Is Base of Support Greater in Unsteady Gait?

Background and Purpose. We investigated dynamic interfoot distance (IFD) throughout the gait cycle in people with unsteady gait caused by vestibulopathy and in people without known neuromuscular pathology. We expected that the subjects with unsteady gait would use a greater IFD than subjects without neuromuscular pathology and that this IFD would be correlated with other measures of locomotor stability. Subjects and Methods. Simultaneous whole-body (11-segment) dynamic kinematic data were collected from 22 subjects with vestibulopathy and 22 subjects without known neuromuscular pathology who were matched for age, height, weight, and body mass index. Two trials each of the participants' gait at preferred speed and paced gait at 120 steps/min were analyzed with a repeated-measures design with multiple dependent variables. Quantitative data were analyzed descriptively and with inferential statistics. Results. Interfoot distance at preferred gait speed did not differentiate unsteady subjects with vestibulopathy from the comparison subjects. Paced gait IFD total range and IFD in single-limb support differed between groups, but IFD at heel-strike did not. However, IFD at heel-strike, the traditional measure of “base-of-support width,” was correlated with measurements of whole-body center-of-gravity stability ($r = .32–.55$). Discussion and Conclusion. Gait at preferred speed permitted the unsteady subjects and the comparison subjects to select similar IFD values, but at the cost of slower gait in the unsteady subjects. When required to walk at a “normal” pace of 120 steps/min, subjects with vestibulopathy increased their IFD. These data suggest that wide-based gait alone cannot differentiate between subjects with and without balance impairments. Base of support and other whole-body kinematic variables are mechanical compensations of vestibulopathic instability. Further studies are needed to determine whether development of active control of these whole-body control variables can occur after vestibular rehabilitation. [Krebs DE, Goldvasser D, Lockert JD, et al. Is base of support greater in unsteady gait? Phys Ther. 2002;82:138–147.]

Key Words: Gait, Interfoot distance, Kinematics, Stability, Vestibular dysfunction.

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A wide base of support (BOS) has long been believed to be a hallmark of unsteady gait.\textsuperscript{1-5} Although static standing stability increases with increases in BOS, such as in tandem versus 30-cm BOS standing,\textsuperscript{6} no reports to date have described BOS during gait and its relationship to dynamic stability, as determined by 3-dimensional kinematics during gait analysis.

Base of support in previous research has been defined as horizontal stride width during the double-support phase when both feet are in contact with the ground and the whole-body center of gravity (CG) remains within the BOS. Weller et al\textsuperscript{7} found that decreased stride width was closely correlated with fall frequency and dynamic stability. Fried et al\textsuperscript{5} and others,\textsuperscript{6,7} however, found increased stride width in elderly people who were prone to falls. The majority of the gait cycle is spent in single-limb support (SLS), during which BOS is minimized to the width of the supporting foot. Therefore, during most of the gait cycle, the whole-body CG is outside this narrowed BOS.\textsuperscript{8} The term “interfoot distance” (IFD), therefore, is a more accurate term to describe stride width throughout the dynamics of gait. Interfoot distance is the mediolateral distance between the center of mass of the left foot and the center of mass of the right foot, which in turn is located along the longitudinal midline of the feet (Fig. 1). Therefore, the IFD represents the purely frontal-plane interfoot separation in global coordinates.

Vestibulopathy impairs one of the 3 primary sensory modalities (vision, proprioception, and the labyrinth structures) responsible for balance information. Vestibu-

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vestibulopathy include sensations of abnormal movements, dizziness, or vertigo and visual disturbances such as nystagmus or oscillopsia. Instability worsens as ataxia, decreased trunk and head rotation, decreased gait speed, and “a widened base of support.” When gait has been studied in other people, most often only lower-extremity kinematics or time-distance variables have been examined. Although the measurements obtained in such studies reveal details of human locomotion, they cannot disclose global balance impairment indicators such as whole-body CG control or IFD throughout the gait cycle.

The purpose of our study was to assess IFD throughout the gait cycle in people with vestibulopathy as compared with gait in people without known neuromuscular pathology. We hypothesized that (1) subjects with vestibulopathy would have greater mediolateral IFD than matched subjects without neuromuscular pathology during gait and (2) mediolateral IFD would be correlated with other indicators of stability (ie, walking speed, mediolateral CG excursion, mediolateral CG/center of pressure [CG/CP] difference, and whole-body CG/CP moment arm) (Fig. 2). In short, if IFD at heel strike, the usual measure of BOS, is a valuable indicator of gait stability, IFD could be used to differentiate subjects with vestibulopathy and unsteady gait from comparison subjects without neuromuscular pathology, and IFD should correlate with other balance indicators during gait.

Method

Subjects

Twenty-two subjects with vestibulopathy (mean age=55 years, range=25–90) and 22 comparison subjects (mean age=53 years, range=26–88) were studied. The comparison subjects had no known neuromusculoskeletal dysfunction and exercised vigorously at least 3 times per week. The groups were matched by age, height, weight, and body mass index (Tab. 1). Subjects with vestibulopathy reported complaints of unsteadiness, for which they requested vestibular rehabilitation, and were diagnosed with vestibulo-ocular reflex abnormalities. These subjects’ symptoms were stable for at least 2 months prior to the study. Onsets of symptoms ranged from 6 months to more than 10 years prior to entry to the study. Etiologies of vestibulopathy included ototoxicity, bilateral inner ear infections, and idiopathic inner ear damage. Subjects were excluded if evidence of other disease processes that might impair balance existed, such as a prior stroke, Ménière disease, perilymph fistula, other central nervous system pathology, peripheral neuropathy, severe arthropathy, orthopedic deformities, or severe visual impairment. All subjects agreed to be studied by written consent and no monetary compensation was offered.

Sixteen of the subjects with vestibulopathy had bilateral vestibular hypofunction, 6 with bilaterally undetectable (“zero”) function (≤0.1 vestibulo-ocular reflex gains during 0.05-Hz sinusoidal vertical axis rotation [SVAR]) and 10 with bilateral dysfunction (0.1–0.25 vestibulo-ocular reflex gains during SVAR). Six subjects had unilateral vestibular dysfunction (unilaterally abnormal SVAR findings or absent caloric responses). No subject had received vestibular rehabilitation prior to testing.

Instrumentation

The kinematic system, an 11-segment whole-body model, that we used to assess posture and balance is described in detail elsewhere and briefly recounted here. Simultaneous bilateral whole-body kinematic data were collected with a motion analysis system using Selspot II hardware and were analyzed with TRACK (Teleme tered Rapid Acquisition of Kinematics) kinematic software and software developed in the Massachusetts Gen-

Figure 1.
Interfoot distance (IFD) is the mediolateral distance between the center of mass of the left foot and center of mass of the right foot (+ located along the longitudinal foot midline in Fig. 1A). (A) IFD at right toe-off, (B) IFD at right heel-strike.

* Selspot AB, Flojelbergsgatan 14, S-431 37 Malmö, Sweden.
† Developed at the Massachusetts Institute of Technology, Cambridge, Mass.
Figure 2.
The whole-body CG-CP moment arm is calculated by the Pythagorean separation as shown. Thus, the CG-CP moment arm represents the “lever arm” that increases directly as the body departs from static equilibrium; therefore, in gait, larger values represent better dynamic balance. The CP position is indicated by the line meeting the foot in both figures; the CG position is indicated by the other line that meets the approximate middle of the body in these schematics.
eral Hospital Biomotion Laboratory (Boston, Mass). Light-emitting diode arrays were mounted on 1 body segments: right and left feet, shanks, thighs, and arms as well as the pelvis, trunk, and head. The system calculated the 6-degree-of-freedom position of each body segment within a $2 \times 2 \times 2$-m$^3$ viewing volume. Segment masses and mass centers were based on regression equations from 2 stereophotogrammetric studies and were expressed as displacements from the initial position in the laboratory global coordinate system. The anthropometric data incorporated into the TRACK software package were acquired from adult male and female subjects without known neuromuscular pathology. The anthropometric data incorporated into the TRACK software package were acquired from adult male and female subjects without known neuromuscular pathology.

Using the segment masses and center-of-mass kinematics, the kinematics of the whole-body center of mass was estimated with regression equations. System precision was within 1 mm in linear displacement, and orientation was within 1 degree in angular displacement.

Kinetic data were recorded with 2 Kistler force platforms together with kinematic data at a sampling rate of 150 Hz. The force platforms were located in the approximate center of our viewing volume, level with the walking surface of the laboratory and hidden from the subjects’ view by carpeting. The CP was calculated from the individual force-plate CPs and the known force-plate locations and orientations. Center-of-pressure displacements also were measured in the laboratory global coordinate system. Another program (SuperPlot2, patent pending) determined individual foot and combined CPs with an accuracy of <3 mm.

### Procedure

All measurements for a subject were obtained in a single session. All subjects were barefoot and walked without assistive devices. At least 1.5 minutes of rest was given between trials. Several practice trials of each activity were performed before data collection. Two trials each of each subject’s gait at preferred speed and paced gait (120 steps/min) over a 10-m walkway were collected where one foot landed completely on one of the force platforms. Gait at preferred speed should reflect movement strategies under self-selected optimal neuromotor control; paced gait at 120 steps/min permits valid cadence-controlled within- and between-subject comparisons. A 120-step/min cadence was chosen according to previous studies and pilot studies to match the average preferred gait speed among the comparison subjects, as well as to include all subjects with functional limitations. Gait at preferred speed was performed first to prevent the preferred cadence from being influenced by the paced cadence.

Prior to gait trials at preferred speed, subjects were instructed as follows: “Move forward in as straight a line as possible, walking at your normal pace, as if you were taking a brisk walk in the park.” For paced gait trials, subjects were first asked to walk in place to the beat set by a metronome set to 120 beats/min. All subjects walked at the same pace so that the different gait variables could be compared without potentially confounding cadence differences. They were then instructed as in the previous gait trials at preferred speed, “but walk to this pace.” Dependent variables measured included the following: IFD, mediolateral CG excursion, maximum mediolateral CG/CP difference, whole-body CG-CP moment arm, and CG forward velocity. Within a gait cycle, walking speed is the CG forward velocity (in laboratory/global coordinates). Mediolateral CG excursion is the difference between maximum and minimum CG positions in the frontal plane. Maximum mediolateral CG/CP difference occurs when CG and CP are farthest apart in the frontal plane. Whole-body CG-CP moment arm is defined as the combined antero-posterior and mediolateral vector root mean square difference between the whole-body CG and CP during SLS. When the CG and CP are most separated, equilibrium is least, and therefore gait stability is most challenged (Fig. 2). The IFD was measured throughout the gait cycle, enabling us to report values during SLS as well as during the more traditionally reported double-support phase. Three IFD values were taken for statistical analysis: IFD range, IFD at heel-strike, and IFD during SLS. The IFD range is calculated as the maximum IFD value minus the minimum IFD value occurring in succession within a stride. The IFD at heel-strike, traditionally termed “base of support,” is thought to be important because it represents the initiation of the

### Table 1. Subject Characteristics

<table>
<thead>
<tr>
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<th>Comparison Subjects*</th>
<th>Subjects With Vestibulopathy (n=22)</th>
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<tr>
<td>BMI* (kg/m$^2$)</td>
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* Subjects without known neuromuscular pathology.

$^*$ BMI=body mass index.
double-support phase of gait when both feet are not flat on the ground and the body is in an inherent state of instability. The value is obtained by measuring the IFD at heel-strike, as determined by force-plate initial deflection. The value for IFD during SLS is obtained by measuring the IFD when the whole-body CG is at its maximum vertical height where traditional BOS values considered only the single supporting foot. The maximum and average values of mediolateral CG excursion, mediolateral CG/CP difference, whole-body CG-CP moment arm, and CG forward velocity were obtained from the same individual gait cycles.

**Data Analysis**

A computer program (SuperPlot2) displays kinematic variables throughout each gait trial. Specific data points for analysis were manually selected by using the computer mouse to point a cursor toward the point of interest on the screen and recorded by reading them from the screen. Values recorded for 2 trials were averaged for statistical analysis. In subjects where only one trial of data was obtained (e.g., they missed the force plate on one trial), that valid trial’s values were used. Because observations with missing values were not included in the statistical analysis, 40 observations (out of 44 possible) were used for statistical analysis of variables for gait at preferred speed and 42 (out of 44 possible) were used for statistical analysis of variables for paced gait. To quantify variability in IFD during gait, the absolute difference between the IFD values of 2 trials was determined for both gait at preferred speed and paced gait.

All quantitative data were analyzed descriptively and with inferential statistics. Graphic results for raw IFD data were studied for shape, timing, and reproducibility (Fig. 3). Descriptive statistics were calculated for all variables during gait at preferred speed and paced gait. The main hypothesis, that subjects with vestibulopathy would have different IFD values from those of the...
walking velocities during both gait conditions (vestibulopathy had slower average and peak CG forward
are presented in Table 2. As a whole, the subjects with
Descriptive statistics of whole-body kinematic variables
during the double-support phase of gait (Fig. 2).

comparison subjects, was tested with repeated-measures multivariate analysis of variance. To test the hypothesis predicting correlations between IFD and kinematic variables, Pearson correlation coefficients were calculated for all subjects combined and for each group separately. All statistical analyses were performed using the Statistical Analysis System 6.04 statistical analysis program. An alpha (significance) level of .05 was used.

Results
During gait at preferred speed, IFD did not differ between groups, varying from 0.86 to 35.85 cm, but the subjects with vestibulopathy used smaller whole-body CG-CP moment arms than the comparison subjects. Only IFD range and IFD in SLS during paced gait were greater (P<.01) in the subjects with vestibulopathy than in the comparison subjects; IFD was not different at heel-strike in paced gait (Fig. 4). Therefore, hypothesis 1 was not supported: BOS did not differentiate subjects who were unsteady from comparison subjects. During paced gait, subjects with VSP increased their IFD range compared with free gait, whereas comparison subjects’ IFD range decreased slightly (mean IFD [±SD] for subjects with vestibulopathy=8.11±4.48 cm; mean IFD for comparison subjects=5.98±3.21 cm; Fig. 4). We found that timing of maximum IFD was more variable among the subjects with vestibulopathy than among the comparison subjects, occurring sometimes during SLS and sometimes during the double-support phase of gait (Fig. 2).

Descriptive statistics of whole-body kinematic variables are presented in Table 2. As a whole, the subjects with vestibulopathy had slower average and peak CG forward walking velocities during both gait conditions (P<.05). Table 3 shows correlations between IFD at heel-strike and CG stability measures; the absolute difference in IFD range between 2 trials of gait at preferred speed was different from that for paced gait when combined across subjects (P=.026). The IFD at heel-strike was inversely related to forward velocity for all subjects. The IFD at heel-strike was directly related to both medio-lateral CG excursion and mediolateral CG/CP difference for the subjects with vestibulopathy during gait at preferred speed, for the comparison subjects during paced gait, and for all subjects combined during both gait conditions (Tab. 3).

Discussion
In contrast to conventional clinical wisdom, our main hypothesis, that unsteady subjects with vestibulopathy would use a greater BOS than subjects without neuromuscular pathology during gait, was not supported. The IFD value at heel-strike corresponded to the “usual” measurement of BOS during gait and did not differ for gait at preferred speed or paced gait. No IFD group differences were found during gait at preferred speed. The IFD range and IFD during SLS, however, were different between groups during paced gait. Apparently, during gait at preferred speed, subjects with vestibulopathy were able to compensate for dynamic instability through adjustment of other kinematic variables such as diminished walking speed (decreased CG forward velocity) and decreased whole-body CG-CP moment arm. Paced gait, however, encouraged subjects with vestibulopathy to walk faster than their preferred speed, apparently challenging the limits of their dynamic stability. The differences in IFD values reported here may be indicative of individuals’ attempts to compensate for dynamic instability.

Previous Studies
Several authors have described BOS data during gait; however, none have studied IFD during the swing phase of gait. Murray et al27 reported stride width measurements on a group of 30 men without known pathology, aged 20 to 65 years, using interrupted light photography. Mean stride width was 7.7 cm (SD=3.5) during free-speed gait (120 steps/min) and 9.1 cm (SD=4.1) during fast-speed gait (138 steps/min). Murray et al,28 in a later study, found similar values. Gabell and Nayak29 measured stride width as determined by a metallic marker on the subject’s shoe landing on a metal runway. Results varied from 5.05 to 17.5 cm (median=9.6 cm) for a group of young subjects and from 3.72 to 15.2 cm (median=7.48 cm) for a group of older subjects. Decreased heel width in the older subjects could have been due to increased toe-out or foot angle, which moves the heels closer together.12,30 These investigators defined stride width as the mediolateral distance between either the medial or lateral malleoli during full floor contact.
Our comparison subjects (SD and Whaley32 reported heel-width measurements of-gravity average forward velocity, CG MAX VEL—center-of-gravity maximum forward velocity. *Subjects without known neuromuscular pathology.

| Table 2. Means and Standard Deviations for Whole-Body Kinematic Variablesa |
|-----------------|-----------------|-----------------|
|                  | All Subjects    | Comparison Subjectsb (n=22) | Subjects With Vestibulopathy (n=22) |
|                  | X        | SD       | X        | SD       | X        | SD       |
| Preferred gait   |           |          |          |          |          |          |
| CG M-L EX (cm)  | 5.09     | 2.35     | 5.05     | 2.73     | 5.14     | 2.00     |
| CG/CP DIFF (cm) | 7.62     | 2.48     | 7.39     | 2.30     | 7.84     | 2.70     |
| CG AVG VEL (cm/s)| 112.99  | 19.85    | 122.11   | 14.01    | 104.33   | 20.98    |
| CG MAX VEL (cm/s)| 122.93  | 24.27    | 135.34   | 15.87    | 112.21   | 25.42    |
| Paced gait      |           |          |          |          |          |          |
| CG M-L EX (cm)  | 5.08     | 2.54     | 5.19     | 2.91     | 4.96     | 2.07     |
| CG/CP DIFF (cm) | 7.93     | 2.78     | 7.64     | 2.59     | 8.30     | 3.04     |
| CG AVG VEL (cm/s)| 120.61  | 19.15    | 127.04   | 18.08    | 113.16   | 18.01    |
| CG MAX VEL (cm/s)| 132.02  | 23.22    | 141.11   | 20.42    | 122.50   | 22.54    |

Two research reports31,32 support the use of BOS to predict falls among elderly subjects. In both studies, stride width values were obtained by measuring mediolateral heel distance during the double-support phase of gait. Heitmann et al31 reported average stride width values of 7.44 cm (SD=2.68) for a group of subjects who were prone to falling and 6.54 cm (SD=3.11) for a group of subjects who were not prone to falling. Gehlsen and Whaley32 reported heel-width measurements obtained for elderly subjects walking on a treadmill at slow and fast speeds. The average heel-width measurements were 7.19 cm (SD=1.58) at 4 km/h and 6.41 cm (SD=1.54) at 6 km/h for subjects without a history of falls and 7.77 cm (SD=1.47) at 4 km/h and 7.39 (SD=1.06) at 6 km/h for subjects with a history of falls. Our comparison subjects’ IFD at initial heel contact was 13 cm, as compared with approximately 7 to 8 cm found in other studies. We believe that the greater IFD in our subjects reflects normal toe-out and toe-in/eversion at the ankle and foot, which affects foot CG location, but may not be reflected in mediolateral heel or malleolar distance measures.

The measurement technique used in our study differs from the techniques of previous studies. Previous researchers used 2-dimensional analysis, but we used a 3-dimensional kinematic analysis system. In previous research of IFD during gait, measurements were taken between the malleoli (medial or lateral) or between the heels. Our system determines IFD by calculating the mediolateral distance between the longitudinal foot midlines. Because feet are usually turned somewhat outward during stance, the part of the foot used as a target will influence the actual IFD measurement. At initial heel contact, the foot is rotated slightly outward and inverted due to supination at the subtalar and midtarsal joints. By mid-SLS, the foot has rotated inward and everted, with the subtalar and midtarsal joints falling into pronation. By toe-off, the foot has reached maximum inversion and is toeing out. Murray28 noted foot angle (toe-in/toe-out) to be an influence on stride width. An increase in toe-out represents a slightly broader mediolateral stride width. We believe our technique more accurately reflects functional BOS during gait.

Our data are consistent with those of previous investigators in revealing substantial variability among individuals without neuromuscular pathology. Perry34 reported that the BOS of subjects without neuromuscular pathology varied from 20 cm to crossing one foot over the other. Our IFD measurements ranged from 0.86 to 35.85 cm during gait at preferred speed. Our data do not support the findings of slight increases in BOS with increases in walking speed noted by Murray et al27 and Inman et al,33 but rather that BOS varied inversely with measurements of preferred walking velocity (r = .39, Tab. 3). In cadence-controlled paced gait, however, IFD at heel-strike did not correlate with walking speed (Tab. 3).

Heel-strike and mid-SLS represent 2 clearly different events during the gait cycle. Differences in IFD between these 2 gait events may suggest that a different degree of dynamic instability occurs at different times during the gait cycle. The values for IFD at heel-strike and SLS, however, were not statistically different; heel-strike usually occurs as the feet draw closer together. In many instances across all subjects, the feet separated as SLS occurred, resulting in similar values for these 2 measures. Thus, heel-strike and SLS measurements may not reflect actual maximum and minimum IFD values, respectively, and we therefore obtained “IFD range” values.

IFD Inconsistency
Between-subject inconsistency (high standard deviations) was evident in both groups of subjects for all IFD measures (Fig. 4). Heitmann et al31 also found high stride-width inconsistency. Winter et al35 and Gabell et al39 reported high between-subject IFD variability.

Our data suggest which method best measures IFD variability. The IFD range was intended to indicate an individual subject’s DIFD inconsistency. The maximum and minimum values were determined by selecting a
curve reversal on the IFD plot (Fig. 3). This measure was intended to isolate a single IFD adjustment. Typically, however, subjects with vestibulopathy show more frequent curve reversals than subjects without pathology, limiting the size of the IFD variable as currently defined. Perhaps selecting the largest and smallest actual values obtained within a gait cycle may reveal a better measure of variability.

**IFD as Indicator of Stability**

Stability during gait and locomotion can be quantified in a variety of ways. Cycle time, double-support time, and stance duration have been shown to vary inversely with balance control. Our data indicate that IFD at heel-strike decreases as walking speed increases, irrespective of subject group (ie, with or without vestibulopathy). The IFD at heel-strike, similar to usual measures of BOS, did not differentiate subjects with unsteady gait from comparison subjects without pathology. Therefore, our data suggest that conventional BOS measures may not be as helpful as simpler measures such as walking speed. Other researchers have identified altered head kinematics and CG control in subjects with vestibulopathy as compared with subjects without pathology. In our study, weak correlations were noted between IFD values at heel-strike and mediolateral CG measurements (mediolateral CG excursion and mediolateral CG/CP difference) (Tab. 3). Other investigators have suggested the existence of multiple dynamic stability determinants. Our results support that contention, unsubstantiated in our second hypothesis, because we found differences between groups for maximum whole-body CG-CP moment arm during gait at preferred speed, IFD range, and IFD at SLS during paced gait, as well as maximum CG forward and average CG velocities and CG mediolateral velocity during both gait conditions ($P<.05$).

**Clinical Implications**

Research suggests that vestibular rehabilitation does not “cure” the vestibulopathy, but rather enhances an individual’s compensation during locomotor activities of daily living, permitting greater adaptation to the dynamically changing challenges to gait stability. Our results suggest that patients with vestibulopathy walk with more variable intersubject IFDs and that changes in IFD were correlated with measurements of walking speed, mediolateral CG excursion, mediolateral CG/CP difference, and whole body CG-CP moment arm. We suggest that changes in all of these whole-body kinematic variables are mechanical compensations of vestibulopathic instability. Because all subjects with vestibulopathy were tested prior to a course of vestibular physical therapy, it is not known whether these indexes of global stability improve after treatment. Development of dynamic active control of these variables, as well as head and eye control, should be investigated in future studies of vestibular rehabilitation efficacy.

The motion analysis system used in our study is not practical for or accessible to most physical therapists. Perhaps the primary utility of our data is to enable us to begin to quantify dynamic instability during gait. These data indicate that although lower-extremity gait kinematic studies including IFD variables are useful, full-body CG and CP variables during gait were more informative indexes of global stability.

Further comparative studies between subjects with vestibulopathy and subjects with other balance impairments are warranted to investigate differences in IFD and full-body CG and CP variables. We studied all subjects with vestibulopathy as a single pathologic group. One suggestion for future study is to compare other subjects with vestibular dysfunction, cerebellar pathology, and other impairments that produce gait unsteadiness.

**Conclusions**

Subjects with vestibulopathy exhibited IFD displacement and timing patterns that were more variable than IFD values of comparison subjects. Subjects with vestibulopa-
thy used more variable IFDs when required to walk (during paced gait) at a faster pace; during free gait, IFD values were correlated with measurements of walking speed, mediolateral CG excursion, mediolateral CG/CP difference, and whole-body maximum moment arm.

When required to walk at a paced gait, subjects with vestibulopathy walked faster than their preferred pace. During preferred gait speed, we found no difference between groups for IFD. Our findings, along with the correlations between IFD measures and other whole-body kinematic variables, suggest that foot trajectory compensation for vestibulopathic instability occurs throughout the gait cycle.

References