Available online at www.sciencedirect.com
ScienceDirect
Journal of Sound and Vibration 298 (2006) 725–754
JOURNAL OF SOUND AND VIBRATION
www.elsevier.com/locate/jsvi

Effect of frequency, magnitude and direction of translational and rotational oscillation on the postural stability of standing people

Naser Nawayseh, Michael J. Griffin*

Human Factors Research Unit, Institute of Sound and Vibration Research, University of Southampton, Southampton SO17 1BJ, UK

Received 1 May 2006; accepted 8 June 2006
Available online 10 August 2006

Abstract

Oscillatory motions can cause injury in transport when standing passengers or crew lose balance and fall. To predict the loss of balance of standing people, a model is required of the relationship between the input motion and the stability of the human body. This experimental study investigated the effect of frequency, magnitude and direction of oscillation on the postural stability of standing subjects and whether response to rotational oscillation can be predicted from knowledge of response to translational oscillation.

Twelve male subjects stood on a floor that oscillated in either horizontal (fore-and-aft or lateral) or rotational (pitch or roll) directions. The oscillations were one-third octave bands of random motion centred on five preferred octave centre frequencies (0.125, 0.25, 0.5, 1.0, and 2.0 Hz). The horizontal motions were presented at each of four velocities (0.04, 0.062, 0.099, and 0.16 ms^{-1} rms) and the rotational motions were presented at each of four rotational angles (0.73, 1.46, 2.92, and 5.85° rms) corresponding to four accelerations (0.125, 0.25, 0.5, and 1.0 ms^{-2} rms), where the acceleration is that caused by rotation through the gravitational vector. Postural stability was determined by subjective methods and by measuring the displacement of the centre of pressure at the feet during horizontal oscillation.

During horizontal oscillation, increases in motion magnitude increased instability and, with the same velocity at all frequencies from 0.125 to 2.0 Hz, most instability occurred in the region of 0.5 Hz. Fore-and-aft oscillation produced more instability than lateral oscillation, although displacements of the centre of pressure were similar in both directions. With the same angular displacement at all frequencies from 0.125 to 2.0 Hz, pitch oscillation caused more instability than roll oscillation, but in both directions instability increased with increased frequency of oscillation. Frequency weightings for acceleration in the plane of the floor during translational and rotational excitation show the significance of low-frequency translational oscillation and high-frequency rotational motion, and show that it is necessary to know whether the measured acceleration is caused by translation or rotation through gravity.

*Corresponding author. Tel.: +44 023 8059 2277; fax: +44 023 8059 2927.
E-mail address: M.J.Griffin@soton.ac.uk (M.J. Griffin).

0022-460X/$ - see front matter
doi:10.1016/j.jsv.2006.06.027

1. Introduction

Standing people are exposed to oscillatory motions in various forms of transport, including buses, trains, and ships. The translational and rotational motions experienced by standing passengers and crew can affect their postural stability. When postural stability is threatened, people attempt to maintain balance by various automatic strategies, such as ankle or hip rotation or body stiffening (e.g. Ref. [1]), or by voluntary stepping, or grasping a support. The adjustment of the body to the motion may cause interference with activities and degrade the performance of tasks. Where such adjustments are not possible or not sufficient, the body may fall and injury may occur.

The effects of various types of perturbations on postural stability have been studied to understand the postural control system and the stepping response occurring during loss of balance. In some studies, subjects have stood on a stationary platform and loss of balance or stepping has been induced by a pull force applied to the waist [2] or by a visual cue [3]. In other studies, loss of balance has been induced by transient forward or backward motion of the floor [3–5] and in a few studies the floor has been oscillated (e.g. Refs. [6,7]).

In transport, loss of balance may occur due to acceleration or deceleration or cornering of a vehicle. The motions may be translational (fore-and-aft or lateral) or rotational (roll or pitch), or a combination of translational and rotational motions.

The prediction of the postural stability of standing persons exposed to external perturbations requires the identification of the relationships between the characteristics of the input motion and the consequent response of the human body. Such knowledge might be used to construct models that predict the probability of losing balance. Models of postural stability in the fore-and-aft direction have been developed for stationary standing and for standing during exposure to fore-and-aft motion of the floor. The models range from conceptual models (e.g. Ref. [8]), to passive models (e.g. Ref. [9]), and to active models with single or multi-link components (e.g. Refs. [10,11]). Conceptual models show the flow of information between the different parts of the control system (i.e. the sensory system, skeletal muscle system and the central nervous system, CNS). The sensory system provides feedback to the CNS on the state of the body and the environment surrounding the body. After processing the information in the CNS, signals are sent to the skeletal muscle system with more than 200 degrees of freedom [12]. Although this type of model provides a framework for understanding the postural control system, it does not quantify the response of the body to external perturbations.

A passive model has been proposed for the prediction of the loss of balance of people standing on the decks of ships [9]. The model has a rigid body with a similar shape, size and mass as the human body, with the centre of mass at a representative height above the base of support. The loss of balance was predicted from the number of motion-induced interruption (MIIs) that are assumed to occur when the centre of pressure at the base threatens to move outside the base of support.

Passive biomechanical models with rigid links have been developed to study the relationship between body movements (e.g. sway angles), and the body reaction (e.g. vertical reaction forces at the feet, or torque at ankles). Some of these models have been used to estimate movements of the centre of mass of the body from movements of the centre of pressure at the floor, or from the displacement of body parts [13,14]. A summary of these models is given in Lewis and Griffin [15]. Being passive, these models do not take into account the active control and regulation of posture in response to motion.

The human postural control mechanisms have also been represented by active biomechanical models. Single and double inverted pendulum systems have been used by Ishida and Imai [10], Peeters et al. [11] and Johansson et al. [16]. Rotations of a pendulum were detected by controllers that apply restoring torques about the pivot (e.g. ankle or hip joint). Such models have been used to investigate relationships between body displacements and torques at the joints in the sagittal plane (e.g. Ref. [17]). Other models have been used to evaluate transfer functions between movements of a floor and movements of the centre of pressure, and the consequent prediction of postural stability [18].

The human body is neither rigid nor symmetrical, and so the postural responses to motion of a floor may be expected to depend on both the frequency and the direction of the applied motion. Furthermore, the body is an active system, so the relation between the motion and the response may depend on the magnitude of the motion. Models of postural stability intended to predict the effects of floor motion on the probability of losing balance should therefore take into account the frequency, magnitude and direction of the applied oscillations.

When measuring acceleration in the plane of the floor of a vehicle during simultaneous fore-and-aft and pitch motion (or simultaneous roll and lateral motion) the measured acceleration comes partially from the translational acceleration and partially from the gravitational component (i.e. $g \sin \theta$, where θ is the angle of pitch or roll). Without resolving the acceleration into its components, it is assumed that human response can be predicted from the measured acceleration without knowing how much of the acceleration is caused by rotation through gravity and how much is caused by translation.

This study investigated the effects of the frequency of oscillation, the magnitude of oscillation, and the direction of oscillation (translational and rotational) on the postural stability of standing people. The results are summarised in models that take into account the effects of frequency, magnitude, and direction of oscillation. It was hypothesised that the same acceleration in the plane of the floor would induce the same stability problems irrespective of whether the measured acceleration was produced by horizontal oscillation or by a gravitational component due to rotational oscillation through gravity.

2. Apparatus, method, and analysis

Two experiments were conducted, one with translational oscillation (fore-and-aft and lateral) and one with rotational oscillation (pitch and roll).

2.1. Apparatus

2.1.1. Translational oscillation

Subjects stood on a flat rigid platform that was caused to oscillate in the fore-and-aft or lateral direction by an electro-hydraulic vibrator capable of producing peak-to-peak displacements of 1 m. A force platform (Kistler Z 13053) consisting of four quartz piezoelectric transducers of the same sensitivity located at the four corners of a rectangular aluminium plate ($600 \times 400 \times 20$ mm) was secured to the vibrator plate and used to measure the vertical reaction force at the feet. The force signals from each cell were amplified using Kistler 5007 charge amplifiers. Acceleration in the horizontal direction (either fore-and-aft or lateral) was measured at the centre of the force platform using a piezo-resistive accelerometer (Setra model 141A). The signals from the accelerometer and the force cells were digitised at 200 samples per second after low pass filtering at 40 Hz.

A metal frame was mounted to the vibrator plate for the subjects to hold in case of falling. The metal frame was also used to support a loose harness that subjects wore throughout the experiment for safety.

2.1.2. Rotational oscillation

Subjects stood on a flat platform that was caused to oscillate in pitch or roll by an electro-hydraulic vibrator capable of producing peak-to-peak displacements of 40°. The 'horizontal' acceleration (i.e. the acceleration measured parallel to the platform as it rotated though $\theta°$, i.e. $g \sin \theta$) was measured using a Setra accelerometer similar to that used to measure translational excitation. The axis of rotation was located on the surface across the middle of the platform on which the subjects stood.

2.2. Method

2.2.1. Subjects and exposure conditions

Twelve male subjects aged 24–41 years participated in both experiments. Their measured statures, weights, foot lengths and foot widths are shown in Table 1. Apart from subject 9, the subjects were the same in both experiments (i.e. with both translational and rotational oscillation). In Table 1, the characteristics are shown for subject 9 who took part in the experiment with translational oscillation. Subject 9 who took part in the rotational experiment was 29 years old and had a stature of 183 cm, weight of 80.8 kg, foot length of 28.5 cm and foot width of 11.0 cm.

In both experiments, subjects were exposed to 60-s periods of each of five one-third octave bands of random motions centred on preferred octave centre frequencies (0.125, 0.25, 0.5, 1, and 2 Hz). In the experiment with translational oscillation, subjects were exposed to four velocities of the platform (0.04, 0.062, 0.099, and $0.16\,ms^{-1}$ rms). In the experiment with rotational oscillation, the subjects were exposed to four accelerations

Table 1
Characteristics of the subjects

Subject	Age (yr)	Stature (cm)	Weight (kg)		Foot length (cm)	Foot width (cm)
			Experiment 1	Experiment 2		
1	25	176	63.2	61.7	24.5	9.6
2	27	176	65.0	65.8	25.6	9.8
3	36	175	74.0	76.0	27.0	10.0
4	41	183	80.9	81.0	25.6	10.3
5	25	167	67.0	71.3	24.0	9.3
6	37	165	63.2	60.0	23.5	9.7
7	24	183	76.3	75.0	26.2	10.0
8	26	176	81.7	81.8	24.5	10.5
9	29	174	73.8	—	25.3	9.8
10	27	173	86.4	91.2	27.3	10.5
11	34	169	65.3	64.5	25.5	9.6
12	24	193	90.4	93.8	29.0	11.0

Table 2
Acceleration, velocity, and displacement for each exposure (rms values)

Frequency (Hz)	Translational experiment			Rotational experiment		
	Acceleration (ms^{-2})	Velocity (ms^{-1})	Displacement (m)	Acceleration (ms^{-2})	Velocity (rad/s)	Displacement (°)
0.125	0.031	0.04	0.051	0.125	0.010	0.730
	0.049	0.062	0.079	0.25	0.020	1.460
	0.078	0.099	0.126	0.5	0.040	2.922
	0.126	0.16	0.204	1.0	0.080	5.851
0.250	0.063	0.04	0.025	0.125	0.020	0.730
	0.097	0.062	0.039	0.25	0.040	1.460
	0.156	0.099	0.063	0.5	0.080	2.922
	0.251	0.16	0.102	1.0	0.160	5.851
0.50	0.126	0.04	0.013	0.125	0.040	0.730
	0.195	0.062	0.020	0.25	0.080	1.460
	0.311	0.099	0.032	0.5	0.160	2.922
	0.503	0.16	0.051	1.0	0.321	5.851
1.0	0.251	0.04	0.006	0.125	0.080	0.730
	0.390	0.062	0.010	0.25	0.160	1.460
	0.622	0.099	0.016	0.5	0.320	2.922
	1.005	0.16	0.025	1.0	0.642	5.851
2.0	0.503	0.04	0.003	0.125	0.160	0.730
	0.779	0.062	0.005	0.25	0.320	1.460
	1.244	0.099	0.008	—	—	—
	2.011	0.16	0.013	—	—	—

(0.125, 0.25, 0.5, and 1.0 ms^{-2} rms) at all frequencies except that with 2 Hz only 0.125 and 0.25 ms^{-2} rms were used. The accelerations, velocities, and displacements of the motions in both directions (i.e. translation and rotation) are shown in Table 2. Notwithstanding the hypothesis that postural stability would be determined by the acceleration irrespective of whether it was translational or due to the gravitational vector during rotation, the same stimuli were not used for both translational and rotational motions because preliminary studies suggested that the stimuli used in the translational experiment (column 2 in Table 2) had little effect on postural stability when used in the rotational experiment, especially at 0.125, 0.25, and 0.5 Hz.

Although all subjects were exposed to the same rms magnitude of oscillation (within 3%) and the same power spectrum in each condition, a unique time history was used for each subject. The order of presentation of the magnitudes and frequencies was randomised. With translational oscillation, half of the subjects commenced with fore-and-aft oscillation and the other half commenced with lateral oscillation. With rotational oscillation, half of the subjects commenced with pitch oscillation and the other half commenced with roll oscillation.

2.2.2. Experimental procedure

Before starting an experiment, each subject was informed of the purpose of the experiment and given instructions to follow during exposure. Subjects were instructed to stand barefoot on the force platform with an upright comfortable posture and with 240 mm separation between the centre lines of the feet. The arms of the subjects hung vertically at their sides. The subjects, who wore a loose safety harness attached to a metal frame, were instructed not to lean against or hold a support unless the motion caused them to fall. During pitch oscillation, the ankles were located directly above the axis of rotation. During roll oscillation, the subjects stood such that the axis of rotation was mid-way between the feet in the mid-sagittal plane of the body.

During exposure to translational oscillation, the vertical reaction forces at the feet were measured and recorded for later analysis. After each 60-s exposure, subjects were asked to report whether they lost their balance and state the probability that they would lose their balance if the same exposure was repeated.

After exposure to all horizontal oscillation stimuli, the vertical reaction forces at the feet were measured while standing stationary so as to measure the sway of the subjects during ‘quiet’ stance. The subjects were instructed to stand still on the force platform adopting an upright posture for 60 s while the vertical forces were measured.

2.3. Analysis

The vertical reaction forces measured at the feet during translational oscillation were used to calculate the displacement of the centre of pressure (COP) as an indicator of body sway. It has been reported that during stationary standing, the centre of pressure represents the vertical projection of the centre of gravity (COG) of the body on the force plate [14]. When the body moves due to external perturbation, the vertical projection of the COG may not coincide with the COP. The position of the centre of pressure, in a coordinate system moving with the force platform, can be obtained from the vertical reaction forces measured at the feet. The position of the centre of pressure in the fore-and-aft direction (during fore-and-aft oscillation) or the lateral direction (during lateral oscillation) as a function of time, with the origin of the moving coordinate system at point A in Fig. 1 can be calculated using the following equation:

$$x(t) = \frac{(F_1(t) + F_2(t))x_p}{F_1(t) + F_2(t) + F_3(t) + F_4(t)},$$

where, $F_1(t)$, $F_2(t)$, $F_3(t)$, and $F_4(t)$ are the measured vertical forces as shown in Fig. 1 and x_p is 400 mm, the distance between the front two force cells (giving $F_1(t)$ and $F_2(t)$) and the rear two force cells (giving $F_3(t)$ and $F_4(t)$) as in Fig. 1. During the stationary condition, the fore-and-aft position of the centre of pressure (COP) was calculated using the above equation while the lateral position of the COP was calculated by replacing $F_1(t)$ in the numerator of the above equation by $F_3(t)$ and using a value of 240 mm for x_p (see Fig. 1).

The displacement of the centre of pressure is given by $x(t) - x_i$, where x_i is the initial position of the COP (i.e. the position of the COP at time $t = 0$ for the exposure).

The variation in the displacement of the centre of pressure with time is a measure of postural sway. The standard deviation (SD) of the displacement of the centre of pressure over the 60-s exposure was used to indicate the postural sway in each condition with translational oscillation.

Data analysis was performed using SPSS (version 12). Non-parametric statistical methods were used throughout: the Friedman analysis of variance was used to test for differences between multiple conditions and the Wilcoxon matched-pairs signed ranks was used to test for differences between pairs of conditions. Spearman’s rank correlation was used to investigate associations between variables.

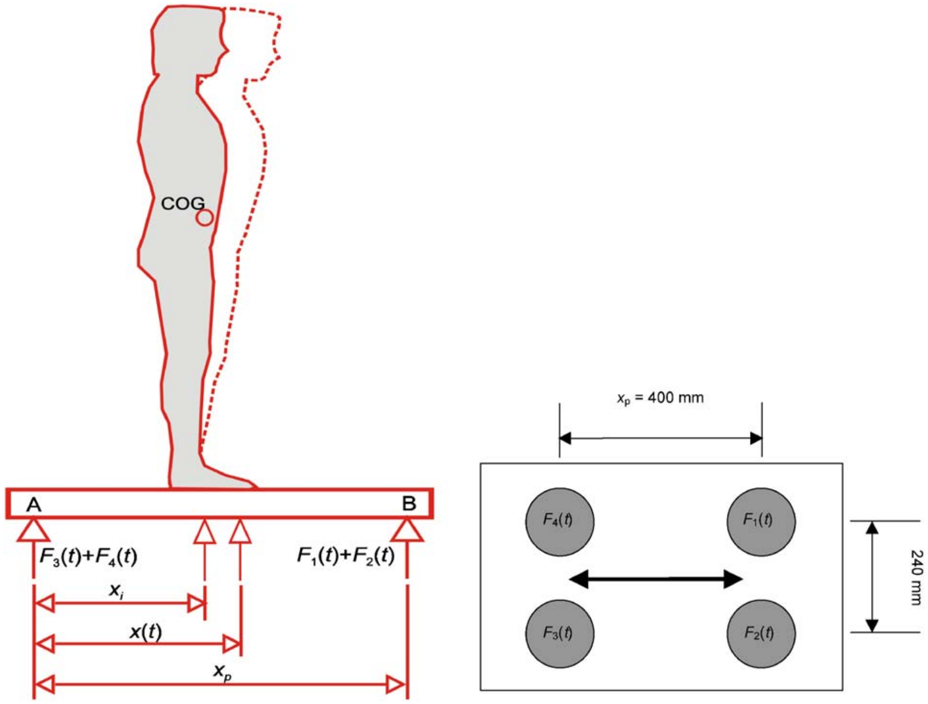

Fig. 1. Arrangement of force cells. The arrow shows the direction of motion; $F_1(t)$, $F_2(t)$, $F_3(t)$ and $F_4(t)$ are the measured vertical reaction forces; x_i is the initial position of the centre of pressure (COP) and $x(t)$ is the instantaneous position of the COP.

3. Results

3.1. Translational oscillation: fore-and-aft and lateral

3.1.1. Centre of pressure (COP)

During fore-and-aft oscillation, the fore-and-aft displacement of the centre of pressure increased with increases in the magnitude of oscillation at each frequency ($p<0.01$, Wilcoxon matched-pairs signed ranks, Fig. 2). At each magnitude of oscillation, there was a significant effect of frequency ($p<0.0001$ Friedman). The standard deviation of the fore-and-aft displacement of the centre of pressure increased with increasing frequency up to 0.5 Hz and decreased at higher frequencies. This trend was seen for the majority of subjects and at all magnitudes of oscillation. No significant differences were found in the standard deviation of the fore-and-aft displacement of the COP between 0.125 and 2 Hz or between 0.25 and 1.0 Hz at any velocity (Table 3).

During lateral oscillation, the standard deviation of the lateral displacement of the centre of pressure also increased with increased magnitude at all frequencies ($p<0.01$, Wilcoxon matched-pairs signed ranks). At each magnitude of oscillation, there was a significant effect of frequency ($p<0.0001$ Friedman). As with fore-and-aft oscillation, the maximum standard deviation of the lateral displacement of the centre of pressure occurred with 0.5 Hz oscillation (Fig. 3). No significant differences were found in the standard deviation of the lateral displacement of the COP between 0.125 and 2 Hz or between 0.25 and 1.0 Hz at any velocity, except at 0.04 ms^{-1} rms between 0.125 and 2 Hz and at 0.099 ms^{-1} rms between 0.25 and 1 Hz (Table 4).

The median standard deviation of the fore-and-aft displacement of the centre of pressure (COP) during fore-and-aft oscillation showed similar values to the median standard deviation of the lateral displacement of the centre of pressure during lateral oscillation (Fig. 4).

The linearity of the centre of pressure displacement was tested for both directions of excitation. Figs. 5 and 6 show the ratio between the standard deviation of the displacement of the COP at 0.062, 0.099, and 0.16 ms^{-1}

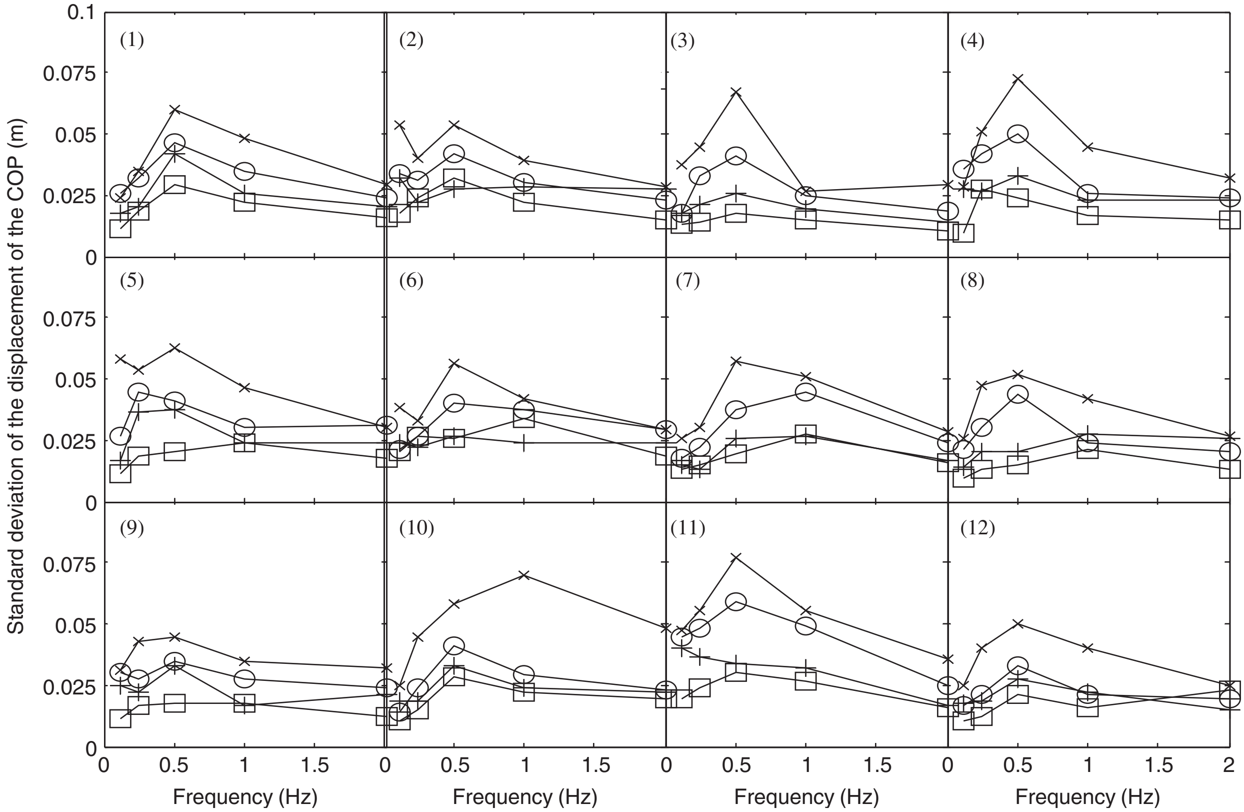

Fig. 2. Standard deviation of the fore-and-aft displacement of the centre of pressure (COP) during fore-and-aft oscillation. □, 0.04; +, 0.062; ○, 0.099; ×, 0.16 ms^{-1} rms. Data from 12 subjects.

Table 3
Effect of oscillation frequency on the standard deviation of fore-and-aft displacements of the centre of pressure

Velocity (ms^{-1} rms)	Frequency (Hz)	0.125	0.25	0.5	1.0	2.0
0.04	0.125	—	**	**	**	↑
	0.25		—	*	↑	ns
	0.5			—	ns	**
	1.0				—	*
	2.0					—
0.062	0.125	—	ns	*	ns	ns
	0.25		—	**	ns	ns
	0.5			—	*	**
	1.0				—	*
	2.0					—
0.099	0.125	—	**	**	*	ns
	0.25		—	**	ns	*
	0.5			—	**	**
	1.0				—	**
	2.0					—
0.16	0.125	—	*	**	↑	ns
	0.25		—	**	ns	**
	0.5			—	**	**
	1.0				—	**
	2.0					—

Wilcoxon matched pairs signed-ranks: ↑ = $p<0.1$, * = $p<0.05$, ** = $p<0.01$, ns = $p>0.1$.

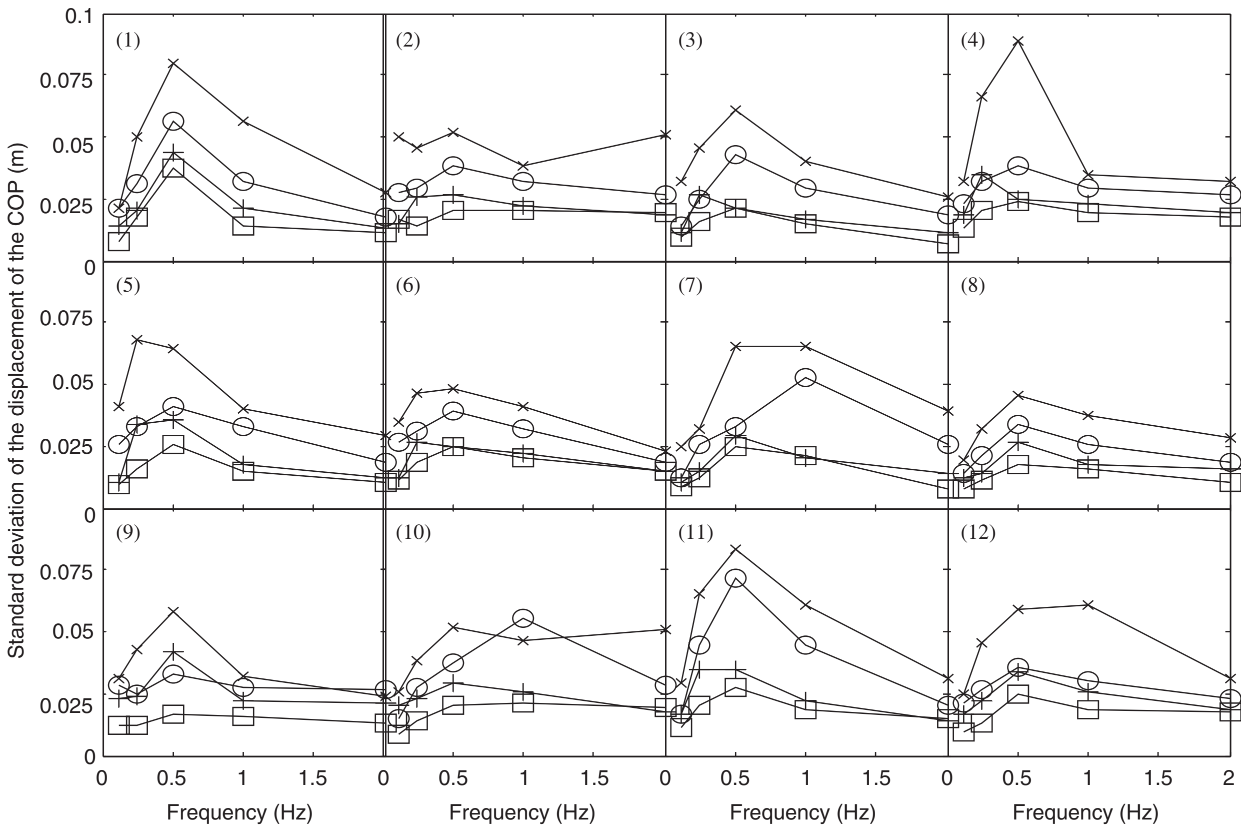

Fig. 3. Standard deviation of the lateral displacement of the centre of pressure (COP) during lateral oscillation. □, 0.04; +, 0.062; ○, 0.099; ×, 0.16 ms^{-1} rms. Data from 12 subjects.

rms to the standard deviation of the displacement of the COP at 0.04 ms^{-1} rms for all subjects. For a linear system, these ratios should be 1.55 (0.062/0.04), 2.48 (0.099/0.04), and 4 (0.16/0.04). Figs. 5 and 6 show variations between subjects and Table 5 shows the medians. The median ratios between the standard deviation of the centre of pressure at two velocities are not equal to the ratios between the velocities, indicating a nonlinear response. The least nonlinearity (if we allow the ratio of the COP to be acceptable within 10 percent of the ratio of the velocities), occurred with 0.125 Hz between 0.04 and 0.062 ms^{-1} rms and between 0.099 and 0.16 ms^{-1} rms, and with 0.5 Hz between 0.062 and 0.099 ms^{-1} rms in the fore-and-aft direction. In the lateral direction, the least nonlinearity occurred with 0.125 and 0.25 Hz between 0.04 and 0.062 ms^{-1} rms, with 1.0 Hz between 0.062 and 0.099 ms^{-1} rms and with 0.125, 0.25, and 2 Hz between 0.099 and 0.16 ms^{-1} rms. There was no significant difference between the ratios of the COP measured at different oscillation velocities and the corresponding ratio of the oscillation velocities in only 7 conditions out of 30 conditions (i.e. with 6 ratios at 5 frequencies) during the fore-and-aft oscillation and in only 10 conditions out of 30 conditions during lateral oscillation (Wilcoxon, $p > 0.05$). When a significant level of 0.01 was used, the number of conditions where there was no significant difference between the ratio of the COP and the ratio of the velocities increased to 11 and 17 during fore-and-aft and lateral oscillations, respectively.

3.1.2. Loss of balance

The percentages of subjects who lost balance (i.e. they fell or they needed to hold a support to prevent falling) during exposure to each stimulus are shown in Fig. 7 for both fore-and-aft and lateral oscillation.

In the fore-and-aft direction at the highest magnitude, loss of balance occurred at all frequencies but the greatest percentage loss of balance occurred at 0.5 Hz where 42% of the subjects lost their balance.

Table 4
Effect of oscillation frequency on the standard deviation of lateral displacements of the centre of pressure

Velocity (ms^{-1} rms)	Frequency (Hz)	0.125	0.25	0.5	1.0	2.0
0.04	0.125	—	**	**	**	*
	0.25		—	**	↑	ns
	0.5			—	**	**
	1.0				—	**
	2.0					—
0.062	0.125	—	**	**	**	ns
	0.25		—	↑	ns	**
	0.5			—	**	**
	1.0				—	**
	2.0					—
0.099	0.125	—	**	**	**	ns
	0.25		—	**	*	**
	0.5			—	ns	**
	1.0				—	**
	2.0					—
0.16	0.125	—	**	**	*	ns
	0.25		—	**	ns	**
	0.5			—	**	**
	1.0				—	*
	2.0					—

Wilcoxon matched pairs signed-ranks: ↑ $= p<0.1$, * $= p<0.05$, ** $= p<0.01$, ns $= p>0.1$.

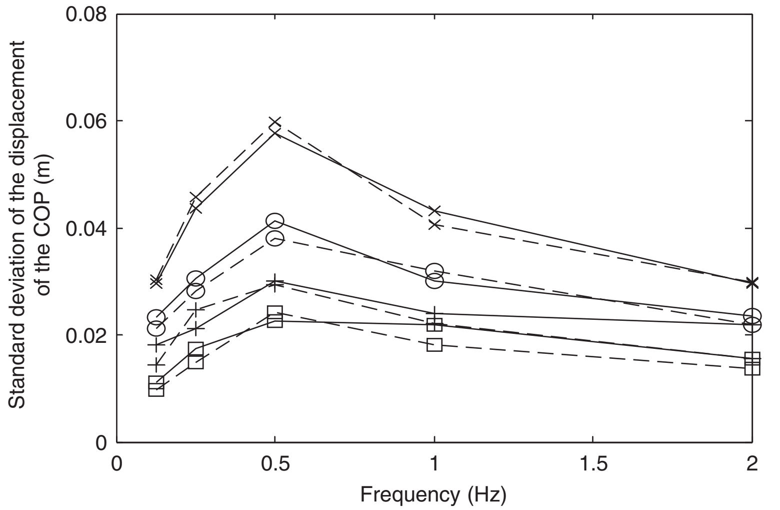

Fig. 4. Median values of the standard deviation of the displacement of the centre of pressure (COP): (——, fore-and-aft COP displacement during fore-and-aft oscillation); (– – –, lateral COP displacement during lateral oscillation); □, 0.04; +, 0.062; ○, 0.099; ×, 0.16 ms^{-1} rms.

In the lateral direction, the greatest percentage loss of balance also occurred at 0.5 Hz at the highest magnitude of oscillation (42% of the subjects lost balance). However, at other frequencies, the percentage of loss of balance was low (less than 9%) or zero.

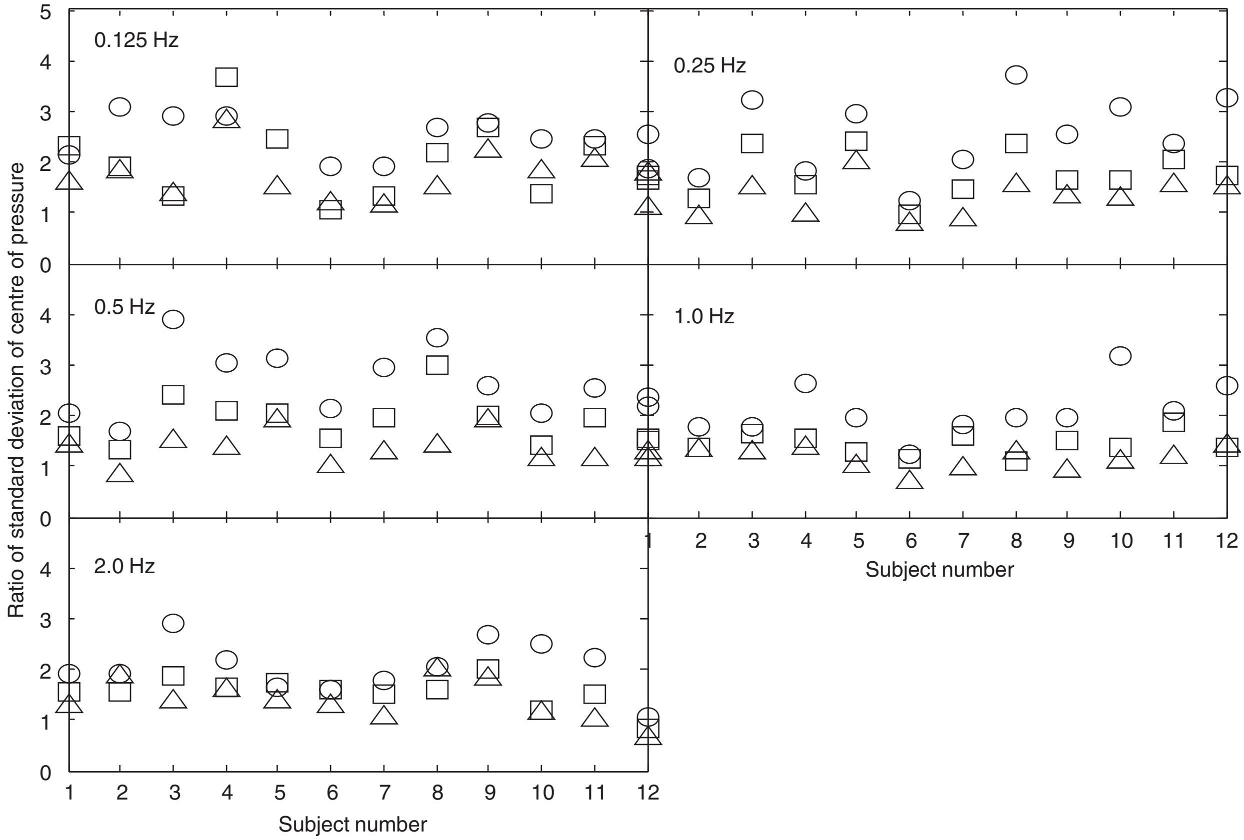

Fig. 5. Ratios of the standard deviation of the fore-and-aft displacement of the centre of pressure (COP) obtained at 0.062, 0.099, and 0.16 ms^{-1} rms to that obtained at 0.04 ms^{-1} rms. Δ, ratio between COP obtained at 0.062 and COP obtained at 0.04 ms^{-1} rms; □, ratio between COP obtained at 0.099 and COP obtained at 0.04 ms^{-1} rms; ○, ratio between COP obtained at 0.16 and COP obtained at 0.04 ms^{-1} rms.

3.1.3. Estimated probability of losing balance

There was high inter-subject variability in the estimates given by subjects for the probability of them losing balance if they were exposed again to the same fore-and-aft stimulus (Fig. 8). Almost all subjects tended to estimate a higher probability of losing balance at high magnitudes than at low magnitudes, except between 0.04 and 0.062 ms^{-1} rms where there was no significant difference in the estimates of the probability of losing balance at any frequency ($p > 0.05$).

The effect of oscillation frequency on the estimated probability of losing balance during fore-and-aft oscillation was significant (Friedman, $p < 0.001$). Wilcoxon matched-pairs signed ranks test showed no significant difference between estimates of the probability of losing balance between 0.5, 1.0, and 2.0 Hz at any magnitude (Table 6).

High inter-subject variability was also found in the subjective estimates of the probability of losing balance during lateral excitation (Fig. 9), but almost all subjects estimated a higher probability of losing balance at high magnitudes than at low magnitudes, except between 0.04 and 0.062 ms^{-1} rms at 0.125, 1.0, and 2.0 Hz where there was no significant difference in the estimates of the probability of losing balance ($p > 0.05$).

The effect of oscillation frequency on the estimated probability of losing balance during lateral oscillation was also significant (Friedman, $p < 0.001$). Wilcoxon matched-pairs signed ranks test showed no significant difference in subject estimates of the probability of losing balance between 0.5, 1.0, and 2.0 Hz at any magnitude ($p > 0.05$) except between 0.5 and 1.0 Hz and between 1.0 and 2.0 Hz at 0.16 ms^{-1} rms (Table 7).

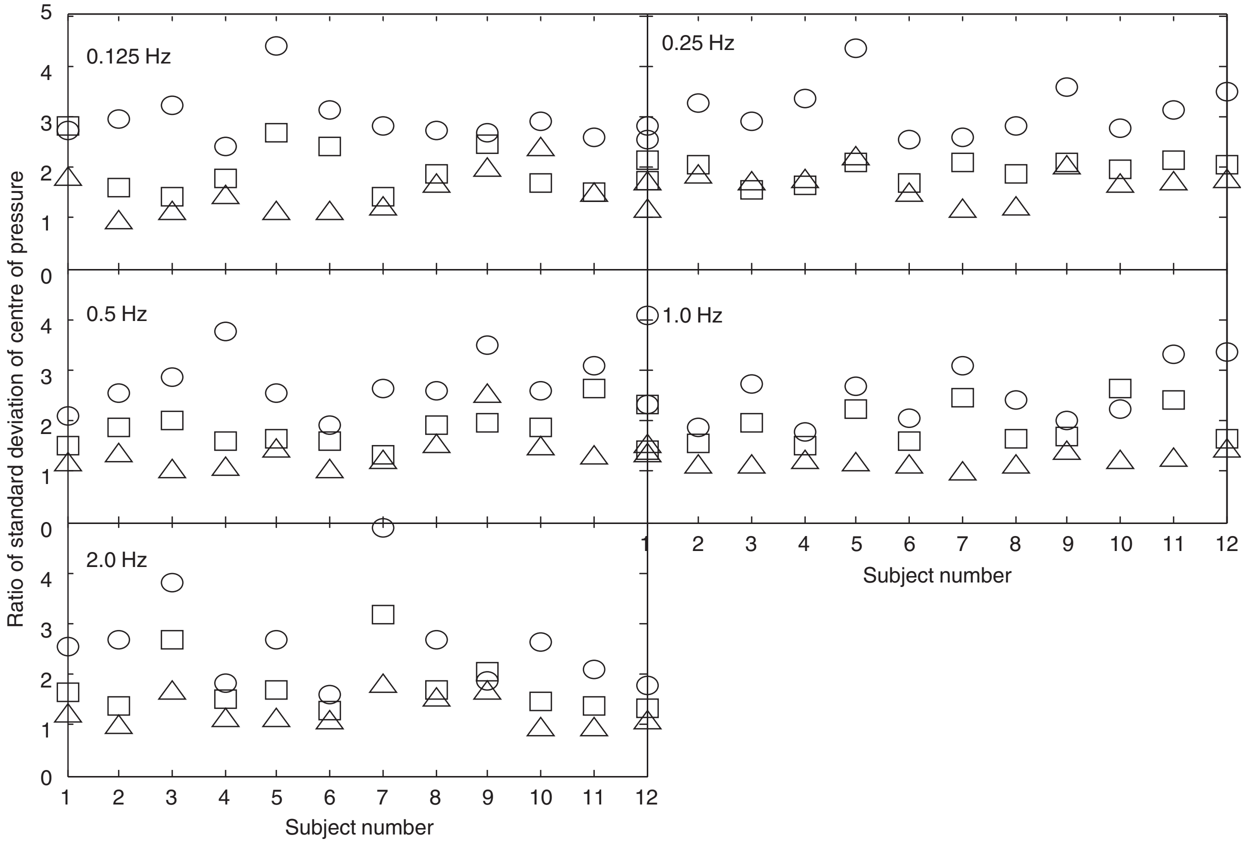

Fig. 6. Ratios of the standard deviation of the lateral displacement of the centre of pressure (COP) obtained at 0.062, 0.099 and 0.16 ms^{-1} rms to that obtained at 0.04 ms^{-1} rms. Δ, ratio between COP obtained at 0.062 and COP obtained at 0.04 ms^{-1} rms; □, ratio between COP obtained at 0.099 and COP obtained at 0.04 ms^{-1} rms; ○, ratio between COP obtained at 0.16 and COP obtained at 0.04 ms^{-1} rms.

Table 5
Median ratios of the standard deviations of the centres of pressure (COP) obtained at different velocities (1: 0.04 ms^{-1} rms; 2: 0.062 ms^{-1} rms; 3: 0.099 ms^{-1} rms; 4: 0.16 ms^{-1} rms)

	Fore-and-aft					Lateral				
	0.125 (Hz)	0.25 (Hz)	0.5 (Hz)	1.0 (Hz)	2.0 (Hz)	0.125 (Hz)	0.25 (Hz)	0.5 (Hz)	1.0 (Hz)	2.0 (Hz)
COP2/COP1	1.69	1.30	1.34	1.17	1.33	1.44	1.70	1.30	1.19	1.10
COP3/COP1	2.06	1.68	1.95	1.46	1.58	1.83	2.00	1.74	1.84	1.58
COP4/COP1	2.59	2.45	2.58	1.97	1.98	2.78	3.02	2.59	2.55	2.59
COP3/COP2	1.15	1.36	1.49	1.28	1.22	1.26	1.17	1.28	1.49	1.40
COP4/COP2	1.51	1.91	2.02	1.86	1.48	1.74	1.89	1.88	2.22	2.17
COP4/COP3	1.48	1.33	1.41	1.35	1.30	1.54	1.59	1.42	1.27	1.52

For example, COP2/COP1 is the median of the ratio between the standard deviation of the COP measured at 0.062 ms^{-1} rms and the standard deviation of the COP measured at 0.04 ms^{-1} rms. The ratio should be compared as follows: COP2/COP1 with (0.062/0.04 = 1.55); COP3/COP1 with 2.48; COP4/COP1 with 4; COP3/COP2 with 1.59; COP4/COP2 with 2.58; and COP4/COP3 with 1.6.

With both fore-and-aft and lateral oscillation, the greatest median estimates of the probability of losing balance were in the region 0.5–1.0 Hz (Fig. 10), which is clearer for high oscillation velocities than for low velocities.

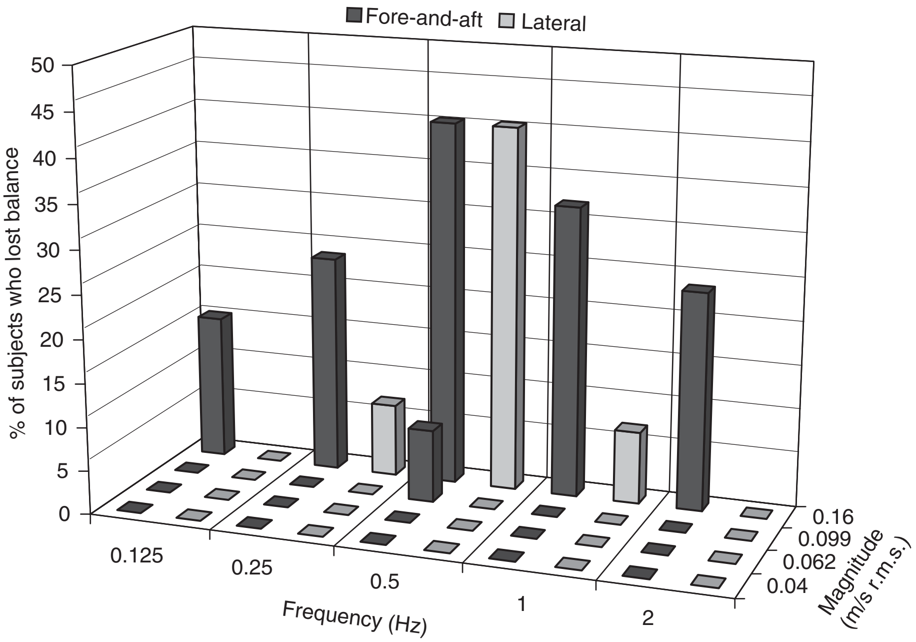

Fig. 7. Percentages of subjects who lost balance during fore-and-aft and lateral oscillation with four magnitudes of velocity at frequencies from 0.125 to 2 Hz.

3.2. Rotational oscillation: pitch and roll

3.2.1. Loss of balance

The percentages of subjects who lost balance during pitch and roll oscillation varied with the magnitude, frequency, and direction of oscillation (Fig. 11). With the lowest frequency (i.e. 0.125 Hz), only one subject (subject 5) was sensitive to the motion: at 0.125 Hz, subject 5 reported loss of balance with all magnitudes of oscillation during pitch motion and with the highest magnitude of oscillation during roll motion. The loss of balance increased with increasing frequency, especially with pitch oscillation and with high magnitudes of oscillation. With 2.0 Hz oscillation, where the highest magnitude was 0.25 ms^{-2} rms, the percentage of subjects who lost balance was greater than at any other frequency for both directions of oscillation.

3.2.2. Estimated probability of losing balance

There was high inter-subject variability in estimates of the probability of losing balance during pitch and roll oscillations (Figs. 12 and 13). In both directions of oscillation, increases in the magnitude of oscillation at any frequency, generally, increased the estimated probability of subjects losing balance (Friedman, $p<0.001$; see median data in Fig. 14).

The effect of oscillation frequency was significant at each magnitude of oscillation (Friedman, $p<0.001$). The median data show an increase in the estimated probability of losing balance with the increase in frequency (Fig. 14). The effect of frequency is shown in Table 8 for pitch oscillation and Table 9 for roll oscillation.

3.3. Correlations with body characteristics

3.3.1. Translational oscillation

The subjects varied in age, stature, weight, foot length and foot width. With translational excitation, age was positively correlated with the standard deviation of the COP in all conditions with significant correlations at 0.125 Hz with 0.062 ms^{-1} rms during fore-and-aft excitation and at 0.125 Hz with 0.04 ms^{-1} rms and at

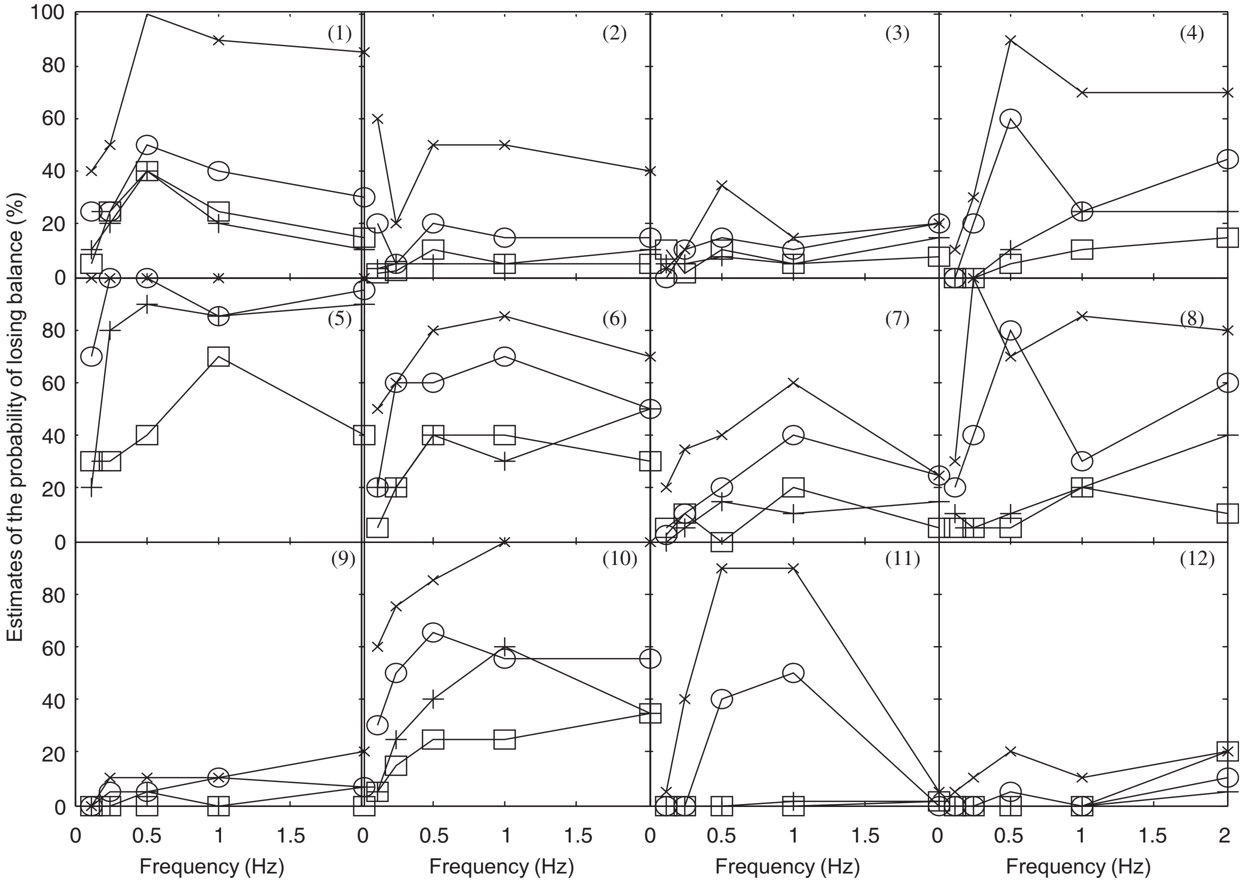

Fig. 8. Estimates of the probability of losing balance during fore-and-aft oscillation. □, 0.04; +, 0.062; ○, 0.099; ×, 0.16 ms^{-1} rms. Data from 12 subjects.

0.25 Hz with 0.062 ms^{-1} rms during lateral excitation indicating more sway in older subjects (Spearman, $p<0.05$).

There was no significant correlation between the standard deviation of the centre of pressure and stature in any condition, except at 2 Hz with 0.16 ms^{-1} rms during fore-and-aft excitation, where there was a negative correlation indicating less sway in taller subjects (Spearman, $p<0.05$). During lateral oscillation, no significant correlation was found between the displacement of the COP and the stature of subjects. However, the Spearman coefficient was negative with frequencies below 1 Hz and positive with frequencies at, and above, 1 Hz, possibly indicating different response strategies and mechanisms at low frequency from those at high frequencies.

Subject weight was negatively correlated with the standard deviation of the centre of pressure in five conditions during fore-and-aft excitation (0.125 Hz with 0.04 ms^{-1} rms, $p<0.01$; 0.25 Hz with 0.04 ms^{-1} rms, 1 Hz with 0.04 ms^{-1} rms, 1 Hz with 0.099 ms^{-1} rms, and 2 Hz with 0.099 ms^{-1} rms, $p<0.05$) and in one condition during lateral excitation (0.5 Hz with 0.099 ms^{-1} rms) ($p<0.05$).

Foot length was negatively correlated with the standard deviation of the displacement of the centre of pressure at 2 Hz with 0.099 ms^{-1} rms during fore-and-aft excitation and positively correlated at 2 Hz with 0.16 ms^{-1} rms during lateral excitation ($p<0.05$). Foot width was negatively correlated ($p<0.05$) with the standard deviation of the centre of pressure in four conditions during fore-and-aft excitation (0.25 Hz with 0.062 ms^{-1} rms, 0.5 Hz with 0.062 ms^{-1} rms, 1 Hz with 0.099 ms^{-1} rms, and 2 Hz with 0.099 ms^{-1} rms) and in five conditions during lateral excitation (0.25 Hz with 0.099 ms^{-1} rms, 0.25 Hz with 0.16 ms^{-1} rms, and 0.5 Hz with 0.04, 0.062, and 0.099 ms^{-1} rms).

Table 6
Effect of the frequency of fore-and-aft oscillation on subject estimates of the probability of losing balance if the same stimulus was repeated

Velocity (ms^{-1} rms)	Frequency (Hz)	0.125	0.25	0.5	1.0	2.0
0.04	0.125	—	ns	*	*	**
	0.25		—	↑	*	*
	0.5			—	ns	ns
	1.0				—	ns
	2.0					—
0.062	0.125	—	↑	**	**	**
	0.25		—	**	*	*
	0.5			—	ns	ns
	1.0				—	ns
	2.0					—
0.099	0.125	—	*	**	**	**
	0.25		—	**	ns	*
	0.5			—	ns	ns
	1.0				—	ns
	2.0					—
0.16	0.125	—	*	**	**	*
	0.25		—	*	*	ns
	0.5			—	ns	↑
	1.0				—	ns
	2.0					—

Wilcoxon matched-pairs signed ranks: ↑ = $p<0.1$, * = $p<0.05$, ** = $p<0.01$, ns = $p>0.1$.

The age, weight and foot width were not correlated with estimates of the probability of losing balance in any condition. However, stature was negatively correlated with estimates of the probability of losing balance at 1 Hz with 0.099 ms^{-1} rms during fore-and-aft excitation and at 0.125 Hz with 0.099 ms^{-1} rms, 0.5 Hz with 0.099 ms^{-1} rms, 1 Hz with 0.04 ms^{-1} rms, and 2 Hz with 0.04 ms^{-1} rms during lateral excitation ($p<0.05$). The length of the feet was negatively correlated with estimates of the probability of losing balance at 0.125 Hz with 0.062 ms^{-1} rms during fore-and-aft excitation and at 0.125 with 0.099 ms^{-1} rms, 0.5 Hz with 0.099 ms^{-1} rms, and 0.5 Hz with 0.16 ms^{-1} rms with lateral excitation.

3.3.2. Rotational oscillation

During rotational excitation, age was positively correlated with estimates of the probability of losing balance with both pitch and roll oscillation in all conditions, although the correlation was not statistically significant in any condition. With both pitch and roll oscillation, estimates of the probability of losing balance were negatively correlated with body weight and foot width in all conditions, but the correlations were statistically significant with weight and foot width only for pitch oscilaltion at 0.125 Hz with 0.125 ms^{-2} rms ($p<0.05$). There were also significant negative correlations between stature and estimates of the probability of losing balance in 11 of the 18 conditions during pitch excitation and in 12 of 18 conditions during roll excitation ($p<0.05$). Foot length was also negatively correlated with estimates of the probability of losing balance in five conditions during pitch excitation but only two conditions with roll excitation ($p<0.05$).

4. Discussion

The subjective and objective methods used in this study indicated similar effects of the magnitude of oscillation: the loss of balance, estimates of the probability of losing balance (with both translational and rotational oscillation), and the displacement of the centre of pressure (with translational oscillation) all increased with increased magnitude of oscillation. Increases in the magnitude of oscillation may be expected to

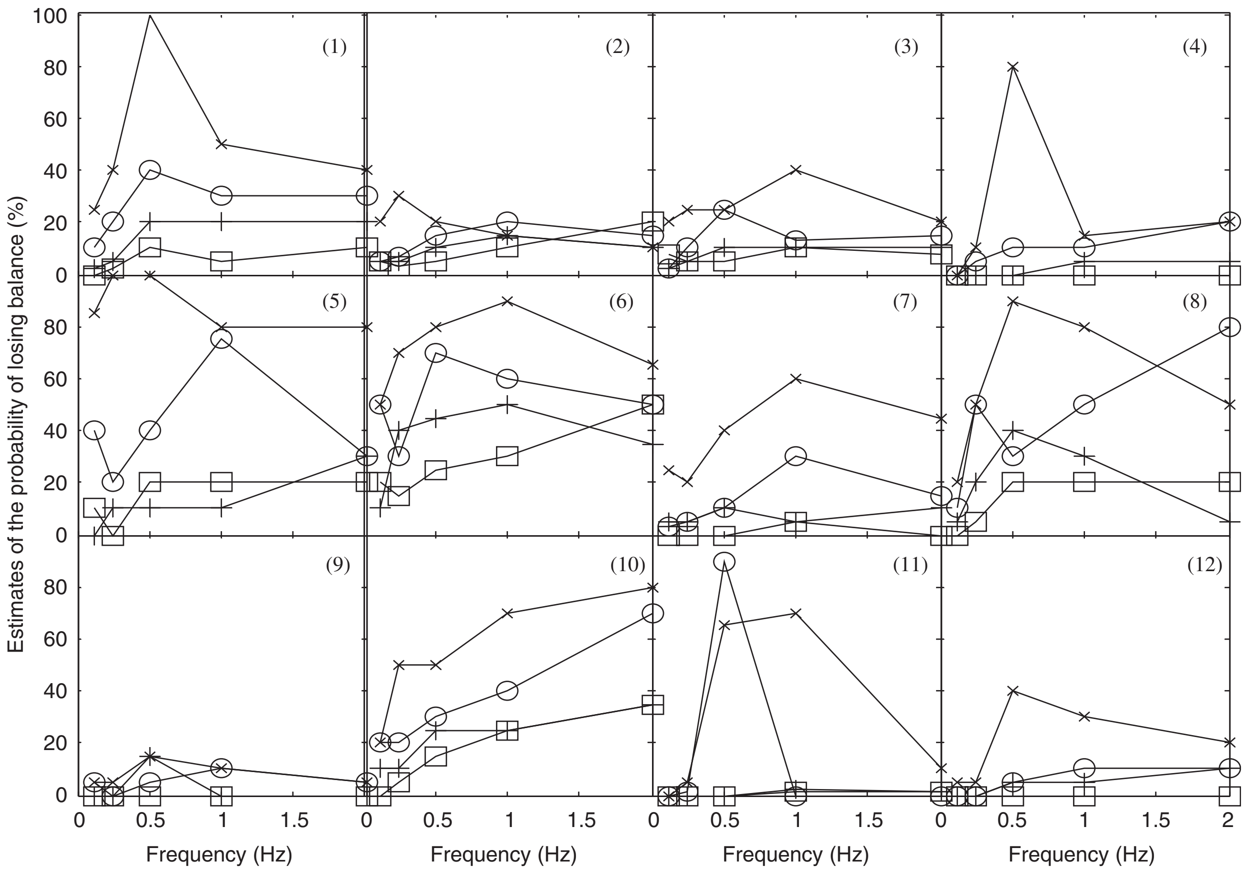

Fig. 9. Estimates of the probability of losing balance during lateral oscillation. □, 0.04; +, 0.062; ○, 0.099; ×, 0.16 ms^{-1} rms. Data from 12 subjects.

increase muscle activation and the movement of various parts of the body, which will be reflected in movements of the centre of pressure and subjects' estimates of the probability of losing balance [19].

Biodynamic responses of seated people to fore-and-aft excitation (apparent mass: the complex ratio between the measured force and the applied acceleration) have been reported to be nonlinear: with increases in the acceleration decreasing the apparent mass at frequencies in the approximate range 0.125–2 Hz [20,21]. In this paper, the relationship between the input velocity and the output COP was also nonlinear, as observed in a few previous studies. Gong [7] measured the transfer function between the input acceleration and the output COP with various motion waveforms (sinusoidal, 1/3 octave random and octave random) at 0.063, 0.125, 0.25, 0.5 and 1.0 Hz with a range of vibration magnitudes. The transfer function for all types of input depended on the vibration magnitude, indicating a nonlinear relationship: the transfer function between the applied acceleration and the displacement of the centre of pressure decreased with increases in the magnitude of the oscillation. Maki [22] also reported that the gain in the transfer function between horizontal platform acceleration and COP displacements decreased as the acceleration magnitude increased. The position of the COP reflects the reaction forces on the platform coming from the inertia of the body and the activation of muscles and, hence, nonlinear response of body muscles. The nonlinear postural response could also be due to variations in the subject strategy used to maintain balance at different motion magnitudes. For example, Nashner and McCollum [23] reported that subjects tended to keep knees, hips, and neck fairly straight and sway predominantly about their ankles when they were exposed to small disturbances but they tended to move at the hip and the ankles when the perturbation placed their centre of pressure near the limits of foot support.

Table 7
Effect of the frequency of lateral oscillation on subject estimates of the probability of losing balance if the same stimulus was repeated

Velocity (ms^{-1} rms)	Frequency (Hz)	0.125	0.25	0.5	1.0	2.0
0.04	0.125	—	ns	*	**	*
	0.25		—	*	**	*
	0.5			—	↑	↑
	1.0				—	ns
	2.0					—
0.062	0.125	—	*	**	**	**
	0.25		—	**	**	↑
	0.5			—	ns	ns
	1.0				—	ns
	2.0					—
0.099	0.125	—	ns	**	**	*
	0.25		—	*	**	**
	0.5			—	ns	ns
	1.0				—	ns
	2.0					—
0.16	0.125	—	**	**	**	*
	0.25		—	*	*	ns
	0.5			—	ns	*
	1.0				—	*
	2.0					—

Wilcoxon matched-pairs signed ranks: ↑ $= p<0.1$, * $= p<0.05$, ** $= p<0.01$, ns $= p>0.1$.

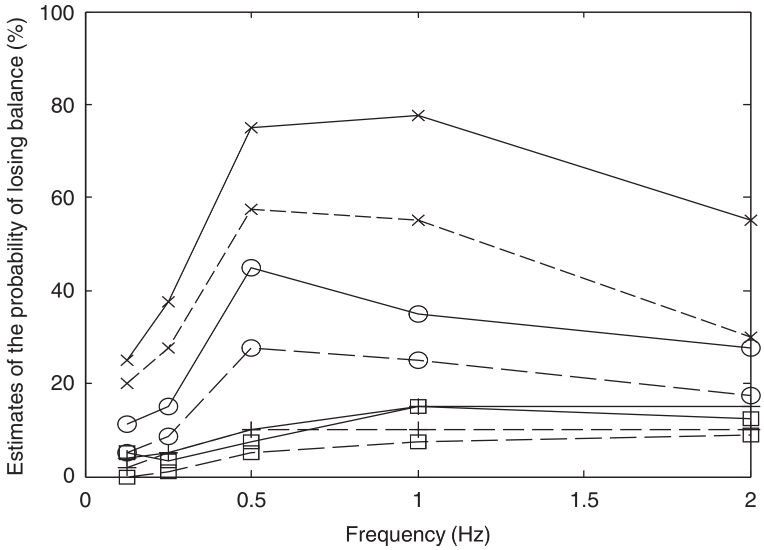

Fig. 10. Median subject estimates of the probability of losing balance: (——, fore-and-aft oscillation); (– – –, lateral oscillation); □, 0.04; +, 0.062; ○, 0.099; ×, 0.16 ms^{-1} rms.

The fore-and-aft displacements of the centre of pressure during fore-and-aft oscillation were similar to the lateral displacements of the COP during lateral oscillation (Fig. 4). However, fore-and-aft oscillation induced greater loss of balance (Fig. 7) and produced higher estimates of the probability of losing balance than lateral oscillation (Fig. 10). The loss of balance and estimates of the probability of losing balance were also higher during pitch oscillation than during roll oscillation. The base of support in the fore-and-aft direction (i.e. the

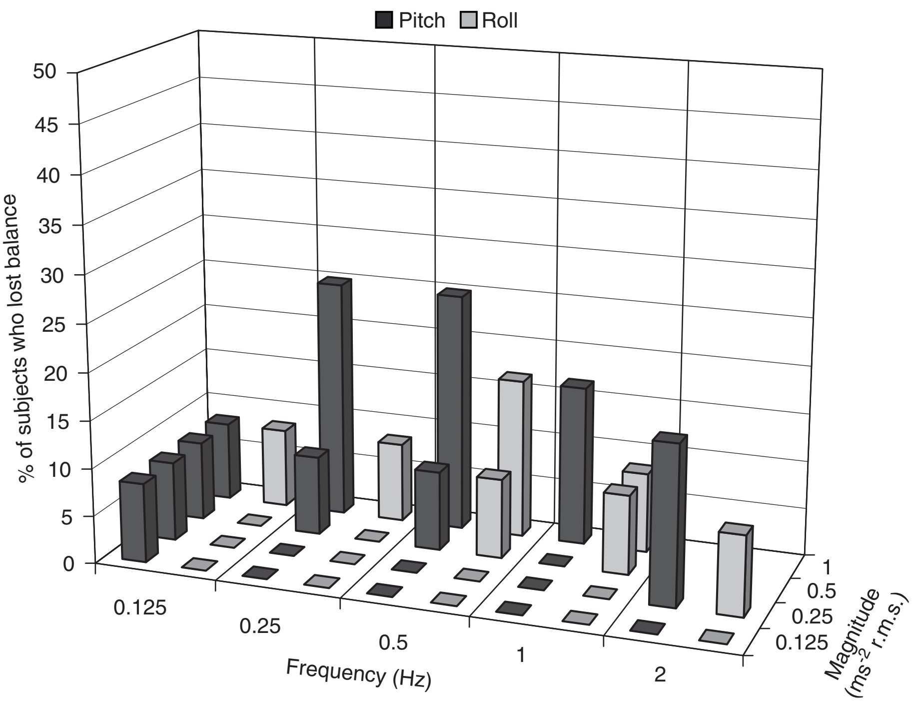

Fig. 11. Percentage of subjects who lost balance during pitch and roll oscillation with four magnitudes of acceleration at frequencies from 0.125 to 2 Hz.

length of the feet) is shorter than the base of support in the lateral direction (i.e. the separation between the feet), so a movement of the COP in the backward–forwards direction will cause more instability than the same movement in the sideways direction. However, in some circumstances the base of support in the lateral direction can be reduced and even smaller than the base of support in the fore-and-aft direction, such as with feet placed together or supported on one foot (such as during walking). Further study is required to extend the results to situations with other foot separations and during walking.

The instability caused by fore-and-aft oscillation and lateral oscillation has been determined separately in the present study, but in many situations there is simultaneous fore-and-aft and lateral oscillation. Vertical acceleration may also alter the reaction forces at the floor and postural stability. In addition, roll and pitch motion may contribute to instability in transportation, where more than one direction of motion occurs simultaneously. The effects of combined motions on the postural stability of standing subjects could be very different from the effect of each axis individually. When exposed to combined fore-and-aft and lateral oscillation, the resolving of the component motions into the fore-and-aft and lateral direction might seem logical, but assumes that postural control is linear and that the body responds independently to motions in these axes. The responses of muscles, such as muscles acting on the legs and hips, which are expected to affect the movements of the COP are highly dependent on the direction of oscillation [24]. The present experiment has shown that response to motion in each direction is nonlinear, so a linear model should not be expected to predict the response to a combination of directions of motion.

Gong [6] measured the transfer function in the frequency range 0.05–1.0 Hz between four acceleration inputs (0.027, 0.054, 0.081, and 0.108 ms^{-2} rms) and the displacement of the centre of pressure. The coherency was high (between 0.55 and 0.9) over the whole frequency range with high acceleration magnitudes but dropped with low magnitudes, especially at low frequencies (less than 0.2 Hz). The high coherency suggested the acceleration and the COP were linearly related and the gain of the transfer functions did not appear to differ between vibration magnitudes. However, Gong [7] reported a nonlinear transfer functions between the acceleration and the displacement of the COP when using narrow bandwidth motions. This led Gong [7] to

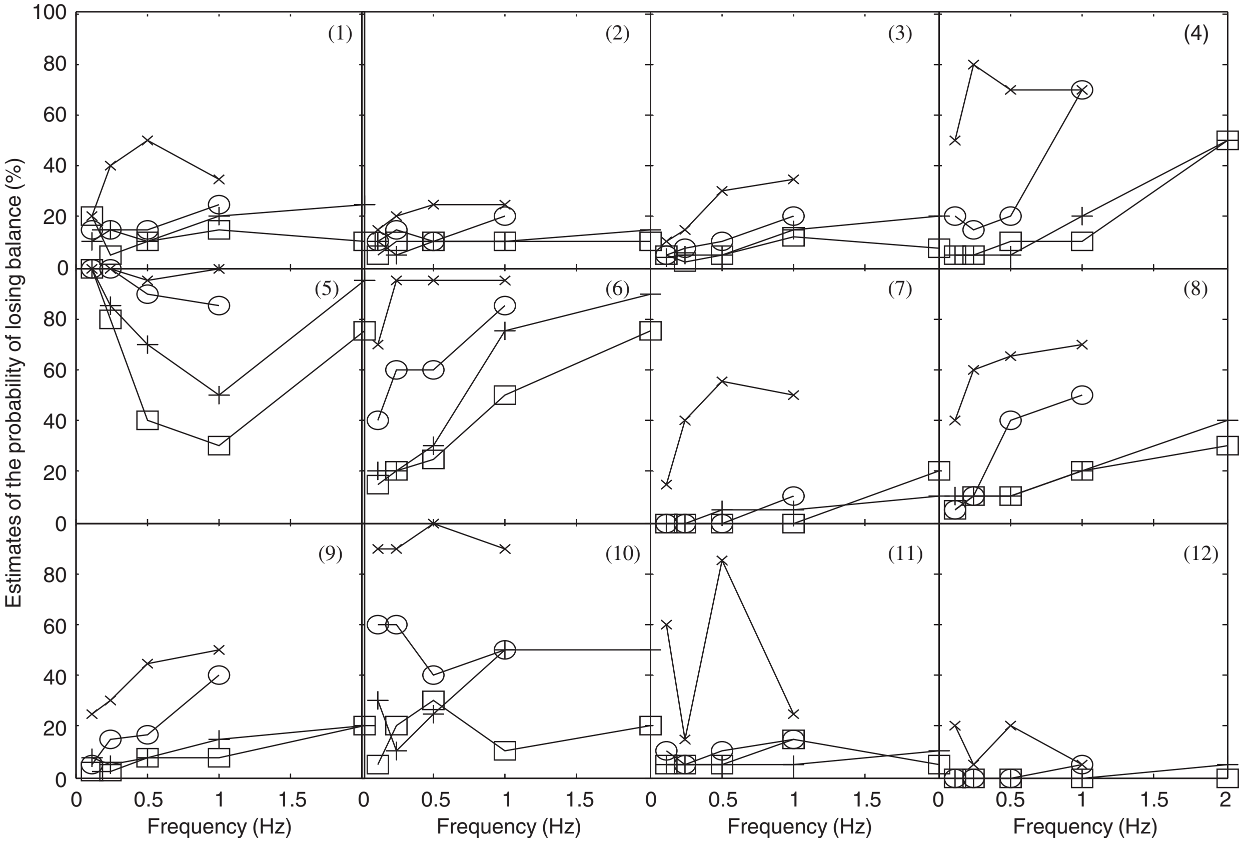

Fig. 12. Estimates of the probability of losing balance during pitch oscillation. □, 0.125; +, 0.25; ○, 0.5; ×, 1 ms^{-2} rms. Data from 12 subjects.

conclude that the nonlinearity of the transfer function between the displacement of the COP and the input acceleration depended on whether the bandwidth was narrow or wide.

With no motion of the floor, the movement of the centre of pressure in the fore-and-aft and lateral directions showed more sway in the fore-and-aft direction than in the lateral direction (Fig. 15). During 'quiet' standing, corrections are required to maintain an upright position of the inherently unstable body [25], with some subjects swaying more than others. Era and Heikkinen [26] reported that the magnitude of the sway (and the deviation of the COP) increased with age. This is consistent with the positive correlation found in this study between the age of the subjects and the displacement of the COP, although the correlation was statistically significant in only one condition with fore-and-aft oscillation and in two conditions with lateral oscillation (see Section 3.3).

Other factors may have contributed to inter-subject variability in both experiments. Biodynamic responses as well as body co-ordination between the parts of the body have been suggested to affect the transfer function between the torque at the ankle joint and the sway angle of the body [11]. There are no known studies of the driving point responses of the standing human body exposed to horizontal or rotational oscillation. High inter-subject variability has been reported in the transmissibility from horizontal floor vibration to the heads of standing subjects (e.g. [27]). Studies with seated subjects exposed to fore-and-aft vibration [20,21] and lateral and roll vibration [28,29] have also reported high inter-subject variability in apparent mass. It may be expected that the biodynamic responses of standing people will also show high inter-subject variability in biodynamic responses and that this will be reflected in their postural stability during exposure to motion.

Estimates of the probability of losing balance during translational oscillation differed from those during rotational oscillation (see Fig. 16 for fore-and-aft and pitch oscillation and Fig. 17 for lateral and roll

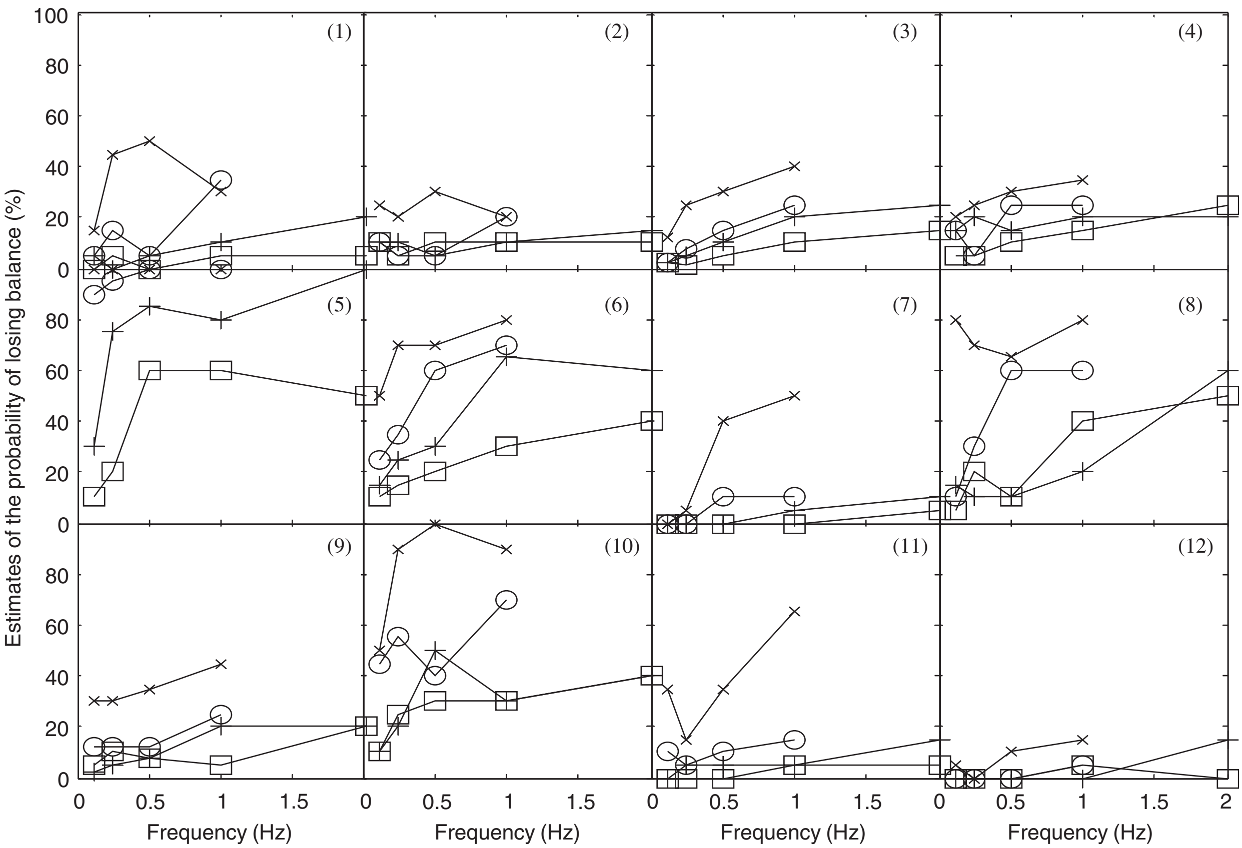

Fig. 13. Estimates of the probability of losing balance during roll oscillation. □, 0.125; +, 0.25; ○, 0.5; ×, $1\,\mathrm{ms}^{-2}$ rms. Data from 12 subjects.

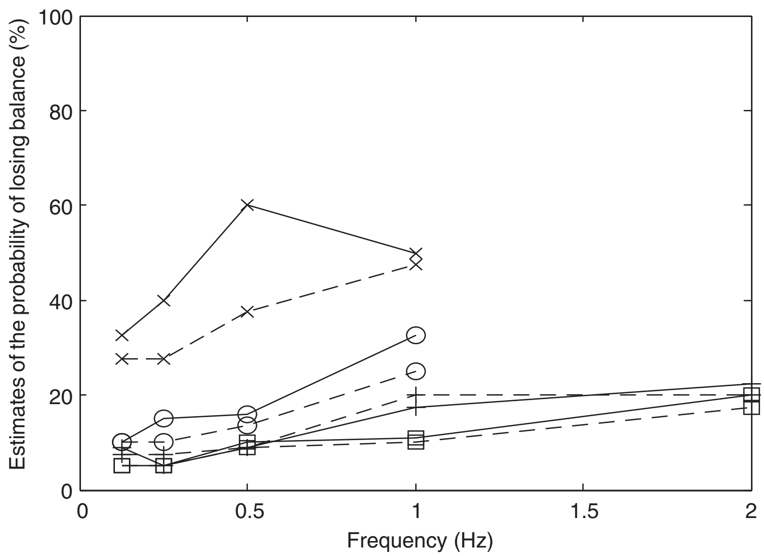

Fig. 14. Median subject estimates of the probability of losing balance: (——, pitch oscillation); (– – –, roll oscillation); □, 0.125; +, 0.25; ○, 0.5; ×, $1\,\mathrm{ms}^{-2}$ rms.

Table 8
Effect of the frequency of roll oscillation on subject estimates of the probability of losing balance if the same stimulus was repeated during pitch motion

Acceleration (ms^{-2} rms)	Frequency (Hz)	0.125	0.25	0.5	1.0	2.0
0.125	0.125	—	ns	ns	ns	↑
	0.25		—	ns	ns	*
	0.5			—	ns	*
	1.0				—	*
	2.0					—
0.25	0.125	—	ns	ns	↑	**
	0.25		—	ns	↑	**
	0.5			—	↑	**
	1.0				—	**
	2.0					—
0.5	0.125	—	ns	ns	*	—
	0.25		—	ns	*	—
	0.5			—	**	—
	1.0				—	—
	2.0					—
1.0	0.125	—	ns	**	ns	—
	0.25		—	*	↑	—
	0.5			—	ns	—
	1.0				—	—
	2.0					—

Wilcoxon matched-pairs signed ranks: ↑ = $p<0.1$, * = $p<0.05$, ** = $p<0.01$, ns = $p>0.1$.

Table 9
Effect of the frequency of pitch oscillation on subject estimates of the probability of losing balance if the same stimulus was repeated during roll motion

Acceleration (ms^{-2} rms)	Frequency (Hz)	0.125	0.25	0.5	1.0	2.0
0.125	0.125	—	↑	*	**	**
	0.25		—	ns	*	**
	0.5			—	*	*
	1.0				—	↑
	2.0					—
0.25	0.125	—	ns	*	**	**
	0.25		—	ns	*	**
	0.5			—	ns	**
	1.0				—	*
	2.0					—
0.5	0.125	—	ns	*	**	—
	0.25		—	ns	**	—
	0.5			—	*	—
	1.0				—	—
	2.0					—
1.0	0.125	—	ns	*	**	—
	0.25		—	*	*	—
	0.5			—	ns	—
	1.0				—	—
	2.0					—

Wilcoxon matched-pairs signed ranks: ↑ = $p<0.1$, * = $p<0.05$, ** = $p<0.01$, ns = $p>0.1$.

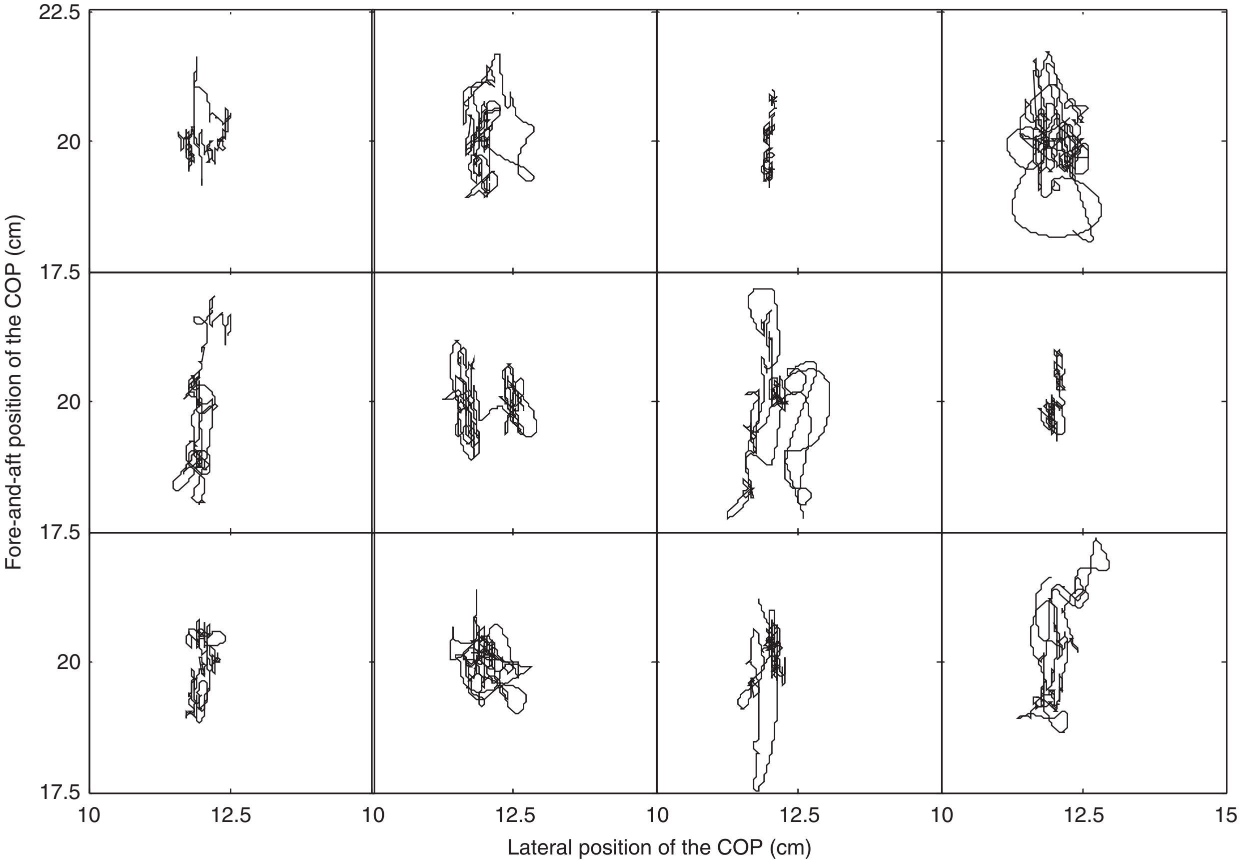

Fig. 15. Fore-and-aft and lateral co-ordinates of the position of the centre of pressure (COP) of 12 subjects during 60 s of static standing. The origin is located at the force cell where $F_3(t)$ was measured (see Fig. 1).

oscillation). The accelerations caused by rotation through gravity during rotational oscillation at low frequencies (below 1 Hz) were greater than those during translational oscillation. Nevertheless, Figs. 16 and 17 suggest that below 1 Hz the postural instability caused by translational oscillation was greater than that caused by rotational oscillation.

All three dependent variables (loss of balance, estimated probability of losing balance, and the displacement of the centre of pressure) indicated that postural stability was most affected by fore-and-aft and lateral oscillation at 0.5 Hz. However, this is partly due to the stimuli used in the experiment (the same velocity at all frequencies). Using stimuli with the same acceleration or the same displacement will give different results (i.e. different shapes for the curves). During pitch and roll oscillation, the estimates of the probability of losing balance, generally increased with increasing frequency of oscillation.

Human responses to oscillatory motions are often 'modelled' by linear frequency weightings of the input acceleration. To construct acceleration frequency weightings from the displacement of the COP and the estimated probability of losing balance obtained with the velocity inputs used in the experiment, the displacement of the COP and the estimated probability of losing balance were divided by the central frequency of that motion, assuming a linear response. The frequency weightings shown in Fig. 18 were normalised to 0.5 Hz by dividing the standard deviation (SD) of the displacement of the COP at each frequency by the standard deviation of the COP with the same input magnitude at 0.5 Hz. The frequency weightings shown in Fig. 19 were normalised to 0.5 Hz by dividing the estimated probability of losing balance at each frequency by the estimated probability of losing balance with the same input magnitude at 0.5 Hz. The figures show broadly similar frequency weightings for fore-and-aft and lateral excitation for both the displacement of the COP and estimates of probability of losing balance. It seems more appropriate to use subjective responses (i.e. estimates

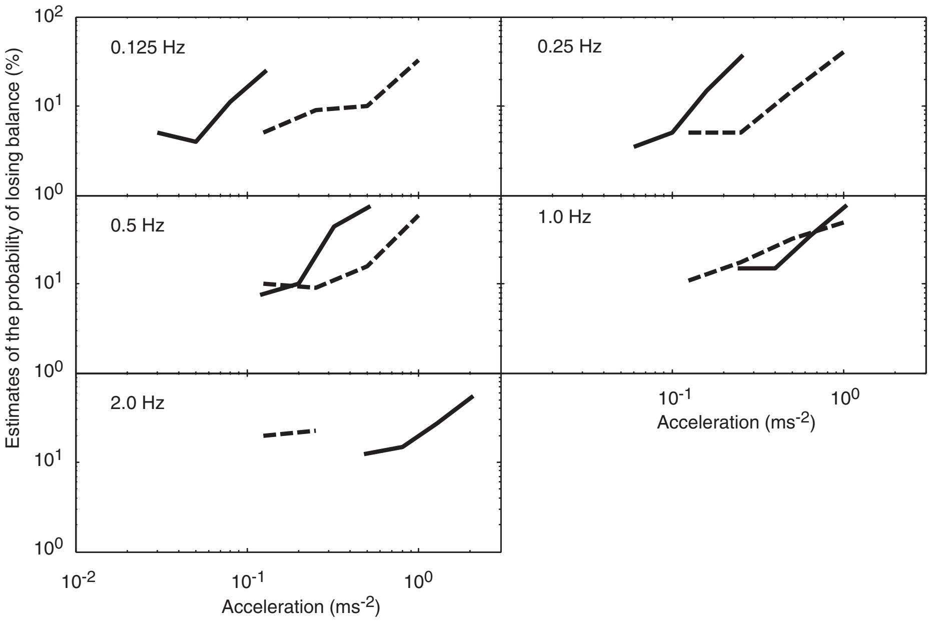

Fig. 16. Effect of oscillation magnitude on the median estimates of the probability of losing balance during fore-and-aft and pitch oscillation at frequencies from 0.125 to 2.0 Hz.——, fore-and-aft; – – –, pitch.

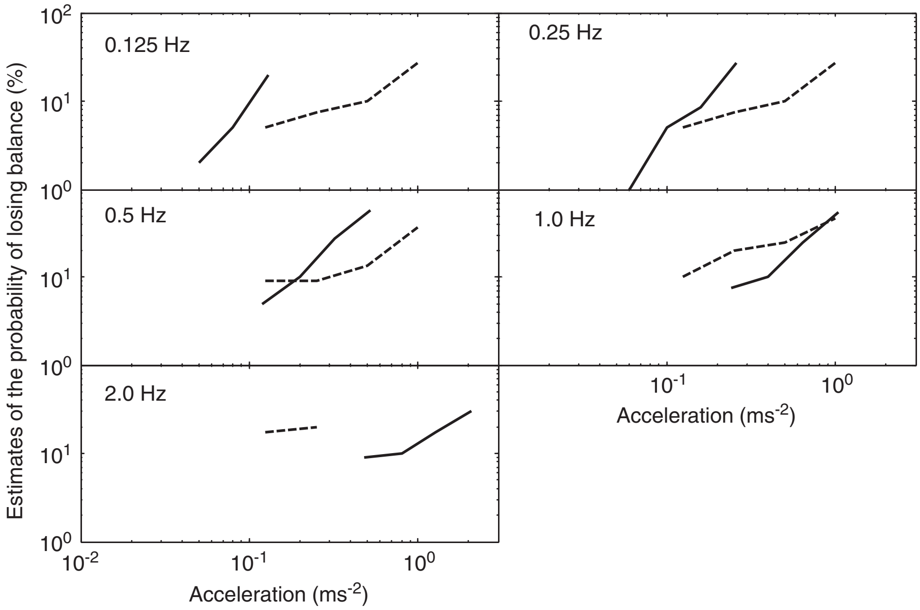

Fig. 17. Effect of oscillation magnitude of the median estimates of the probability of losing balance during lateral and roll oscillation at frequencies from 0.125 to 2.0 Hz.——, lateral; – – –, roll.

of the probability of losing balance) for a frequency weighting because it represents feelings and reflects the different sensitivity to fore-and-aft and lateral displacement of the COP.

The acceleration frequency weightings depend on the applied magnitude of translational oscillation (Figs. 20 and 21). Due to this nonlinear behaviour, different frequency weightings might be needed to predict the probability of losing balance with different magnitudes of oscillation.

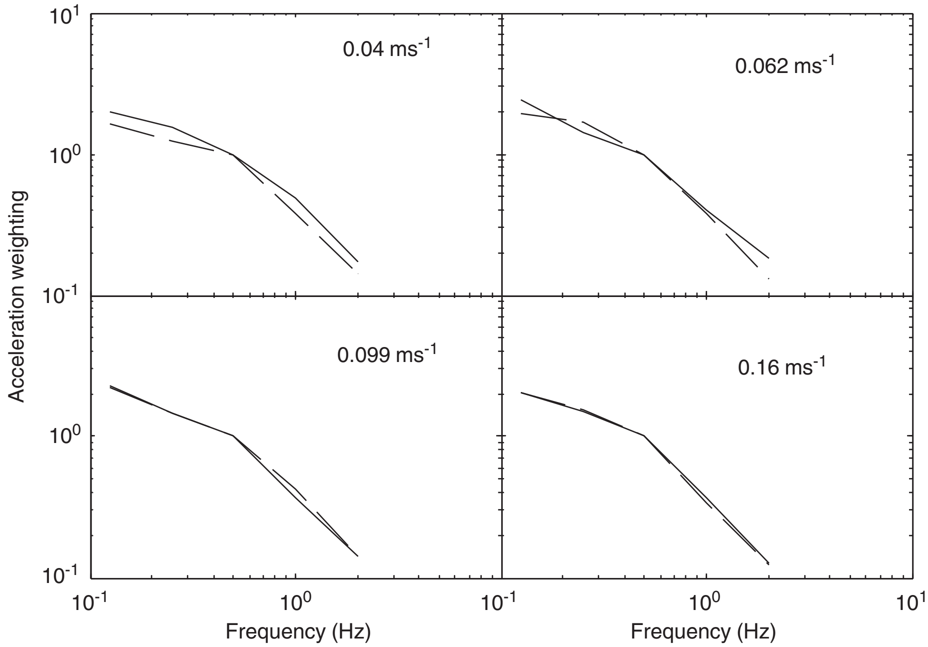

Fig. 18. Acceleration frequency weightings based on the standard deviation of the centre of pressure in the fore-and-aft and lateral directions during fore-and-aft and lateral oscillation, respectively. The data are referenced to the standard deviation of the COP measured at 0.5 Hz.——, fore-and-aft; – – –, lateral.

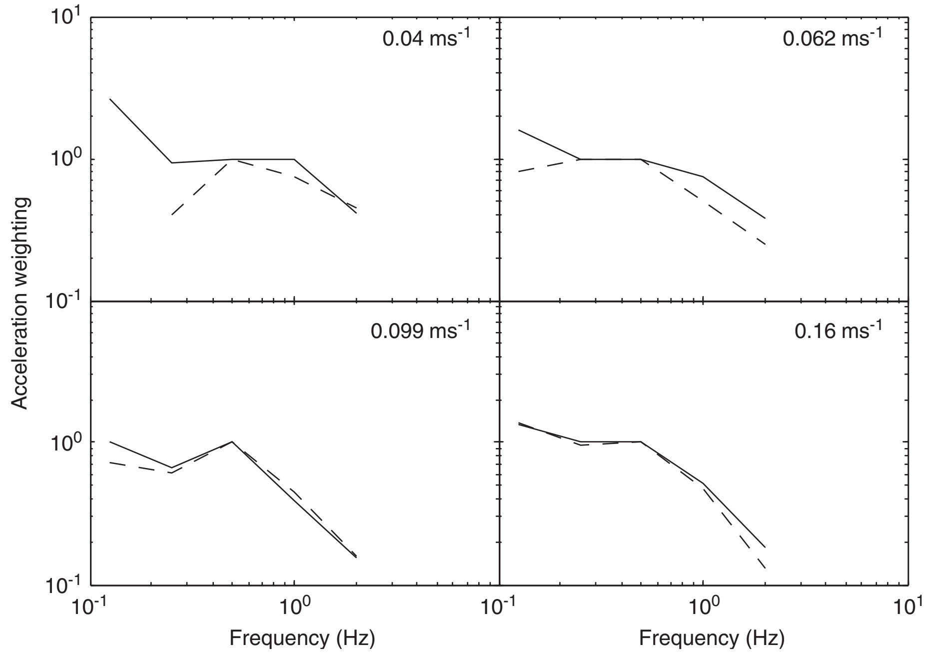

Fig. 19. Acceleration frequency weightings based on subject estimates of the probability of losing balance during fore-and-aft and lateral oscillation. The data are referenced to the estimates at 0.5 Hz.——, fore-and-aft; – – –, lateral.

Frequency weightings for roll and pitch oscillation based on subjects estimates of the probability of losing balance are shown in Fig. 22. The figure shows broadly similar weightings for the two axes of motion. It is clear from the figures that with rotational motion, unlike the response to translational motion, the subjects

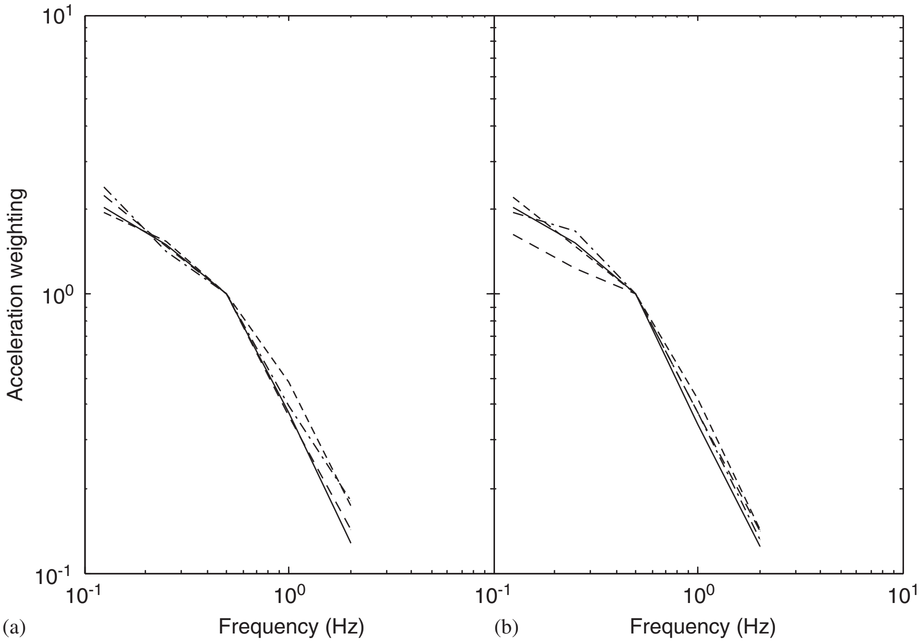

Fig. 20. Acceleration frequency weightings based on the standard deviation of the displacement of the centre of pressure at four oscillation velocities. · · · · , 0.04 ms^{-1}; – · – · –, 0.062 ms^{-1}; – – – –, 0.099 ms^{-1}; ———, 0.16 ms^{-1}. (a) fore-and-aft oscillation; (b) lateral oscillation.

were more sensitive to high-frequency acceleration than low-frequency acceleration. As was found with translational oscillation, the frequency weightings depend on the applied magnitude of oscillation (Fig. 23) indicating the need for different frequency weightings with different magnitudes of oscillation.

The results indicated that fore-and-aft oscillation caused more instability than lateral oscillation (see Fig. 10) and pitch oscillation caused more instability than roll oscillation (see Fig. 14). However, the frequency weightings shown in Figs. 19 and 20 indicate similar frequency weightings during fore-and-aft and lateral oscillation and during pitch and roll oscillation. The normalisation to 0.5 Hz mentioned in the previous paragraphs had the effect of concealing the effect of direction of oscillation. The median estimated probabilities of losing balance at 0.5 Hz during fore-and-aft oscillation were 7.5, 10, 45 and 75 compared to 5, 10, 27.5 and 57.5 during lateral oscillation at 0.04, 0.062, 0.099 and 0.16 ms^{-1} rms (corresponding to 0.126, 0.19, 0.31, and 0.5 ms^{-2} rms), respectively. During rotational oscillation, the median estimated probabilities of losing balance were 10, 9, 16, and 60 during pitch oscillation compared to 9, 9, 13.5, and 37.5 during roll oscillation at 0.125, 0.25, 0.5, and 1.0 ms^{-2} rms, respectively.

The standard deviation of the COP and, to a greater extent, estimates of the probability of losing balance, changed nonlinearly with oscillation magnitude during fore-and-aft and lateral oscillation (Fig. 24 and 25) and during pitch and roll oscillation (Fig. 26). Notwithstanding the nonlinearities, it is interesting to compare estimates of the probability of losing balance normalised to the magnitude of motion (i.e. the acceleration) during translational and rotational oscillation. Fig. 27 compares fore-and-aft and pitch oscillation while Fig. 28 compares lateral and roll oscillation. From these figures, it is clear that standing people are more sensitive to low-frequency acceleration than high-frequency acceleration when exposed to translational oscillation but they are more sensitive to high frequencies than low frequencies when exposed to the gravitational acceleration arising from rotational oscillation.

The frequency weightings do not model the mechanisms in the sensory and musculoskeletal systems controlling postural stability, but they do reflect the sensitivity of people to different frequencies of oscillation. Rigid body models do not take into account the effects of the frequency of oscillation or the durations of the forces applied to the body [15]. The response of a rigid body model depends only on the instantaneous force applied, without taking into account the preceding force time history: it does not depend on the position of the COG of the body.

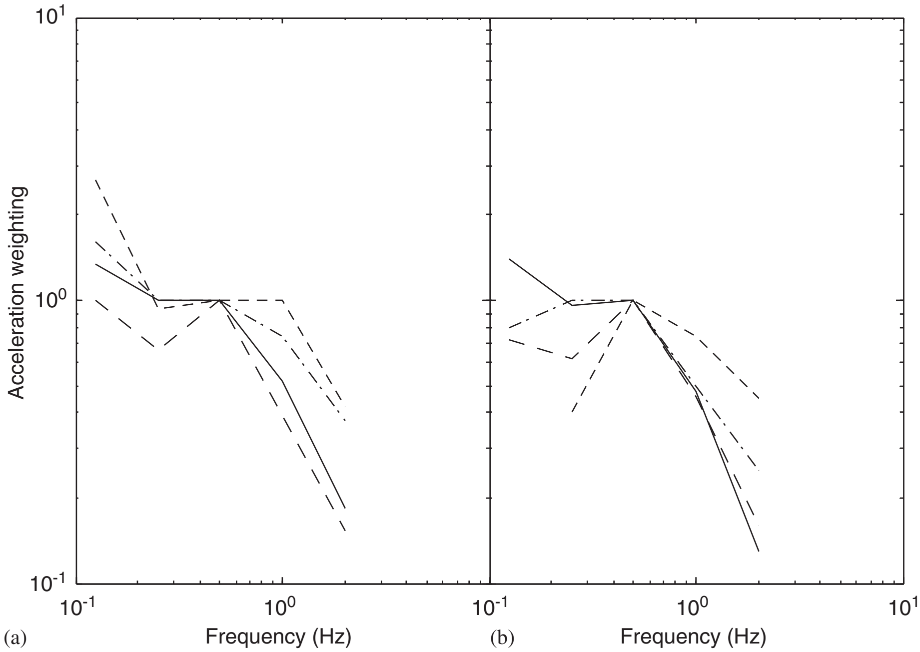

Fig. 21. Acceleration frequency weightings based on estimates of probability of losing balance at four oscillation velocities. · · · · , $0.04\,ms^{-1}$; – · – · –, $0.062\,ms^{-1}$; ––––, $0.099\,ms^{-1}$; ———, $0.16\,ms^{-1}$. (a) fore-and-aft oscillation; (b) lateral oscillation.

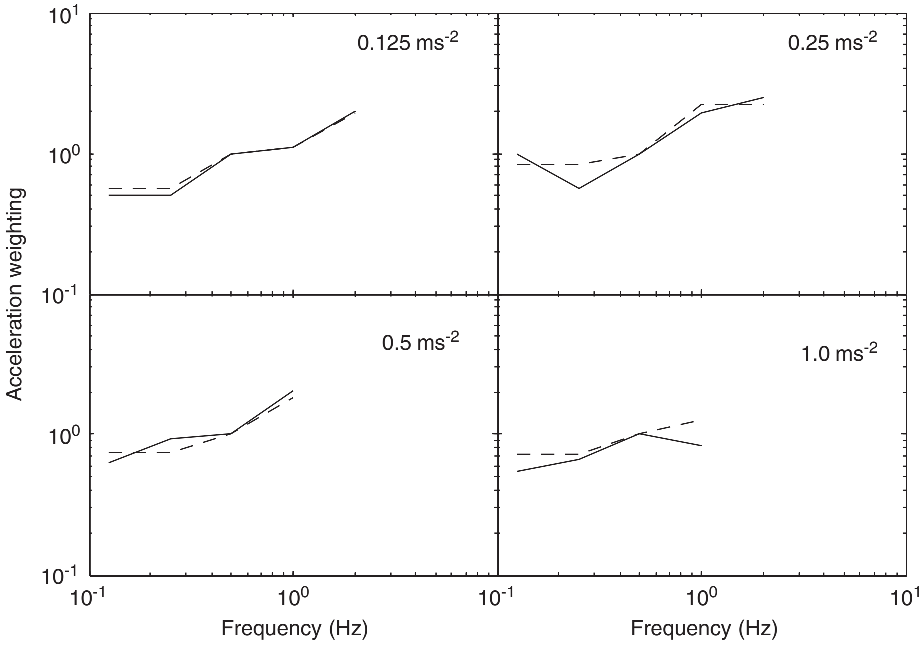

Fig. 22. Acceleration frequency weightings based on subject estimates of the probability of losing balance during pitch and roll oscillation. The data are referenced to the estimates at 0.5 Hz.——, pitch; –––, roll.

The single- and multi-link (passive and active) models seek to reflect understanding the postural control systems and quantify the relationships between input variables (such as the applied acceleration or body movements) and output variables such as the COP and the torque at a joint. However, these models do not currently provide accurate predictions of the probability of losing balance. Models that take account of the

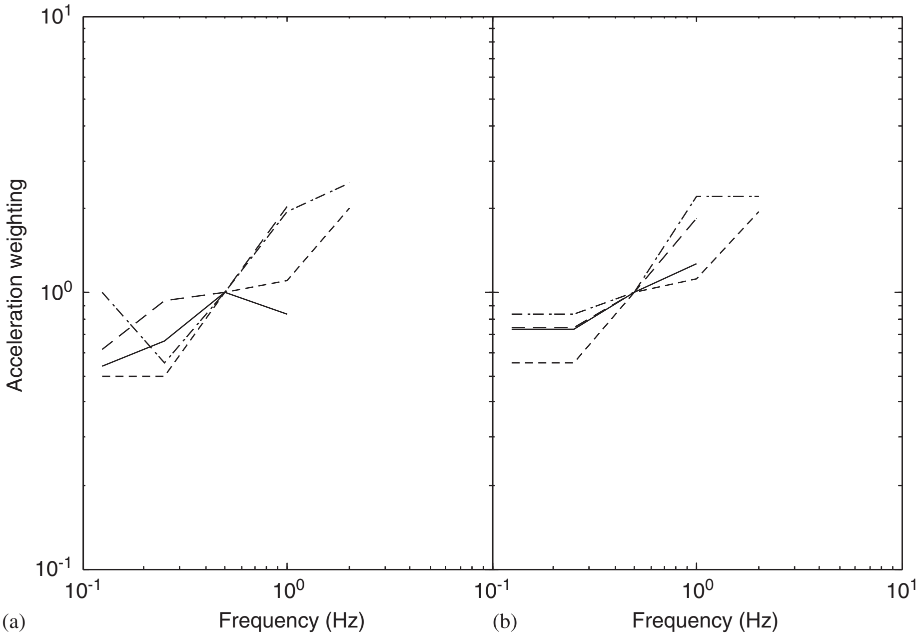

Fig. 23. Acceleration frequency weightings based on estimates of probability of losing balance at four oscillation accelerations. · · · · , $0.125\,\mathrm{ms^{-2}}$; – · – · –, $0.25\,\mathrm{ms^{-2}}$; – – – –, $0.5\,\mathrm{ms^{-2}}$; ———, $1.0\,\mathrm{ms^{-2}}$. (a) pitch oscillation; (b) roll oscillation.

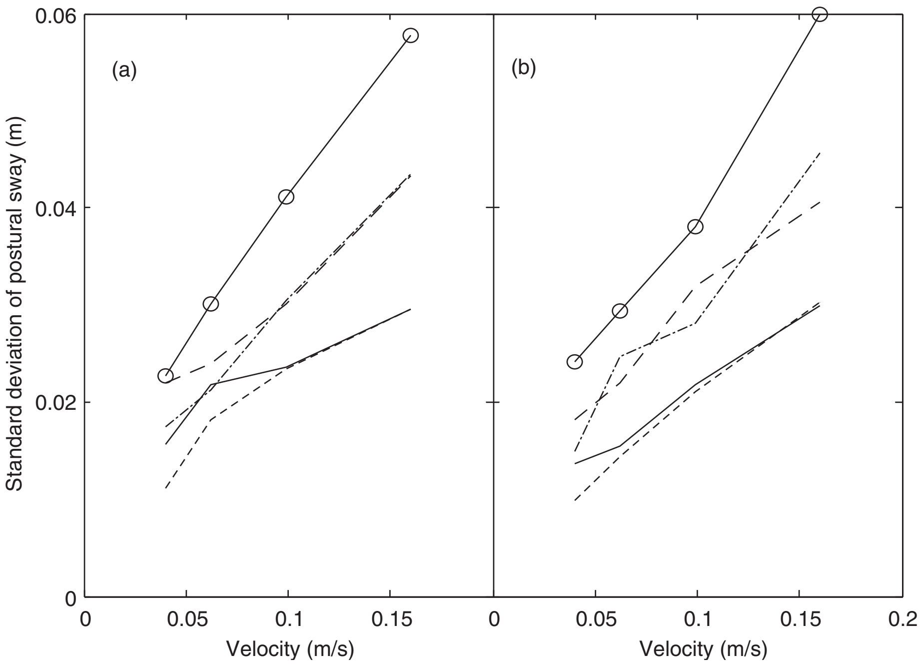

Fig. 24. Effect of oscillation magnitude on the standard deviation of the centre of pressure at 4 frequencies of translational oscillation. · · · · , 0.125 Hz; – · – · –, 0.25 Hz; —o—, 0.5 Hz; – – – –, 1 Hz; ———, 2 Hz. (a) fore-and-aft oscillation; (b) lateral oscillation.

subjective evaluation of motions (e.g. estimations of the probability of losing balance) may be more appropriate for the prediction of the probability of losing balance.

The present experiments were conducted with movement of only the platform on which the subjects stood while everything else around the subjects was stationary. Having a stationary scene around the subjects may

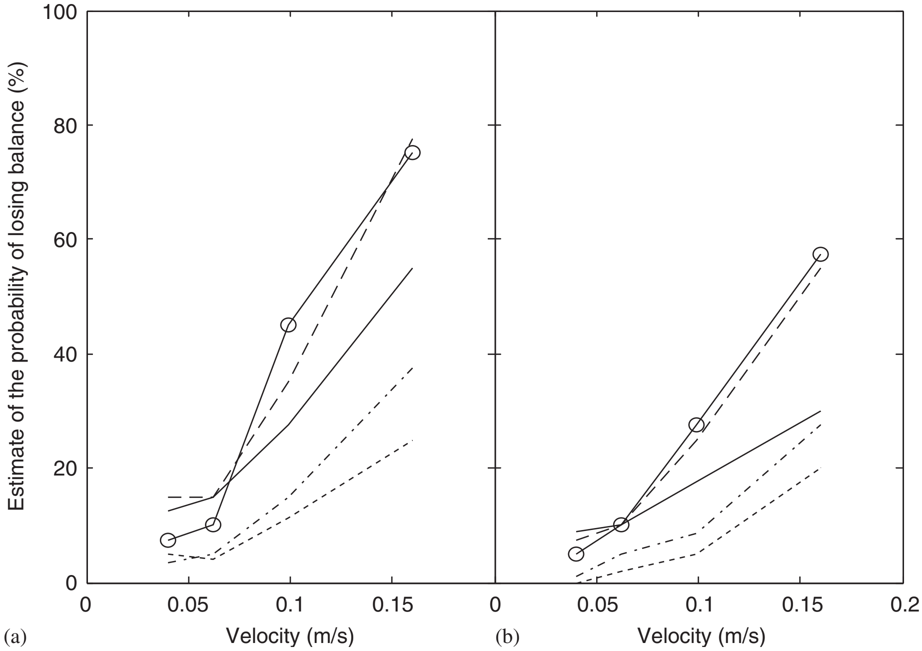

Fig. 25. Effect of oscillation magnitude on estimates of the probability of losing balance at 4 frequencies of translational oscillation. · · · · ·, 0.125 Hz; – · – · –, 0.25 Hz; —o—, 0.5 Hz; – – – –, 1 Hz; ———, 2 Hz. (a), fore-and-aft oscillation; (b) lateral oscillation.

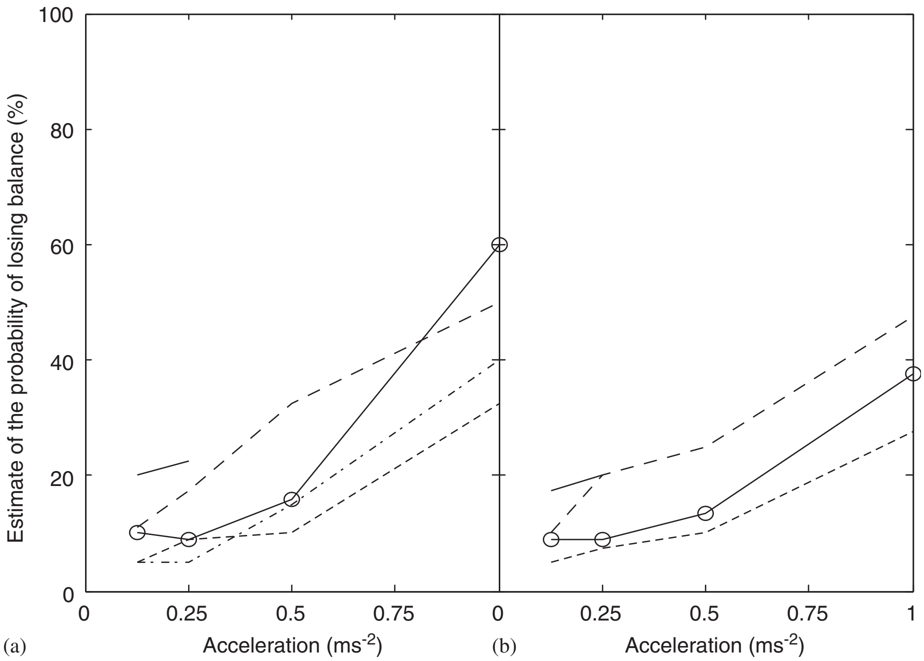

Fig. 26. Effect of oscillation magnitude on estimates of the probability of losing balance at 4 frequencies of rotational oscillation. · · · · ·, 0.125 Hz; – · – · –, 0.25 Hz; —o—, 0.5 Hz; – – – –, 1 Hz; ———, 2 Hz. (a) pitch oscillation; (b) roll oscillation.

have affected the results. On ships, and in trains, etc., the internal scene moves with the floor. Visual information can provide feedback to the postural stability control system. Further study is required to understand and model the influence of the visual scene on the results presented here.

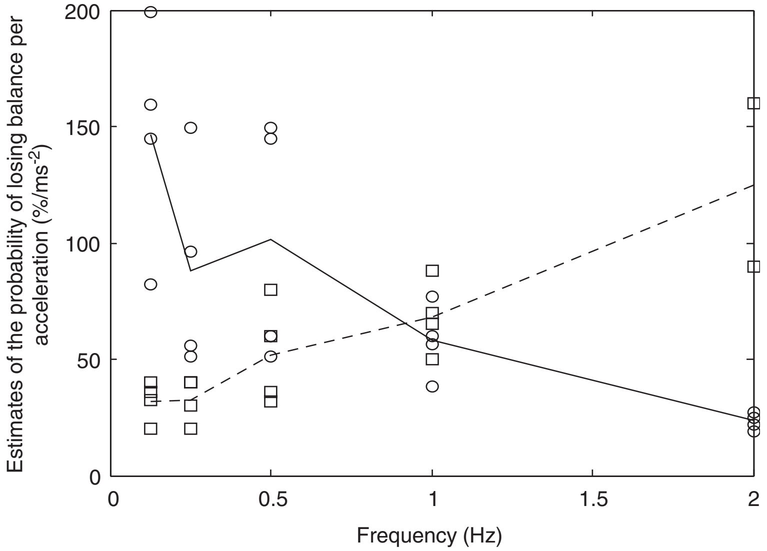

Fig. 27. Estimate of the probability of losing balance per acceleration. ○, during fore-and-aft oscillation; □, during pitch oscillation. ———, mean of estimates at 4 oscillation magnitudes during fore-and-aft oscillation; – – – –, mean of estimates at 4 oscillation magnitudes during pitch oscillation.

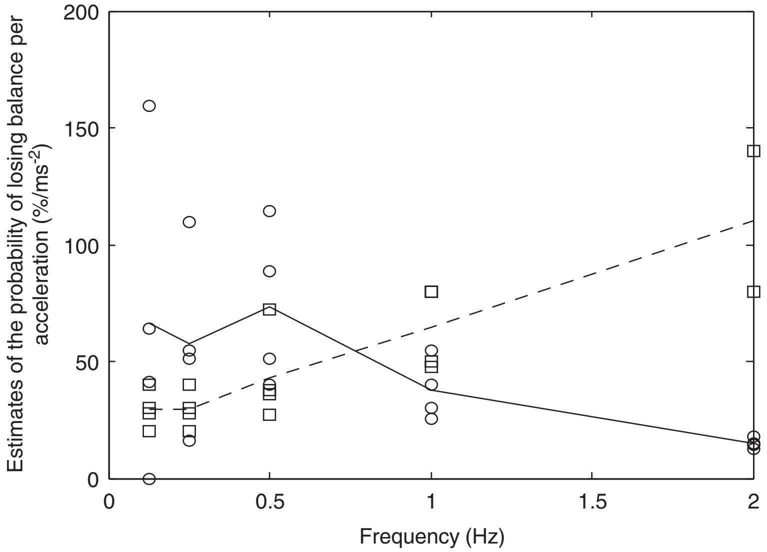

Fig. 28. Estimate of the probability of losing balance per acceleration. ○, during lateral oscillation; □, during roll oscillation. ———, mean of estimates at 4 oscillation magnitudes during lateral oscillation; – – – –, mean of estimates at 4 oscillation magnitudes during roll oscillation.

5. Conclusions

The postural stability of standing body depends on the magnitude, frequency, and direction of the applied oscillation.

The displacement of the centre of pressure, the loss of balance, and subject estimates of the probability of losing balance all increased with increasing magnitude of horizontal (fore-and-aft and lateral) oscillation. The displacement of the centre of pressure (COP) increased nonlinearly with increases in the magnitude of horizontal oscillation. Loss of balance, and subject estimates of the probability of losing balance, also increased with increases in the magnitude of rotational (i.e. pitch and roll) oscillation.

During fore-and-aft and lateral oscillation with the same velocity at all frequencies from 0.125 to 2 Hz, the displacement of the centre of pressure, the loss of balance, and subjective estimates of the probability of losing balance all peaked at around 0.5 Hz. During pitch and roll oscillation with the same angular displacements at all frequencies, there was a trend towards subjects being more likely to become unstable with the higher frequencies of oscillation.

Fore-and-aft oscillation caused more instability than lateral oscillation for the conditions used in this study. Similarly, pitch oscillation caused more instability than roll oscillation.

Standing people are more sensitive to low-frequency acceleration than high-frequency acceleration when exposed to translational (i.e. fore-and-aft or lateral) oscillation but they are more sensitive to high frequencies than low frequencies when exposed to the gravitational acceleration arising from rotational (i.e. pitch or roll) oscillation.

It is concluded that the postural stability of people standing on a floor exposed to both translational and rotational oscillations cannot be accurately predicted without separate consideration of the translational and rotational components of the motion.

Acknowledgement

This research was supported by the European Commission under the Framework 5 Growth programme—Project No. G3RD-2002-00809 (COMPASS).

References

[1] C.F. Runge, C.L. Shupert, F.B. Horak, F.E. Zajac, Ankle and hip strategies defined by joint torques, *Gait and Posture* 10 (1999) 161–170.

[2] Y.C. Pai, M.W. Rogers, J. Patton, T.D. Cain, T.A. Hanke, Static versus dynamic predictions of protective stepping following waist-pull perturbations in young and older adults, *Journal of Biomechanics* 31 (1998) 1111–1118.

[3] W.E. McIlroy, B.E. Maki, The control of lateral stability during rapid stepping reactions evoked by antero-posterior perturbation: does anticipatory control play a role?, *Gait and Posture* 9 (1999) 190–198.

[4] W.E. McIlroy, B.E. Maki, Early activation of arm muscles following external perturbation of upright stance, *Neuroscience Letters* 184 (1995) 177–180.

[5] Y.C. Pai, B.E. Maki, K. Iqbal, W.E. McIlroy, S.D. Perry, Thresholds of step initiation induced by support surface translation: a dynamic centre-of-mass model provides better prediction than static model, *Journal of Biomechanics* 33 (2000) 387–392.

[6] D. Gong, The effect of the magnitude and the bandwidth of horizontal floor vibration on postural responses in standing posture, Presented at the 32nd UK Group Meeting on Human Responses to Vibration, held at ISVR, University of Southampton, Southampton, England, 17–19 September 1997.

[7] D. Gong, Effect of narrow-band platform motions on induced postural responses in a standing posture, Presented at the 34th UK Group Meeting on Human Responses to Vibration, held at Ford Motor Company, Dunton, Essex, England, 22–24 September 1999.

[8] G.C. Agarwal, B.M. Berman, L. Stark, Studies in postural control systems part I: torque disturbance input, *IEEE Transactions on Systems Science and Cybernetics* 6 (1970) 116–121.

[9] R. Graham, Motion-induced interruptions as ship operability criteria, *Naval Engineers Journal* 102 (1990) 65–71.

[10] A. Ishida, S. Imai, Responses of the posture control system to pseudorandom acceleration disturbances, *Medical and Biological Engineering and Computing* 18 (1980) 433–438.

[11] H.P.M. Peeters, H.B. Caberg, J.M.F. Mol, Evaluation of biomechanical models in posturography, *Medical and Biological Engineering and Computing* 23 (1985) 469–473.

[12] S.H. Koozekanani, C.W. Stockwell, R.B. McGhee, F. Firoozmand, On the role of dynamic models in quantitative posturography, *IEEE Transactions on Biomedical Engineering* 27 (1980) 605–609.

[13] L. Kodde, H.B. Caberg, J.M.F. Mol, C.H. Massen, An application of mathematical models in posturography, *Journal of Biomedical Engineering* 4 (1982) 44–48.

[14] P.O. Riley, R.W. Mann, W.A. Hodge, Modelling of the biomechanics of posture and balance, *Journal of Biomechanics* 23 (1990) 503–506.

[15] C.H. Lewis, M.J. Griffin, Modelling the effects of deck motion on postural stability, ISVR Contract Report No. 95/12, Institute of Sound and Vibration Research, University of Southampton, UK, 1995.

[16] R. Johansson, M. Magnusson, M. Akesson, Identification of postural dynamics, *IEEE Transaction on Biomedical Engineering* 35 (1988) 858–869.

[17] A. Ishida, S. Miyazaki, Maximum likelihood identification of a posture control system, *IEEE Transactions on Biomedical Engineering* 34 (1987) 1–5.

[18] B.E. Maki, P.J. Holliday, G.R. Fernie, A posture control model and balance test for the prediction of relative postural stability, *IEEE Transactions on Biomedical Engineering* 34 (1987) 797–810.

[19] P. Era, M. Schroll, H. Ytting, I. Gause-Nilsson, E. Heikkinen, B. Steen, Postural balance and its sensory-motor correlates in 75-year-old men and women: a cross-national comparative study, *Journal of Gerontology: Biological Sciences and Medical Sciences* 51 (1996) M53–M63.

[20] N. Nawayseh, M.J. Griffin, Non-linear dual-axis biodynamic response to fore-and-aft whole-body vibration, *Journal of Sound and Vibration* 282 (2005) 831–862.

[21] N. Nawayseh, M.J. Griffin, Tri-axial forces at the seat and backrest during fore-and-aft whole-body vibration, *Journal of Sound and Vibration* 281 (2005) 921–942.

[22] B.E. Maki, Selection of perturbation parameters for identification of the posture-control system, *Medical and Biological Engineering and Computing* 24 (1986) 561–568.

[23] L.M. Nashner, G. McCollum, The organisation of human postural movements: a formal basis and experimental synthesis, *Behavioral and Brain Sciences* 8 (1985) 135–172.

[24] S.P. Moore, D.S. Rushmer, S.L. Windus, L.M. Nashner, Human automatic postural responses: responses to horizontal perturbations of stance in multiple directions, *Experimental Brain Research* 73 (1988) 648–658.

[25] L.M. Nashner, Practical biomechanics and physiology of balance, in: G.P. Jacobson (Ed.), *Handbook of Balance Function Testing*, Moskey-New Book, 1993.

[26] P. Era, E. Heikkinen, Postural sway during standing and unexpected disturbance of balance in random samples of men of different ages, *Journal of Gerontology* 40 (1985) 287–295.

[27] G.S. Paddan, M.J. Griffin, The transmission of translational floor vibration to the heads of standing subjects, *Journal of Sound and Vibration* 160 (1993) 503–521.

[28] T. Fairley, M.J. Griffin, The apparent mass of the seated human body in the fore-and-aft and lateral directions, *Journal of Sound and Vibration* 139 (1990) 299–306.

[29] T.P. Gunston, The apparent mass of the seated human body in response to lateral and roll vibration, in: 38th United Kingdom Conference on Human Response to Vibration, Institute of Naval Medicine, Alverstoke, Gosport, PO12 2DL, England, 17–19 September 2003.