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Trunk sway measures of postural stability during clinical balance tests: effects of a unilateral vestibular deficit

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Abstract

This research evaluated whether quantified measures of trunk sway during clinical balance tasks are sensitive enough to identify a balance disorder and possibly specific enough to distinguish between different types of balance disorder. We used a light-weight, easy to attach, body-worn apparatus to measure trunk angular velocities in the roll and pitch planes during a number of stance and gait tasks similar to those of the Tinetti and CTSIB protocols. The tasks included standing on one or two legs both eyes-open and closed on a foam or firm support-surface, walking eight tandem steps, walking five steps while horizontally rotating or pitching the head, walking over low barriers, and up and down stairs. Tasks were sought, which when quantified might provide optimal screening for a balance pathology by comparing the test results of 15 patients with a well defined *acute* balance deficit (sudden unilateral vestibular loss (UVL)) with those of 26 patients with less severe chronic balance problems caused by a cerebellar-pontine-angle-tumour (CPAT) prior to surgery, and with those of 88 age- and sex-matched healthy subjects. The UVL patients demonstrated significantly greater than normal trunk sway for all two-legged stance tasks especially those performed with eyes closed on a foam support surface. Sway was also greater for walking while rotating or pitching the head, and for walking eight tandem steps on a foam support surface. Interestingly, the patients could perform gait tasks such as walking over barriers almost normally, however took longer. CPAT patients had trunk sway values intermediate between those of UVL patients and normals. A combination of trunk sway amplitude measurements (roll angle and pitch velocity) from the stance tasks of standing on two legs eyes closed on a foam support, standing eyes open on a normal support surface, as well as from the gait tasks of walking five steps while rotating, or pitching the head, and walking eight tandem steps on foam permitted a 97% correct recognition of a normal subject and a 93% correct recognition of an acute vestibular loss patient. Just over 50% of CPAT patients could be classified into a group with intermediate balance deficits, the rest were classified as normal. Our results indicate that measuring trunk sway in the form of roll angle and pitch angular velocity during five simple clinical tests of equilibrium, four of which probe both stance and gait control under more difficult sensory conditions, can reliably and quantitatively distinguish patients with a well defined balance deficit from healthy controls. Further, refinement of these trunk sway measuring techniques may be required if functions such as preliminary diagnosis rather than screening are to be attempted.

Keywords: Clinical balance tests; Unilateral vestibular loss; Cerebellar pontine angle tumours; Balance disorders; Trunk sway; Screening balance disorders

1. Introduction

The need for a simple quantified test to screen for balance disorders is widely recognized. Several authors have suggested that scoring a combination of stance and gait tasks might fulfil this need [1–4]. To date, however, there are no criteria for selecting which tasks

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and which measurements provide optimal screening of balance disorders such as those resulting from an inner ear or vestibular nerve deficit.

Patients with peripheral vestibular deficits often show instability during stance tasks, particularly following the acute onset of the deficit. Making the stance task more difficult by removing visual inputs or reducing the efficacy of lower-leg proprioceptive inputs using servo-control to destabilize the support surface has been claimed as a means to identify a vestibular deficit [5–8]. A number of authors have, however, suggested that the original technique described by Nashner et al. [5] is not effective in identifying balance disorders [9,10], is not related to a patients dizziness handicap [11], or could be just as effectively tested using a foam mat as the unstable support surface, rather than a mechanically driven one [12]. Such problems are not unexpected with Nashner's technique [5] given that the measurements of the forces imposed on the support surface were only made in the pitch plane and designed to estimate the pitch angular sway of the body. It is well known that the sway of patients with a vestibular deficit is mainly side to side (in the roll plane) and they fall to the side of the deficit. Instability in other directions such as the fore-aft (pitch) plane are also seen [13]. Stance tests appear to achieve an improved level of identification of a balance disorder in the elderly only when force plate measurements in the roll direction as well as video motion analysis of the body are added to the measurement battery [14]. However, motion analysis systems are expensive to purchase and need trained personnel to operate, thereby defeating the purpose of screening: cheap and easy to operate.

Another method of testing balance is to use one-legged stance or tandem walking. Both tasks reduce the base of support in the roll direction. Roll instability is seen more clearly when performed with eyes closed. Under these circumstances, there is a greater reliance on vestibular inputs and subjects with vestibular deficits have greater difficulty with this type of task. For this reason these tasks feature in task batteries focussing on sensory inputs [15] where, for example, the time a patient can stand on one-leg eyes-closed is timed [16]. Time standing on one leg is not just a criterion for testing the sensory contributions to balance stability. Several authors have documented the relationship of quadriceps muscle strength to the duration of one-legged stance, with a decrease in duration occurring particularly in institutionalized women and other elderly prone to fall [17–19]. Thus, reliance on single leg stance tests to screen for a vestibular deficit may lack specificity unless it can be shown that the characteristics of sway movements are fundamentally different for vestibular deficit patients and those with weak quadriceps muscles.

Gait performance, specifically gait velocity, has been described as a gold standard for assessing balance and functional performance in vestibular deficit subjects [10]. A number of authors have recommended using several gait tasks such as tandem walking, gait with horizontal head rotation and gait with eyes closed to assess balance control [16,20]. Deviations in gait from a straight path, task duration and loss of balance are the usual qualitative means of scoring these tasks. It is not known whether, when quantified, these gait tests could reliably distinguish vestibular deficit subjects from normals. Recent findings [21] have in fact suggested that presence of neural pacemakers underlying gait help provide more stable posture control during gait for vestibular deficit subjects than that observed when standing. In summary, it is an open question which components of standard clinical stance and gait tasks best provide screening of balance deficits.

The purpose of the current study was to determine a set of trunk sway measurements and balance control tasks which could provide a quantified screening battery for a balance deficit. By measuring the angular deviations of the trunk directly, without relying on indirect measures from force-plate measurements embedded within a support surface, we were in a position to make direct comparisons between the amplitudes of trunk instabilities noted in the same subject for both stance and gait tasks. By including task duration as a measurement, we were also in a position to weight duration relative to trunk sway variations as a screening measurement. Our use of a system, which measured angular deviations of the trunk directly, avoided both the limited accuracy and frequency band-width of angular measurements derived from video linear-motion analysis systems and assumptions about whether the body moves as an inverted pendulum during body sway [22–24] or not. Our conclusion that two-legged stance and gait tasks are needed to optimally screen for balance disorders builds on earlier suggestions that both types of tasks are required for this purpose [10,16] and focuses attention on those trunk sway measures which are most sensitive to a balance disorder.

2. Methods

2.1. Subjects

Two groups of vestibular deficit subjects were enrolled in this study. One group consisted of 15 vestibular loss patients admitted on an emergency basis with symptoms typical of an acute unilateral peripheral vestibular deficit (UVL). The symptoms were nausea, a spontaneous nystagmus beating towards the healthy ear, and a falling tendency towards the deficit side which was established by eye movement responses (or lack thereof) to bithermal caloric irrigation of one of both irrigated ears. Subsequent examinations, that is a complete electronystagmyographic examination and in cases with an asymmetric hearing loss, recording of auditory brainstem potentials, established that these patients were suffering from an acute peripheral vestibular deficit with a canal paresis (CP) value greater than 50% in the caloric examination (range 56–100%, mean 83%) and vestibular-ocular reflex responses to horizontal whole body rotation with abnormal asymmetry [25]. Patients within this group were tested within 5 days of their admittance as an in-patient. The second group of 26 patients consisted of those with a confirmed unilateral cerebellar-pontine-angle-tumour (CPAT). The diagnosis was made on the basis of magnetic resonance imaging (MRI) generally after recordings of auditory brainstem potentials showed pathological response latencies. This group of patients was tested 1–3 days prior to surgery. At that time, their tumour size estimates from the MRI ranged from 8 to 50 mm (mean 24 mm) for the maximum diameter.

Normal subjects were selected from a pool of healthy subjects from whom we had previously acquired data [26]. Selection criteria for matching subjects with patients were based on having the same age and gender as each of the patients and yielded a total of 88 subjects from our pool of 250 normal subjects. All normal subjects were free of any known disorder that could influence balance control including orthopaedic, vestibular or somatosensory disorders. Each subject provided witnessed informed consent prior to carrying out the experimental tasks.

2.2. Measurement system

Trunk sway was measured with two digitally-based angular-velocity transducers (SwayStar system developed with Nicolet Biomedical Inc) mounted onto the hardened part of an elasticized motor-cycle belt which normally provides the wearer a lower back support. By inverting the belt so that the hardened part was directed up the lower back, it was possible to measure trunk angular deviations at the level of the lower back (lumbar 2–3) without restricting movement of the hips and pelvis with respect to the trunk. One transducer was oriented to measure angular velocity deviations in the roll (side-to-side) plane and the other measured angular velocity deviations in the pitch (fore-aft) plane. Angular deviations were calculated on-line using trapezoid integration of angular velocities. Because, the maximal task duration was 20 s and baseline drift of the transducers was 0.01 deg/s (1 S.D.), measurement error had a standard deviation of maximally 0.2 deg. Angular velocity values were transferred to a PC interface at rates of 102.4/s with 16 bit accuracy over the range of ± 256 deg/s via a 10 m cable. The long cable enabled the subject to move freely during all tasks.

2.3. Measured tasks

All tasks were performed without shoes. The tasks could be classified into four categories.

- *Two-legged stance tasks* (for a duration of 20 s or until the subject had to be prevented from falling by spotters) were performed either eyes open with and without a foam support surface, or eyes closed, again with and without a foam support surface. The subject was asked to stand naturally, not with the feet together, in order to test the roll stability of normal stance. The foam used had a height of 10 cm, length 204 cm, width 44 cm and a density of 25 kg/m^3 (40% DIN 53577).
- *One-legged stance tasks* had a maximum duration of 20 s or until the subject touched the ground with the non-support leg. If a touchdown of the non-stance leg occurred before 20 s, the trial was repeated and the trial with the longest duration used for analysis. These tasks were performed either eyes open with and without a foam support surface, or eyes closed on a normal support surface.
- *Semi-stance tasks* comprised walking eight tandem steps (looking at the feet if wished). Two types of support surface were used: foam and a normal floor. Duration was the time to complete the task or until a fall occurred.
- *Gait tasks*. These included three walking trials without obstacles. Walking approximately five steps (over 3 m) with the eyes closed, walking 3 m while horizontally rotating the head in rhythm with stepping, and walking 3 m while pitching the head in rhythm with stepping. Two gait trials with obstacles were also employed: Walking over a set of four low (24-cm high) barriers placed 1-m apart. Walking up a set of stairs with two upward and two downward steps. Step height was 23 cm. The stairs had no handrails. Trial duration was the time to complete the task.

2.4. Data analysis

Five measurements were calculated from the data collected during each task. These were the total task duration and the peak-to-peak excursions in the roll and pitch directions for both angular displacement and angular velocity. Before the peak-to-peak movements were calculated data from the first 1 s of each task was excluded from consideration to avoid initial stabilising movements entering the calculations, for example, during the onset of one-legged stance. Data from the last 2 s of tasks was also excluded. If the task duration was < 20 s, only the last 10% of data were excluded. This was done to exclude the effects of falls entering the measurements other than as a shortened task duration. Differences in the population means were first examined for each task using ANOVA techniques after the measurements had been logarithmically transformed to yield a Gaussian-like distribution of measurement values [26]. Before transformation the population distributions of the measurements were more Poisson-like. Once a measure was found to be significantly different for the populations, differences between the individual population means were explored with Bonferroni tests. The level for significance for both the ANOVA and the Bonferroni tests was set at $P < 0.05$.

To rate the significant differences in population means in the order of importance in classifying the severity of the balance deficit, a stepwise linear discriminant analysis was performed using the measurement variables [27]. The variables that were entered into the analysis were those found to be significant for the ANOVA analysis (Table 1). Stepwise linear discriminant analysis as a first step repeats the ANOVA on the individual variables and then selects the variable with the highest F value (clearly, pitch velocity for standing on two legs eyes closed on a foam support; Table 1) to enter the analysis. Before the next ANOVA is repeated on the remaining variables, the covariation of the entered variable explaining the variation across populations in the remaining variables is first removed. This procedure is repeated until the F values of the remaining variables are no longer significant (we used a level of $P = 0.05$). Finally, ANOVAs are repeated on the entered variables to ensure that only variables with a significance greater than $P < 0.05$ remained. A variable among the entered variables was removed if it failed to satisfy this criterion. Of the 40 variables entered into the discriminant analysis five were used as classification discriminators (Fig. 5). Subjects were then classified into the group the subject's individual values of the classification variables lay closest to. The closeness was measured in terms of the distance of the subjects' individual values in multi-dimensional space to that of the population mean. The accuracy of the classification was inversely proportional to the square of the distance to the population mean that is near values yielded a high probability of a correct classification.

3. Results

Many of the tasks which have been specifically designed to be more difficult for subjects with vestibular loss lead to clear significant differences in trunk sway with respect to normals (Table 1). Specifically tasks that involved reducing or disturbing additional sensory inputs (eye closure or a foam support surface) produced large sway magnitudes. Some of the tasks, though, such as standing on one leg eyes closed, proved near to impossible for unilateral vestibular loss (UVL) subjects to perform. Surprisingly, tasks like walking over a set of barriers or up and down stairs could be performed reasonably well by the patients.

3.1. Stance tasks

Fig. 1 shows typical examples of the trunk sway recorded for the most unstable two-legged task of standing on a foam support with eyes closed. It could have been predicted from previous studies [5,12,14], where the efficacy of lower-leg proprioceptive inputs has been reduced, or decorrelated with upper body sway, that vestibular loss subjects would be considerably more unstable than normals for this task. Figs. 1 and 2 show that trunk sway was less for the patients

Table 1
Significant F statistics and levels of significance (in parentheses) observed when testing for differences between UVL, CPAT and normals (tests with only duration as a significant variable are not listed)

Task	Duration	Roll angle	Pitch angle	Roll velocity	Pitch velocity
s2eo	–	7.09 (0.0012)	3.28 (0.0408)	–	–
s2eom	3.94 (0.0219)	3.68 (0.0279)	5.28 (0.0063)	4.20 (0.0172)	6.83 (0.0015)
s2ec	3.97 (0.0214)	4.13 (0.0184)	4.66 (0.0111)	4.25 (0.0164)	11.02 (0.0001)
s2ecm	13.14 (0.0001)	17.32 (0.0001)	17.06 (0.0001)	15.88 (0.0001)	27.73 (0.0001)
s1eo	10.81 (0.0001)	–	5.37 (0.0058)	4.50 (0.0130)	5.46 (0.0053)
s1eom	4.67 (0.0111)	4.68 (0.0111)	5.23 (0.0066)	3.66 (0.0285)	5.24 (0.0041)
s1ec	18.78 (0.0001)	–	–	–	–
w8tan	3.62 (0.0295)	–	3.32 (0.0395)	–	5.25 (0.0064)
w8tanm	–	11.06 (0.0001)	9.49 (0.0001)	5.50 (0.0051)	9.48 (0.0001)
w5ec	5.26 (0.0064)	–	5.06 (0.0077)	3.83 (0.0243)	11.84 (0.0001)
w5hr	5.61 (0.0047)	–	7.52 (0.0008)	3.54 (0.0319)	9.19 (0.0002)
w5hp	4.04 (0.0200)	3.58 (0.0308)	5.49 (0.0052)	4.92 (0.0088)	8.33 (0.0004)
stairs	3.55 (0.0316)	4.66 (0.0112)	5.68 (0.0043)	–	6.61 (0.0019)
barriers	5.06 (0.0077)	–	–	–	–

s means standing, 2 on two legs, 1 on one leg, eo eyes open, ec eyes closed, m on foam support, w means walking, 8 eight, 5 five steps, tan tandem, hr means head rotating, and hp head pitching.

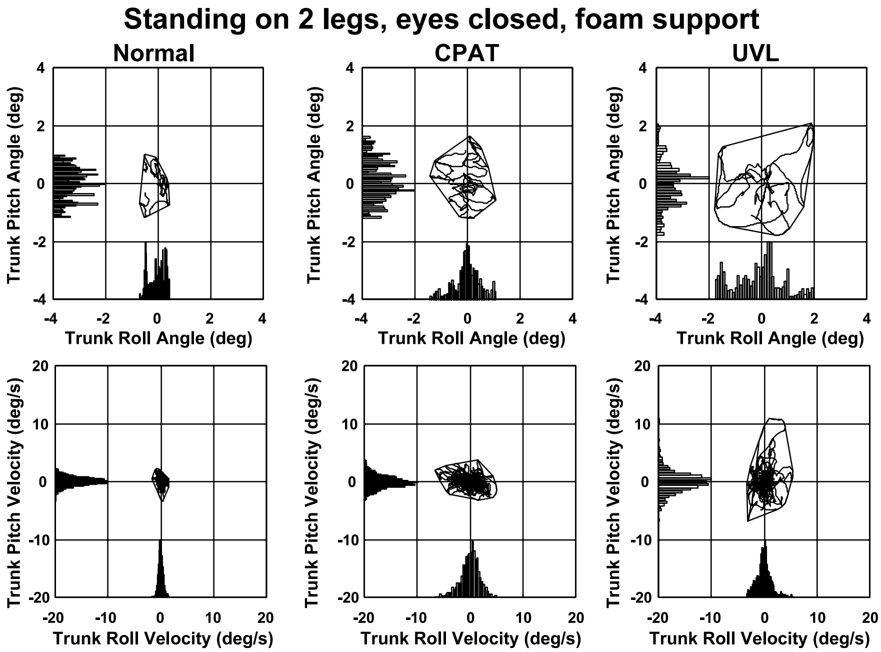


Fig. 1. Representative examples of trunk sway recorded from a normal, a UVL and a CPAT subject during two-legged stance on a foam support surface with eyes-closed. The displayed ranges of sway are those of a subject with values closest to those of the population median value. The upper set of traces is for the angle displacements, the lower for angular velocity deviations. Rightward roll and forwards pitch is plotted positive, along the abscissa and ordinate, respectively. Along each axis of the plot a histogram of roll and pitch angle and velocity values is plotted based on the total displacement range being divided into 40 bins. Each sample of trunk angle (or angular velocity) was then assigned to the appropriate pitch and roll direction histogram bins as a single count for the respective histograms. For plotting purposes the bin with maximum count was scaled to 1/4 the width of the inserts shown in the figure. Depending on the closeness of the bins and printer resolution the individual histograms appear black or black and white in the figure. The envelope of the angle and velocity values (or convex hull) is also shown. The duration of the plots is 20 s in all cases. Note the progressive increase in pitch and roll sway amplitudes from left to right.

with a chronic vestibular deficit compared to those with an acute deficit. That is the CPAT patients had more sway than that of normals, but less than that of UVL subjects. All four of the trunk sway measurements and task duration showed this effect for the task of standing eyes closed on a foam support, however, the peak-to-peak trunk roll and pitch angle were significantly different from normal across all two-legged stance tasks (Fig. 2; Table 1). The eyes-closed two-legged stance task on foam caused UVL subjects to fall most. Thereby the mean duration of stance was significantly shorter than that of normal subjects. The mean durations were 16.1 s (standard error of mean (SEM) 1.6 s) and 19.9 s (SEM 0.16 s) for UVL, and normal subjects, respectively. Some UVL subjects also fell before 20 s for the eyes open foam and eyes closed normal surface two-legged stance task. For these tasks the difference in task duration with respect to normal and CPAT subjects was also significant (Table 1). Although not as pronounced as the results on the foam support, eyes-closed task, other two-legged stance tasks also showed differences in the amount of trunk sway observed for normal subjects. Standing on a normal surface, eyes closed, revealed that all five measured variables were different across the three populations (Table 1). Both roll and pitch angles were significantly different across populations even while standing eyes open on a normal surface. Across all two-legged stance tasks, it was the amplitude of pitch angular velocity and roll angle, which provided the clearest indication of greater trunk sway following vestibular loss.

For one-legged stance the amplitudes and velocities of trunk sway measured from vestibular loss subjects prior to a fall were significantly greater than that of normals, except for eyes-closed trials. This latter lack of significance may have resulted from our exclusion of the falling component of the sampled data from the analysis. The duration of one-legged stance was however considerably shorter when visual inputs were not available (eyes closed conditions). The *F* value was 18.8 ($P < 0.0001$) for the comparison of mean durations. The mean duration of one-legged eyes closed stance was 2.8, 9.6, and 13.2 s for UVL, CPAT and normal subjects, with SEMs ranging from 0.4 to 1.0 s. The very short duration the UVL subjects could maintain one-legged stance eyes closed (almost all of them fell towards the side of the deficit) meant that once we excluded the data of the fall, little of the time course of standing remained to analyse. This raised the issue of how to score a fall when it occurred other than as a reduced duration (Section 4).

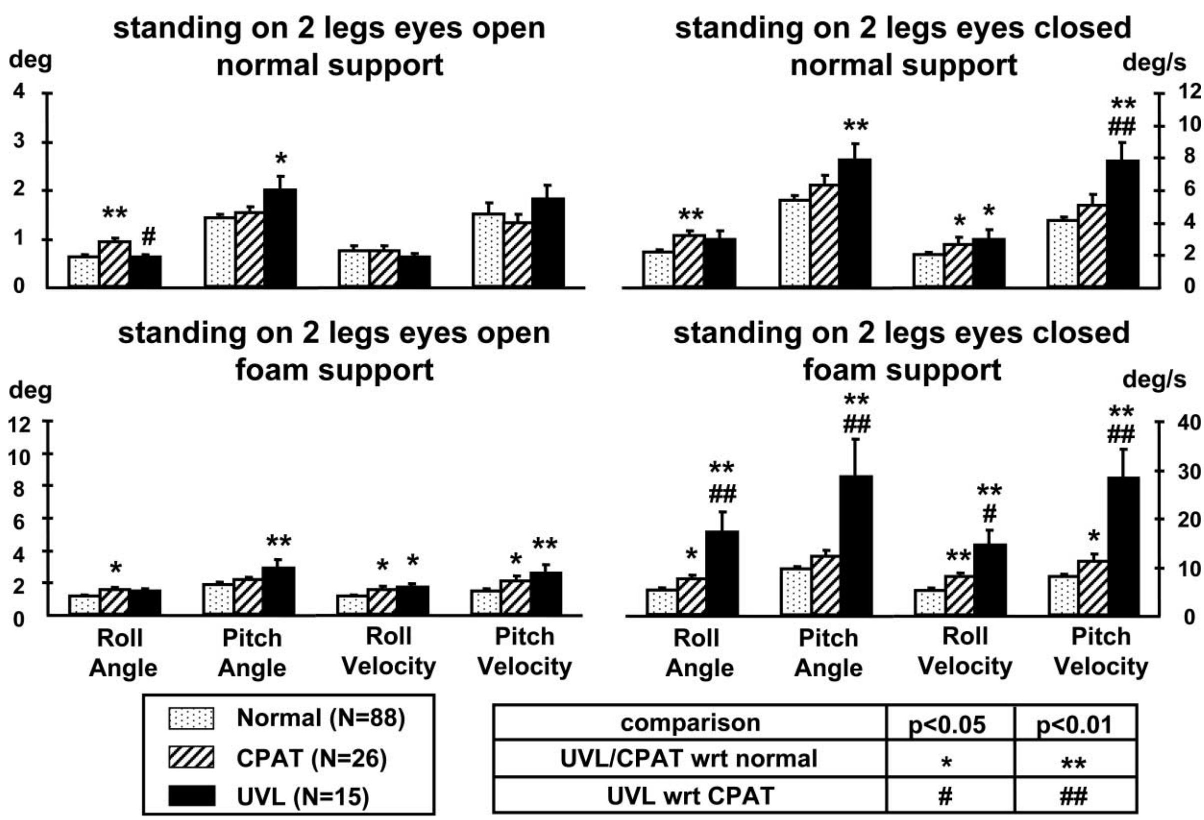
3.2. Gait tasks

Walking tasks that involved more difficult sensory and motor control conditions lead to an increase in trunk sway in vestibular deficit subjects compared to that of normals. Fig. 3 shows some examples of this for the task of walking five steps while rotating the head from left to right in the horizontal plane. The examples in this figure were selected from subjects with sway values closest to that of the median value of the population means. As these example plots show the UVL subjects had larger trunk sway for all four measures with pitch angle and angular velocity being the most significant (Table 1). The example plots in Fig. 3 and plots of means and standard errors in Fig. 4 indicate there was no significant difference between the trunk sway amplitudes of the CPAT patients and normal subjects. CPAT patients required only a longer duration for the walking with head rotation task. In fact trunk sway measures for almost all of the gait tasks showed little difference between normal and CPAT subjects. The exceptions were generally restricted to task durations which were longer for CPAT subjects.

The trunk sway during the gait tasks was significantly greater in UVL subjects for almost all gait tasks except walking over barriers. Thus, for walking five steps eyes closed, pitch and roll velocity as well as pitch angle were greater in magnitude in UVL patients than normals (Fig. 4). Walking five steps with eyes closed also yielded increases in the same variables for UVL patients with respect to CPAT patients, but not between CPAT patients and normals. In addition, the amplitude of pitch and roll velocity and angle was greater than normal in UVL patients when walking five steps while pitching the head up and down (Table 1; Fig. 4). Also the duration was longer.

For the semi-gait tasks of walking eight tandem steps on a foam support the UVL patients were quite unstable with roll angle and pitch velocity being considerably larger than normal. Again the values of CPAT patients had magnitudes in between those of normal and UVL patients, however, the size of the population variances caused a lack of statistical significance for the CPAT means with respect to normal but not with respect to UVL subjects.

The task of walking up and down stairs also led to increased sway in UVL patients with respect to nor-



comparison	p<0.05	p<0.01
UVL/CPAT wrt normal	*	**
UVL wrt CPAT	#	##

Fig. 2. Population sway ranges for two-legged stance tasks following a vestibular deficit. The columns have a height equal to the mean peak-to-peak sway of the normal, CPAT and UVL populations. The standard error of each mean is indicated by a vertical bar on the column. The scales for the columns are shown on the left for trunk angle and on the right for angular velocity. Significant differences between the population means (after subject values have been logarithmically transformed) as determined by Bonferroni t-tests are indicated at the foot of the figure. Note that the most significant differences are obtained for the task of standing eyes closed on the foam support surface.

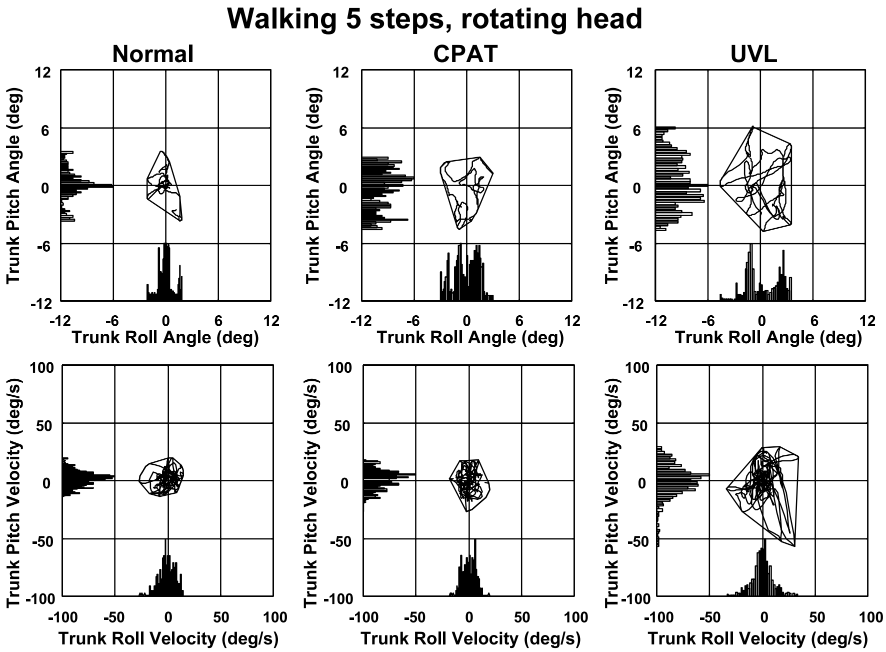


Fig. 3. Typical examples of trunk sway recorded as x–y plots for normal, CPAT, and UVL subjects for the task of walking five steps with concurrent head rotation. Details of the plots may be found in the legend to Fig. 1. The task durations are approximately 5 s, with the CPAT and UVL subjects having longer durations. Note the increase in the range of roll amplitudes in the order normal, CPAT and UVL subjects.

mals. However, the task of walking over barriers led only to a marginal increase in sway in the vestibular deficit patients, without statistical significance being reached (Table 1). The CPAT patients did, however, have significantly longer durations for the barrier task (8.5 s (SEM 0.6) compared to 7.2 s in normals).

3.3. Ranking the significance of the measured variables

The stepwise discriminant analysis performed on 40 most significant variables of Table 1 produced an interesting mix of tasks variables for classification purposes. Thereby we could confirm the efficacy of some parts of the task structure originally developed by Shumway-Cook and Horak [4] to distinguish vestibular loss patients from normals. Fig. 5 shows these variables and ranks them in order of their significance when entered into the classification vector. The mix of tasks is interesting because two-legged stance is represented for both conditions with reduced sensory inputs (eyes closed with and without a foam support surface), and with normal sensory conditions. Gait is represented by the task performed under the sensory conditions which can be expected to be difficult for UVL patients with primarily a horizontal semicircular canal deficit [28], that is those tasks with head rotations and on a foam surface. The mix of variables shows that trunk pitch velocity and roll angle appear to be most useful in identifying vestibular deficit patients thereby following the trends seen in Table 1. However, durations also feature, namely the longer duration for walking five steps with head rotation was the third most significant classification variable. The most significant classification variable was, as expected from Table 1, the velocity of trunk pitch when standing eyes closed on the foam support surface.

The classification accuracy achieved with the five variables listed in Fig. 5 is illustrated by an insert to the figure. Almost all (97%) of the normals were classified accurately. The UVL patients were separately classified with an accuracy of 93%. The classification accuracy for CPAT patients was much lower, 54%. The other 46% of these patients were classified as normal. Interestingly false classifications of normal and UVL patients was as CPAT patients. In this respect the quantified CTSIB protocol as reduced here to five tests for classification purposes seems to identify both normal and acute vestibular deficit patients very accurately for a screening procedure. The screening appears more likely to produce false negative for subjects with borderline balance problems as typically observed with CPAT patients. Furthermore, the underlying basis the screening procedure adopts, fits clinical observations that CPAT patients have minor balance symptoms compared to the major balance problems of acute UVL patients.

4. Discussion

This study has shown that by using trunk roll and trunk pitch velocity measurements during a combination of five stance and gait tasks it is possible to identify vestibular deficit subjects with an acute deficit separately from normals and from those with a chronic deficit as occurs for CPAT patients. The accuracy of individual classifications within the UVL and normal groups was very high with almost all subjects having high probability of a correct assignment. With the same tasks, those with a borderline balance deficit such as experienced by CPAT patients, were just as likely to be identified as normal than as CPAT patients indicating that further refinement of the technique is necessary if the technique is to be used for more than just screening purposes. The borderline 12 out of 26 CPAT subjects had probabilities near 0.4 and 0.6 of being classified as normal or as a CPAT patient. Thus, using trunk sway measures from the two stance and three gait tasks separates those with clear abnormalities from normals well, but the same technique tends to classify those with borderline balance problems either as such or as normal, suggesting the screening approach is conservative concerning false positive assignments.

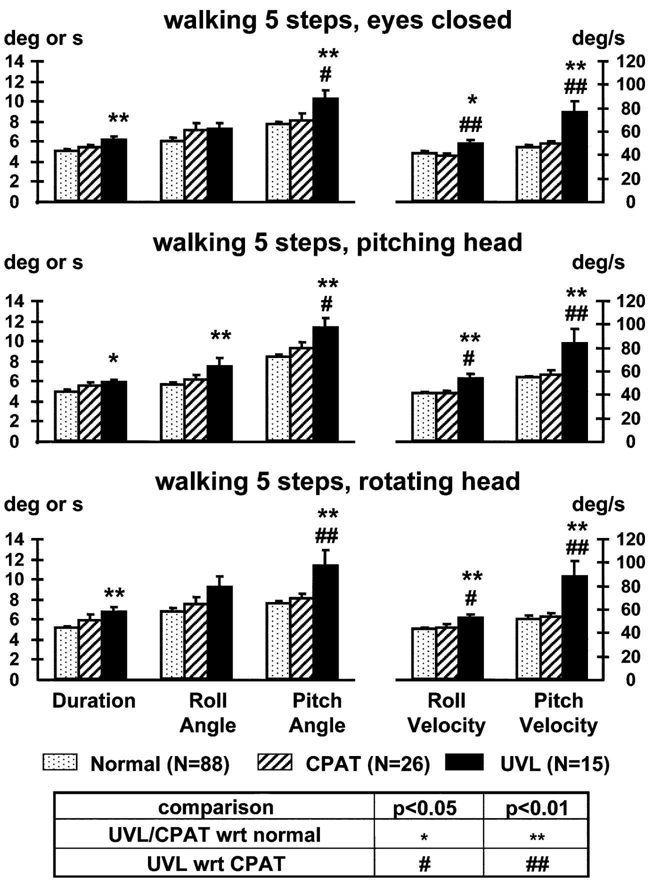


Fig. 4. Population trunk sway displacements and task durations for normal, CPAT and UVL subjects expressed as mean peak-to-peak displacement and time to complete five steps of walking under different conditions. Details of the figure are described in the legend to Fig. 2. Note the significantly greater population means of the UVL subjects when the head is rotated left to right during gait.

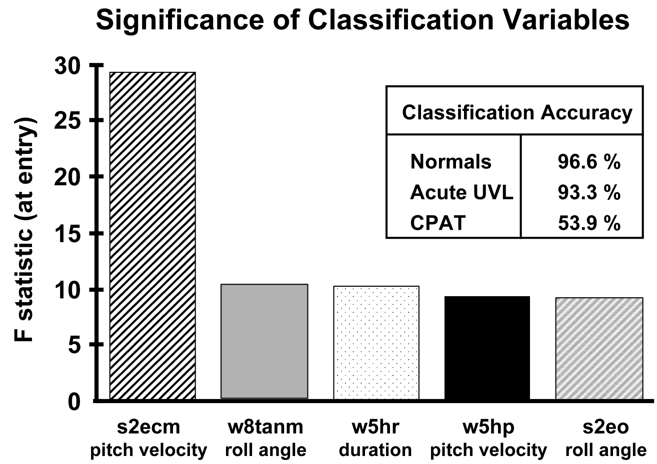


Fig. 5. Significance of each of five classification variables used to separate out members of normal, CPAT and UVL populations into one of these populations. The level of significance at entry of the measurement variable into the classification vector is indicated by the height of the column above the variable name. The final classification accuracy achieved when all variables have been entered is listed in the insert.

We have used the same trunk sway measures for both stance and gait tasks. Therefore, we are in the unique position of being able to compare the amount of trunk sway for both these types of tasks thereby providing an estimate of the efficacy of similar measurements taken from both stance and gait tasks in the identification of balance deficits. Other quantification techniques either measure support surface reaction forces using strain gauges imbedded in the support surface in order to estimate the location of the centre of body mass [29] or use video-motion analysis [14] to compute the location of the centre of mass of the body more accurately. The force-platform has the major disadvantage that determining the angular displacement of the trunk is not possible with this technique unless a major simplification about body sway is assumed. It must be assumed that the body moves as single inverted pendulum, an assumption which has been criticized by several authors [24,30,31]. Movement of the trunk about the hips, or for that matter, the amount of knee flexion cannot be determined from such measurements. Furthermore, if there is a movement of the feet on the force plate such as occurs with the lateral rocking of the foot rotation point [32] during one-legged stance the crucial assumption of a fixed foot on the platform is violated. Clearly, any movement of the foot that occurs along or especially away from the platform surface also invalidates such measurements. Thus, techniques that rely on force platforms [6] cannot reliably measure trunk sway during stance tasks and certainly cannot be used for any gait tasks unless the force platform is extended considerably in length.

Our results confirmed the effectiveness of a reduced form of the CTSIB protocol in screening for a vestibular deficit. With the trunk measures we used, we did not find a bias towards identifying normals as either a severe (UVL) or less severe (CPAT) vestibular deficit subjects as El-Kashlan et al. [16] found when using the CTSIB protocol in a semi-quantitative way. It was interesting to note that only two of the four two-legged stance tasks we employed (normal support surface and the foam support surface eyes closed) featured among our classification tasks. This suggests that recording sway during two-legged stance under more conditions than these two is probably unnecessary for balance screening purposes even though other authors have suggested using two-legged stance tasks with a stabilized visual field too [4,6,16,22]. However, when malingering is suspected it would prove useful to have easy-to-perform stance tasks as part of the test battery because generally these subjects perform such tasks with abnormal sway, and the harder to perform tasks with normal sway [33].

For the two two-legged stance tasks different directional variables were chosen, pitch velocity for the foam support eyes closed, roll angle for the eyes open and normal support. Thus if measurements were restricted to the pitch plane as they are in the Nashner and Peters [6] protocol, where the unstable surface is mechanically driven then roll measurements would not have been employed. The presence of pitch and roll measures for both stance and gait tasks among our balance deficit identifying variables also indicates that a restriction to pitch plane measures alone is not advisable. Other studies [34,35] have in fact provided an experimental basis for using both roll and pitch trunk sway measures in order to adequately define balance deficits. In both normal and vestibular loss subjects, responses to balance perturbations appear to be differently controlled for the pitch and roll directions. Trunk roll occurs earlier and is corrected faster than pitch when the upright stance is perturbed. Also the balance control deficit is greater in the roll direction following total vestibular loss [35]. Clinically, it would be expected that a trunk roll deficit occurs following a UVL. Surprisingly, the same deficit causes pitch control of the trunk to be associated with higher angular velocity amplitudes which in turn must reduce the overall stability in the pitch direction. Other authors have shown that in the pitch plane hip angular velocity and ankle angular velocity (or possibly the respective accelerations) are coordinated to move in opposite directions during quiet stance [24] with the hip angular velocity changes being three times faster than those of the ankle joint. We presume that this coordination becomes disrupted following a vestibular loss [35] and is most apparent for pitch velocity rather than pitch angle.

Previous authors have used task duration as a means to quantify performance for gait and stance tasks of the CTSIB protocol. The limited duration of one-legged stance is a well-known aspect of poor balance control. A shorter than normal duration of one-legged stance is not however, indicative of only a vestibular deficit. The shorter duration could be due to a weakness in the quadriceps muscles (especially in the elderly), but also due to an orthopaedic deficit in the knee joint [18,36]. The duration for many tasks is so short in acute vestibular deficit patients because they simply fall, we were not surprised that it entered as a screening variable. More surprising was the presence of duration as a screening variable for the task of walking five steps while rotating the head, rather than for one-legged stance. We expected that a trunk roll variable from at least one task would enter because roll angle amplitude was one of the significant differences observed for many gait measurements (Table 1). Nonetheless the fact that duration and roll angle and pitch angular velocity (Fig. 5) should be considered for the gait task component of the classification procedure indicates that including gait measures provide a significant improvement in identification power for a vestibular loss. Other authors [2] have emphasized this point more powerfully indicating that even when stance is made unstable in the pitch plane using a moving support surface [6] rather than a foam support, changes in balance control which are more or less like that of normals cannot be usefully assessed with two-legged stance tasks alone. Gait tasks are, according to O'Neil et al. [2], essentially for judging such changes. Of the five screening variables our statistical procedures selected, two were from stance tasks one from a semi-stance task (tandem walking) and two were from the simultaneous walking and head movement gait tasks. Even if we were to reduce the number of variables for screening purposes, the gait measures obtained during head rotation and tandem walking tasks would feature as a very necessary component of a screening procedure.

The absence of significant trunk sway variables for the one-legged eyes-closed stance trials raises the question about how to score a fall. Even though almost all UVL patients fell while performing this task our analysis techniques caused the fall only to be present as a shorter duration. When analysing sway we excluded either the last 2 s or the last 20% of the data stream depending on which yielded the shorter period to be excluded (Section 2). Thereby, we only analysed non-falling trunk sway. In fact, in developing our protocol we were more interested in tasks that were difficult but could be completed by almost all patients without falling. One alternative would have been to include the trunk sway of falling in the analysis yielding in the case of UVL patients a bias towards large values with large variances depending on the roll angle when the non-support-foot touched the floor. Another alternative would be to assign an arbitrary maximum of pitch and roll amplitudes when a fall occurs as some authors have done for pitch displacement using centre of foot pressure measurements [5,6].

An interesting question arises whether for stance tasks our measurement technique using trunk angular sway is more selective of a vestibular deficit than using centre-of-foot pressure (CFP) calculations obtained from support-surface strain gauges. Interestingly the best stance task for discriminating between the groups was with use of the foam support surface. The introduction of this surface between the foot and the strain gauge surface introduces a measurement error into the CFP calculations as the distance between the foot and the strain gauges becomes non-negligible. Another methodological problem with CFP measurements is the lack of correlation between CFP and hip pitch angle (related to the trunk angles we measured) for frequencies below 2 Hz in normal subjects [24]. This would suggest that CFP measurements cannot be used to reliably calculate the best discriminating variable we found during stance, trunk pitch velocity. For these reasons, we believe that trunk pitch angular velocity and trunk roll angle measurements offer considerable advantages over CFP calculations as an indication of body sway produced by a balance deficit, quite apart from the added advantage of being able to quantify trunk sway during gait tasks. The only authors that have tried to improve CFP calculations, used static rather than dynamic measures of trunk sway. Benvenuti et al. [14] looked at the average orientation of the body segments during a 40 s trial in addition to CFP and found that average hip and leg angles provided a significant improvement in the identification in the frail elderly from normal coherts with minimal balance problems. These authors did not however rank the variables in their discriminant analysis. They did investigate the repeatability of measures and found angle measures of the hip, knee and ankle to be more reliable than CFP measures. Whether this would still prove to be the case if they had examined peak-to-peak measures as we did rather than average angles of joint flexion remains to be investigated.

Because, we have used the same instrumentation to measure balance performance gait and stance tasks we are hopeful that our technique will prove useful for assessing a wide variety of balance disorders. Furthermore, because we have provided a basis for focusing on those tests which provided the optimal screening information we hope to have taken a step to help the clinician to reduce the number of multiple approaches and instruments used [20] to assess balance deficits. Thus, the tests and measurements that we would recommend using in a clinical setting to screen for a balance disorder are in order of significance:

Task	Surface	Measurement
Stand on two legs eyes closed	Foam	Trunk pitch velocity
Walk eight tandem steps	Foam	Trunk roll angle
Walk 3 m (five steps) while horizontally rotating the head	Normal	Duration
Walk 3 m (five steps) while vertically pitching the head	Normal	Trunk pitch velocity
Stand on two legs eyes open	Normal	Trunk roll angle

In new studies [37,38] that have validated these tests on a separate group of UVL patients, we have confirmed that the magnitude of trunk sway recorded during these tasks is correlated with the state of compensation [25] of an acute UVL. For this reason, we are hopeful that the screening procedures described here will also prove useful in quantifying the degree of improvement in a balance deficit brought about by a treatment procedure.

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