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Recovery trajectories of vestibulopathic subjects after perturbations during locomotion

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Abstract. We compared the mediolateral (M/L) responses to perturbations during locomotion of vestibulopathic (VP) subjects to those of controls. Eight subjects with unilateral vestibular loss (100% Reduced Vestibular Response from the caloric test) resulting from surgery for vestibular schwannoma and 11 controls were selected for this study. Despite their known vestibulopathy, all VP subjects scored within the normal range on computerized dynamic posturography Sensory Organization Tests. During gait, subjects were given surface perturbations of the right support-phase foot in two possible directions (forward-right and backward-left) at two possible magnitudes (5 and 10 cm) that were randomly mixed with trials having no perturbations. M/L stability was quantified by estimating the length of the M/L moment arm between the support foot and the trunk, and the M/L accelerations of the sternum and the head. The VP group had greater changes ($p < 0.05$) in their moment arm responses compared to controls. The number of steps that it took for the moment arm oscillations to return to normal and the variability in the moment arms were greater for the VP group. Differences in the sternum and head accelerations between VP and control groups were not as consistent, but there was a trend toward greater response deviations in the VP group for all 4 perturbation types. Increased response magnitude and variability of the VP group is consistent with an increase in their sensory noise of vestibular inputs due to the surgical lesion. Another possibility is a reduced sensitivity to motion inputs. This perturbation approach may prove useful for characterizing subtle vestibulopathies and similar changes in the human orientation mechanism after exposure to microgravity.

Keywords: Balance, postural stability, vestibular, microgravity, vestibular testing, locomotor perturbations, impulse response

1. Introduction

In a previous study [30], we compared estimates of mediolateral (M/L) moment arm length with M/L displacements of the sternum in subjects' responses to single, mechanical perturbations delivered to the support phase foot during paced walking. This methodology may be considered as evoking the impulse response of the balance control system during gait. Figure 2A shows the M/L displacement of the support surface, which makes an abrupt change during the support phase of the right foot during the 5th step of a trial. The application of surface perturbations has been widely used to investigate the properties of balance control during quiet standing and in-place walking, but has less frequently been used to study the dynamic responses of the balance control system during natural locomotion [9, 11,12,22,24–26,28,29,31,35].

Previous studies suggest that responses of vestibulopathic (VP) subjects during perturbations differ from healthy controls mostly in response magnitude instead

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of response timing. Allum [2] and Carpenter [4] have reported that with the exception of soleus muscle during toe-down perturbations balance perturbations (translation or rotation of support surface during quiet standing) produced no changes in EMG onset latencies, in subjects with bilateral vestibular hypofunction (BVH), relative to controls, but did affect the amplitudes of the EMG responses. Allum concluded that the role of the vestibular system was that of response modulation of the muscles controlling postural control synergies [2]. This finding agrees with the conclusion of Dietz et al., in their study concerning which sensors control posture during stance, which concludes that the vestibular system does not seem to be strongly involved in the generation of the EMG response patterns [6]. Macpherson showed that labryrinthecomized cats had an acute hypermeteria in response to perturbations as compared to preoperative controls during standing, but this overresponse returned to control values after about 10 days. Support surface studies of BVH subjects perturbations by Shupert produced increases in head acceleration level, compared to controls, and also larger changes in their trunk angle, particularly for forward perturbations of the body [21]. Pozzo, et al., investigated the ability of BVH subjects to stabilize their heads while hopping with eyes closed, and showed a decrease in head stability compared to normal controls [34].

Oddsson et al., chose to concentrate on the effects of the perturbation on the frontal plane kinematics and the associated mediolateral (M/L) stability [30]. The hypothesis that study addressed was based upon the findings by Powell, et al., for unperturbed locomotion at various stance widths. Powell, et al., showed that for small, voluntary, steady-state changes in M/L stance width, a subject can be modeled as a single inverted pendulum in which the body center of mass (CoM) pivots about the subtalar joint [33]. They showed a linear relationship between M/L single leg displacement and M/L average body acceleration during steady locomotion in normal subjects. Thus, it would be of interest to study further the relationship between these variables.

Our previous results showed that the magnitude of the M/L moment arm estimates during the first few steps after a perturbation during steady walking varied in a nearly linear fashion with the magnitude and direction of the perturbation. The normalized M/L sternum displacements were similarly related to the perturbations. All subjects had normal vestibular function and all responses were sampled during the same event of the gait cycle (same A/P displacement of both shanks, called "Coincident A/P Shank"). Thus, despite the si-

multaneous sampling of M/L moment arm and M/L body motion, our results agreed with those of Powell.

In his modeling of M/L control of locomotion, Kuo has suggested that an energy-efficient way to stabilize M/L sway is to vary the moment arm of the foot during the support phase [19]. His experiments introduced different amounts of noise disturbances into the M/L sensorimotor control system by changing the sensory input conditions from eyes open to eyes closed [3]. They showed that step width increased slightly with the eyes closed condition. The variability of lateral step placement also increased with eyes closed. From this Kuo suggests that the body could use sensory information, including vestibular inputs, to control M/L stability during locomotion. This control scheme implies a sequence of events in which sensory inputs, including M/L accelerations of the head/body, that are estimated during the support phase of one step would be used to control the M/L moment arm during the next support phase of the contralateral foot.

Several other interesting differences between VP and control subjects during locomotion were found by Ishikawa using gait analysis. These include: decreased path integration ability and changes in the time from heel strike to forefoot strike [16–18]. Peruch reported further deficits related to path integration, and concluded that acute vestibular disorders produced a transient disorganization of spatial memory during locomotor tasks that required navigation between multiple locations [32]. From this and from studies of responses of vestibulopathic subjects to perturbations during quiet standing and walking in place, it is reasonable to think that a change in sensory input due to a change in vestibular sensory inputs will affect locomotor performance in response to surface perturbations.

In the present study, we varied the sensory input by using two groups of subjects, those with normal vestibular function, and those with documented peripheral vestibular lesions, but with no abnormalities in postural control as measured by computerized dynamic posturography. In contrast to our previous study, we also used more than one sampling time per gait cycle to reflect the sequential nature of the putative sensory motor control scheme. We estimated the M/L accelerations of the head and sternum during the support phase of one foot and compared them with the M/L moment arms for the next support phase of the contralateral foot as sampled at Coincident A/P Shank. The overall hypothesis for the present study was that vestibulopathic subjects would have larger changes in both their M/L moment arms and their M/L accelerations as compared to normal subjects.

Another objective of this study was to provide further insight into the utility of the paradigm and the set of metrics developed by Oddsson, et al., to quantify locomotor stability [30]. If it can be shown that there exist significant quantitative differences between balance parameters of the healthy and vestibulopathic population, then the developed paradigm would have potential usefulness as a clinical measure for diagnosing vestibulopathies, and for evaluating rehabilitative strategies.

2. Methods

The methods used in this experiment were developed by Oddsson, et al., and are described in greater detail in the original manuscript [30]. A brief description is provided here.

2.1. Instrumentation

Perturbations to gait were applied at the feet with the use of a custom-built, high-performance balance platform called BALDER (BALance DisturbER). The BALDER platform is a computer-controlled, 2.1 m square, 2-axis device, containing a center-mounted force plate and triggering mechanism. For the present experiment, the device was programmed to deliver all perturbations along a single direction that was $45°$ to the subjects' direction of travel.

Kinematic data from 6-marker rigid infrared emitting diode (IRED) arrays, located on the sternum, shanks (midway between ankle and knee), and head were collected at a rate of 40 Hz using an Optotrak 3020 optical tracking system (Northern Digital, Waterloo, Ontario). Each rigid array has 3 columns of IREDs separated by 44 mm, and 2 rows that are separated by 30 mm. The root-mean-square position error is on the order of 1 mm. The location of the camera unit was such that the viewing volume began 2.5 m prior to the center of BALDER and extended 8.3 m past the center of BALDER (Fig. 1).

2.2. Experimental set-up

An overview of the experimental set-up is shown in Fig. 1. The complete walkway consisted of a 3.7 m staging area leading up to BALDER, followed by an additional walkway extension of 10 m. Both the extension and staging area were 1.2 m wide. Thin (5 mm) aluminum plates were fixed to the walkways adjacent to BALDER to bridge the 7 cm gaps on either side of the platform. A thin (5 mm) rubber mat, aligned with the walkway across BALDER, was used to cover the embedded force plate. In this way, the subjects were unaware of the exact location of the triggering mechanism for the perturbation.

2.3. Subjects

We tested 8 vestibulopathic subjects whose demographics and vestibular test results are shown in Table 1. All subjects had undergone surgery for vestibular schwannoma and had 100% Reduced Vestibular Response (RVR) asymmetry in the caloric test portion of the electronystagmography test battery (ENG). All subjects except one also had absent nystagmus responses to an ice water caloric test in the ear corresponding to the side of the surgery. A score of 100% RVR plus an absent ice water caloric response on the side of no response to binaural, bithermal testing is strongly suggestive of a complete loss of, or greatly reduced lateral semicircular canal function in one ear. The time interval between the surgery and the experimental test session averaged 59 months, and the subjects considered themselves to have compensated well to the surgical procedure.

All vestibulopathic subjects had normal composite scores for sensory organization test and motor control test of Equitest®, computerized dynamic posturography (CDP). The gains and time constants of these subjects' vestibuloocular reflex (VOR), as estimated by sinusiodal harmonic acceleration testing from 0.01 Hz to 1.0 Hz were consistent with a unilateral peripheral lesion [7,8], in that the mid-range gain tended to be below normal means, and the VOR time constants were significantly below normal. Subjects 1 and 8 may have had a slight to moderate hypofunction in the ear opposite the lesion as well as the ipsilateral hypofunction [8].

Thus, while the clinical tests of the vestibuloocular reflex are abnormal for these subjects, the most commonly used test of vestibulospinal function is normal for all these subjects. The reason for this inclusion criterion is that we wish to develop a test of vestibulospinal reflexes that is more sensitive than existing clinical tests of postural stability.

The impact of subjects' balance impairment on their daily life was evaluated with a six level function scale (in Table 1) designed for Meniere's disease by the American Academy of Otolaryngology – Head and Neck Surgery [10]. Subjects scored themselves in a range of 1 to 3 out of 6 levels, where level 6 represents
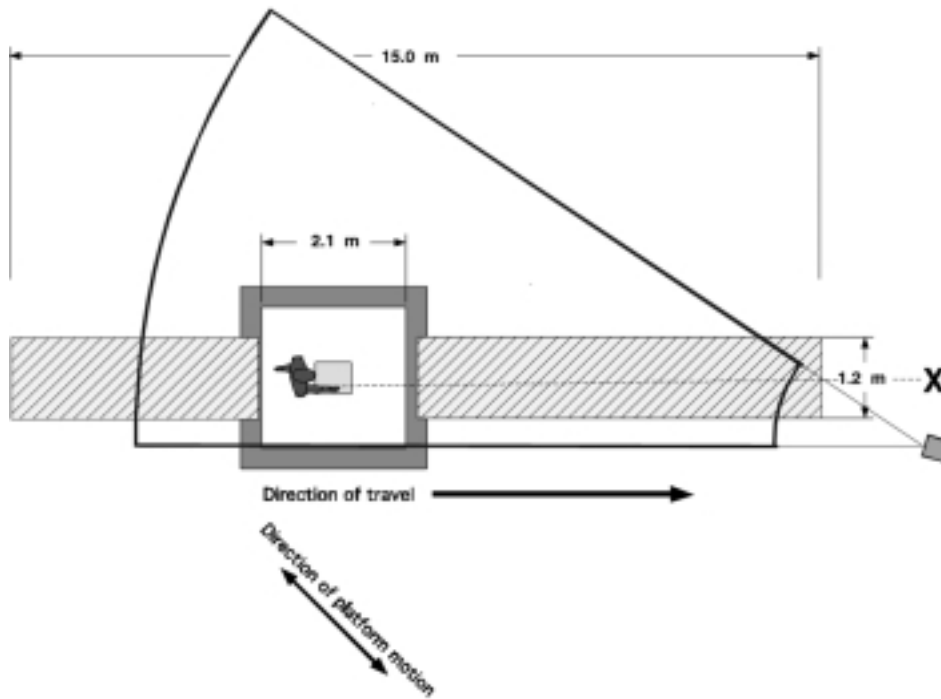
Fig. 1. A plan view of the experimental set-up showing the Balder platform with walkway, and the Optotrak 3020 viewing volume.

Table 1
Vestibulopathic subject demographics

ID	Age	Gender	Side of tumor	Time since surgery (mo)	SOT score	MCT score	RVR	VOR midrange gain	VOR time constant (sec)	Probability of normal VOR*	UVH or (pBVH)	AAO-HNS function level (1 to 6)	Other diseases
1	58	female	right	98	69	155	100	0.79	3.3	1.26e-10	(0.036)	3	allergy
2	56	male	left	24	70	155	100	0.86	6.5	0.0057	UVH	1	none
3	56	male	right	107	73	148	100	0.79	7.8	0.00928	UVH	1	asthma
4	40	male	left	13	70	129	100	0.71	8.0	5.73e-10	UVH	1	none
5	68	male	left	16	68	146	100	N/A	N/A	N/A	N/A	3	back problems
6	31	male	left	5	71	154	100	0.76	6.3	4.16e-08	UVH	2	none
7	51	female	right	90	74	144	100	0.61	9.9	5.21e-05	UVH	1	hypertension
8	67	female	right	126	78	128	100	0.49	4.7	1.32e-07	(0.0129)	2	vision problem

Legend:

RVR – Reduced vestibular response to bilateral, bithermal caloric stimulation. All subjects except #7 had a zero deg./sec. nystagmus response to ice water in the side-of-tumor ear. Subject #7 had a 5 deg./sec. response.

SOT – Sensory organization test: Normal mean composite scores are 80.2 for 20-59 year olds (yo) & 76.9 for 60–69 yo.

5th percentile (abnormal) limits are 68.5 for 20–59 yo & 70.0 for 60–69 yo.

MCT – Motor control test: Normal mean composite scores are143.0 for 20–59 yo & 151.8 for 60–69 yo.

5th percentile (abnormal) limits are 161.0 for 20–59 yo & 170.8 for 60–69 yo.

VOR – Vestibuloocular reflex.

N/A – Not available.

* See Dimitri et al., 1996.

UVH or (pBVH) – Unilateral (UVH) or bilateral vestibular hypofunction, based upon Dimitri et al., 2002. If patients is scored as bilateral hypofunction (BVH), then the probability that of this occuring by chance is given in parentheses.

complete disability. Three subjects scored themselves as level 1: activities not affected by dizziness. Two scored themselves as level 2: cessation of some activities during spells of dizziness, but otherwise no change in any plans or activities to accommodate dizziness. The remaining 3 subjects scored themselves as level 3: having to change some plans or activities to allow for dizziness. In summary, the objective and subjective

measures generally indicate only mild balance deficits in the VP subjects.

The control population consisted of 11 healthy subjects (10 male, 1 female) with a mean age of 35 ± 9 years (age range 26–57 years), and was tested by Oddsson, et al., None of the control subjects had a previous history of orthopedic, neurological, or vestibular disorders [30].

2.4. Protocol development

A series of pilot tests were conducted to develop a standardized, repeatable and safe perturbation protocol that could be tolerated by both the healthy and vestibulopathic subject populations [30]. All perturbations were applied to the right foot. The perturbations were applied in two directions: 45 degrees forward and to the right (FR) of the subject's trajectory and 45 degrees backward and to the left (BL) of the subject's trajectory. The purpose of the perturbation was to induce an abrupt change in the dynamics of the mediolateral control during steady locomotion. This abrupt change is somewhat like applying an impulse to the system in order to observe its so-called impulse response. The perturbations were applied at two magnitudes (5 and 10 cm), thus resulting in four possible perturbation types. The protocol consisted of 3 trials of each of the four perturbation types and 12 trials with no perturbation, delivered in random order. Since half the perturbations occur in one direction, while the other half occur in another, and there are no perturbations at all in half of the runs, it is very difficult for the subject to predict what will happen in any particular run. All perturbation types had a constant acceleration of 9.81 m/s^2 and a constant deceleration of 5 m/s^2. Maximum velocities were 0.5 and 0.7 m/s for the small (5 cm) and large (10 cm) perturbations, respectively. The detection of heel-strike from the force plate embedded in BALDER was used to trigger the onset of the perturbation. The onset of perturbation occurred at 180 and 200 ms following heel-strike for the FR and BL perturbation types, respectively. The onset time for BL perturbations was 20 ms longer to avoid tripping subjects by getting their swing leg caught behind the stance leg.

Subjects walked barefoot and at a pace of 100 steps per minute (1.67 Hz), kept by an electronic metronome. Pacing was used to decrease variability between subjects and to assist subjects in their return to normal gait following the perturbation. Each trial consisted of an end-to-end trip down the 12 meter long walkway during paced locomotion, and included between 13 and 16 steps lasting between 7 and 10 seconds. Subjects' arms were allowed to swing naturally. Subjects were instructed to maintain visual fixation on a target located past the end of the walkway at eye level. All trials began with a step of the left foot, and the starting points were adjusted so that the third right heel strike landed on the force plate. Three trials of paced walking were collected prior to the introduction of any perturbations. Before the actual test protocol began, subjects were allowed as many practice trials of each perturbation type as desired.

2.5. Data analysis

The sampled data from the Optotrak 3020 system were stored to disk in real-time and were later formatted into files that were compatible with MATLAB (The MathWorks, Natick, MA). All further analysis was performed using custom MATLAB routines.

2.6. Moment arm estimates

2.6.1. Coincident A/P shank displacement calculation

To obtain a single estimate of the moment arm during the support phase of each foot, we re-sampled the already-processed 40 Hz kinematic data for the sternum and shank at a specific event called Coincident A/P shank displacement. The kinematic data were sampled at a rate of once per swing phase (twice per gait cycle), at the instance when the A/P positions of the two shanks were equal. This event was selected as the sampling criteria because it occurs once, and only once, during each swing phase in normal gait. Figure 2 shows an example of the time series plots of the M/L sternum and shank trajectories in response to a large, FR perturbation for a control subject (A) and a vestibulopathic subject (B). The abscissa has been reversed so that the horizontal plane of motion may be considered as if viewed from above, with the subject proceeding from left to right. The dotted vertical lines mark the times at which the A/P positions of the shanks are equal. The values of each time series plot at these events make up the 12 to 16 samples that are used for all further data analysis.

2.6.2. Moment arm calculation

At each above mentioned sampling event, the M/L distance between the sternum and the shank of the stance leg was calculated to provide an estimate of the M/L moment arm of the inverted pendulum system (Fig. 3). A correction was made to account for the

difference between the M/L displacement of the shank displacement and the M/L displacement of the foot. Our estimate is a simplified one that is based upon the assumption of a single inverted pendulum model, and does not account for factors such as the rotations of the hip and pelvis. These moment arms are shown by the vertical solid lines connecting the sternum position and the circular markers. Hollow markers indicate data prior to the perturbation, while filled markers show the data following the perturbation. The moment arm values for each perturbation type were averaged for each subject on a step-by-step basis. These values were then normalized using the mean values from the non-perturbation trials, in order to reduce the individual differences in moment arm magnitude between subjects. To normalize, we divided the mean value for the perturbation trial by the mean value for the non-perturbation trial on a step-by-step basis. The normalized, subject mean values were then averaged on a step-by-step basis over each subject population.

2.7. Mediolateral acceleration estimates

2.7.1. Heel strike calculation

The heel strike event during each gait cycle was estimated from the 40 Hz kinematic data at the instance when differences between the A/P shank velocities equaled zero. This event was determined to be an accurate representation of heel strike based on video analysis and force plate data from one step in the gait cycle. Heel strike sampling was reconfirmed on a step-by-step basis for each subject.

2.7.2. Head and sternum average acceleration calculation

The M/L head and sternum acceleration was estimated during the single support phase prior to the contralateral heel strike event. A smoothed, phaseless double differentiator was implemented using the MATLAB "Filtfilt.m" function, and yielded regularly sampled acceleration estimates (40 Hz) for the entire test run. From these, the mean M/L head and sternum acceleration values during single support phase for each perturbation type were calculated for each subject on a step-by-step basis. We averaged the data over the last two thirds of each support phase to obtain one single estimate per support phase. These values were then normalized for each subject on a step-by-step basis using the non-perturbation trials and were subsequently averaged separately over each subject population.

2.8. Statistical tests

We used a 2-way analysis of variance (ANOVA) to look for interactions among the responses of different steps and between the two subject groups. If an interaction was discovered, a post-hoc comparison of the corresponding steps between the two subject populations was then made using the Tukey honest significant difference (HSD) test. This adjusts the p values taking into account multiple comparisons. We set the adjusted significance level at 0.05 for rejection of the null hypotheses that there were no significant differences between groups or among steps All statistical analysis was done using Statistica.

3. Results

3.1. Response trajectory

The M/L trajectories of the sternum and shanks of a healthy subject in response to a large, forward-right perturbation applied to the right foot are shown in Fig. 2A. The M/L platform displacement is shown versus time as an inset on the bottom of Fig. 2A. The onset of the perturbation occurred in the support phase of the fifth step (marked as 5). This moved the right foot laterally, thus increasing the effective moment arm between the support foot and the center of mass of the body. The increase is shown, approximately, as the M/L distance between the sternum (e) and the shank [5]. This increase in the approximate moment arm, in turn, induced a leftward acceleration in the trunk, as indicated by the M/L sternum trace. To compensate for increased trunk acceleration, the subject increased the length of the moment arm of the contralateral foot during the next support phase, as approximately indicated by the M/L distance between the sternum (f) and the shank [6], and produced a peak to peak M/L sternum sway (f–e) of approximately 85 mm. The M/L moment arm then returned to baseline approximately 5 or 6 steps following the perturbation [11].

The response of a vestibulopathic subject to the large forward-right perturbation, shown in Fig. 2B, follows the same general trends as the control subject. Interesting differences include: 30% greater peak-to peak M/L sternum displacement (f–e), 110 mm for this subject compared to the control subject, and changes in the M/L distances between the sternum and the shank in response to the perturbation. Specifically, the change in the displacement (5–e) during the 5th step compared
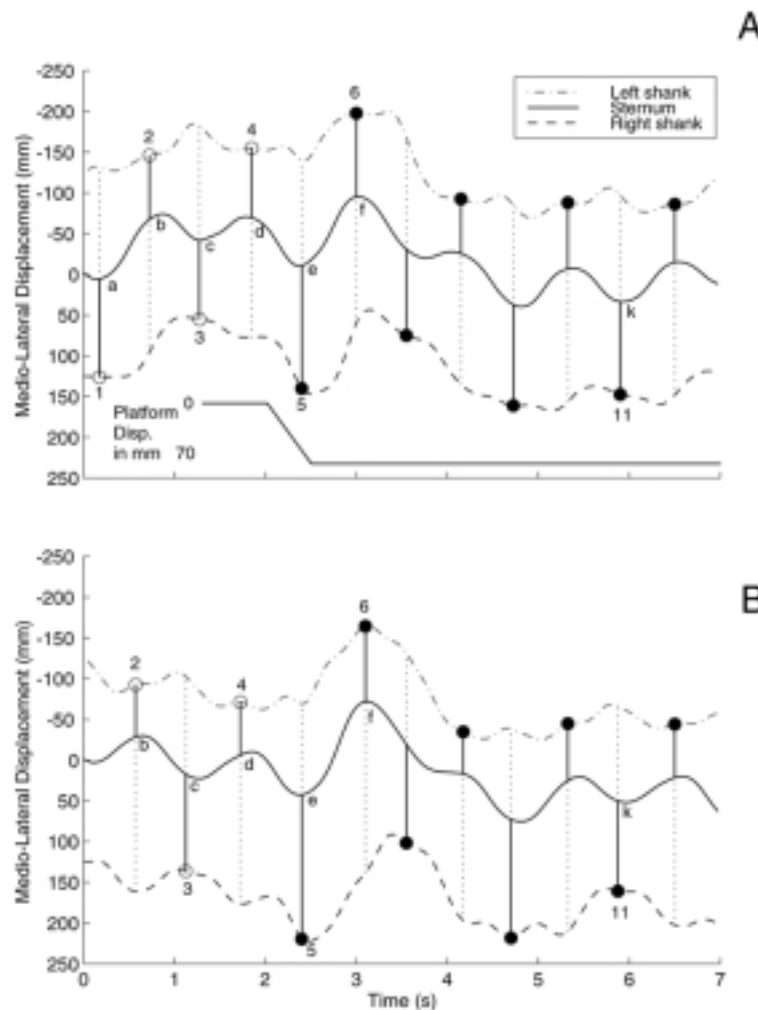
Fig. 2. Time series plots of M/L displacement of the sternum and shanks for a typical healthy (A) and vestibulopathic (B) subject during a large, forward-right perturbation trial. The M/L displacement of the platform used to perturb the foot is shown at the bottom of (A). The vertical, dotted lines indicate when the two shanks had equal A/P positions, which is when the time series data was sampled for analysis. The solid vertical bars indicated the step-by-step estimate of the moment arm between the sternum and the shank in stance that controls M/L motion of the body. Hollow circles mark pre-perturbation samples, and solid circles mark post-perturbation samples. The solid horizontal bar indicates the time over which the platform was translating. Numbers and letters label specific events that are referred to in the text.

to the 3rd step (3–c) is greater (about 33%) for the vestibulopathic subject than it is for the control. Similarly, the change in displacement during the 6th step compared to the 4th step is about 11% greater for the vestibulopathic subject versus the control.

A backwards-left perturbation (not shown) tends to move the support foot more nearly under the center of mass, thus decreasing the M/L distance between shank and sternum. This decreases the effective moment arm, which results in a decreased leftward acceleration of the trunk, so that the moment arm during the next support phase of the contralateral (left) foot would theoretically not need to be very large to counter the decreased

acceleration. This reaction is exactly what happens in the actual response

Four out of the eight vestibulopathic subjects tested in this current study took a quick, corrective step with the left (swing) foot in response to the large, forward-right perturbation. This response, which was only seen for the larger forward right perturbation, has been previously reported for the 3 subjects in the normal subject group [30]. To avoid mixing two fundamentally different responses, these subjects were analyzed separately from the four subjects that did not respond with a quick, corrective step in response to the 10 cm, forward-right perturbation. The other three perturbation types did not
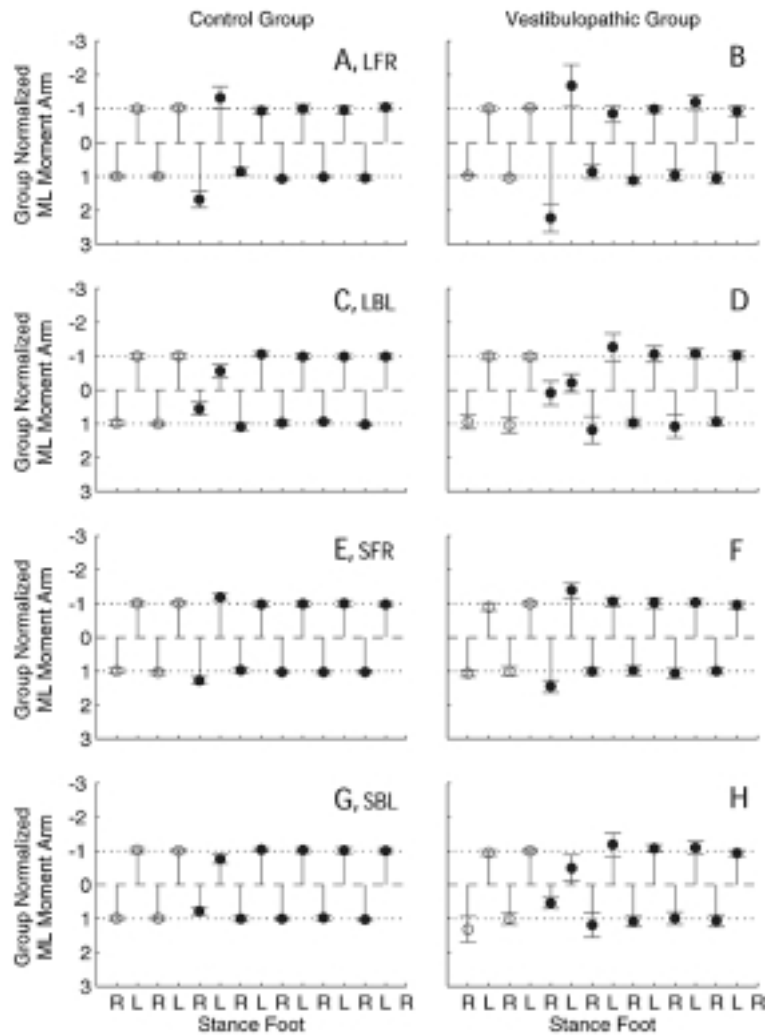
Fig. 3. Mean normalized moment arm and standard deviations for each perturbation type for the group of healthy subjects in the left-hand column, vestibulopathic subjects in the right-hand column. A) and B) Forward-right, 10 cm perturbation. C) and D) Rearward-left, 10 cm perturbation. E) and F) Forward-right, 5 cm perturbation. G) and H) Rearward-left, 5 cm perturbation. Hollow circles mark pre-perturbation samples, and solid circles mark post-perturbation samples Error bars show ± one standard deviation.

elicit these quick, corrective steps from any of the subjects. Thus, the quick step response for vestibulopathic subjects occurs in 4 subjects for one test condition. The remaining 87% of the runs had a response pattern of continued, regularly paced locomotion, and will be the focus of this study.

3.2. Amplitude of M/L moment arm response

Figure 3 shows the normalized, averaged moment arm responses to each of the four perturbation types for the control group in the left column of panels, and the vestibulopathic group in the right column of panels.

The responses to the large (10 cm) perturbations in the forward-right (FR) direction are shown in the first row of panels (A,B), and in backward-left (BL) direction are shown in the second row of panels (C,D). The responses to the small (5 cm) perturbations in the FR and BL directions are shown in the third (E,F) and fourth (G,H) rows of panels, respectively. Hollow circles show data before the perturbation was applied, and filled circles show the response after the perturbation. The direct effect of the perturbation occurs during the 5th step and is followed by the reactive response to the perturbation, beginning on the 6th step and continuing for several steps. For the FR perturbations, which moved the right stance foot forward and to the right, the ensuing left

Table 2
P values for differences in vestibulopathic and control groups' M/L moment arm, sternum acceleration and head accelerations

Perturbation type	M/L Response ANOVA p values		
	Moment arm	Sternum acceleration	Head acceleration
LFR	0.01835	0.23419	0.02258
LBL	0.00016	0.34122	0.55250
SFR	0.00069	0.93530	0.12221
SBL	0.00019	0.00555	0.05382

step was widened to counteract the leftward acceleration of the trunk initiated by the perturbation. Therefore, the FR perturbations led to increases in the M/L moment arm amplitudes for both the 5th and 6th steps (see Figs 3A, 3B). This behavior was seen in all but one subject for both magnitudes of the FR perturbations. The application of the BL perturbations, which moved the right stance foot backward and to the left, caused the ensuing left step to be brought inward to resist the rightward acceleration of the trunk. Consequently, the BL perturbations led to decreases in the M/L moment arm amplitudes for the 5th and 6th steps (see Figs 3C, 3D). All subjects displayed this behavior for the large, FR perturbations while only one subject did not for the small, FR perturbations. Larger perturbation magnitudes resulted in larger changes in moment arm values, for both perturbation directions and in both subject populations.

The results (Table 2) of a 2-way ANOVA and HSD test showed that the increases in M/L moment arm values caused by the direct effect of all perturbations were statistically greater during the first two steps after the perturbation. A step-by-step comparison of VP with controls using a Bonferoni corrected t test revealed significant (at $p < 0.05$ level) differences between groups for the first step following the perturbation for all 4 perturbations. For the next step, the difference between groups was significant for only the large-back left and the small-forward right perturbations.

3.3. Sternum acceleration

Figure 4 shows the mean of the normalized acceleration data of the sternum averaged across the healthy subject population in the left column of panels and across the vestibulopathic subject population in the right column. Note that the scaling on the abscissa is reversed to conform to an overhead view of the subject traveling in the horizontal plane from left to right (Fig. 1). This section refers to the heel strike sampled events. The perturbation occurred immediately (180–

200 ms) following the 5th step and therefore is not shown until the 6th step. The perturbation occurred following the 5th step, and the reactive response to the perturbation began on the 6 th step. The sternum acceleration magnitude initially increased for the forward-right perturbations, and initially decreased for the backward-left perturbations. The larger perturbations elicited larger magnitude changes. As in the case of M/L moment arm length, a 2-way ANOVA revealed an interaction between subject population and step number for sternum acceleration. A comparison of the sternum acceleration between the healthy and vestibulopathic subject populations, however, revealed no statistically significant differences in any of the steps following the perturbation, except for the small back left case.

3.4. Head acceleration

Figure 5 shows the mean of the normalized acceleration data of the head averaged across the healthy subject population in the left column of panels and across the vestibulopathic subject population in the right column. The perturbation occurred on the 5th step, and the reactive response to the perturbation showed up on the 6th step. The head acceleration magnitude initially increased for the forward-right perturbations, and initially decreased for the backward-left perturbations. The larger perturbations elicited larger magnitude changes, regardless of direction. There was a statistically significant difference in the first step following the perturbation for the large forward right case, and a borderline difference ($p = 0.053$) for the small backward left case.

3.5. Effect of subject age upon response estimates

Elderly subjects have been shown to have a decreased control of lateral stability [23]. Since the normal and vestibulopathic groups were not completely age-matched, we looked for correlations between sternum accelerations, head accelerations, and moment arm estimates with subject age, but found they were
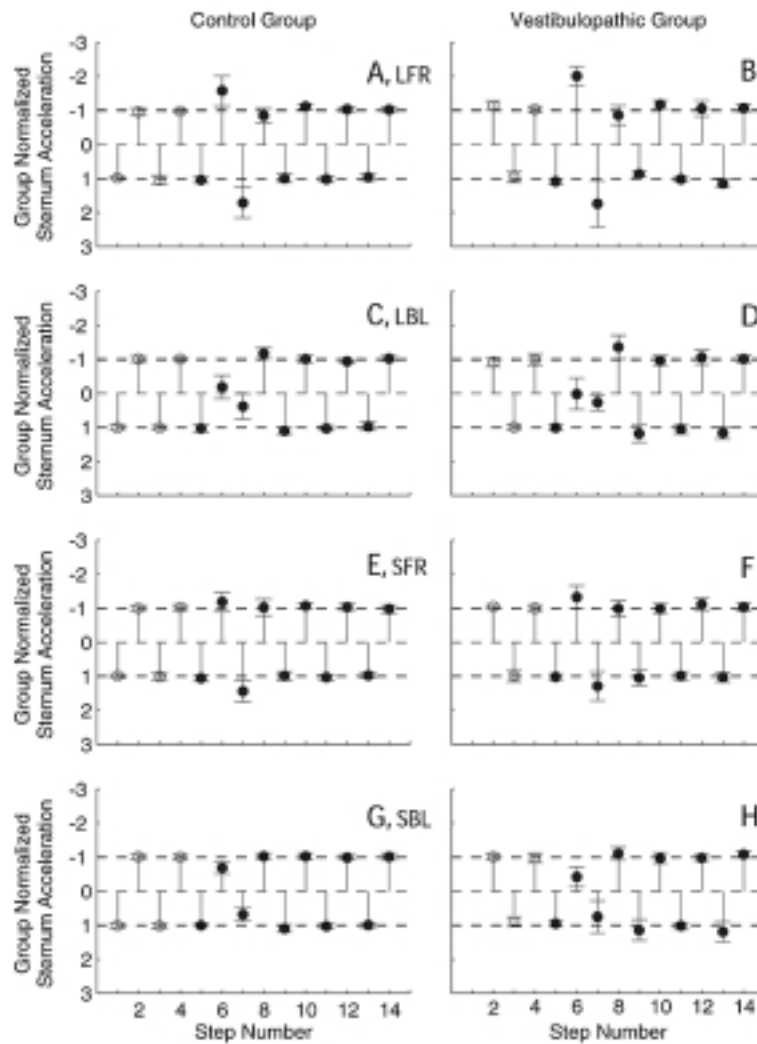
Fig. 4. Mean normalized M/L sternum acceleration and standard deviations for each perturbation type for the group of healthy subjects in the left-hand column, vestibulopathic subjects in the right-hand column. A) and B) Forward-right, 10 cm perturbation. C) and D) Rearward-left, 10 cm perturbation. E) and F) Forward-right, 5 cm perturbation. G) and H) Rearward-left, 5 cm perturbation. Hollow circles mark pre-perturbation samples, and solid circles mark post-perturbation samples.

not strongly correlated. Average R-squared values for normal subjects was 0.09 for the normal subjects and was 0.24 for the vestibulopathic subjects.

Elderly subjects have also been reported to have an increased M/L postural stiffness in quiet stance [5], to perturbations of quiet stance [1], and during locomotion [13]. The general trend is for younger subjects to exhibit more relative motion between torso and pelvis, while elderly subjects behave more like a single inverted pendulum. These changes in body roll could be indirectly reflected in the vertical velocity of the sternum and the head, using the large backward trials of control subjects and pooled VP and controls. (We used

this trial because we had responses from all 8 VP subjects.) The velocity parameters were the root-mean-square velocity for 10 steps of the trial (Vrms), and the peak maximum (Vmax), and peak minimum (Vmin), velocities that occurred in the step immediately following the perturbation. Using Pearson's correlation coefficient we determined the linear correlation for each of these three velocity parameters with subject age, and found only moderate R-squared values, Table 3. (Pooling VP with control subjects always gave smaller R-squared values compared to those using just controls.) Nonetheless, the trend was for increasing values of Vrms and Vmax with subject age and decreasing
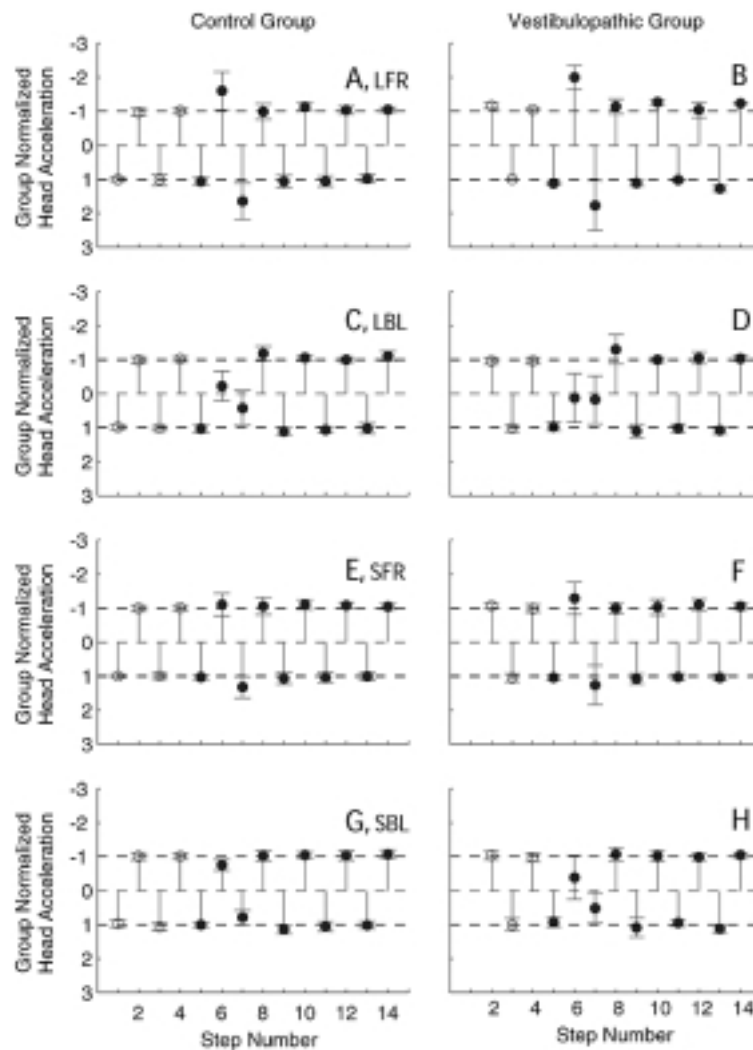
Fig. 5. Mean normalized M/L head acceleration and standard deviations for each perturbation type for the group of healthy subjects in the left-hand column, vestibulopathic subjects in the right-hand column. A) and B) Forward-right, 10 cm perturbation. C) and D) Rearward-left, 10 cm perturbation. E) and F) Forward-right, 5 cm perturbation. G) and H) Rearward-left, 5 cm perturbation. Hollow circles mark pre-perturbation samples, and solid circles mark post-perturbation samples.

values of Vmin with subject age for both the sternum and head velocities.

3.6. Response asymmetries due to side of lesion

Four vestibulopathic subjects had tumors removed from their right sides, while the remaining four had surgery on the left. We compared the normalized sternum accelerations, head accelerations moment arm data for the right sided tumor subjects with the left sided ones to look for possible left-right response asymmetries on a step-by-step basis for the first three steps following the perturbation, using a 2-way ANOVA. We

found no significant ($p < 0.05$) differences between the left and the right sided tumor subjects for any of the three steps.

3.7. Decay of M/L responses

The absolute value of the M/L moment arm responses to a surface perturbation resemble the impulse response of a linear, second-order system (Fig. 6). In order to compare the decay responses of the healthy and vestibulopathic populations to each of the four perturbation types, we calculated the mean and standard deviation (SD) of the absolute values of the responses

Table 3
Correlation of vertical velocity of the sternum and trunk with subject age

Parameter	Sternum velocity			Head velocity		
	Vrms	Vmax	Vmin	Vrms	Vmax	Vmin
Pooled R^2	0.000	0.06	0.05	0.04	0.04	0.12
Controls R^2	0.16	0.09	0.06	0.21	0.08	0.27

for the three steps prior to the perturbation. These are meant to estimate a pre-perturbation baseline. The "nominal range" of this baseline was set to have an upper limit of the mean plus 1 SD and a lower limit of the mean minus 1 SD.

The number of steps following the perturbation before the M/L moment arm length returned to and remained within the nominal region were then determined to quantify the decay response. This method was not highly sensitive to the actual size of the bounds as long as the region was not too large that it encompassed all oscillations, or too small that the oscillations never settled. The results of this analysis (Table 4) showed that the number of steps required for the complete decay of the moment arm responses to the large-FR, large-BL, and small-BL perturbations were greater for the vestibulopathic population than the healthy population. For sternum acceleration, the number of steps required for decay to near baseline for vestibulopathic subjects exceeded the number for controls in the large-FR and the small-BL perturbations. For head accelerations, this decay difference was apparent only for the large-FR perturbation.

3.8. Mean deviations of M/L responses following perturbations

Referring to Fig. 6, we calculated the mean of the absolute value of the deviation of the normalized moment arm responses, using just the steps following the perturbation, for each subject for each perturbation type. We refer to this as the mean absolute normalized deviation (MAND) Using 2-way ANOVA and the HSD test, we next compared the average MAND of the vestibulopathic group with that of the control group for each perturbation type. The MAND values between groups were significantly different ($p < 0.05$) for all four perturbation types. For sternum acceleration, the MAND values were significantly different between groups for all but the small forward right perturbation. For head acceleration, the MAND values between groups were not significant for any perturbation.

4. Discussion

The position and orientation of the trunk has been proposed to be critical to the maintenance of dynamic equilibrium [15]. Approximately two thirds of the mass of the body is contained within the head, arms, and trunk, thus making the trunk a good approximation of the CoM. Analysis of frontal plane kinematics during locomotion has shown that the CoM is in a perpetual state of M/L instability and oscillates within the base of support defined by the two feet [20]. The CNS is thought to use information about the M/L acceleration of the CoM during these oscillations to determine the appropriate positioning of the swing foot prior to heel strike [37]. Experiments of Powell, et al., that imposed several fixed step widths during non-perturbed gait illustrated a distinct linear relationship between the M/L acceleration of the CoM and the M/L position of the stance foot [37]. Vestibulopathic (VP) subjects exhibit poor control of the CoM during dynamic activities [27, 36]. Responses to surface perturbations of VP subjects have normal latency, but increased amplitude as compared to those of healthy subjects [14]. It stands to reason that if VP patients have difficulty correctly detecting M/L acceleration during a surface perturbation, then the corresponding M/L placement of the stance foot would be distinct from that of healthy individuals. Reduced vestibular gain has been used to explain increased A/P head accelerations in elderly subjects during locomotion [37].

We initially hypothesized that the amplitudes of the responses of the M/L moment arm length and M/L sternum and head accelerations to the various perturbations would be greater for the VP population than for controls. For the case of M/L moment arm, our hypothesis proved to be true for all 4 types of perturbations we applied. The amplitudes of the responses of the M/L sternum accelerations, however, showed a significant difference between the control and VP populations for only one type of perturbation. The M/L head accelerations showed significant differences between VP and controls for one type of perturbation and a borderline ($p = 0.053$) difference for one more. Nonetheless, there was an overall trend for mean VP normalized M/L head and sternum accelerations to show larger changes
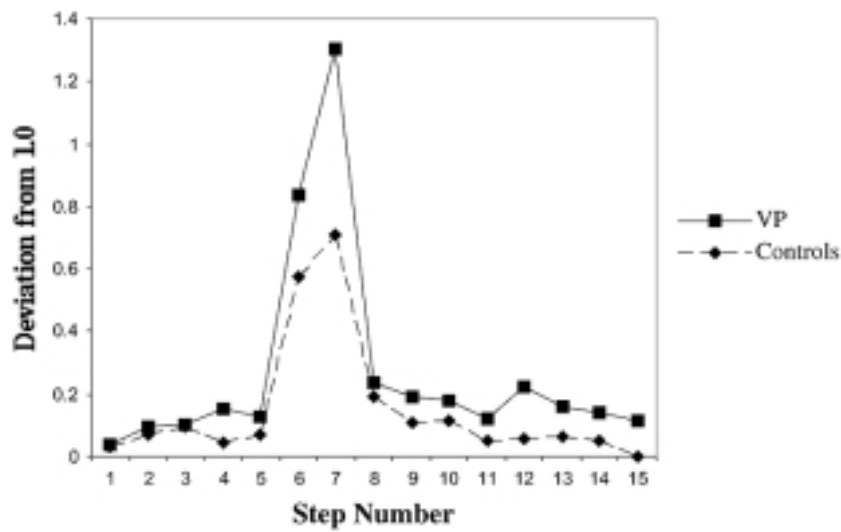
Fig. 6. Mean, Absolute, Normalized M/L Moment Arm Deviations for the Large, Forward-Right Perturbation.

Table 4
The number of steps required for the M/L responses to return within the normal-walking range for both subject populations and each perturbation type

| Perturbation Type | Number of steps for M/L responses to decay | | | | | |
| | Moment arm | | Sternum acceleration | | Head acceleration | |
	Control population	Vestbulopathic population	Control population	Vestbulopathic population	Control population	Vestbulopathic population
LFR	2	4 (9)*	3	4	5	11
LBL	2	4	5	4	4	3
SFR	2	2	2	2	5	2
SBL	2	4	2	4	2	2

*Note: a small peak breached normal limits for the 9th step (see Fig. 6, point labeled step #12).

in response to perturbations when compared to mean changes in the control group. Despite this trend, we do not think our results, as presently analyzed, support the original hypothesis. This may be due to experimental factors such as non age-matched VP and control subject groups. Maki reported age-related differences in subject's strategy in recovering to lateral perturbations while standing or waking in place [22,23]. Older subjects were more likely to take extra steps and also to use their arms more in attaining their equilibrium [25] Furthermore, Allum et al. [1] recently reported age-related differences in trunk roll stiffness following a support surface roll perturbation during quiet standing. Elderly subjects showed reduced trunk motion in the early passive part of the response compared to younger subjects essentially causing a larger instability in the elderly subjects for a given perturbation since the trunk followed the direction of the roll perturbation. We found a corresponding age effect in our subjects during gait perturbations as a weak but detectable correlation

between the vertical velocity of the sternum and head with age (Table 3) suggesting that elderly subjects may have a higher trunk stiffness in response to M/L perturbations. These differences in velocity or roll rate are somewhat more difficult to detect during locomotion, compared to standing because they are superimposed upon the periodic muscle activity pattern related to locomotion. This may account for the relatively small effects we observed. Nevertheless the findings are in agreement with those of others [1,5,13], and deserve further investigation in future experiments. Thus we provide a caveat that at least some of the differences between VP and control responses may be due to age effects. We are planning future experiments designed to address this issue. Other experimental factors include the possible need for a larger VP group to increase the statistical power, or a less-than-optimal choice for the moment arm sampling time (co-incident A/P shank displacement) for each step cycle.

An alternate explanation may be that the original model of Powell, et al., who studied small, steady-state

changes in M/L moment arm in normal subjects does not apply to the present experiment. In Powell's study, M/L moment arm was strongly correlated with M/L sternum acceleration. But, our abrupt perturbation may evoke a different strategy that is designed to control M/L sway via changes in M/L moment arm while simultaneously maintaining the head as a stable platform to provide clear vision.

The acceleration sensors in persons with unilateral vestibular function (presumably the otolith organs of one inner ear) should have enough motion sensitivity to control both these tasks. Indirect evidence to support this is that the angular VOR gains in our VP subjects range from low normal to normal (0.49 to 0.86). We speculate that a unilateral lesion may not significantly reduce the sensitivity to motion, but instead may increase the measurement noise, resulting in a relatively larger error in estimating the true acceleration. One effective strategy for these subjects would be to slightly increase their corrective moment arm in response to a perturbation to insure against making too small an adjustment that might require a disruptive maneuver, such as a quick corrective step. This strategy would allow them to maintain a normal, rhythmic pattern of walking, albeit with larger transient deviations in their M/L moment arms compared to normals. Bauby and Kuo have shown theoretically, that the deviation away from the normal state in response to a perturbation is related to upon the measurement error due to sensor noise, during lateral control of walking [3]. An increased error produces an increased deviation.

Additionally, we found that the number of steps required for the response of the M/L moment arm length to decay back within nominal range was greater in the cases of the large-FR, large-BL, and small-BL perturbations. Our mean absolute normalized deviation (MAND) results point to an increase in the average absolute "error" that occurs in VP subjects after the perturbation. These findings support the concept for an increased measurement noise in the estimated M/L acceleration signal. They also agree with experimental results which show an increase in variability in subjects' lateral foot placement during locomotion when tested eyes closed versus eyes open [3]. This speculation of reduced sensor noise does not definitively rule out the possibility that the response changes in the vestibulopathic subjects are just due to a decrease in the sensitivity or "gain" of the motion input. In our opinion, however, this would logically lead to vestibulopathic subjects making smaller corrections relative to controls, which is the opposite from our results.

Another aim of this study was to provide further insight into the utility of the surface perturbation paradigm and M/L stability metrics developed by Oddsson, et al., to quantify locomotor stability and gauge vestibular rehabilitation techniques. Vestibulopathic subjects were tested using this paradigm so their results could be compared to those of the healthy subject population. Significant differences between VP and control group M/L moment arm response, especially considering the VP group had normal SOT scores in computerized dynamic posturography, further support the potential utility of this paradigm for clinical balance testing and for evaluation of space flight crew after exposure to microgravity.

Finally, subject #8 said she had been avoiding slippery surfaces and challenging walking tasks ever since her surgery 10 years ago. She stated that she was able to discover "survival strategies" to this challenging situation. She said she felt much more comfortable about moving around after participating in the perturbation protocol, than before. This anecdotal evidence suggests a possible rehabilitative role for perturbations of foot position during locomotion to train for prevention of falls during slips.

Acknowledgements

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