts and a T-shirt. Each of the 11 marker arrays was attached to its respective body segment. The array location and methods of attachment have been fully described elsewhere (Jevsevar, Riley, Hodge, & Krebs, 1993; Krebs, Wong, Jevsevar, Riley, & Hodge, 1992). Once the subject was fully instrumented, he or she was asked to walk in a straight line on a 10-m walkway at a cadence of 120 steps per minute. Cadence was monitored with an audible metronome. Prior to testing, subjects practiced at the set cadence until they could walk comfortably without "marching" or other "forced" appearance. Mean horizontal speed of progression while in the viewing volume of the camera was measured from the displacement of the center of gravity over time. All subjects walked at average speeds of between 1.15 and 1.37 m/s.

To ensure steady-state gait when data were collected, the space volume for data collection is located in the center of the 10-m walkway to permit the subjects to take at least three strides prior to entering the cameras' fields of view. Several practice trials were given in order for the subject to familiarize himself or herself with the desired walking cadence and the protocol of testing. Each trial was sampled for 3 s with a rest period of 1 to 2 min between trials. For a trial to be accepted, a full gait cycle was needed with proper foot contact with the force platform. The force platform was hidden under a carpet in order to prevent the subject from purposely targeting its location.

Data Processing

Individual segmental linear momenta were calculated as the vector product of the segments' mass and linear velocity (Hutchinson, Riley, & Krebs, 1994), expressed in segmental coordinates. Segmental angular momenta were calculated as the vector product of the segments' moment of inertia and angular velocity about three axes originating at the center of gravity of each segment. Whole-body momenta, reported here in global coordinates, were computed from the instantaneous kinematics of the 11 segments used in our whole-body model (Hutchinson, Riley, & Krebs, 1994; Riley, Mann, & Hodge, 1990). Details of the model and the specific mathematical procedures to calculate whole-body momenta are provided in Hutchinson et al. (1994).

In addition to the description of momentum characteristics, the joint torques for the ankle and knee in the sagittal plane were calculated (Hutchinson, Riley, & Krebs, 1994). To standardize all values to subject mass, all momenta and torques are expressed as a function of body mass. Therefore, linear momenta are in kg · m/s/kg of body mass (i.e., m/s) and angular momenta are in kg · m^2 · rad/s/kg of body mass (i.e., m^2 rad/s).

Results

The results are presented descriptively. Due to the small number of subjects, no attempt was made to determine statistically significant differences. Linear and angular momenta for all subjects are presented in graphical representation of time versus magnitude for the stance phase of gait, from right heel contact to right toe-off (Figure 2). Ankle and knee range of motion, torques, and powers for all subjects are presented in Figure 3.

Linear Momenta for Elderly Individuals Without a History of Falls

Whole-body momentum in the horizontal anteroposterior direction follows a sinusoidal pattern that, as expected, remains positive (Figure 2, linear momentum along the X-axis). The body achieves this maximum forward momentum just after heel contact (0% of right stance phase for right heel contact and approximately 85% of the stance phase on the right for left heel contact). Minimum whole-body forward momentum occurs at approximately 50% of the stance phase, when the body is essentially directly over the small base of support formed by the supporting lower extremity in single limb support. These results are in agreement with earlier theories that maximum kinetic energy levels during gait occur during the double limb support phases of the gait cycle, while minimum kinetic energy levels occur at the time of single leg support (Inman, Ralston, & Todd, 1981).

Vertical whole-body momentum also follows a sinusoidal pattern. However, unlike the anteroposterior momentum, which is always forward directed, vertical momentum alternates between the upward and downward direction (Figure 2, linear momentum along the Y-axis). The vertical linear momentum is directed downward (negative values) at heel contact. This downward momentum continues, although it is get-

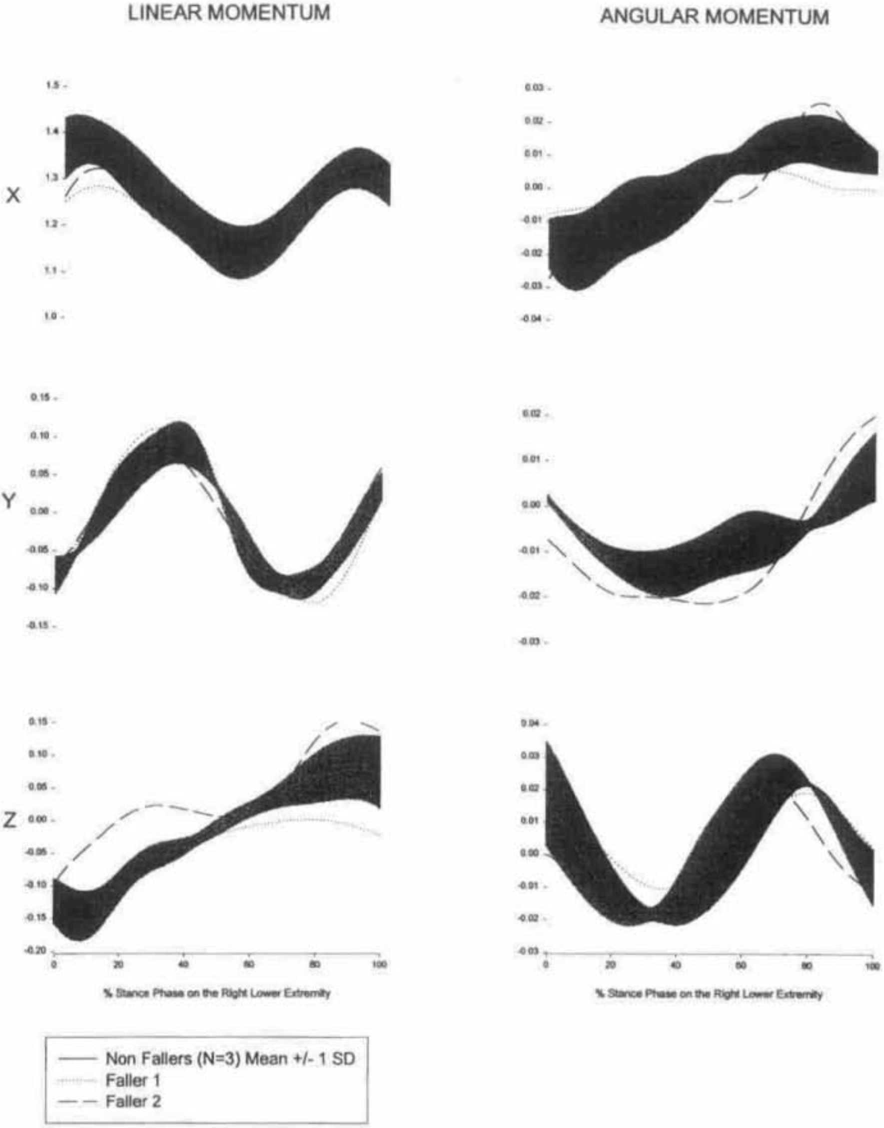

Figure 2 — Graphical representation of whole-body linear and angular momenta for 3 healthy elderly individuals and two elderly individuals with a history of falls walking at a cadence of 120 steps per minute and a speed of between 1.15 and 1.37 m/s. The full line represents the ensemble average for the 3 healthy (without a history of falls) subjects. The shaded area around this line represents ±1 standard deviation around that mean. The 2 individuals with a history of falls are represented separately. The dotted line represents the undiagnosed faller, and the dashed line represents the individual with vestibular hypofunction. The X-axis is in the horizontal anteroposterior direction, with the positive values indicating forward momentum. The Y-axis is in the vertical direction, with the positive values indicating an upward momentum. The Z-axis is in the horizontal mediolateral direction, with the

ting progressively smaller, until just about toe-off of the opposite foot (at approximately 15% of stance phase). Then, vertical whole-body momentum is directed upward until approximately 50% of the stance phase, where momentum is again directed downward until just about toe-off. Maximum downward momentum occurs just prior to heel contact, while maximum upward momentum occurs in the period between approximately 35% and 45% of the stance period.

In the mediolateral direction (side-to-side motion of the body), whole-body linear momentum (along the Z-axis) also follows a sinusoidal pattern that is centered about zero (Figure 2). The peak linear momentum occurs just slightly after heel contact with the ground, with the direction of that momentum being directed toward the lower extremity that has just made contact with the ground. Subsequent to heel contact, the momentum continues to be directed toward the foot on which the body mass is being transferred. This momentum gets progressively smaller until the middle of the stance phase is reached, at which time a momentum of progressively greater magnitude in the direction of the swing leg occurs.

Angular Momenta for Elderly Individuals Without a History of Falls

Whole-body angular momentum around the anteroposterior axis (X) refers to the tendency of the whole body to rotate in the frontal plane around a horizontal axis located at the approximate center of mass of the body. From a posterior view, negative values indicate a clockwise rotation. Overall, angular momentum of the whole body about the X-axis is of relatively small magnitude (Figure 2). The clockwise angular momentum of the whole body is maximum just around the time of right heel contact. This clockwise angular momentum of the whole body continues (although becoming progressively smaller) during the initial 50% of the stance phase. In the second half of the stance phase, a counterclockwise angular momentum of progressively larger magnitude occurs up to the time of left heel contact, which occurs at approximately 85% of the stance phase (Figure 2).

Whole-body angular momentum around the vertical axis is also of relatively small magnitude. Figure 2 illustrates that at the time of right heel contact (0% of the stance phase), the angular momentum about the vertical axis of the body (Y-axis) is near zero. A negative angular momentum (momentum directed clockwise when looking down from above the person) takes place throughout most of the stance phase on the right limb. This clockwise angular momentum reaches its peak value between 30% and 40% of the stance phase. At the time of left heel contact (approximately 85% of the stance phase), the angular momentum is again near zero.

Whole-body angular momentum around the mediolateral axis is graphically illustrated in Figure 2 as the angular momentum about the Z-axis. When viewing from the right side of the body, a positive momentum is in the clockwise direction. At heel contact

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positive values indicating momenta directed toward the left side of the body, away from the right stance leg. A negative angular momentum about the X-axis indicates that the whole body is rotating clockwise (from a posterior view) about the anteroposterior axis fixed at the center of mass of the body. A negative angular momentum about the Y-axis (from a superior view) indicates that the whole body is rotating clockwise about a vertical axis. A positive angular momentum about the Z-axis indicates that the whole body is rotating clockwise (looking at the body from the right side) about the axis fixed at the center of mass of the body. Linear momenta (normalized to kilograms of body mass) are in m/s, and angular momenta (also normalized to kilograms of body mass) are in m^2 rad/s.

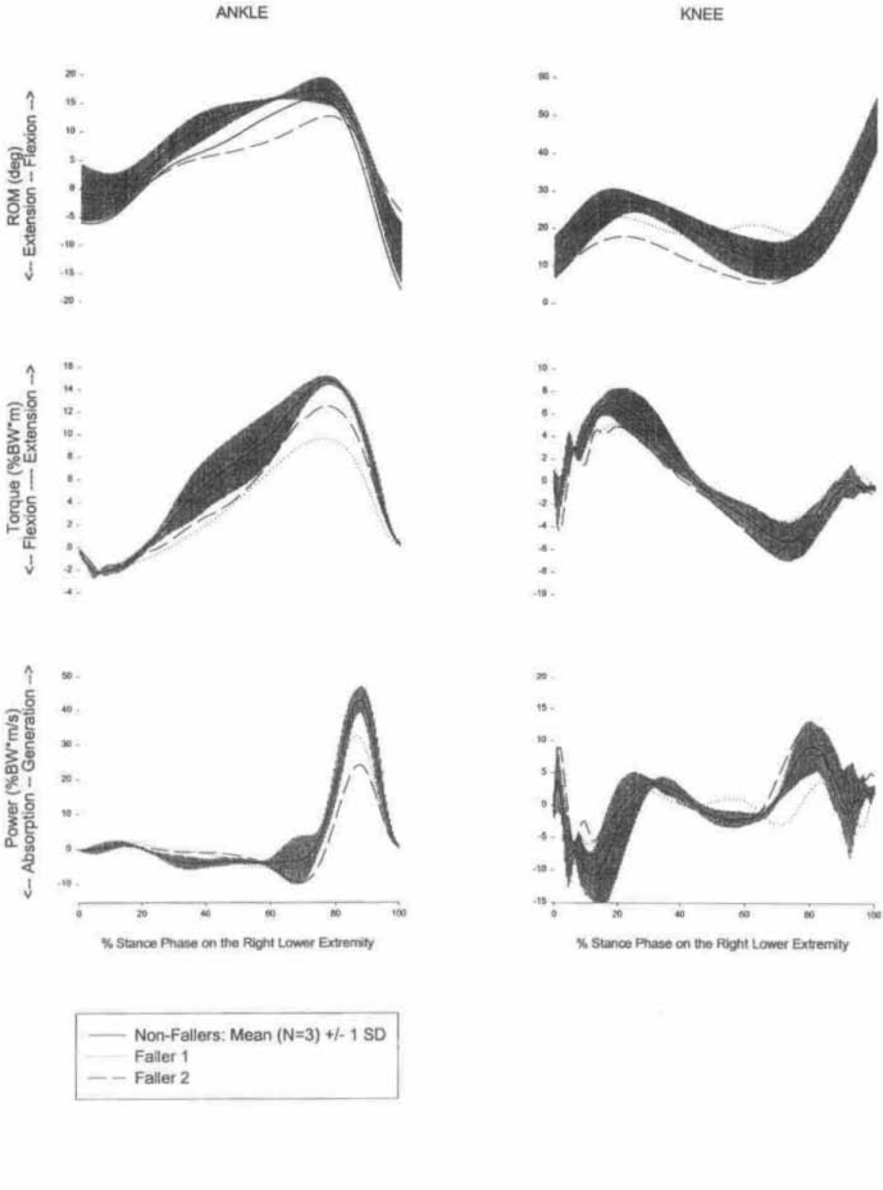

Figure 3 — Graphical representation of ankle and knee range of motion, torque, and power during ambulation for 3 healthy elderly individuals and 2 elderly individuals with a history of falls. Torque and power (normalized per kilogram of body mass) are presented for the stance phase of the gait cycle. The solid line represents the ensemble average for the 3 healthy (without a history of falls) subjects. The shaded area around this line represents ±1 standard deviation around the mean. The 2 individuals with a history of falls are represented separately. The dotted line represents the undiagnosed faller, and the dashed line represents the individual with vestibular hypofunction. Units are in %BW · m (or kg · m/kg of body mass) for joint torques and %BW · m/s (or kg · m/s/kg of body mass) for joint power.

(0% of the stance phase), a clockwise momentum of the whole body is present (Figure 2). This clockwise momentum progressively decreases until about 15% of the stance phase (or toe-off of the opposite foot). From 15 to 50% of the stance phase, a counterclockwise momentum of the whole body is created before reversing and assuming a clockwise direction again for the remainder of the stance phase. Similar to the angular momentum about the X- and Y-axes, the angular momentum about the Z-axis is relatively small.

Linear Momenta for Elderly Individuals With a History of Falls

Whole-body linear momenta for the 2 individuals with a history of falls are represented in Figure 2. The dotted line represents the 83-year-old female with a history of falls of unknown cause, and the dashed line represents the 61-year-old male with a history of falls related to bilateral vestibular hypofunction. Overall, with the exception of the linear momentum about the Z-axis for the individual with vestibular hypofunction, both the magnitude and the general pattern of the curves are similar to those of the non-fallers.

Angular Momenta for Elderly Individuals With a History of Falls

Similar to the graphics for linear momenta, each individual with a history of falls is represented by their own curve on each of the angular momentum graphics (Figure 2). Only some small differences in the magnitude and the general pattern of the curves are noted when comparing the fallers to the control group.

Ankle and Knee Joint Moments and Powers

Ankle and knee joint range of motion, moments, and powers are illustrated in Figure 3. These curves are overall very similar to published data in the literature (Inman, Ralston, & Todd, 1981; Winter, 1983). As compared to the non-fallers, ankle plantarflexion moment was found to be approximately 37% and 17% lower for the subject with an undiagnosed cause of falls and the subject with vestibular hypofunction, respectively. Similarly, peak knee extension moment that occurs shortly after heel contact was approximately 29% smaller for both subjects with a history of falls as compared to our group of non-fallers. These differences in peak torque values are also reflected in the joint powers (Figure 3).

Discussion

Momenta for Elderly Individuals Without a History of Falls

Peak values for all three components of whole-body linear momenta and two of three components of whole-body angular momenta occur near heel contact (Figure 2). Conversely, minimum values for five of the six components of linear and angular momenta occur at the middle of stance phase. At that time, only whole-body angular momentum around the vertical axis is near its peak value.

Whole-body momentum in the forward direction is maximum just after heel contact (0% of right stance phase for right heel contact and approximately 85% of the stance phase on the right for left heel contact). These periods of maximum forward linear momentum (the first and last 15% of the stance phase) coincide with the timing of the propulsive torque generated by the ankle plantarflexors just prior to toe-off (Figure 3).

It is noteworthy that whole-body linear momentum in the forward direction is maximal during the double limb support phase of the gait cycle when the base of support is large. The peak whole-body linear momentum values reached in the forward direction are in the order of 10- to 15-fold greater than the peak whole-body linear momentum reached in the vertical and mediolateral directions (Figure 2). This difference in magnitude clearly establishes forward momentum as the dominant component during gait.

The characteristics of the linear momentum in the forward direction may have implications with regard to fall events occurring during gait. First, when its magnitude is compared to the magnitude of the linear momentum in the vertical and mediolateral directions, it is clear that fall events related to loss of control of whole-body momentum would likely occur in the anterior direction (as when falling forward after tripping over an object). Second, it would be intuitive that such fall events would be more likely to occur at times in the gait cycle when whole-body momentum is the greatest. This may be true; however, the magnitude of the momentum is not the only factor: The likelihood (or available strategy) of recovering from a trip must also be considered. Eng et al. (1994) demonstrated that when the swing leg is interrupted near the end of the swing phase, just prior to heel contact, an overall extension response of the swing leg is generated. This extension response places the foot of the swing leg on the ground in a location anterior to the body's center of mass, which provides support as well as a partial or full recovery of balance. When disruption of the swing leg occurs in early swing, a flexion response of the swing leg occurs. This flexion response is aimed at stepping over the obstacle in order to move the foot in a position to control the forward momentum of the body (i.e., in front of the body's center of mass). Therefore, the body has apparently developed strategies that take into consideration the location of the base of support as well as the location of the center of mass to control the whole-body forward momentum during challenges arising in steady-state gait.

In the vertical direction, maximum downward whole-body momentum occurs just prior to heel contact, and maximum upward momentum occurs in the period between 35% and 45% of the stance phase. Therefore, for a full gait cycle, there are four times when the direction of the vertical momentum reverses and is zero: at 50% of stance for the right and left lower extremity and near left and right toe off (15% and 100% of stance phase on the right) (Figure 2). These points of reversal correspond to the points of highest (at mid-stance) and lowest (end of double limb support) location of the center of mass of the body.

This pattern of vertical linear momentum is probably important in balance control during gait. Because of the upward momentum of the body in the interval between 15 to 50% of stance phase, perhaps an interruption or disruption of the contralateral swing leg is more easily accommodated because the body is actually moving upward at that time. This hypothesis would be partially supported by the observation made by Eng et al. (1994) that disruption of the swing leg in early swing leads to an upward (flexion) adjustment of the swing limb in order for the foot to go over the obstacle. This adjustment, which requires a relatively long time, is aimed at positioning the foot of the swing leg in front of the body's center of mass. In contrast, disruption of the swing leg near the end of the swing phase (85% of contralateral stance), at a time when the vertical linear momentum is near maximum and directly downward, leads to a quick extension reaction of the lower extremity. This action is aimed at quickly positioning the foot on the ground (Eng, Winter, & Patla, 1994).

While it is more difficult to establish precisely the role of mediolateral whole-body linear momentum on fall events, it appears that two key events must occur for a smooth walking pattern to take place. First, linear momentum must change direction at a time during the gait cycle where the base of support is extremely small (only the width of the foot of the stance leg). Second, appropriate foot contact with the ground must be made with the swing leg in order to control (and slow down) the linear momentum, which is increasing and directed toward the swing leg in the second half of the stance phase.

Smooth control of linear momentum in the mediolateral direction requires that the center of mass of the body is kept medial to the center of the foot (Winter, 1995). It is therefore critical that the linear momentum be stopped and reversed in direction at mid-stance. Unlike forward and downward momentum, which may be controlled through muscular actions, the primary method of control of mediolateral momentum likely occurs through appropriate foot placement resulting in a larger or narrower step width (Winter et al., 1995).

When considering all three linear momenta simultaneously, we note that for all three directions, peak linear momentum occurs just prior to or after heel contact with the ground. This overall momentum is being directed mostly forward (the contributor of largest magnitude), but also downward and toward the leg making new contact with the ground (Figure 2). At that time, adequate control of momentum is highly dependent on proper foot contact with the ground. Conversely, for all three linear momentum components, the point where whole-body momentum is minimal occurs at mid-stance. In fact, at that point, momentum is primarily taking place in a single direction—forward—since mediolateral and vertical momenta are near zero. Interestingly, the period of time where momentum is smallest matches well with the midpoint of stance phase when the base of support is limited.

Momenta for Elderly Individuals With a History of Falls

Overall, although some small differences exist, neither individual presents with gross abnormalities in either the pattern or the magnitude of the linear momenta along the X- and Y-axis. The subject with vestibular hypofunction fails to demonstrate a smooth control of momentum along the Z-axis (mediolateral). Similar to the 3 healthy subjects, the subject's whole-body momentum at foot contact is directed toward the foot making contact with the ground. But unlike the control group, which shows a smooth decrease and reversal of momentum direction, this subject displays an abrupt decrease of momentum followed by an extended period of time (between 20% and 60% of stance phase) where the momentum is stabilized around 0 m/s. In this individual, the latter part of stance phase also shows a rapid increase in linear momentum directed toward the foot that is in late swing phase. Future studies may wish to consider the rate of change of the momenta over time (i.e., the slope) as an additional possible indicator of the control or "smoothness" of gait.

The overall pattern of all three angular momenta for both individuals with a history of falls is similar to those of the non-fallers. However, the individual with a history of falls of unknown cause presents with smaller peak magnitude of angular momentum in the frontal plane (about the X-axis) and the sagittal plane (about the Z-axis). Also, for the individual with vestibular hypofunction, angular momentum is greater than expected in the transverse plane (around the Y-axis). In this subject, we also observe the lack of smoothness of the angular momentum curve around the X-axis.

Ankle and Knee Joint Moments and Powers

While whole-body momenta were of similar magnitude for all subjects, the peak torque values for ankle plantarflexion and knee extension were 17 to 37% lower in the subjects with a history of falls. These differences in peak torque values were also reflected in the joint powers (Figure 3). These findings are in agreement with earlier work that has shown that older individuals with a history of fall events have decreased strength of the ankle plantarflexors (Schultz, 1995; Wolfson, Judge, Whipple, & King, 1995). As Schultz (1995) points out, while most tasks of daily living do not require joint torque strength of a very large magnitude, larger physical demands may occur when trying to recover from a loss of balance.

For the two individuals with a history of falls, similar whole-body momenta were maintained with smaller torque values at the ankle and knee. Therefore, these momenta, present in conjunction with smaller torque values, could possibly result in a gait pattern less well controlled and therefore more prone to disruption by external circumstances such as slick flooring or poor foot-ground interaction.

Limitations of the Study

The primary limitation of this preliminary work is the small number of subjects who were included in the study. But, this work should contribute to the body of literature on gait by exposing the interaction that may exist between whole-body momentum and lower extremity torque production during gait.

Conclusion

We present a description of whole-body momenta during gait for a small group of healthy elderly individuals and two elderly individuals with a history of falls. Whole-body momentum during gait has been ignored as a possible contributor to fall events. During ambulation, all three linear momenta and two of three angular momenta reach a maximum value during the latter part of the stance phase, which corresponds to the time when the contralateral lower extremity is nearing heel contact with the ground. This appears to be the crucial time for the control of momentum during gait, which must be accomplished by proper foot positioning and momentum control by the opposite lower extremity. Our 2 subjects with a fall history had lower ankle torque and power in late stance and lower knee torque and power at weight acceptance. Therefore, it is likely that their ability to control their high/near normal momentum is impaired, threatening their locomotion stability.

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