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Postural stability when walking and exposed to mediolateral oscillatory motion: effect of oscillation waveform

Hatice Mujde Ayık[1] and Michael J. Griffin[1]

[1]Human Factors Research Unit, Institute of Sound and Vibration Research, University of Southampton, Southampton, SO17 1BJ, England

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Correspondence address:

Emeritus Professor Michael J. Griffin

Human Factors Research Unit

Institute of Sound and Vibration Research

University of Southampton

Southampton, SO17 1BJ

England

Tel: +44 (0) 2380 592277

Fax: +44 (0) 2380 592927

Email: M.J.Griffin@soton.ac.uk

Abstract

Postural stability can be threatened by the low frequency motions in transport that are usually quantified by their root-mean-square (r.m.s.) acceleration. This study investigated how the stability of walking people depends on the waveform of 1-Hz and 2-Hz mediolateral oscillations of the surface on which they walk. Walking on a treadmill, 20 subjects were perturbed by random oscillations of the treadmill with one-third octave bandwidths: different waveforms with the same r.m.s. acceleration and different waveforms with the same peak acceleration. Stability was measured subjectively and objectively by the velocity of the centre of pressure in the mediolateral direction. Subjective and objective measures of walking instability increased with increasing r.m.s. acceleration of oscillations having the same peak acceleration. These same measures of instability were also affected by the peak acceleration when the r.m.s. magnitude of the oscillations was constant, especially with 1-Hz oscillations. It is concluded that r.m.s. measures of acceleration are insufficient to predict the postural stability of walking passengers exposed to mediolateral oscillations and that peaks in the oscillations should also be taken into account.

Keywords: motion waveform; vibration; mediolateral surface oscillation, walking, perturbed locomotion

Word count: 3585

Introduction

When travelling by land, sea, or air, standing and walking passengers and crew experience various types of oscillatory perturbations that disturb their postural stability. To minimise instability caused by motion of a floor on which people are supported it is necessary to understand how the magnitude, the frequency, the direction, and the duration of the oscillations combine to cause loss of balance. In trains, buses, aircraft, buildings, etc., the motions causing instability are often transient, so it is desirable to understand whether instability can be predicted from a peak measure of the motion or from a short-term average measure of the motion. Although discomfort when exposed to whole-body vibration has been studied extensively for seated and standing subjects, the effects of transients have not been investigated systematically to predict the stability, discomfort, or difficulty of people walking in transport.

The effects of vibration on seated people in transport can be predicted using motion evaluation procedures defined in national and international standards.[1,2] Currently, the same procedures are used for seated and standing people, with the root-mean-square (r.m.s.) average acceleration one of the measures of the magnitude of a vibration. However, the discomfort of seated people exposed to whole-body vibration depends on the motion waveform, with greater sensitivity to random vibration than to sinusoidal vibration of the same frequency and the same r.m.s. magnitude[3]. The r.m.s. magnitude is more obviously unsuitable for predicting the discomfort of transient motions, since an r.m.s. average depends on when the averaging starts and finishes, and this is often not easily defined for transients. The 'crest factor' (ratio of the peak value of a motion to the r.m.s. value of the motion) has been advocated to identify whether r.m.s. average can be used. However, a crest factor limit of 3.0 in the original 1974 version of ISO 2631 prevented it being applied to the vibration in very many vehicles, and a crest factor limit of 9 in the current ISO 2631 is overly lenient and allows the apparent severity of transients to be lessened by extending the period over which the r.m.s. average is measured.[2] The use of r.m.s. values, with or without a crest factor limit, is therefore recognised in current standards as an inappropriate indicator of the human response to transient motions.

With transient oscillations of the same peak magnitude, the discomfort increases as the duration of the transient increases.[4] Average measures of the acceleration (e.g., the r.m.s. value) do not reflect this increase. A measure that increases with increasing duration (e.g., the vibration dose value) may reflect the increasing discomfort. [5-9]

The characteristics of transient perturbations of a floor supporting stationary standing subjects have been reported in terms of the peak velocity or peak displacement.[10,11] Some suggest that perturbations are better quantified by acceleration because this provides the destabilizing input to the postural control system and because stabilizing joint moments are triggered in response to acceleration[12,13], although the kinetics of postural recovery have been reported to depend on the velocity of platform motion.[14] With sinusoidal oscillation of a platform in the mediolateral axis of walking subjects, their postural stability was similar when motions had the same velocity, irrespective of the frequency of oscillation in the range 0.5 to 2 Hz.[15]

Apart from quantifying motions to predict walking instability, discomfort or difficulty, postural stability research has also sought metrics to identify walking balance. Understanding how people maintain stability when walking, particularly when exposed to perturbations, is key to preventing falls, but there is no universally accepted measure of stability. Studies of perturbed walking have identified perturbation-induced changes in specific measures of walking stability. Trunk sway and gait variability measures of mediolateral kinematics while walking have been found to be frequency-dependent and sensitive to visual stimuli [21]. Foot placement as quantified by step width and step length variability and variability in postural sway were greater during perturbation than during normal walking[21]. The ‘margin of stability’ as a metric to identify postural stability has been defined as the distance between a velocity adjusted or ‘extrapolated’ position of the COM (center of mass) and the edge of an individual’s BOS (base of support) at any given instant in time[22]. A similar metric is used to identify walking stability when exposed to visual or platform oscillations in anterior-posterior and mediolateral directions[23,24]. During normal walking, the velocity of the centre of pressure (COP) velocity follows a predictable pattern and has been suggested as a potentially useful measure in gait research[25]. Although an increase in COP velocity is associated with decreased ability in postural control during upright stance[,27], like other objective measures used in the assessment of postural control, it may be inadequate to explain the complete nature of postural control[28].

The experimental study reported here was designed to investigate the effect of the waveform of oscillations of the supporting surface in the mediolateral axis on the postural stability of walking subjects and to determine whether postural stability can be predicted from either the r.m.s. magnitude or the peak magnitude of oscillations. It was hypothesized that within acceleration waveforms having the same duration, same frequency, and same r.m.s.

magnitude, postural stability would decrease (as measured subjectively by self-reported probability of losing balance and objectively by measures of COP) as the peak magnitude of the oscillations increased. Similarly, it was hypothesized that within acceleration waveforms having the same duration, same frequency, and same peak magnitude, postural stability would decrease with increasing r.m.s. magnitude of the oscillations. Motions with different frequency but the same r.m.s. velocity and the same peak velocity (and therefore the same crest factor) were expected to produce similar postural instability[15]. Mediolateral COP velocity was expected to increase during perturbed walking due to stepping reactions to recover from postural instability caused by the perturbations.

Method

Subjects: Twenty healthy male subjects with median age 29.5 years (range 25 to 41), stature 175 cm (range 165 to 182 cm), weight 71.2 kg (range 47.2 to 92.2 kg) participated in the study. Subjects completed a questionnaire to exclude those with relevant disorders (balance related problems like regular falls, vertigo, dizziness, light headedness; ear related diseases like benign paroxysmal positional vertigo, labyrinthitis, trauma, Ménière's disease, Perilymph fistula, superior canal dehiscence syndrome, bilateral vestibulopathy; brain-related diseases including meningitis, encephalitis, epidural abscess, syphilis, cerebral or cerebellar ischemia or hypoperfusion, stroke, lateral medullary syndrome (Wallenberg's syndrome), Cogan syndrome, Arnold-Chiari malformation, hydrocephalus, multiple sclerosis, Parkinsons, CNS or Posterior Neoplasms; musculoskeletal problems in the back or lower limbs) or using drugs that may affect postural stability (benzodiazepines and/or triazolam)). Informed consent was obtained prior to participation in the experiment that was approved by the Human Experimentation Safety and Ethics Committee of the Institute of Sound and Vibration Research.

Apparatus: A treadmill (Kistler Gaitway®) incorporating eight force sensors was used to provide the walking task and measure vertical ground reaction forces. Subjects were secured by a loose safety harness that did not restrict walking but prevented knees and hips contacting the floor during a fall (Fig. 1). Although hand supports were available, subjects did not use them during the experiment – the effects of hand supports has been studied separately.[16]The treadmill was supported on a six-axis motion simulator in the Human Factors Research Unit at the Institute of Sound and Vibration Research.

FIGURE 1 ABOUT HERE

Acceleration was recorded by accelerometers attached to the simulator platform (FGP model FA101-A2-5G). Data acquisition via the treadmill software was triggered at the moment the simulator acceleration commenced. The acceleration and vertical force signals collected by the Gaitway® data acquisition system were sampled at 100 samples per second and stored in a personal computer.

Procedure: While walking on the treadmill, subjects were perturbed by mediolateral oscillations of 8-s duration that were cosine tapered for 1.5 s at the beginning and the end. The oscillatory motions were one-third octave bandwidth random motions centred on either 1 Hz or 2 Hz.

Each subject was exposed to 28 different test motions: seven different waveforms having the same r.m.s. magnitude and seven different waveforms having the same peak magnitude, at both frequencies (1 Hz and 2 Hz) (Table 1). The seven different waveforms were selected to have specific values for the crest factor (the ratio of peak value to the r.m.s. value of the acceleration waveform): 1.6, 1.8, 2.0, 2.24, 2.5, 2.8, and 3.15. Motions at 1 Hz and 2 Hz with the same crest factors had the same r.m.s. velocity and the same peak velocity, and were expected to produce similar postural instability. [15]Examples of the waveforms centred at 1 Hz are shown for these crest factors in Figure 2.

The peak lateral acceleration of a train can only increase with an increase in the duration over which the motion is measured, whereas the r.m.s. acceleration can increase or decrease but mostly remains more-or-less constant. The crest factor of the acceleration therefore increases as the duration of measurement increases. When measured over minutes or hours, the crest factors in a train can be greater than studied here. However, when measured over any 8-s period, the lateral acceleration of a moving train will normally have a crest factor within the range of crest factors studied here (1.6 to 3.15).

FIGURE 2 ABOUT HERE

Subjects were asked to walk on the treadmill at a comfortable walking speed (0.7 ms^{-1}) throughout the experiment. This was the average comfortable walking speed preferred by subjects in a preliminary study.[15] While subjects walked on the treadmill, oscillatory motions of the simulator supporting the treadmill were applied in their mediolateral axis at random unpredictable times.

Subjects were exposed to pairs of motion stimuli having the same frequency and separated by a 10-s pause. The first stimulus, a reference motion, was a sinusoidal motion of

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8-s duration with 1.5-s cosine tapering at either end. The crest factor of this motion measured over 8 s was 1.6. The 1-Hz reference motion had a magnitude of 0.7 ms^{-2} r.m.s. and the 2-Hz reference motion had a magnitude of 1.4 ms^{-2} r.m.s. Because these two reference motions had the same r.m.s. velocity and the same peak velocity they were expected to cause similar postural instability.[15] After experiencing the test motion, subjects were invited to rate the 'discomfort or difficulty' in their walking task caused by the second motion, assuming the 'discomfort or difficulty' caused by the first motion was 100. They were also asked to reply to the question: "what is the probability that you would lose balance if the same test motion were repeated?''

At each frequency, the experiment involved two parts (Table 1). Within Part A, the test motions had the same r.m.s. magnitudes but differing peak values. Within Part B, the test motions had the same peak magnitudes but differing r.m.s. magnitudes. At both frequencies, all the motions were presented in a different random order to each subject. The order of presenting the two parts was balanced across subjects. For convenience, the stimuli at each frequency having the same r.m.s. magnitudes but differing peak values are labelled as 'Part A' and the stimuli having the same peak magnitudes but differing r.m.s. magnitudes are labelled as 'Part B'.

Subjects were given experience of walking on the treadmill without perturbation before the experiment commenced. The gait during normal walking without perturbation was measured over a period of 8 s after the subjects had been given this experience.

TABLE 1 ABOUT HERE

The COP measures of the subjects were calculated from the reaction forces under the feet during each motion.[15,16]

Analysis: The force time-histories (from the 8 vertical force sensors in the treadmill) were processed to determine COP time histories during each motion. The COP velocity in the medio-lateral direction was obtained by differentiating the mediolateral position of the COP after filtering the position of the COP using an 8-Hz low-pass Bessel filter.

An example of the gait of a subject exposed to a one-third octave band of 1.0-Hz mediolateral oscillation (0.7 ms^{-2} r.m.s. and 2.0 ms^{-2} peak) is shown in Fig. 3. The COP position shows the mediolateral location of the resultant of the ground reaction forces and is indicative of mediolateral foot placement (Fig. 3a). The COP velocity indicates the rate of change of COP position (Fig. 3b). The acceleration waveform of the mediolateral oscillation is shown in Fig. 3c.

FIGURE 3 ABOUT HERE

The dependence of postural stability on both the r.m.s. magnitude and the peak magnitude of the oscillations was tested for both the subjective and the objective measures of postural stability in terms of reported probability of losing balance, discomfort or difficulty rating, and mediolateral COP velocity. Non-parametric statistical tests were performed using SPSS (version 17), as the normality of the distribution could not be guaranteed with 20 subjects. The Friedman analysis of variance tested for differences between multiple conditions and the Wilcoxon matched-pairs signed ranks test investigated differences between pairs of conditions. Associations between variables were investigated using Spearman's rank correlation.

The relation between 'discomfort or difficulty' ratings and the peak and r.m.s. magnitudes of oscillation were quantified using Stevens' power law, which shows the rate at which sensation increases with an increase in the magnitude of the physical stimulus causing the sensation: [17]

$$\Psi = k_{rms} * (\Phi_{rms})^{n_{rms}} \quad \text{Equation 1}$$

$$\Psi = k_{peak} * (\Phi_{peak})^{n_{peak}} \quad \text{Equation 2}$$

where ψ is the sensation magnitude (i.e., 'discomfort or difficulty' rating), Φ_{rms} is the r.m.s. magnitude of the lateral acceleration of the treadmill, Φ_{peak} is the peak magnitude of the acceleration, k_{rms} and k_{peak} are constants, and n_{peak} and n_{rms} are the rates of growth of sensation with respect to the peak acceleration and the r.m.s. acceleration.

Equations1 and 2 can be rewritten in logarithmic form:

$$\log\Psi = \log k_{rms} + n_{rms} \log\Phi_{rms} \quad \text{Equation 3}$$

$$\log\Psi = \log k_{peak} + n_{peak} \log\Phi_{peak} \quad \text{Equation 4}$$

By performing linear regression between the experimental values of log ψ and log φ, estimates of the constants k_{rms} and k_{peak} and the exponents n_{rms} and n_{peak} were obtained for each subject.

Results

With acceleration waveforms having the same frequency and same r.m.s. magnitude, postural stability decreased as the peak magnitude of the oscillations increased. With 1-Hz oscillations, the reported probability of losing balance and the 'discomfort or difficulty' ratings increased with increasing peak acceleration (p<.01, Spearman; Fig. 4a,4b) and the peak mediolateral COP velocity also increased with increasing peak acceleration (p<.05, Spearman; Fig. 4c). With 2-Hz oscillations, the 'discomfort or difficulty' ratings increased with increasing peak acceleration (p<.01, Spearman, Fig. 4d), but there was no significant change in the self-reported probability of losing balance (p=.157, Friedman, Fig. 4e) or the peak COP velocity (p=.258, Friedman, Fig. 4f).

FIGURE 4 ABOUT HERE

With acceleration waveforms having the same frequency and same peak magnitude, postural stability decreased as the r.m.s. magnitude of the oscillations increased. With both 1-Hz and 2-Hz oscillations, the reported probability of losing balance, the 'discomfort or difficulty' ratings, and the r.m.s. magnitude of the mediolateral COP velocity increased with increasing r.m.s. acceleration (p<.01, Spearman; Fig.5a, 5b, 5c, 5d, 5e and 5f).

During oscillation at 1 Hz or 2 Hz with any r.m.s. magnitude and any peak magnitude of oscillation, the r.m.s. and peak COP velocities in the mediolateral direction were greater than during normal walking without oscillation (p<.01, Wilcoxon, Fig. 4c, 4f, 5c and 5f).

FIGURE 5 ABOUT HERE

The rates of growth of 'discomfort or difficulty' differed between 1-Hz and 2-Hz oscillations according to whether the r.m.s. magnitude or the peak magnitude of the oscillations was held constant. With 1-Hz oscillations, the rate of growth of 'discomfort or difficulty' with increasing r.m.s. acceleration when the peak acceleration was held constant (n_{rms}: median 0.597, inter-quartile range 0.368 to 0.804) was not significantly different from the rate of growth with increasing peak acceleration with the r.m.s. acceleration held constant (i.e.,n_{peak}: median 0.508, inter-quartile range 0.212 to 0.993) (p=.852, Wilcoxon). However, with 2-Hz oscillations, the rate of growth of 'discomfort or difficulty' with increasing r.m.s. acceleration with the peak acceleration held constant (i.e. n_{rms}: median 0.844, inter-quartile range 0.678 to 1.215) was significantly greater than the rate of growth with increasing peak acceleration with the r.m.s. acceleration held constant (i.e. n_{peak}: median 0.270, inter-quartile range 0.131 to 0.497) (p<.001, Wilcoxon). The reduced rate of growth with increasing peak acceleration in the 2-Hz motions is consistent with the absence of significant correlations between peak acceleration and both the self-reported probability of losing balance and the peak COP velocity.

With the r.m.s. velocity of the pertubing motions the same at 1 Hz and 2 Hz, the reported probability of losing balance was similar (p>.05, Wilcoxon, Fig. 6a) and the mediolateral r.m.s. COP velocity was similar (p>.05, Wilcoxon, Fig. 6b) at each crest factor.

FIGURE 6 ABOUT HERE

Subject stature and subject weight had no statistically significant effects on the subjective or objective measures of postural stability used in the study.

Discussion

With sitting subjects exposed to complex waveforms with a fundamental frequency of 8 Hz and crest factors in the range 2.1 to 8.5, Griffin and Whitham found that motions with the same r.m.s. magnitude caused more discomfort when the peak magnitudes were greater (i.e. when the crest factor was greater).[4] Increasing discomfort with increasing crest factor with unchanged r.m.s. values has also been reported by Howarth and Griffin.[5] This is consistent with the findings of the current study with walking subjects: with the r.m.s. acceleration constant, the 'discomfort or difficulty' ratings, the reported probability of losing balance, and the peak mediolateral COP velocity all increased with increasing peak acceleration of 1-Hz oscillations (Figs 4a, 4b and 4c). With 2-Hz oscillation, although there was a slight effect of peak acceleration on the 'discomfort or difficulty' ratings, the peak reported probability of losing balance and the mediolateral COP velocity were not affected by changes in the peak magnitude (Figs. 4d, 4e and 4f).

The subjective ratings and the objective measurements of COP velocity suggest postural instability when walking is sensitive to the peak magnitude of 1-Hz mediolateral oscillations but not the peak magnitude of 2-Hz mediolateral oscillations. Arranging the placement and timing of stepping reactions, subjects might have better adapted to 1 Hz oscillations being closer to their walking frequency under experimental conditions and overcome the perturbations by wider and more synchronized stepping. The strategy to synchronize with the motion might make it important when and how much the peak oscillations occur during the gait cycle. However, when exposed to 2 Hz oscillations, higher than walking frequency in experimental conditions, subjects might not have found the time to overcome the effects of perturbation on their stability and peaks might lose their importance. With high magnitude oscillations at 2 Hz it is also likely that discomfort rather than postural stability is a concern for walking subjects (Fig 4d vs Fig 4e). With higher frequencies of oscillation, the displacements are smaller even though the velocity is the same. So for a given velocity, there is a frequency above which the peak displacement will be insufficient to cause instability.

With 1-Hz oscillations, there were similar positive associations between 'discomfort or difficulty' ratings and both the peak acceleration and the r.m.s. acceleration, (n_{rms}=0.597 and n_{peak}=0.508), suggesting that both the peak value and the r.m.s. value contributed to the 'discomfort or difficulty' caused by 1-Hz oscillations. With 2-Hz oscillation, the association between 'discomfort or difficulty' was much stronger for the r.m.s. magnitude of the acceleration than the peak acceleration (compare Figs 4d and 5d). This suggests that with this higher frequency of oscillation the r.m.s. acceleration provides a reasonable indication of the 'discomfort or difficulty' of a walking task, even when there are transients with dominant motions around this frequency.

The 'discomfort or difficulty' ratings, the reported probability of losing balance, and the mediolateral r.m.s. COP velocity increased with increasing r.m.s. acceleration at both 1 Hz and 2 Hz (Figs 5a, 5b,5d and5e). The r.m.s. acceleration therefore seems to be an appropriate measure to quantify the magnitude of oscillations at both frequencies if the peak magnitudes is the same.

The mediolateral velocity of the COP was influenced by the perturbations to walking used in the current study and was strongly associated with the subjectively reported probability of losing balance. With greater magnitudes of oscillation, the risk of fall increases and subjects adjust the placement and timing of successive steps to try to overcome the effects of external perturbations.[18] The increased mediolateral COP r.m.s. velocity with increasing magnitude of oscillations suggests wider stepping or faster stepping, or both. The greater mediolateral COP r.m.s. velocity during mediolateral oscillation than during normal walking without oscillation (Figs. 4c, 4f, 5c and 5f) also indicates that subjects adapted their stepping strategies in response to the perturbations. McAndrew *et al*. found that walking people took wider and faster steps during continuous random oscillation than during normal walking without oscillation.[19]

The mediolateral COP r.m.s. velocity and the reported probability of losing balance were similar for oscillations having similar r.m.s velocity (Fig. 6). This is consistent with Sari and Griffin who found that, over the frequency range 0.5 to 2 Hz, postural instability was better predicted by the r.m.s. velocity of sinusoidal motions than either the displacement or the acceleration of the motion.[15]However, the current findings show that the problems caused by mediolateral oscillation when walking cannot be predicted solely from the r.m.s. velocity and that the peak velocity should also be considered.

The walking speed of 0.7 ms^{-1} used in this study is slower than normal walking but it was the preferred speed for subjects walking in the laboratory environment on a treadmill attached to a 6-axis motion simulator that was expected to move. The findings of the study will mostly be applicable to environments within transport (e.g., trains, buses, and ships) where walking speed is slower than when walking in stationary environments.

In this study, each subject provided a single response to each 8-s perturbation. Previous experience with such stimuli suggested a single response was sufficient and avoided both familiarisation with individual stimuli and fatigue due to long periods of perturbed walking [15,16]. Each subject was exposed to 56 motions during the experiment in addition to prior training on the experimental method and experiencing example motions.

It can be concluded that walking stability while exposed to various waveforms of mediolateral oscillations of the supporting surface is influenced by the peak magnitude of the oscillation, and not only the r.m.s. magnitude. The r.m.s. magnitude is therefore not the optimum method of predicting the postural stability of walking subjects exposed to mediolateral oscillations, especially with low frequency motions around 1 Hz.

In experimental studies, it would be advantageous to report the waveform of platform perturbations. If motion velocity is used to quantify the perturbation, both the peak magnitude and the r.m.s. magnitude of the velocity should be reported. If the perturbation is specified in terms of acceleration, the frequency of the motion should also be provided, together with the peak and r.m.s. magnitudes of the oscillations.

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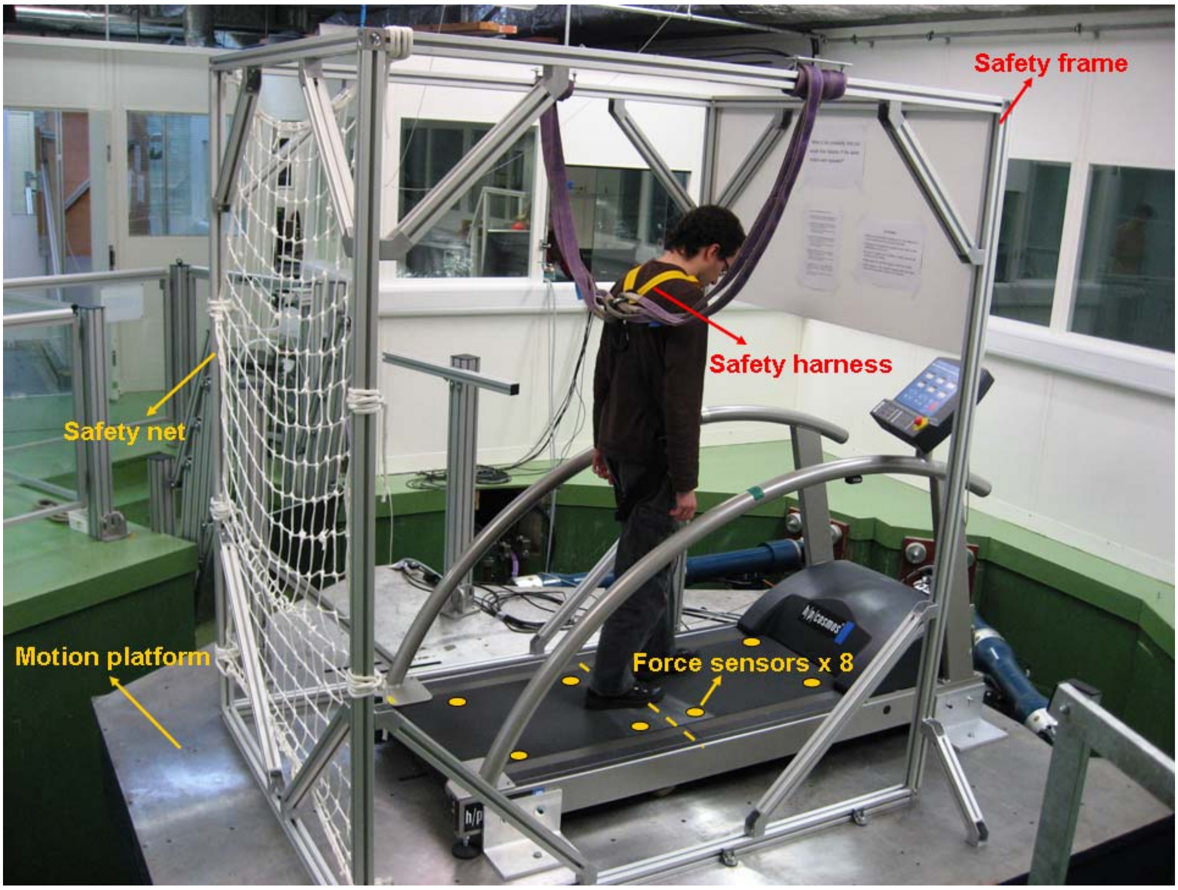

Figure 1 Experimental apparatus.

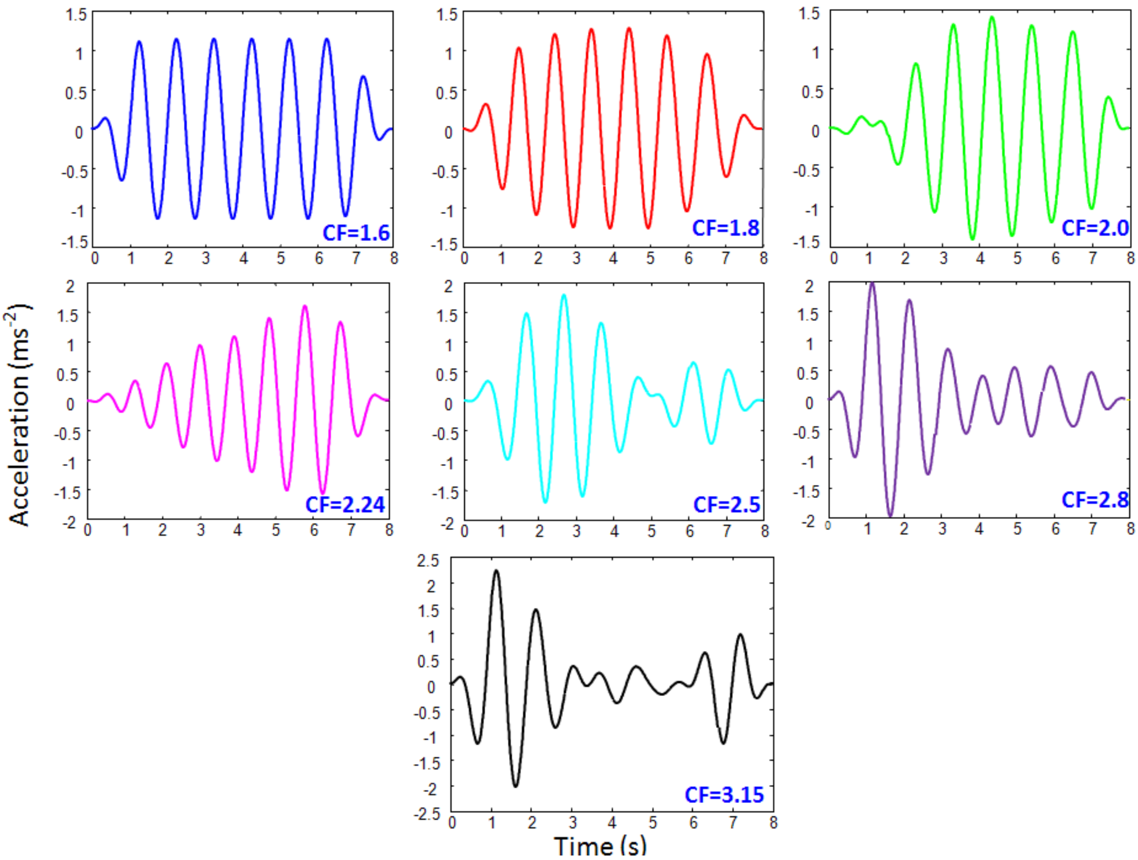

Figure 2 One-third octave bandwidth random waveforms centred at 1 Hz. All motions have a magnitude of 0.7 ms^{-2} r.m.s. but different crest factors (CF).

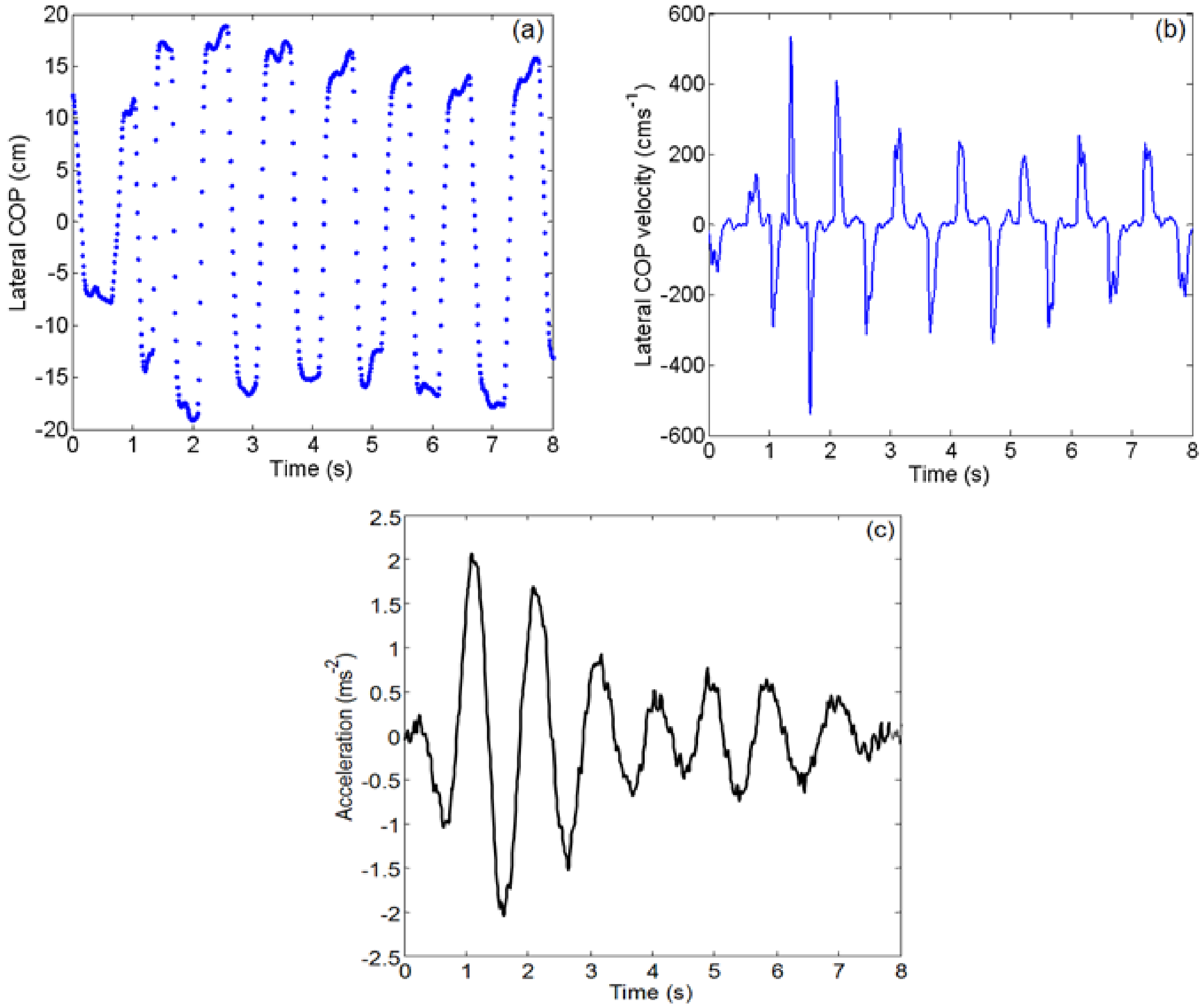

Figure 3 Example time histories of the centre of pressure (COP) for a subject walking while exposed to 0.7 ms^{-2} r.m.s. mediolateral oscillation at 1 Hz: (a) COP position in the mediolateral direction; (b) COP velocity in the mediolateral direction; (c) measured mediolateral acceleration.

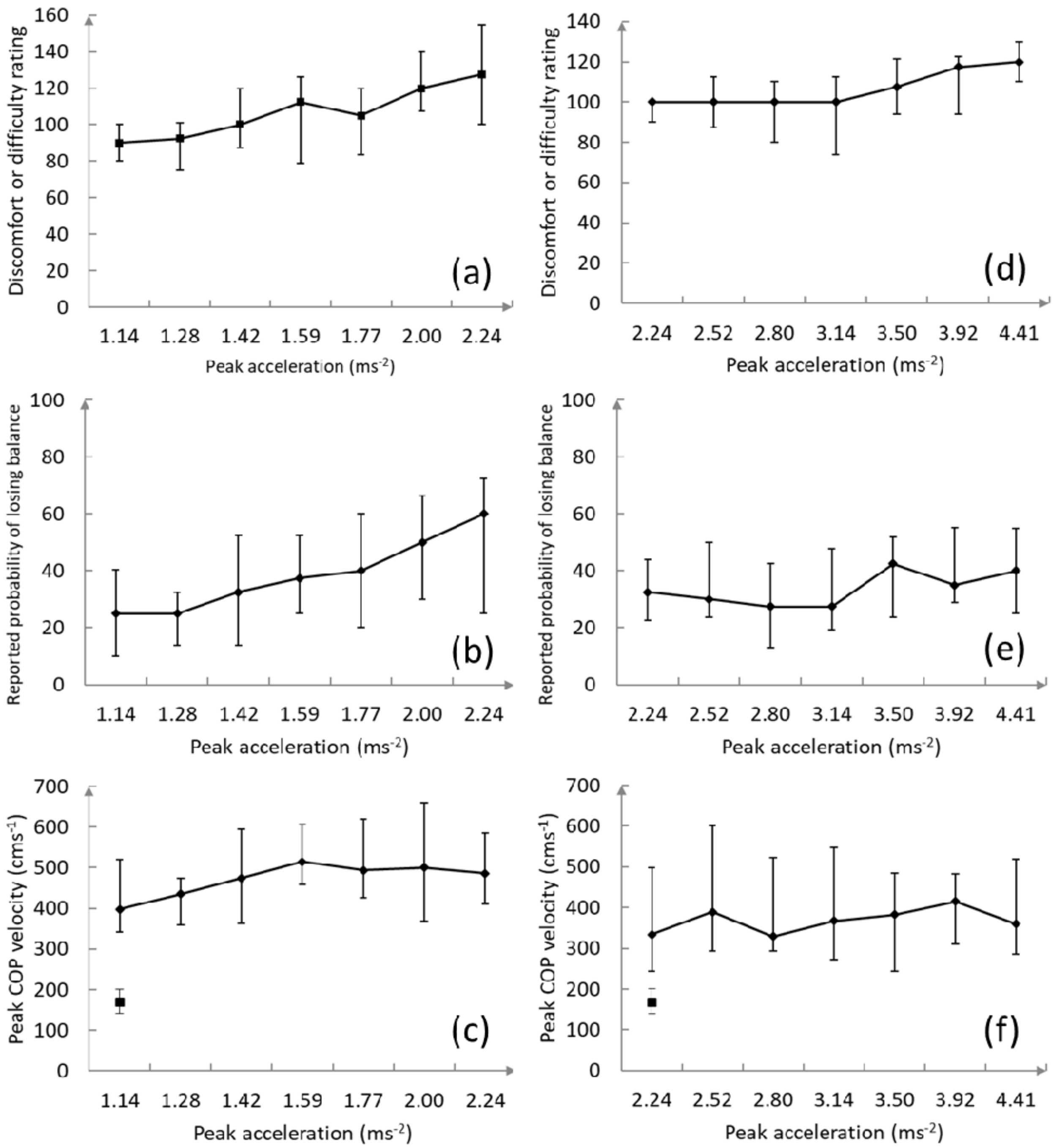

Figure 4 For oscillations having the same r.m.s. acceleration, with increasing peak magnitude of acceleration at 1 Hz (left figures) and 2 Hz (right figures): (a, d) ‘discomfort or difficulty’ ratings increased (p<0.01, Spearman); (b, e) reported probability of losing balance increased at 1 Hz (p<0.01, Spearman) but was not significantly affected at 2 Hz (p=0.157, Friedman); (c, f) peak mediolateral r.m.s. COP velocity increased at 1 Hz (p<0.05, Spearman) but was not significantly affected at 2 Hz (p=0.258, Friedman). The medio-lateral peak COP velocity was less during normal walking without oscillation (■) than during any oscillation at either 1 Hz or 2 Hz. Medians and inter-quartile ranges from 20 subjects.

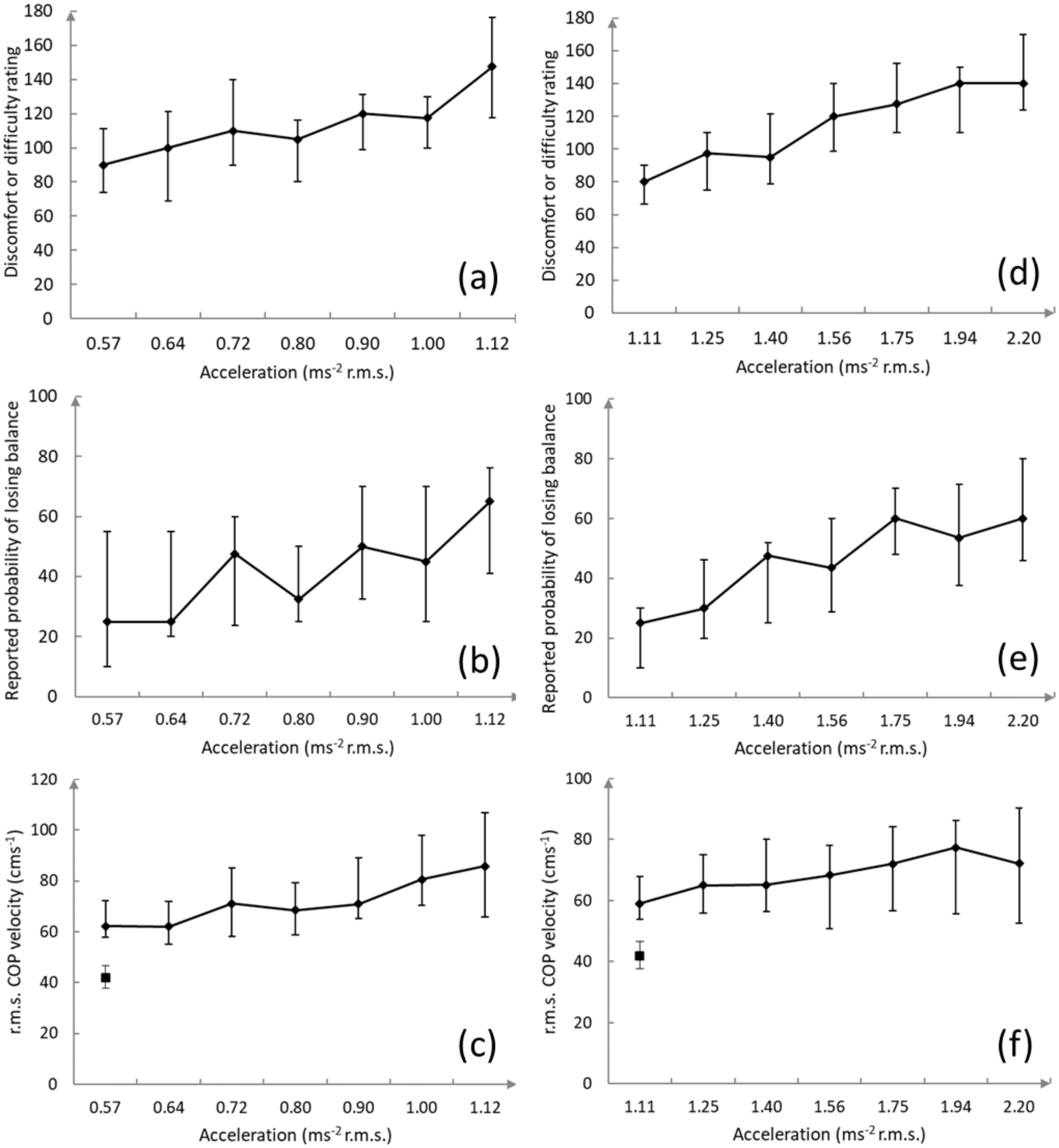

Figure 5 For oscillations having the same peak acceleration, with increasing r.m.s. magnitude of acceleration at 1 Hz (left figures) and 2 Hz (right figures): (a, d) 'discomfort or difficulty' ratings increased (p<0.01, Spearman); (b, e) reported probability of losing balance increased (p<0.01, Spearman); (c, f) peak mediolateral r.m.s. COP velocity increased (p<0.01, Spearman). The mediolateral r.m.s. COP velocity was less during normal walking without oscillation (■) than during any oscillation at either 1 Hz or 2 Hz. Medians and inter-quartile ranges from 20 subjects.

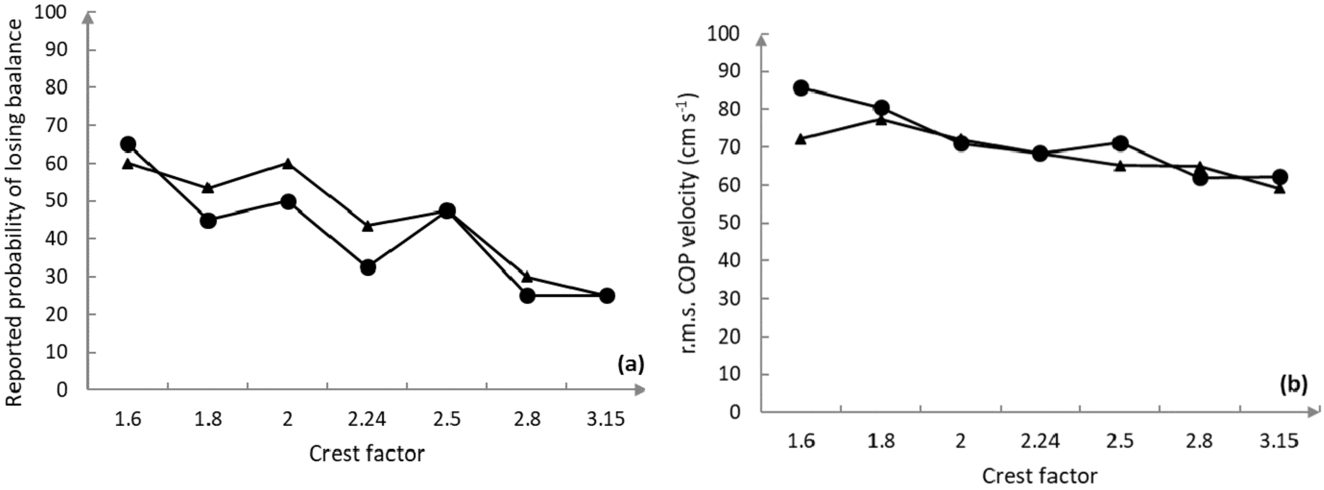

Figure 6 For all crest factors, with the r.m.s. velocity the same at 1 Hz or at 2 Hz, the reported probability of losing balance (a) was not significantly different (*p*>0.05, Wilcoxon) and the mediolateral r.m.s. COP velocity (b) was not significantly different.: —●—1.0 Hz, —▲—2.0 Hz. Medians from 20 subjects.

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Table 1: Characteristics of the mediolateral acceleration used in the study (CF = crest factor).

f=1 Hz PART A						**f=1 Hz PART B**					
waveform	r.m.s. acc (ms^{-2})	peak acc (ms^{-2})	r.m.s. vel (ms^{-1})	peak vel (ms^{-1})	CF	waveform	r.m.s. acc (ms^{-2})	peak acc (ms^{-2})	r.m.s. vel (ms^{-1})	peak vel (ms^{-1})	CF
sinusoidal	**0.71**	1.14	**0.12**	0.21	1.60	random	0.57	**1.80**	0.09	**0.30**	3.15
random	**0.71**	1.28	**0.12**	0.22	1.80	random	0.64	**1.79**	0.11	**0.30**	2.80
random	**0.71**	1.42	**0.12**	0.24	2.00	random	0.72	**1.80**	0.12	**0.30**	2.50
random	**0.71**	1.59	**0.12**	0.27	2.24	random	0.80	**1.79**	0.13	**0.30**	2.24
random	**0.71**	1.77	**0.12**	0.29	2.50	random	0.90	**1.80**	0.15	**0.30**	2.00
random	**0.71**	1.99	**0.12**	0.34	2.80	random	1.00	**1.80**	0.16	**0.30**	1.80
random	**0.71**	2.24	**0.12**	0.38	3.15	sinusoidal	1.12	**1.79**	0.18	**0.30**	1.60
f=2 Hz PART A						**f=2 Hz PART B**					
waveform	r.m.s. acc (ms^{-2})	peak acc (ms^{-2})	r.m.s. vel (ms^{-1})	peak vel (ms^{-1})	CF	waveform	r.m.s. acc (ms^{-2})	peak acc (ms^{-2})	r.m.s. vel (ms^{-1})	peak vel (ms^{-1})	CF
sinusoidal	**1.40**	2.24	**0.12**	0.21	1.60	random	1.11	**3.50**	0.09	**0.30**	3.15
random	**1.40**	2.52	**0.12**	0.22	1.80	random	1.25	**3.50**	0.11	**0.30**	2.80
random	**1.40**	2.80	**0.12**	0.24	2.00	random	1.40	**3.50**	0.12	**0.30**	2.50
random	**1.40**	3.14	**0.12**	0.27	2.24	random	1.56	**3.50**	0.13	**0.30**	2.24
random	**1.40**	3.50	**0.12**	0.29	2.50	random	1.75	**3.50**	0.15	**0.30**	2.00
random	**1.40**	3.92	**0.12**	0.34	2.80	random	1.94	**3.50**	0.16	**0.30**	1.80
random	**1.40**	4.41	**0.12**	0.38	3.15	sinusoidal	2.19	**3.50**	0.18	**0.30**	1.60