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Quantifying ataxia: Ideal trajectory analysis—A technical note

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Abstract—We describe a quantitative method to assess repeated stair stepping stability. In both the mediolateral (ML) and anterio-posterior (AP) directions, the trajectory of the subject's center of mass (COM) was compared to an ideal sinusoid. The two identified sinusoids were unique in each direction but coupled. Two dimensionless numbers—the mediolateral instability index (I_{ML}) and AP instability index (I_{AP})—were calculated using the COM trajectory and ideal sinusoids for each subject with larger index values resulting from less stable performance. The COM trajectories of nine nonimpaired controls and six patients diagnosed with unilateral or bilateral vestibular labyrinth hypofunction were analyzed. The average I_{ML} and I_{AP} values of labyrinth disorder patients were respectively 127% and 119% greater than those of controls ($p<0.014$ and 0.006, respectively), indicating that the ideal trajectory analysis distinguishes persons with labyrinth disorder from those without. The COM trajectories also identify movement inefficiencies attributable to vestibulopathy.

Key words: *ataxia, balance assessment, body sway, stability, vestibular subject.*

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INTRODUCTION

Reduced stability during standing and gait, dizziness, and other symptoms are common among patients with vestibular labyrinth defects (LD; 1–4). The loss or reduced function of this important sensory mechanism can significantly reduce a patient's ability to perform functional everyday tasks such as walking and stair climbing. There are a number of tests designed to quantify ataxia as a consequence of such diseases; however, many rely on ordinal scaling or an "all-or-none" quality (5–7). These qualities make such tests less than ideal for monitoring progressive changes. We report a method that produces unbounded scalar values in quantitatively assessing a patient's ability to perform a simple task, repeated stair stepping.

To simplify some of the complexities in studying the dynamics of human locomotion, fluidal and semirigid components of the body are often mapped together as rigid segments (8,9). The aggregate behavior of these body segments is often investigated by calculating and studying their resulting center of mass (COM) kinematics and dynamics (e.g., momentum). Dynamic stability, in relation to ataxia, is thus the ability to control the position and momentum of one's COM (4).

In normal human walking and stair climbing, an individual's COM oscillates side to side sinusoidally as

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the body's weight is transferred from one foot to the other as one advances forward (10). In repeated stair stepping, the process of ascending and descending a single step while continuously facing the same direction, the body's COM also oscillates side to side. The COM also moves anteriorly and posteriorly in a similarly sinusoidal fashion, albeit at half the frequency (**Figure 1**).

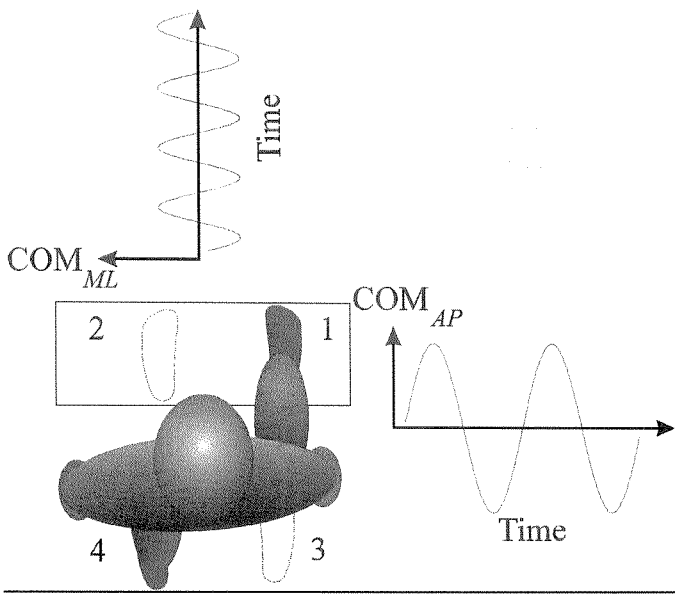

Figure 1.
Sketch indicating the order of a subject's footfalls onto step and floor and how the COM oscillates ideally during the stepping protocol.

The oscillations of an individual's COM during repeated stepping can be thought of as having a periodic component that is related to the frequency of stepping and a residual sway component, which does not occur at the frequency of stepping. Averaged power spectral estimates of mediolateral (ML) COM trajectories for groups of nonimpaired controls and LD patients immediately reveal that there are notable differences in the spectral signatures between the groups (**Figure 2**). It can be observed from the ML power spectrum of LD patients in **Figure 2A** that additional power exists at frequencies outside and principally below that of the stepping component. It is the energy content in these nonstepping-related frequencies that are the focus of the present study.

The goal of the method described is to assess the functional ambulatory stability of patients. This is considerably different from applying tests to screen individuals for labyrinth defects; however, it does not preclude

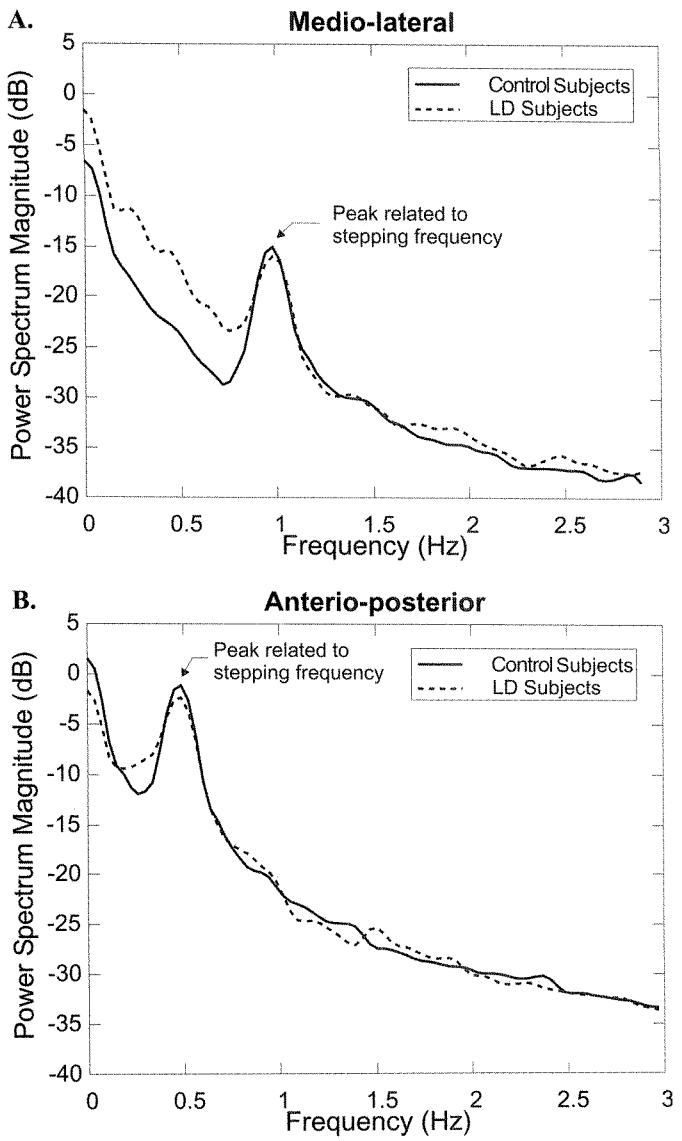

Figure 2.
Averaged power spectra of COM ML (A) and AP (B) displacements. Displacements were normalized by static standing COM heights prior to calculation of individual spectra ($n_{control}$=9, n_{LD}=6).

the method from being incorporated into a vestibular diagnostic test battery. A typical test battery for diagnosing both unilateral and bilateral vestibular hypofunction may include such examinations as caloric testing and electronystagmography. The results of such screenings, however, cannot predict the success of performing ambulatory tasks, in part because they do not account for developed compensatory strategies.

In this method we identify an ideal displacement trajectory of an individual's COM while executing the

repeated stepping task *a posteriori*, that is, a trajectory that accounts for sway only related to the stepping frequency. Once the ideal trajectory is identified, it is used to develop two dimensionless numbers that describe the dynamic instability of the individual in his or her respective ML and anterio-posterior (AP) directions. We hypothesize that the observed displacement trajectory of controls will more closely match their individual ideal trajectories than do those of LD subjects. Because a considerable difference has been observed in the power spectra between the control and LD groups in the ML direction in pilot data, and not necessarily so in the AP direction, we further predict that analysis of ML translations alone will prove most effective in exemplifying instability levels between groups.

To illustrate the efficacy of this objective technique, we compared the calculated numerical results of the trajectories studied, to their phase plane portraits. Phase plane portraits are plots of the velocity of a trajectory versus its displacement in a single degree of freedom. These plots are often used as a subjective approach to evaluate nonlinear systems and graphically provide information on the stability and organization of human locomotion (2,10–15).

METHODS

Subjects

Six subjects comprised the LD group (three males, three females). The mean age of these subjects was 55 years with a standard deviation (SD) in age of 24 years (**Table 1**). Three had bilateral vestibular hypofunction (BVH) and the remaining three had unilateral vestibular hypofunction (UVH) as diagnosed by an otoneurologist. Diagnostic tests to evaluate the subjects included electronystagmography, visual-vestibular interaction rotation and calculation of vestibular-ocular reflex (VOR) gains, phase, and asymmetry on a computerized sinusoidal vertical axis rotation (SVAR) device. Reduced VOR gains, at least three SDs below normal values, during SVAR rotations between 0.01 and 0.10 Hz, were the principle diagnostic criterion of the BVH patients. The UVH patients had unilateral reduced caloric responses and confirmatory SVAR abnormalities, including asymmetries.

Nine nonimpaired subjects (three males, six females), with a mean age of 30 years and an SD of 10 years, were incorporated in the study as a control group. All were free from vestibular or otoneurological pathologies by history and physical examination (**Table 1**). All LD and control subjects were capable of performing the stepping task and ambulating without assistance.

Table 1.
Subject characteristics.

No	Cond	Sex	Age	Hgt
1	H	F	30	92.1
2	H	F	28	92.0
3	H	F	25	90.2
4	H	F	52	91.7
5	H	M	26	95.8
6	H	F	25	91.0
7	H	F	25	90.1
8	H	M	39	92.7
9	H	M	25	88.2
10	UVH	F	73	87.2
11	BVH	M	25	94.7
12	UVH	F	72	91.6
13	BVH	F	70	90.5
14	BVH	M	25	100.0
15	UVH	M	69	102.0

No=subject number, Cond=condition; Age in years; Hgt=static center of mass height, in cm; H=healthy; UVH=unilateral vestibular hypofunction; BVH=bilateral vestibular hypofunction.

Instrumentation

Data collection instrumentation, determination of kinematic data, and COM position estimates are described in detail elsewhere (9,16) and are briefly presented here. Whole body kinematic data were collected using a four-camera motion-capturing system (Selspot II). The kinematics of eleven body segments, incorporating the head, thorax, pelvis, upper arms, thighs, legs, and feet, were measured by collecting three-dimensional (3-D) position data from arrays of strobed infrared light emitting diodes (LEDs) affixed to each segment. At least three LEDs were mounted to each rigid disk to form an array. The arrays were firmly attached to each body segment. Thus, translations and orientations of each body segment can also be estimated. The displacements and rotations of each segment were sampled at 150 Hz.

Measuring anthropomorphic details of each subject allows us to estimate the relative COM location of individual segments (17,18). We can then calculate the position of the whole-body COM knowing the COMs of the individual body segments and their locations in space.

Procedure

The stair-stepping task involved having the subject step repeatedly on and off a single 7.6-cm-high platform.

The width of the platform was 57.6 cm with a forward depth of 23.0 cm. Four consecutive foot placements constituted a stepping cycle (**Figure 1**). Subjects were asked to step at a cadence of 120 beats per minute (BPM) kept by a metronome. In this protocol, subjects step forward up onto the platform with the right foot and then with the left. Continuing with the cadence, the subject then steps backward off the platform with the right foot and then with the left, and repeats this pattern until instructed to stop. Data collection was initiated after the subjects had completed at least two stepping cycles and proceeded thereafter for 10 s. Subjects were barefoot and allowed to swing their arms freely at their sides. No instruction was given to maintain any visual fixation during the trial.

Data Analysis

We were concerned only with horizontal translations of the COM in the plane of the floor of the testing facility. The horizontal trajectory was separated into two orthogonal components and AP displacements of the COM were delimited as COM_{AP} and ML displacements (frontal plane) as COM_{ML}. To separate a subject's extraneous COM translations from their ideal translations for a given experiment the data were analyzed as follows.

This analysis assumes that the COM trajectories were sinusoidal (**Figure 3**) with the AP sway frequency equal to one half the ML sway frequency.

$$f_{AP} = \frac{1}{2} f_{ML} \qquad [1]$$

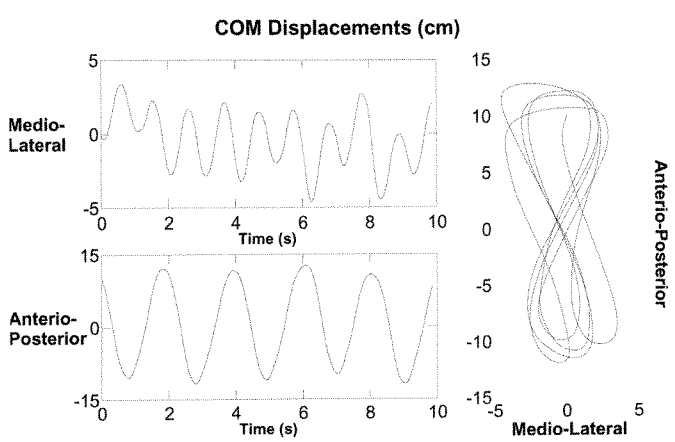

Figure 3.
Example of COM horizontal trajectories for a control subject from a ten-second stepping trial. Left: Orthogonal components of COM displacements along ML (upper) and AP directions. Right: Combined COM displacements depicting actual path over floor.

The expected frequency for f_{ML}, 1 Hz, was the stepping frequency (120 BPM, i.e., 2 Hz) divided by the number of steps (two) required for a single oscillation of COM_{ML}.

We consider that an ideal trajectory exists for both COM_{ML} and for COM_{AP}. Determination of ideal trajectories for a given trial was a multistage process encompassing several optimization steps. The first procedure was to remove the majority of the low-frequency trends in the time-series data. This was done by using a fourth-order infinite-impulse response high-pass filter implemented bidirectionally with a cut-off frequency set at one half the trajectory's expected frequency of oscillation.

The goal of the second procedure was to simultaneously identify the frequency and phase shifts (f_{ML}, Φ_{ML}, and Φ_{AP}, respectively) of the three parameters sought in the basis functions of Equation 2. A description of all parameters is given in **Table 2**.

$$\hat{x}_{ML}(t) = \sin(2\pi \cdot f_{ML} t + \Phi_{ML})$$
$$\hat{x}_{AP}(t) = \sin(2\pi \cdot f_{ML} t + \Phi_{AP}) \qquad [2]$$

Table 2.
Parameter descriptions.

Parameter	Description (unit)
x	Measured COM displacement trajectory (cm)
f	Ideal oscillation frequency (Hz)
Φ	Phase lag (degrees)
$\hat{x}$	Trajectory of ideal basis function (cm)
$\tilde{x}$	High-pass filtered COM trajectory (cm)
r	[Pearson's] Correlation coefficient*
A	Ideal trajectory gain
b	Ideal trajectory offset (cm)
J	Optimization cost*
X	Ideal COM displacement trajectory (cm)
e	Error trajectory (cm)
h_{static}COM	Height of COM during static standing (cm)
I	Instability index*

COM=center of mass; *=dimensionless quantity; parameter subscripts employed throughout the article indicate an association to a specific direction.

The cost function J to be minimized in this procedure was a function of the Pearson's coefficient of correlation of the filtered trajectories $\tilde{x}(t)$ to their respective basis function (Equation 3). Because we wanted to maximize the sum of two correlations, we defined the cost

function as the negative of the sum and employed a quasi-Newton optimization approach to minimize Equation 4 (19).

$$r_{ML} = \text{corr}\left(\tilde{x}_{ML}(t), \hat{x}_{ML}(t)\right)$$
$$r_{AP} = \text{corr}\left(\tilde{x}_{AP}(t), \hat{x}_{AP}(t)\right) \quad [3]$$

$$J = -(r_{ML} + r_{AP}) \quad [4]$$

Using the identified parameters from the minimization of Equation 4, a simple least-squares fit was performed separately on each $x(t)$ to calculate a gain A and an offset b as used in Equation 5 for each basis function to produce the ideal trajectories $X(t)$.

$$X_{ML}(t) = A_{ML}\hat{x}_{ML}(t) + b_{ML}$$
$$X_{AP}(t) = A_{AP}\hat{x}_{AP}(t) + b_{ML} \quad [5]$$

$X_{ML}(t)$ and $X_{AP}(t)$ were the unique ideal trajectories of a subject for a specific execution of the stepping task. Subtracting a subject's actual trajectory from their ideal (Equation 6) produced error trajectories e.

$$e_{ML}(t) = X_{ML}(t) - x_{ML}(t)$$
$$e_{AP}(t) = X_{AP}(t) - x_{AP}(t) \quad [6]$$

Taking the root-mean-square (RMS) of the error vectors produces two performance values with units of centimeters. To transform these values to dimensionless numbers useful for inter-subject comparison as well as to normalize as a function of anthropomorphic characteristics, we divide the RMS values by the subject's nominal COM height in centimeters. The COM height was estimated from data collected during quasi-static standing with feet set parallel and 30 cm apart.

$$I_{ML} = \frac{\sqrt{\overline{e_{ML}(t)^2}}}{h_{static\ CG}}$$
$$I_{AP} = \frac{\sqrt{\overline{e_{AP}(t)^2}}}{h_{static\ CG}} \quad [7]$$

Though the analysis procedure may appear complex, it was easily implemented in the high-level programming language, MATLAB®.

Low numerical values for the instability indices will denote that a subject differed little from an ideal execution of the repeated stepping task. Large values will indicate that the subject differed significantly from his or her potential ideal. Thus high instability indices, which represent poor performance, also signify ataxia or a lack of uncoordinated power in movement.

To determine whether the method distinguished between controls and LD patients, an analysis of variance (ANOVA) for unequal sample sizes was applied to the resulting directional stability parameters I_{ML} and I_{AP}, as well as to the identified correlations (20).

RESULTS

The identified ideal trajectory parameters, mediolateral sway frequencies (f_{ML}), trajectory amplitudes (A) and offsets (b) are listed in **Table 3**. The average values for the estimated ideal frequencies, f_{ML}, for both control and LD groups were identical (0.96 Hz) but were less than the expected frequency of 1 Hz. It was originally thought that f_{ML} could be directly determined as a function of the indicated stepping frequency from the electromechanical metronome as estimated above. This was unacceptable because the indicated frequency was not equal to the actual frequency. Furthermore, some subjects in both groups did not necessarily keep their footfalls in rhythm with the metronome. The SD in f_{ML} among the control subjects (0.008 Hz) was quite small and was less than one percent of the average value. For LD subjects, this value (0.039 Hz) increased to greater than four percent of their average.

The correlation values r_{ML} and r_{AP} obtained from minimizing Equation 4 are also listed in **Table 3**. As a comparative measure between groups, only the correlation values associated with COM_{ML} produced a statistically significant difference $p<0.007$).

Table 4 lists the instability index parameters, I, for each subject. On average, I_{ML} and I_{AP} are approximately twice as large for LD subjects as compared to controls (**Table 4**). The LD subjects, on average, have significantly more ML and AP sway ($p_{ML}<0.014, p_{AP}<0.006$) at frequencies that are not related to the stepping frequency compared to controls (**Table 5**). An example of an ideal trajectory as fit to COM_{ML} for a control subject is given in **Figure 4A**. The error trajectory for this case is shown in **Figure 4B**. In contrast, **Figures 4C** and **D** show similar plots produced for an LD subject.

The scatter plot of all AP and ML instability indices in **Figure 5** shows the moderate correlation ($r=0.84$, $p<0.001$)

Table 3.
Identified parameters and trajectory correlations.

No	Cond	f_{ML}	A_{ML}	b_{ML}	A_{AP}	b_{AP}	r_{ML}	r_{AP}
1	H	0.955	2.21	-0.35	11.43	-9.17	0.95	0.97
2	H	0.941	1.89	4.18	10.30	3.21	0.96	0.97
3	H	0.955	2.05	0.40	8.48	-1.58	0.93	0.98
4	H	0.956	1.64	1.21	13.50	-12.27	0.95	0.96
5	H	0.963	2.48	3.32	11.57	8.55	0.96	0.97
6	H	0.950	2.38	-0.85	9.00	9.94	0.94	0.94
7	H	0.959	2.89	1.93	12.19	-7.87	0.97	0.95
8	H	0.964	1.86	1.22	11.00	-1.26	0.95	0.96
9	H	0.970	2.55	-4.37	12.55	6.97	0.93	0.93
10	LD	0.961	2.40	-3.11	9.86	-1.46	0.96	0.96
11	LD	0.972	2.41	-1.12	13.55	-8.07	0.94	0.94
12	LD	0.934	2.29	1.15	9.37	-3.32	0.85	0.97
13	LD	1.026	2.22	10.56	9.90	-9.17	0.91	0.95
14	LD	0.912	1.81	2.27	9.99	-1.45	0.82	0.93
15	LD	0.970	1.75	4.18	9.47	3.91	0.72	0.88

No=subject number; Cond=subject condition: H=healthy; LD=labyrinth defect (UVH or BVH); ML=medio-lateral; AP=antero-posterior; f_{ML}=ideal medio-lateral sway frequency, in Hz; *A* and *b* are gains and offsets of respective ideal trajectory, in cm; *r*=maximized correlation values.

Table 4.
Instability indices of healthy subjects (Nos 1-9) and subjects with labyrinth defects (Nos 10-15).

	Healthy			LD	
No	I_{ML}	I_{AP}	No	I_{ML}	I_{AP}
1	1.21	1.58	10	1.11	1.55
2	1.12	1.10	11	1.55	1.60
3	0.87	1.47	12	2.20	4.21
4	0.98	1.43	13	4.51	2.47
5	1.27	1.77	14	4.51	3.91
5	1.28	1.71	15	6.76	6.61
7	3.18	1.61			
8	1.32	1.36			
9	2.41	1.90			
Mean	**1.51**	**1.55**		**3.44**	**3.39**
SD	*0.76*	*0.24*		*2.18*	*1.94*

No=subject number; ML=medio-lateral; AP=antero-posterior; all values × 100; SD=standard deviation.

Table 5.
Results of ANOVA between healthy and labyrinth defect groups.

Parameter	*p*<
r_{ML}	0.007
r_{AP}	0.060
I_{ML}	0.014
I_{AP}	0.006

r=maximized correlation values for each direction; *I*=instability indices for each direction; ML=medio-lateral; AP=antero-posterior.

between the two indices. Two control subjects clearly stand out as outliers, with notably higher values of I_{ML}, from the otherwise tight grouping. However, the values of I_{AP} for the same two subjects do not appear to be significantly different than others in the group. Additionally, two LD subjects lie within the grouping of values for normal subjects. The clinicians observing the experiments considered these individuals to be well-compensated LD patients. Moreover, one of these two subjects was an active collegiate baseball player at the time of testing. It should be noted here that there was no significant correlation between values of I_{ML} and I_{AP} for control subjects (r=0.44, p>0.05) but a modest correlation was found for LD subjects (r=0.80, p<0.05). Additionally, no significant correlation was found between age and I_{ML} or I_{AP} in either group. Examples of velocity *versus* displacement phase plane portraits of COM_{ML} and COM_{AP} for several subjects are presented in **Figure 6**.

We also found that the ML and AP displacement trajectories of each subject's COM and estimate of their pelvic center were highly correlated in our experiments. The average Pearson correlation between the COM and pelvic center trajectories among all subjects was greater than 0.98 in each axis.

DISCUSSION

A subject's "ideal" trajectory is dependent on his or her anthropomorphic characteristics as well as on the pro-

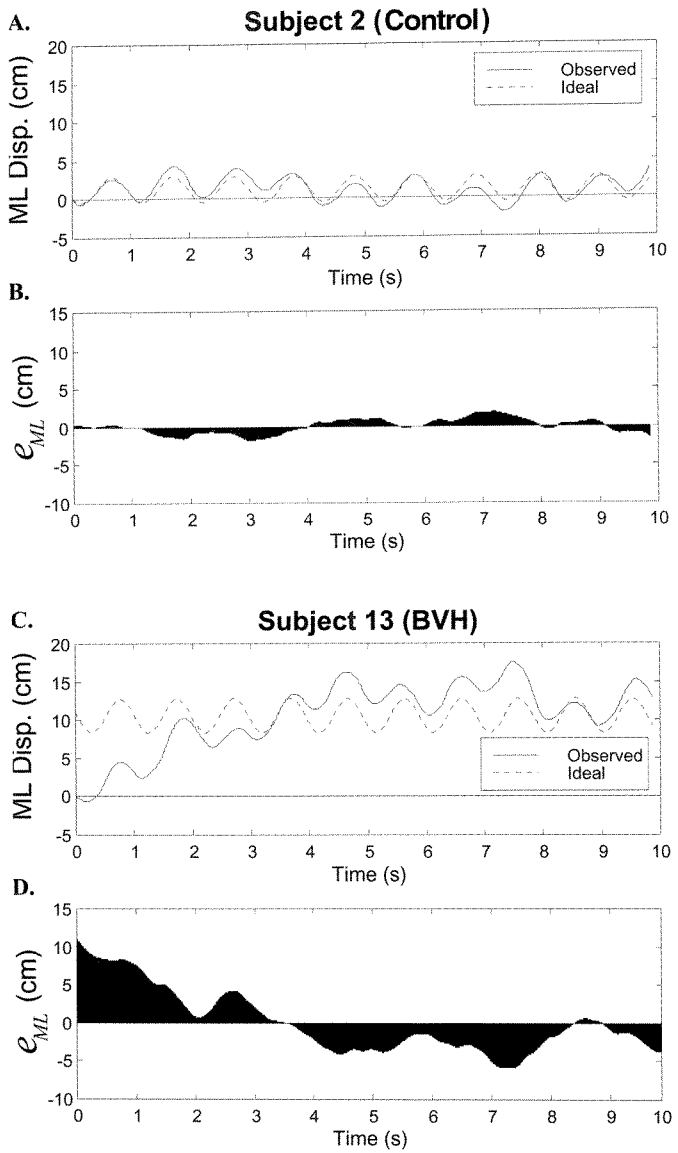

Figure 4.
COM_{ML} observed and calculated ideal displacement trajectories for control (A) and LD (C) subjects. Mediolateral error traces, e_{ML}, are plotted below in plots B and D, respectively, exemplifying the differences between observed and ideal trajectories.

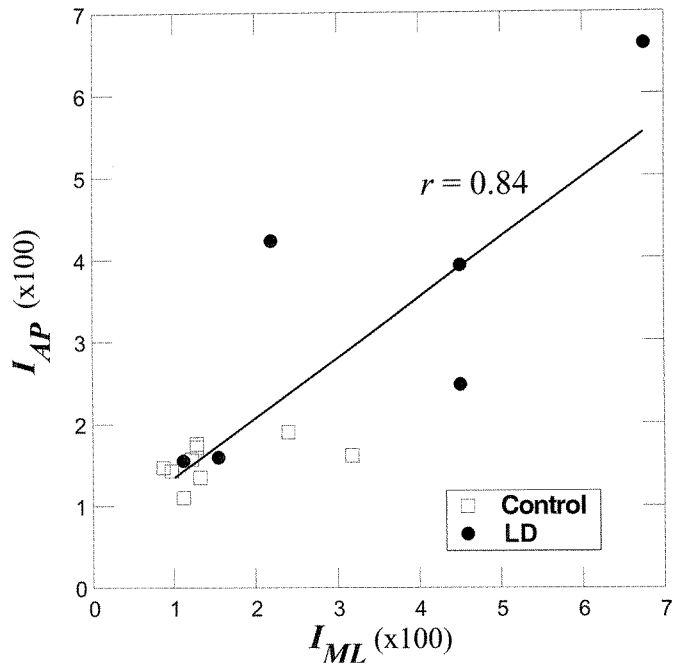

Figure 5.
Scatter plot and regression line of instability indices for control and LD subjects.

tocol of the assigned stepping task. Body mass distribution and segment lengths contribute to the height of the COM which in turn contributes to its 3-D trajectory. Additionally, the stepping rate and height of the platform also affect the ideal trajectory. The first by affecting the COM velocity along the path, the later by affecting the general gait pattern by defining the required range of motion for the limbs. Once the ideal trajectory was determined, the extraneous translations were readily separated and used to evaluate the performance. Scalar values were calculated, normalized, and transformed into dimensionless numbers for easy comparison with other subjects.

The ideal trajectory technique discriminates between control and pathologic groups in ML and AP whole-body COM displacements. Because control subjects have less variation in their displacement trajectories than the LD subjects in this time-paced task, we can assert that they also had superior control of their center of mass velocity. Therefore, we conclude that subjects with lower instability indices also had superior control of their whole-body momentum and kinetic energy during the task.

Due to the frequency coupling stated in Equation 1, the phase plane portraits (**Figure 6**) of COM_{ML} show twice as many orbits as that for COM_{AP}. This can give the potentially misleading impression that the COM_{ML} trajectories are inherently more complex.

The first assumption of this analysis technique was that ideal trajectories in both the ML and AP directions are sinusoidal. Therefore, a phase plane portrait of an ideal trajectory is effectively a plot of the derivative of a basis function in Equation 2, a cosine, *versus* that basis function. Because these functions will have the same frequency and phase, the portrait will be of a single ellipse

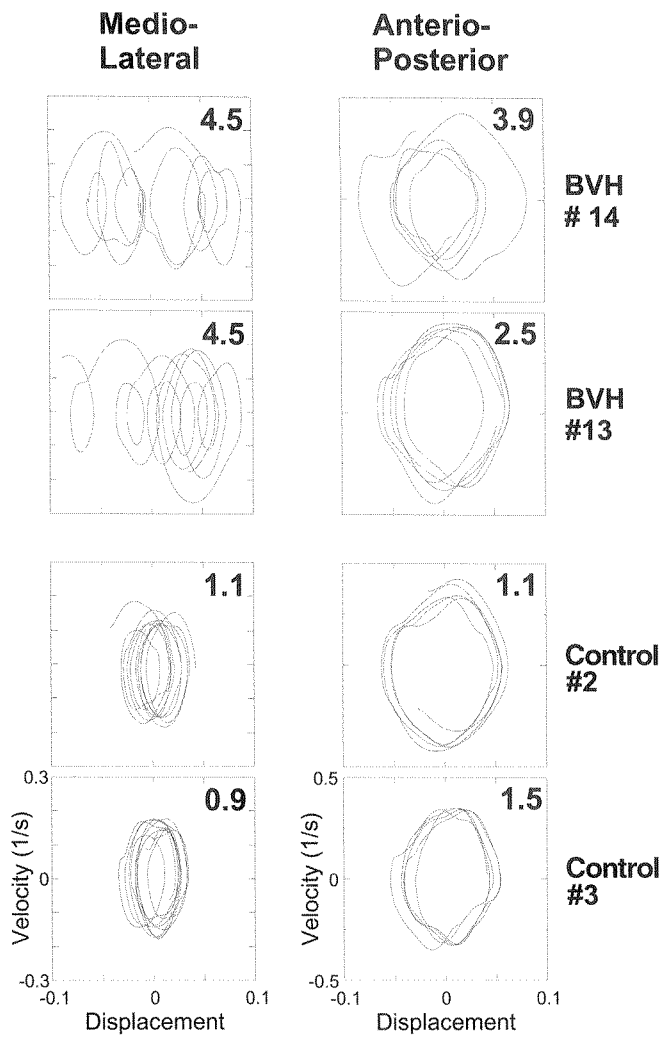

Figure 6.
Phase planes portraits for four subjects in their ML (left column) and AP (right column) directions. Trajectories are normalized by subject's static COM height. The associated instability indices I_{ML} or I_{AP} ($\times 100$) for each trajectory are displayed as bold numbers inside each graph. Subjects are sorted row-wise by value of I_{ML} with the least stable subjects towards the top. Plotting scales are constant along columns.

with its major and minor axes aligned with the plot axes. The less elliptical and, more importantly, the less repeatable a measured trajectory is, the less stable the subject tends to be (11). Instability indices I_{ML} and I_{AP} calculated from the data for each phase plane portrait are also displayed in **Figure 6**. The magnitudes of I_{ML} and I_{AP} relate well with the performance of the subject as visually perceived by the investigators. Our findings tend to indicate that BVH subjects are not necessarily less stable than UVH subjects. However, our population size for these two groups is too small to statistically support or refute this inference. The stability level of each individual is more likely dependent on the level of developed compensatory control strategies as compared to the level of labyrinth impairment (21).

Assuming that all individuals naturally attempt to perform a given physical activity such that they minimize their metabolic energy cost for that activity, it is reasonable to assert that a subject's ideal displacement trajectory is the minimum energy time series path to complete the stepping task (11,12,22). Deviation from the ideal displacement trajectory also means deviation from an ideal velocity trajectory, because this is a time-based task. Such deviations produced by LD subjects indicate that their absolute COM trajectories are greater than their ideal trajectories. Therefore, under the assumption that stair stepping is an energy-dissipating process (23), LD subjects are expending additional energy to complete the same task as that of control subjects.

The excessive energy use by LD subjects is further substantiated by review of the power spectra in **Figure 2**. The plots represent the normalized (for COM height) and averaged power spectra for both the control and LD groups. The area between the two curves is proportional to the relative difference in the magnitude of movement between the two groups. Because such movement is an energy-dissipative process, this also means that the area must be proportional to the relative difference in energy expended by each group to complete the same task. Consider this analogous to the power spectra of an alternating current through an ideal resistor, where the time integral of the spectra is proportional to the total energy dissipated.

The integral of the spectra in **Figure 2A** (5.8 dB-Hz) indicates that the LD group will dissipate notably more energy than the control group in the ML direction. There is only a negligible difference in area between the spectra of **Figure 2B** (-0.2 dB-Hz), indicating that both groups will expend similar amounts of energy to ambulate in the AP direction. We believe there is a greater difference between the ML spectra because subjects are not physically constrained in their frontal plane and can progress to the left and right on the step as much as necessary as long as their foot placements stay within the step width. Some LD subjects have been observed traveling nearly the full extent of the step width during a single trial, though all trials are initiated in the middle of the step. Motion in the AP direction has a biomechanical constraint, in that in order to continue stepping on and off the step, the median location of the COM_{AP} must be near the edge of the step.

We had anticipated that COM_{ML} trajectories, by our original inspection of their power spectra and phase plane portraits, as well by anecdotal evidence, would be more telling about a subject's stability and provide a better discriminator between the groups than COM_{AP}. We found conflicting evidence relating to this prediction when comparing both the correlation values and the instability indices. Although I_{AP} has higher discriminating power, due to the small sample sizes this finding should not be considered definitive. Moreover, for these data, r_{ML} by itself was capable of discriminating between groups but r_{AP} was not (**Table 5**).

The instability indices, as expected, are superior to the correlation values for intersubject comparison for two reasons. First, I can discriminate between groups in both directions investigated. Second, the scales of the values are tremendously different from those of r. The between-group differences for the averaged values of r_{ML} and r_{AP} were only nine and two percent, respectively. For I_{ML} and I_{AP} the averaged differences were 127 and 119 percent, respectively.

The resultants of Equations 2 and 7 are reasonably sensitive to the values of f_{ML}. Therefore, an iterative optimization approach was employed to obtain this parameter and not a more common numerical approach such as the discrete Fourier transform (DFT). The DFT proved to be too coarse in its frequency intervals for application in this technique, given the sampling characteristics of the data.

We examined two degrees of freedom (DOFs) of a three-DOF trajectory. Omitting the vertical trajectory and limiting the analysis to the two remaining DOFs allowed for easy comparison to phase plane portraits. As stated above, a greater difference can be seen between the two groups by analysis of COM_{AP} alone. However, I_{AP} only describes the performance in a single DOF and does not necessarily describe the comprehensive functional performance of an individual ambulating in 2-D space. A potential way to address this would be to produce the magnitude from the two values I_{ML} and I_{AP}. Additional work is required to develop and justify the most equitable procedure of this multi-dimensional analysis. Furthermore, it might be prudent in future research to analyze an individual's 3-D functional performance by incorporating the vertical COM translations omitted in this study.

The number of LD subjects used in this study was limited. This shortcoming was in part due to the inherent difficulty of stepping at 120 BPM. Not all LD subjects tested were included in the analysis as several were unable to perform the task at the indicated stepping frequency of 120 BPM. Future work with this method should test control subjects and patients at slower stepping rates. A procedure that would allow the quantification and normalization of a task's relative difficulty would be useful to compare data from different subjects tested at different stepping frequencies.

The age distribution of control and LD subjects was limited but does not affect the proof-of-concept for ideal trajectory analysis. Our future work will include precisely age-matched samples as well as an increased number of subjects. Future work to corroborate or contradict this study's finding that there is no correlation of I with age would be useful.

High correlations between COM and pelvic center trajectories during the stepping task imply that the instrumentation and some data reduction used in our analysis can be simplified or eliminated if we elect to analyze the pelvis center instead of the COM. For instance, to analyze the pelvic center, one would not need to calculate segmental mass properties. It is expected that measurement of displacements of a single point on a subject, e.g., a tracked marker or the end of a mechanical tracker placed near the L4 or L5 vertebra, would likely be sufficient to estimate the relative displacement of the pelvic center. Saini et al. have shown that tracking a marker place on the sacrum can accurately estimate the vertical displacement of the COM (24). A more practical approach, in particular if employing passive reflective markers, may be to track two markers, one on each anterior superior iliac crest, to describe the relative displacement. Thus, a whole-body motion tracking system could be substituted by motion tracking devices that are more suitable for clinical or field use. Acoustical, mechanical boom, or electromagnetic motion tracking systems are examples of such devices.

We found that the repeated stepping task is a simple and efficient task to implement, and that it models a quasi-functional activity. The approach is sufficient to excite the dynamics of interest while requiring little testing area. The analysis technique provides useful numerical values that will be advantageous for intersession comparison of groups and individuals (3). The instability indices coincide well visually with the ordering of performance by examination of subjects' phase plane portraits.

We conclude that ideal trajectory COM analysis during repeated stair stepping is useful and practical, and

should be included in balance assessment profiles. Additionally, this technique can be employed for quantitative gait assessment of astronauts to evaluate the postflight influence of neurovestibular adaptation to microgravity.

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