UNIVERSITY OF SOUTHAMPTON

FACULTY OF ENGINEERING AND THE ENVIRONMENT

INSTITUTE OF SOUND AND VIBRATION RESEARCH

POSTURAL STABILITY WHEN WALKING AND EXPOSED TO LATERAL OSCILLATIONS

by

Hatice Mujde SARI

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ABSTRACT

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The role of accelerations encountered in daily life in causing postural instability or falls is not well understood, especially when walking. Lateral oscillations disturb walking stability during train journeys but it is desirable that passengers feel comfortable and they do not fall due to loss of balance. This research was designed to improve understanding of the mechanisms of walking stability and to construct a model for predicting the probability of losing balance in walking railway passengers. Postural stability was assessed using both a subjective measure (the reported probability of losing balance) and objective measures of centre of pressure (COP).

The first of four experiments investigated how postural stability when walking depends on the frequency (0.5 to 2.0 Hz) and the magnitude (0.1 to 2.0 ms^{-2} r.m.s.) of transient lateral oscillation. The probability of losing balance reported by 20 subjects was used to obtain stability thresholds for the lateral accelerations experienced in trains. It was shown that postural stability cannot be predicted solely from either the peak or the r.m.s. value of lateral acceleration but can be predicted from the peak or the r.m.s. velocity of sinusoidal lateral oscillation..

The second experiment with 20 subjects investigated the extent to which a hand support (rigid vertical bar) modifies postural stability when walking during lateral oscillation. The hand support improved postural stability at all frequencies (0.5 to 2 Hz) and at all velocities (0.05 to 0.16 ms^{-1} r.m.s.). The improvement in postural stability from holding the support and the forces applied to the hand support were independent of support height and were greater during perturbed walking (30-50% when the support was held throughout the oscillation, 20-30% when the support was held if required) than during normal walking (15%). When it was required, subjects preferred to hold the hand support at a height of 126 cm above the surface supporting the feet.

The third experiment investigated how the postural stability of walking people is influenced by the waveform of lateral oscillations. Twenty subjects were exposed to a range of 1 Hz and 2 Hz lateral oscillations having the same r.m.s. magnitude but different waveforms. The reported probability of losing balance and the lateral COP velocity was found to be sensitive to the peak magnitude of the oscillations especially at 1 Hz. It was concluded that the r.m.s. value is not an optimum method for predicting the postural stability of walking subjects exposed to low frequency lateral oscillations and that peaks in the motion should also be considered.

The influence of subject characteristics (age, gender, weight, stature, shoe width, fitness) on postural stability was investigated in a fourth experiment with 100 subjects. Age had the greatest influence on postural stability, with an increase in COP measures with increasing age. There was no significant effect of any subject characteristic on self-reported probability of losing balance. The stability thresholds of young males (determined in the first experiment) can therefore be applied to a wider age range (18 to 70 years) of fit and healthy people, including females.

The subjective experimental findings have been used to develop an empirical model for predicting the probability of losing balance in walking people exposed to lateral oscillation. Analysis of the objective measure of COP revealed that the ‘stepping strategy’ is the principal means of maintaining postural stability when walking is perturbed by lateral oscillation.

The developed model can be used to predict the perceived risk of fall when walking and exposed to lateral oscillations from the peak and r.m.s. velocity of oscillations. The model predicts the perceived probability of losing balance during exposure to various waveforms of oscillations and is applicable to males and females with variety of ages (18 to 70 years).

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DECLARATION OF AUTHORSHIP

I, HATICE MUJDE SARI

declare that the thesis entitled

POSTURAL STABILITY WHEN WALKING AND EXPOSED TO LATERAL OSCILLATIONS

and the work presented in the thesis are both my own, and have been generated by me as the result of my own original research. I confirm that:

- this work was done wholly or mainly while in candidature for a research degree at this University;
- where any part of this thesis has previously been submitted for a degree or any other qualification at this University or any other institution, this has been clearly stated;
- where I have consulted the published work of others, this is always clearly attributed;
- where I have quoted from the work of others, the source is always given. With the exception of such quotations, this thesis is entirely my own work;
- I have acknowledged all main sources of help;
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Signed: …………………………………………………………………………..

Date:……………………………………………………………………………….

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Definitions and abbreviations

Anterior-posterior direction The axis connecting the front and back in humans. Analogous to back-and-forth direction.

BOS Base of support. The area on which the body is supported. It is defined by the length of the foot in the sagittal plane and separation of the feet in the frontal plane.

COM Centre of mass. The point where the whole body mass is concentrated.

COG Centre of gravity. The vertical projection of COM on the ground.

COP Centre of pressure. The point location of the vertical ground reaction forces under the feet.

CF Crest factor

CNS Central nervous System. The part of the nervous system that coordinates the activity of all parts of the body. Several parts of the central nervous system (CNS), which consists of the spinal cord and the brain, take part in controlling posture.

Dorsiflexion Movement reducing the angle between the foot and the leg.

Double support phase The period of time in gait cycle in which the body is supported on both feet. The time from the heel strike of the one foot up to the toe-off of the other foot.

Frontal plane Coronal plane. The plane formed by the vertical gravity axis and the medio-lateral axis of the human body.

f Frequency

Gait Movement pattern during locomotion.

Gait cycle Also referred as stride time. It is the time duration from the initial heel contact of one foot to the next initial heel contact of the same foot. Suppose the gait cycle starts with the heel strike of the right foot, it will follow by toe-off of the left foot, single support phase on the right foot, heel strike of the left foot, toe-off of the right foot, single support phase on the left foot and will finish by the heel strike of the right foot.

Heel-contact The instant at which the foot hits the ground with heel.

k 'Constant' in Stevens power law

Medio-lateral direction Lateral axis connecting the right and left in humans. Analogous to lateral direction.

n 'Exponent' in Stevens power law

Perturbation Externally applied inputs to human body in order to stimulate human posture and evoke automatic postural responses.

PLB Probability of losing balance

PLB$_{s1}$ Probability of losing balance when using hand support continuously throughout the oscillation.

PLB$_{s2}$ Probability of losing balance when using hand support if required during exposure to oscillation.

r.m.s. root-mean-square.

Sagittal plane The plane parallel to the plane of progression. It is formed by the vertical gravity axis and the anterior-posterior axis of the human body.

Single support phase Single stance phase. The period of time during gait cycle in which the body is supported by only one foot.

Step time The time duration from the initial heel contact of one foot to the next initial heel contact of the opposing foot.

Stretch reflex Short latency spinal reflex which results in muscle contraction in response to stretching of the muscle.

Toe-off The instant at which the toe is about to leave the ground.

Somatosensory system The sensory system that responds to contact forces and gives information on body orientation with respect to the ground or support surface. The somatosensory system is defined as a combination of cutaneous, kinaesthetic (proprioceptive) and visceral sensory systems.

Proprioceptive system The sensory system detecting the relative movements of body parts with respect to each other.

Vestibular system The sensory system that is responsive to the angular and the linear accelerations of the head with respect to an earth-fixed inertial frame of reference.

v_{peak} peak velocity of oscillation

v_{rms} r.m.s. velocity of oscillation

ϕ Objective magnitude (in general, r.m.s. acceleration)

ψ Subjective magnitude (discomfort)

Chapter 1

Introduction

Fit and healthy people maintain their postural stability without noticeable effort while performing daily tasks. Keeping balance may be more difficult when people encounter disturbances (e.g. slips, trips, etc.) during daily activities (e.g. standing, walking, reaching an object, etc.). External disturbances are also encountered during transport where they may result from improper road and/or driving conditions. Disturbances to postural stability during transport (e.g. in ships, trains) are especially challenging for standing and walking postures.

Inability to maintain balance may be caused by many factors like balance related illnesses, incapability of the individual for an instant of time (e.g. attention problems) and severity of the external perturbation (e.g. magnitude, frequency and direction). Some of these factors are controllable while others totally depend on the ability of the person. Although attention or muscular strength are specific to each individual, balance problems related with diseases (e.g. musculoskeletal diseases, vestibular loss) may be treated by medical doctors or physiotherapists through diagnosis, treatment or rehabilitation programs. Road and rail conditions can be improved by engineers to control the severity of motions (e.g. to eliminate high magnitudes). Safer designs of walking aids and supports in transport can be provided by research in engineering and ergonomics.

Multidisciplinary research in postural stability aims to control or eliminate at least some sources of disturbances resulting in loss of balance and falls. There have been many attempts to quantify postural stability for the purpose of identifying fall risk and taking protective measures before falls occur. Other than increasing the quality of living standards and safety of people especially the elderly, and preventing falls, there has been also scientific curiosity behind the attempt to understand the physiological and biomechanical aspects of balance. This understanding may be useful to develop predictive models of postural stability which may be a valuable tool to estimate the effects of disturbances on postural stability and also identify balance related deficits via standardized methods (e.g. moving platform balance tests)

Experiences with accelerations encountered in daily life and their role in causing postural instability or falls have not been well understood especially for walking people. The belief that acceleration in the lateral direction is a dominant source of postural instability for walking passengers in trains is the driving force for the current research.

The overall aim of the research is to get a better understanding of the postural stability of walking subjects when perturbed by lateral oscillations and to represent this understanding in a postural stability model that can predict the effects of lateral oscillations on the postural stability of walking people. The application of the research and the model (e.g. the prediction of stability thresholds) aims to improve the postural stability of walking passengers in trains in terms of both the optimization of the motions on trains and the design of supports for passengers.

Four separate experiments have been designed to fulfill the objectives of the research. The four experiments are reported in Chapters 4 to 7. The first experiment investigated the effects of magnitude and frequency on the postural stability of walking people perturbed by lateral oscillatory motions. The second experiment determined the extent to which supports help to improve postural instability, and the optimum height for hand supports. The third experiment was conducted to determine whether the magnitude dependency found in the first experiment could apply to other types of waveforms. The fourth experiment aims to investigate the subject's physical characteristics (e.g. age, gender, weight, stature, fitness) that could affect the subjective and objective measurements of postural stability in response to lateral oscillations.

Following this introduction, Chapter 2 includes a review of the literature related to the postural stability of standing and walking people. In Chapter 3, the methods used in this thesis (i.e. the equipment, testing conditions, methods, and analysis tools) are presented. In Chapters 4 to 7, four experiments are reported. Chapter 8 contains a discussion of the methods and results of all experiments and a proposed preliminary predictive model of the postural stability during perturbed walking. Chapter 9 concludes this thesis.

Chapter 2

Literature review

2.1. Introduction

This chapter reviews previous research on postural stability. Many studies have been conducted to investigate the mechanisms involved in postural stability and develop standardized methods for identifying fall risk and diagnosing postural instability problems. Although some of the postural stability research has been conducted on walking people, most studies were conducted with standing people. Results of research on standing stability are also reported in this review as they may also help to understand walking stability.

In Section 2.2, an introduction is made to the concept of postural stability, discussing the biomechanical and sensory components of balance together with postural strategies. In Section 2.3, the methods of quantifying postural stability and measures of postural stability are mentioned. Section 2.4 presents an overview of the effects of factors such as the characteristics of vibration (i.e. frequency, magnitude, waveform), supports and subject characteristics (i.e. age, gender, weight) on postural stability. In Section 2.5, models of postural stability are discussed.

2.2. Postural stability

The human body is an inherently unstable system – "two thirds of our body mass is positioned two thirds of our body height above the ground (Winter, 1995)" – due to the continuous destabilizing effect of gravity. Postural stability can be defined as the ability to keep the body in an upright position by compensating the destabilizing effects of gravity and other disturbances (e.g. slips, mis-steps, obstacles, etc.). The degree to which postural stability is maintained is dependent on whether employed postural control strategies keep the body in equilibrium so as not to fall.

The human postural control system involves many subsystems, including biomechanical, sensory, neural, and muscular systems. The complexities of these subsystems have been reduced by researchers making simplifying assumptions: 2-dimensional single link inverted pendulum models (Cenciarini and Peterka, 2006; Yutaka *et al.* 2001, Mergner *et al*., 2006) have been used to represent complex 3-dimensional multi-segment dynamics of human body; sensory systems have been modelled by constant gain values (Mergner *et al.*, 2006; Peterka, 2003); a simple proportional-integral-derivative (PID) control structure has been used in human postural control models to represent the complex neural control structure of the central nervous system (Peterka, 2003, van der Kooij *et al*., 2001). Simplifying assumptions may help to develop simple and practical quantitative models of human balance based on the experimental data. However, it is important to be aware of the functional importance of the underlying simplified mechanisms for a better understanding of human balance and sensible interpretation of the experimental data.

The biomechanics of human balance, the sensory systems involved in balance, and postural strategies are briefly introduced in Sections 2.2.1, 2.2.2, and 2.2.3.

2.2.1. Biomechanics of human balance

The static and dynamic characteristics of multi-body segments provide humans with the ability to adopt the postural configurations (e.g. sitting, standing, walking, reaching) that are necessary to perform daily activities. Successful execution of these daily tasks requires maintenance of postural stability.

Figure 2.1 shows a simple representation of a standing human body posture with five body segments (head, trunk, upper leg, lower leg and foot) in the sagittal plane and in the frontal plane.

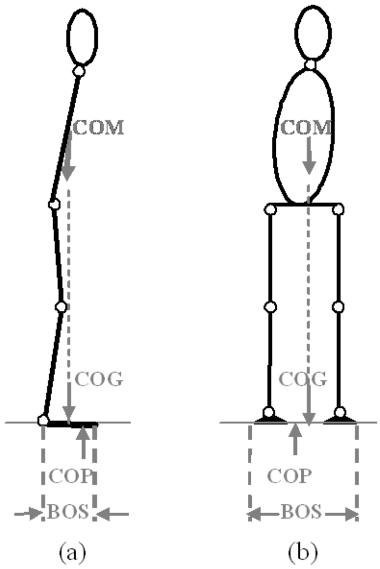

Figure 2.1: Five-segment representation of a standing subject in the (a) sagittal plane (b) frontal plane.

The centre of pressure (COP, Figure 2.1) is the point location of the vertical ground reaction forces under the feet (Winter, 1995). The centre of gravity (COG, Figure 2.1) is the vertical projection of the centre of mass (COM, Figure 2.1) on the ground (Winter, 1995). The base of support (BOS, Figure 2.1) is the area defined by the length of the foot in the sagittal plane and the distance between the feet in the frontal plane.

For a standing person, if the centre of gravity (COG) remains within the perimeter of the base support (BOS), postural stability is maintained. If, on the other hand, the COG goes beyond the limits of this perimeter, balance cannot be maintained (Jacobson, 1993). The centre of pressure (COP) is adjusted with respect to the movements of centre of gravity (COG) to maintain postural stability while standing.

Figure 2.2 shows a walking subject during the single support phase of the gait cycle. While standing subjects are continuously supported on two legs, walking subjects are supported on only one leg during 80% of the gait cycle (Woollacott and Tang, 1997).

Maintaining postural stability is quite different during standing and walking due to the differences in the dynamics of multi-body segments. Although the base of support (BOS) is stationary for a standing subject, it is continuously moving for a walking subject. While the objective of postural control during standing is to maintain the vertical projection of the COM within the limits of the base of support (BOS) which is stationary, the COM must move outside the non-stationary base of support limits when walking (Winter, 1995).

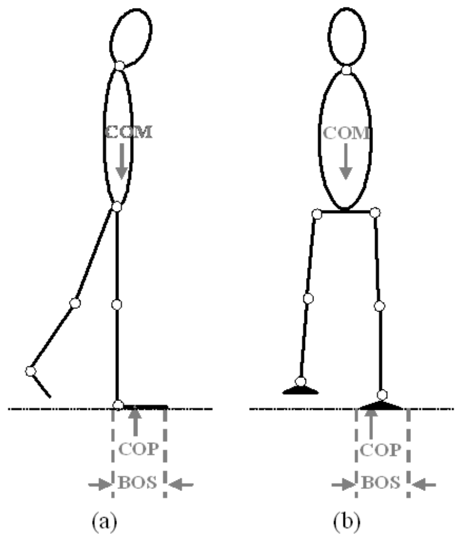

Figure 2.2: Five-segment representation of a walking subject in the (a) sagittal plane (b) frontal plane.

For a walking subject, postural stability is determined mainly by stepping strategies (Horak and Nashner, 1986; Nashner, 1980, Hof *et al*., 2007) where a stable base of support (BOS) is provided by the next step. While the main concern of postural stability during stance is the maintenance of the centre of gravity (COG) within the limits of stability determined by the BOS, dynamic balance during locomotion is maintained by adjusting the timing and placement of successive steps (Nashner, 1980). Deviations in the normal stepping trajectory are the main triggering stimuli for activating the reactive control mechanisms during walking (Nashner, 1980).

Using this control mechanism, the parameters like speed of forward progression, placement of successive steps, and height of the COM are adjusted.

2.2.2. Sensory components of human balance

As mentioned in the previous section, human balance is mechanically related to biomechanical measures of COP, BOS, COM and their interactions. To determine the state of postural stability and take proper actions in terms of automatic coordinated movements, these parameters are estimated by means of sensory systems.

The sensory components involved in human postural control are the visual, vestibular and somatosensory systems. Vision provides information on the relative position of the head with respect to the environment. The vestibular system is responsive to the angular and the linear accelerations of the head with respect to an earth-fixed inertial frame of reference. Otoliths respond to translational forces whereas semicircular canals detect the angular motion of the head. The somatosensory system responds to contact forces and gives information on body orientation with respect to the ground or support surface. It is defined as a combination of cutaneous, kinaesthetic (proprioceptive) and visceral sensory systems. The somatosensory system is responsive to force and displacement and so it can detect the relative orientations of body segments and applied forces.

Sensory systems do not all respond at the same time to the applied stimuli since each system can sense the motion of the body over different ranges of frequencies (Griffin, 1990). The somatosensory information supplied from end organs may be dominant for the perception of vibration at intermediate and high frequencies. For standing subjects on a moving platform, the initial sensory input is provided by the plantar mechanoreceptors that are excited by shear forces under the feet at the start of platform acceleration (Maki and Ostrovski, 1993b). Visual and vestibular systems are responsible for the perception of vibration at low frequencies. The contribution of the visual system to postural stabilization has been shown to occur in the low frequency range of about 0.03-0.3 Hz (Diener *et al.*, 1982). The otoliths are suggested as being responsible for some motion perception thresholds below 1 Hz (Griffin, 1990). High frequency thresholds may be determined by the somatosensory system rather than the visual or vestibular systems.

Although the somatosensory system has been claimed as the most dominant among all other senses (Diener and Dichgans, 1988; Diener *et al*., 1986), this probably depends on circumstances. For example, with eyes open and viewing a close stationary visual field, low frequency motion of the body is generally detected by the visual system before the vestibular or somatosensory system (Griffin, 1990). However, if the visual field is far away from the subject or moving with the subject, then vision does not provide consistent information on motion of the body. When the eyes are closed, the vestibular system can detect some low frequency translational oscillations below the thresholds of the somatosensory system (Griffin, 1990).

2.2.3. Postural strategies

Three different automatic postural strategies are used when human balance is perturbed while standing. Suppose that standing subjects are trying to maintain their balance on a moving support surface. When the support surface is firm and the centre of gravity COG is positioned within the limits of the BOS, an ankle strategy is mostly preferred (Jacobson, 1993; Nashner, 1986). A hip strategy is preferred in more risky conditions such as reduced limits of stability (e.g. narrow support surface) and when the centre of gravity (COG) is positioned near the extremes of the BOS (Jacobson, 1993; Nashner, 1986). A combined strategy with ankle and hip synergies can also be utilized with support surfaces that are intermediate between short and long (Nashner, 1986). In summary, the ankle strategy is preferred in the case of small postural disturbances whereas a hip strategy is preferred when the limit of stability is challenged to a greater extent (Horak and Nashner, 1986). A further strategy is either stepping or stumbling, which occurs when the centre of gravity (COG) moves beyond the limits of stability. This balance-recovery strategy is called compensatory stepping (McIlroy and Maki, 1993).

The postural strategies of standing subjects in the sagittal plane and the frontal plane are not very different. Medio-lateral stability of standing subjects has been suggested to be primarily controlled by hip abductors and adductors (Winter *et al*., 2003). The strategy accompanied with the muscle activity in hip abductors-adductors is referred as loading and unloading strategies (Winter, 1995). The hip strategy is common both in the sagittal and frontal plane but different muscles (hip extensors-flexors in sagittal plane, hip abductors- adductors in frontal plane) are activated.

The control of automatic postural movements may be both closed loop and open loop. In closed loop postural control (feedback control), errors between the desired (reference) and the actual postural state are sensed by the sensory systems. The errors are then minimized and corrective motor activities are performed accordingly. Open loop control is a feed-forward control in which corrective motor action is generated based on prediction rather than minimization of errors. Anticipatory postural adjustments (e.g. leaning, co-contraction, etc.) preceding voluntary movements is an example of open loop control. These types of adjustment are used to compensate for the postural perturbation that a voluntary movement is likely to cause (Maki, 1993).

The main strategy to maintain balance during locomotion is the stepping strategy (Horak and Nashner, 1986; Nashner, 1980, Hof *et al*., 2007); further strategies are used for fine tuning purposes. Active ankle moment via the ankle subtalar joint is used only for fine tuning purposes in the frontal plane whereas large errors in the foot placement are corrected by the hip moment via hip abductors and adductors (MacKinnon and Winter, 1993; Winter, 1995, Hof *et al*., 2007). Although the stepping strategy seems to be quite different from the postural strategies developed during standing, the general principle of the adjustment of the COP with respect to movements of the COM during standing (Murray *et al*., 1967; Prieto *et al*., 1993; Maki and McIlroy, 1996) still applies during locomotion: with an upcoming step the COP moves to provide a new stable BOS that can compensate the COM movement. The movement of COM determines the placement of the foot in successive steps so that the COM will remain within the stable BOS area formed during the double support phase of the gait cycle.

Adjusting step width via foot placement has frequently been referred to as an important strategy to maintain postural stability in the frontal plane while walking. It has been suggested that the step width, determined by the foot placement, regulates the COM trajectory to maintain balance in the frontal plane (Townsend, 1985). As a reactive control strategy in response to an external perturbation, Oddsson *et al.* (2004) concluded that, the CNS (central nervous system) adjusts the step width (moment arm) to compensate for the lateral acceleration induced by the perturbation. Oddsson *et al.* (2004) observed two strategies when gait was disturbed by transient perturbations with lateral components. The first strategy relied on the alteration of step width (moment arm) to compensate for the destabilizing moment caused by lateral perturbation. The second strategy involved aborting the swing leg and abruptly hitting the ground with this foot to increase the ground reaction force which contributes to the stabilizing moment.

The necessity for active control of lateral foot placement to maintain frontal balance while walking has been pointed out in many studies (Winter, 1992 and 1995; Redfern and Schumann, 1994; Bauby and Kuo, 2000). Bauby and Kuo (2000) developed a 3-dimensional passive dynamic model of walking. Stability analysis of the developed model revealed an unstable mode confined to lateral motion and this instability decreased as the step width increased.

2.3. Quantifying postural stability

One third of the elderly older than 65 years experience at least one fall per year (Tinetti *et al.*, 1988; Campbell *et al*., 1990). Falls are among the leading causes of injury and death especially for the elderly aged 85 and above (Overstall *et al.*, 1977, Hindmarsh and Estes, 1989). Falls are also mentioned to be the second leading cause of fatalities next to motor vehicle accidents world-wide (Courtney *et al*., 2001). Identifying the risk of falls, especially for the elderly, has been one of the main motivations of researchers quantifying postural stability. Another motivation is to get a better understanding of the developed postural strategies and the mechanisms involved in human balance. If a measure can be found to identify the state of postural stability, this measure can be used to identify the risk of falling before a fall occurs. A measure of postural stability can also be used to investigate the effects of several independent variables of interest (e.g. the magnitude of external perturbations, age, gender, etc.) on postural stability via modelling.

2.3.1. Methods of quantifying postural stability

2.3.1.1. Balance tests without perturbations

For the analysis of human postural control in quiet standing, spontaneous sway tests have commonly been used (Fernie and Holliday, 1978; Black *et al.*, 1982). Spontaneous sway measures have been generally based on COP measurements during unperturbed standing. These balance tests are easy to apply and give a basic understanding of postural stability. Spontaneous sway tests have also been used to investigate the effects of some important parameters (e.g. vision, support, age) on postural sway in quiet standing.

Unperturbed balance tests during locomotion have been carried out by asking subjects to walk on a level surface in laboratory environments where kinematic (e.g. position, velocity and acceleration of body segments) and kinetic measurements (e.g. COP) have been gathered (MacKinnon and Winter, 1993; Winter, 1995). Treadmills incorporated with force plates have also been used (Owings and Grabiner, 2004; Barak *et al.*, 2006; Hof *et al.*, 2007) to provide a walking task in a more controlled environment. Treadmills incorporated with force sensors are especially useful to avoid the challenge of force plate targeting in conventional walkway systems. Although there are differences between treadmill walking and walking over ground, the biomechanics of walking on a treadmill and over ground have been shown to be similar (Wagenaar and Beek, 1992). Compared to the traditional walkway experiments, there may be advantages of using treadmill in terms of controlled walking speed, reduced volume for movement recording and collecting data from more than a few strides.

The analysis of postural stability in unperturbed locomotion has been used to develop models of walking stability (MacKinnon and Winter, 1993; Winter, 1995). Gait measures (e.g. step length, step width, double support time) obtained in normal walking experiments have been used to compare the postural stability of young and elderly, healthy and unhealthy people (Kaufman *et al.*, 2006) or fallers and non-fallers (Barak *et al.*, 2006). An objective measure of postural stability which best predicts the risk of fall has been investigated among those gait measures (Maki, 1997). Obstructed walking experiments have been performed (Hahn and Chou, 2003; Lee and Chou, 2006) to identify potential fallers based on biomechanical measures (e.g. COM).

2.3.1.2. Balance tests with perturbations

Although quiet standing tests mentioned in Section 2.3.1.1 give a basic understanding of human postural control, they are not effective methods for analysing postural control behaviour in terms of an input-output model. Perturbations are helpful to evoke automatic postural strategies so that input-output models of postural stability can be constructed. Postural responses to small and large amplitude perturbations may be quite different due to nonlinear behaviours like sensory thresholds or nonlinear muscle stiffness (Maki and Fernie, 1988). However, spontaneous sway studies do not represent the input types that are larger in typical fall conditions (Maki and Fernie, 1988). That is why postural responses measured in quiet standing may not be useful in understanding human postural control in response to large amplitude perturbations. Disturbances encountered in real life (e.g. missteps, trips, slips, self-disturbing activities, transport) should be taken into account if postural stability in actual fall conditions is the concern. Perturbation parameters should be designed so as to represent real fall conditions. Meanwhile, experimental safety and ethics should be considered.

Moving (tilting and or translating) platform perturbations (Blümle *et al*., 2006, Maki *et al*., 1987, Peterka, 2002, Nawayseh and Griffin, 2006) have been widely used to evoke automatic postural strategies in standing subjects. Figure 2.3 shows an example of a moving platform (Blümle *et al*., 2006) used to perturb a standing subject. Maki and Ostrovski (1993a) used a moveable platform to compare the effects of transient and continuous stimuli on the postural stability of standing subjects in the sagittal plane. Peterka (2002, 2003) used the postural responses to moving platform perturbations to develop a simple control model of postural stability and identify sensory contributions to balance. Maki *et al*. (1987) used small amplitude continuous random or pseudorandom perturbations of a moving platform to identify an input-output model of postural stability in standing subjects.

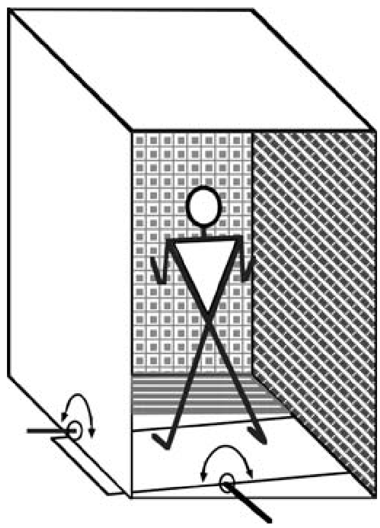

Figure 2.3: Schematic presentation of a moving platform to which a moving visual scene is attached (Blümle *et al*., 2006).

Various other types of perturbations to standing subjects have been summarized by Bortolami *et al*. (2003). These methods involve pushes and pulls on the body, release from a leaning posture to observe recovery strategies, self-generated perturbations (e.g. movement of arms), and vertical drops to simulate actual falling. A hold-and-release paradigm has been used by Bortolami *et al*. (2003) to simulate unexpected loss of balance such as tripping or loss of footing.

In this method, a horizontal force has been applied to the sternum of standing subjects. While the subjects resist to this horizontal force, it is suddenly withdrawn.

Perturbations to standing subjects have also been applied to detect stability thresholds (Jongkees and Groen, 1942; Graaf and Weperen, 1997). Jongkees and Groen (1942) perturbed standing subjects by sudden constant accelerations in terms of a step input. With eyes closed and feet together, subjects were able to maintain their stability up to an acceleration level of 0.76 ms^{-2} in a backward direction, 0.48 ms^{-2} in a forward direction, and 0.33 ms^{-2} in a sideward direction. Standing subjects were exposed to sudden acceleration and deceleration by means of a computer-controlled treadmill (Graaf and Weperen, 1997). Subjects were turned in the desired direction on the treadmill and sudden acceleration or deceleration was applied at a random instant. The stability thresholds were noted as the subjects managed to stand without holding the handrails, taking a protective step or stabilizing their body by large body sways or arm movements. The threshold values obtained (0.54 ms^{-2}, 0.45 ms^{-2}, and 0.61 ms^{-2} for forward, sideward and backward acceleration, respectively) were similar to the ones obtained by Jongkees and Groen (1942) and threshold values were lower for older subjects. People may be expected to be more stable when standing and supported on two legs than when walking and supported on only one leg for 80% of the gait cycle (Woollacott and Tang, 1997). Stability thresholds for walking subjects may therefore be different from those of standing subjects, but they have not been previously reported.

Moving platforms have also been used to investigate the effect of motion direction on the postural stability of standing subjects. Although experimental perturbations to human subjects have been generally applied in anterior-posterior (back and forth) direction, perturbations in daily lives also include lateral components. Maki *et al.* (1996) used a multi directional platform to investigate the effects of independent parameters (e.g. perturbation direction, perturbation magnitude, gender, prediction) on the compensatory stepping strategy. Perturbations with lateral components have also been applied to walking subjects. Oddsson *et al.* (2004) asked subjects to walk barefoot on a 12-m walkway along which a translating platform equipped with a force plate was incorporated (Figure 2.4). An impulsive mechanical perturbation was applied 45° forward and to the right, 45° rearward and to the left of walking subjects. Perturbation was applied immediately (i.e., 180 to 200 ms) after right foot heel strike. Medio-lateral distance between the sternum and supporting foot was used to investigate postural stability. The hypothesis regarding the alteration of step width in response to lateral perturbation was verified.

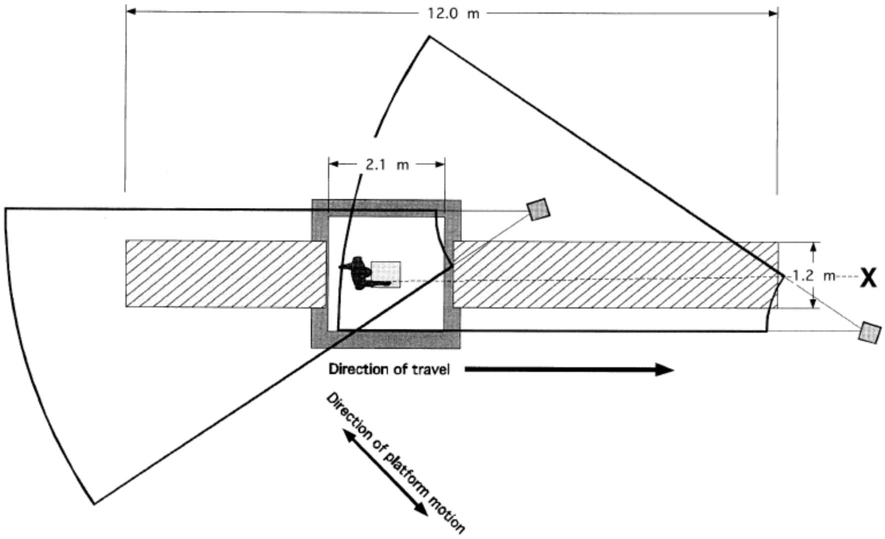

Figure 2.4: An overview of the experimental setup showing the translating platform embedded in a walkway (Oddsson *et al*., 2004).

A moveable platform (a separately moving surface for each foot) embedded in a 4-meter walkway was used to perturb the stepping of walking subjects (Nashner, 1980). Nashner (1980) changed the longitudinal positions of two platforms by a half cycle sinusoidal displacement. The perturbations were applied at different phases of the gait cycle. Berger *et al.* (1984) applied randomly timed impulsive accelerations (2.5-14 ms^{-2}) or decelerations of a treadmill to the walking subjects. Postural synergies as indicated by EMG activities developed in response to small gate perturbations were found to be similar to those developed during stance as suggested by Nashner (1980). However, when the perturbation amplitude increased, the dynamic postural control problem during locomotion was reported to become more complex.

Slip perturbations, as a representation of actual falls caused by slips, have been commonly used to investigate walking stability (Bhatt *et al*., 2005 and 2006; You *et al*., 2001, Lockhart *et al*., 2003). Bhatt *et al*. (2005) examined the effects of walking speed on postural instability caused by slips and compensatory stepping strategies developed for recovery. Walking subjects were perturbed by a slip induced by a computer-controlled moveable platform embedded in a 7-meter long walkway. Three force plates were used to measure ground reaction forces for the analysis before and after perturbation. The COM position and velocity with respect to the BOS were obtained using the kinematic data and 12-segment body dynamics. The authors (Bhatt *et al*., 2005) concluded that slower walking speed resulted in an increased risk of falling due to the decreased postural stability at slip onset.

Perturbations have also been applied for training and rehabilitation purposes. Mansfield *et al.* (2007) applied perturbation-based balance training to the elderly to investigate whether age-related impairments in compensatory stepping and grasping reactions can be improved. Unpredictable and multi-directional moving platform perturbations were applied to older adults (64-80 years) with fall or instability history. Subjects were either standing or walking in place.

The main objectives of the 6-week balance training program was to reduce the frequency of collision between the stance leg and the stepping foot, reduce the frequency of multiple-step responses, and increase the speed of grasping reactions.

Moving platform perturbations are quite common to investigate perturbed balance during standing. There have been few studies in which moving platform perturbations were applied to investigate walking stability. The effect of support surface perturbations on human walking was examined by Brady *et al.* (2009) who reported increased step width and increased step width variability in response to sinusoidal translations of a six-axis motion platform on which a treadmill was attached. Step width variability in response to sinusoidal platform perturbations has also been reported by O'Connor and Kuo (2009). McAndrew *et al*. (2010) examined the postural responses to pseudo-random oscillations (0.16-0.5 Hz) of a visual scene and support surface (Computer Assisted Rehabilitation Environment, CAREN, Figure 2.5). A moving six-axis motion platform on which a treadmill was fixed was used to assess ride comfort in railway vehicles (Ride comfort simulator, Japan, Suzuki *et al*., 2006). Apart from moving platforms, walking stability has also been perturbed by sudden pushes or pulls applied to the waists of subjects while walking on a treadmill (Hof *et al*., 2010).

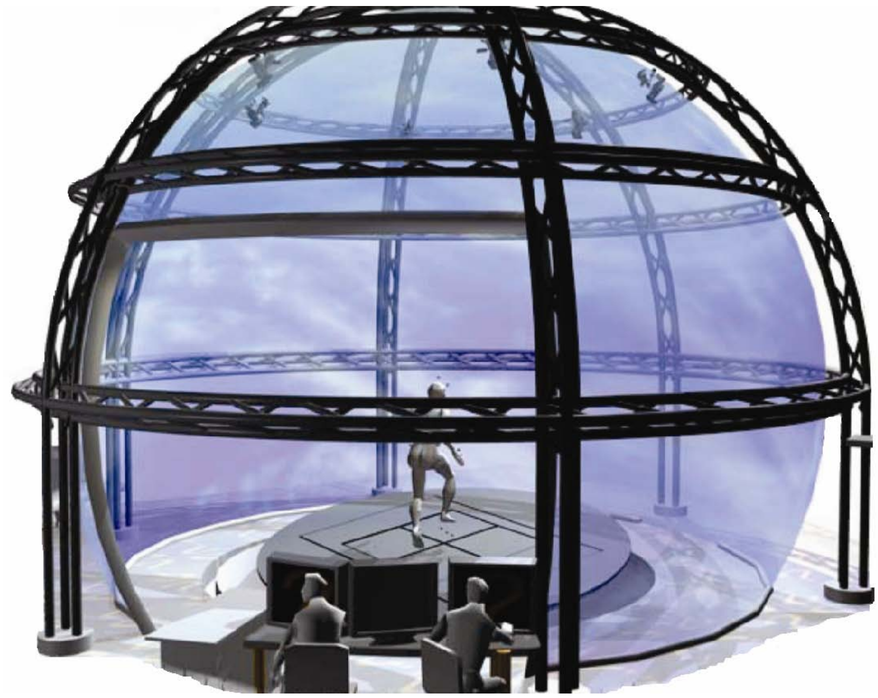

Figure 2.5: The CAREN: virtual reality system contained in a 7-m diameter dome with a six degree of freedom platform with a built-in instrumented treadmill (McAndrew *et al*., 2010).

2.3.2. Measures of postural stability

Although many parameters have been suggested to be used as a measure of postural stability, research is still continuing to determine the best measure that can represent how well the postural stability can be maintained in certain circumstances or how much a human subject is under the risk of fall.

2.3.2.1. Objective measures

Since COP movements are adjusted according to the movements of the COM, the COP has been used as an objective measure of postural stability during quiet standing. There are several advantages of using the COP as a measure of postural stability in standing subjects. First, the COP displacement is approximately proportional to the net ankle torque, so it is representative of the stabilizing effect of ankle muscle activity (Maki and Ostrovski, 1993a). In other words, ankle muscle activity performed during quiet standing adjusts the position of the COP with respect to the COG to cope with the destabilizing effect of gravity (Winter, 1995). Secondly, the COP can be interpreted as the degree to which stability limits are approached (Maki *et al*., 1987), since postural stability is lost as the COP reaches the limits of the BOS (i.e. the perimeter of the feet).

The COM motion has also been used to investigate human postural control. Estimating the whole body COM requires the use of a 3-dimensional complex whole body biomechanical model. It would be useful to identify the state of postural stability via simpler measures (e.g. motion of individual body segments like pelvis, trunk or head). As an individual body segment, head movement has been shown to be an appropriate parameter to identify balance during quiet standing (Berthoz and Pozzo, 1988; Alexander *et al*., 1992). Motion of the COM has been reported to be a more consistent and sensitive measure than the kinematics of individual body segments in identifying dynamic instability in elderly people (Hahn and Chou, 2003).

The mean COP speed (cumulative distance of COP over sampling period) representing the overall amount of activity to maintain balance was shown to be a sensitive parameter for predicting postural stability during quiet standing (Barotto *et al.* 2002, Hue *et al.* 2007).

Variability in step kinematics has been found to predict falls in elderly (Maki, 1997; Hausdorff *et al*., 2001). Gait tests without any perturbation were carried out on 75 elderly subjects (82±6 years) and spatial-temporal measures of foot placements were obtained (Maki, 1997). Prospective falls data were collected on a weekly basis for a 1-year follow-up period. Correlations of the objective measures with future falls and pre-existing fear of fall were analysed. Increased stride-to-stride variability in stride length, speed and double-support time were found to be correlated with falling but showed little evidence of fear. Reduced stride length, reduced speed, increased double support time, as all reported previously for elderly (Murray *et al*., 1969; Imms and Edholm, 1979; Pavol *et al*., 1999), and poorer clinical gait scores were associated with fear of falling but provided little indication of future falls. Among all other objective measures, stride-to-stride variability in speed has been suggested to be the best indicator of falling.

Variability in spatial and temporal characteristics of foot placement was also tested by Owings and Grabiner (2004) in healthy young and elderly subjects. Similar to findings of Bauby and Kuo (2000), step width variability was found to be much larger than step length or step time variability and a better discriminator between young and elderly. The authors (Owings and Grabiner, 2004) concluded that step width variability is a more meaningful descriptor of postural control during unperturbed walking than step length and step time variability.

In response to perturbed balance, reactive control strategies involving continuous adjustment of the COP according to the movements of the COM, are believed to be developed (Murray *et al*., 1967; Prieto *et al*., 1993; Maki and McIlroy, 1996). So, relative motion between the COM and the COP has often been used to identify dynamic postural stability during locomotion (Kaya *et al*., 1998, Lee and Chou, 2006). An interaction between the COM and the COP has also been used by others as indicators of dynamic stability (Prince *et al*., 1994; Tucker *et al*., 1998).

When the COP and COM are connected by a line at an instantaneous time during the dynamic gait cycle, the inclination angles (Figure 2.6) between this line and the vertical COP line, both in the sagittal and the frontal plane, were used as stability measures by Lee and Chou (2006). Temporal distance gait parameters like stride length, gait velocity and step width were also measured. Gait velocity and stride length decreased significantly in elderly patients with balance problems as previously reported (Alexander, 1996; Wolfson *et al*., 1992). Peak medial COM-COP inclination angles were significantly greater for elderly patients with balance problems whereas their peak anterior COM-COP inclination angle were significantly smaller than normal elderly. The results are inconsistent with the results of Hahn and Chou (2003) since, in unobstructed walking trials, both patients and healthy elderly showed similar COM displacements in the medio-lateral direction (Hahn and Chou, 2003). Among the two subjects having the same COM displacements with respect to their COP, the taller one has a smaller COM-COP inclination angle. Therefore, Lee and Chou (2006) suggested that COM-COP inclination angle was a more suitable parameter (compared to the relative COM-COP displacement) for quantifying postural stability as it takes into account the inter-subject variability.

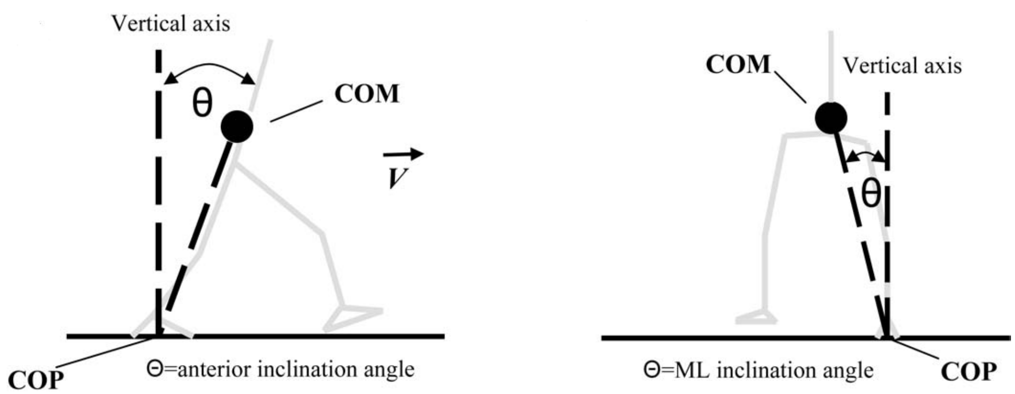

Figure 2.6: Lateral COM-COP inclination angle in the sagittal and the frontal plane (Lee and Chou, 2006).

The relative displacement and velocity of the COM with respect to the BOS have been recognized as important parameters to predict the postural stability of walking subjects in response to slips (You *et al*., 2001; Pai and Iqbal, 1999). Sagittal plane analysis of postural stability in response to slip perturbations revealed that smaller excursions and a greater COM velocity with respect to the BOS were advantageous in regaining balance from slips (You *et al*., 2001). Rather than relative COM-COP motion, Lockhart *et al.* (2003) showed that horizontal heel contact velocity and transitional acceleration of COM are significant measures of postural stability that can be used to identify slip related falls in the elderly.

In response to perturbations with lateral components (±45° to the line of progression of walking subjects), the step width has been used as a parametric measure to investigate postural strategies employed in perturbed locomotion (Oddsson *et al*., 2004). The adjustment of step width has been developed as a reactive control strategy to minimize the lateral destabilizing effect of the perturbation. Alteration of step width with active control of foot placement has been previously pointed out by many researchers (see Section 2.2.3).

Objective EMG measurements have been used to investigate muscular activation patterns in standing and walking subjects. Muscular activities have not been reported as a measure of postural stability that can identify fall risk. The amplitude and onset of activation in muscles have been rather used to examine the nature of employed postural strategies (e.g. the type of strategy, coordination of muscular activities) (Nashner, 1980; Berger *et al*., 1984; Nashner *et al*., 1979).

Increase in step width and step width variability in response to platform perturbations have been reported previously (Brady *et al*., 2009; O'Connor and Kuo, 2009). Walking subjects took wider, shorter and faster steps during lateral oscillations of the CAREN platform (Figure 2.5) than during back-and-forth oscillations and normal walking without oscillations (McAndrew *et al*., 2010). Walking subjects also showed greater variability in step length and step width during perturbed walking than during normal walking. Consistent responses to perturbations suggest that step width and step width variability are potential parameters for gait training and patient assessment (McAndrew *et al*., 2010).

Standing balance has also been clinically tested by a computerized dynamic posturography platform (Equitest, NeuroCom International Inc., Clackamass, Oregon, USA). The scores obtained by various balance tests are based on COP measurements and these scores have been used as objective measures to identify the state of postural stability. The system consists of a moveable dual force platform that can translate or rotate along with a moveable visual surround. The Sensory Organization Test (SOT) also assesses the contribution of three sensory systems (visual, somatosensory, and vestibular inputs) to balance under a variety of altered visual and surface support conditions.

2.3.2.1. Subjective measures

Several tests have been used for the clinical assessment of gait and balance disorders (e.g. Get up and Go Test, Timed up and Go Test). The Tinetti Performance-Oriented Mobility Assessment (POMA) test is one of the most frequently used test. The scores of Tinetti test which are below 20 have been associated with fivefold increased risk of falling (Rubenstein and Trueblood, 2004).

Clinical evaluations of postural stability do not involve objective measurements and the outcomes depend on the type of clinical test used. Elderly people who report a single fall may be examined using the 'Get up and Go Test' (Melzer *et al*., 2004) whereas for the elderly who demonstrate balance and gait abnormalities or who have recurrent falls, a more comprehensive fall evaluation is required (Melzer *et al*., 2004).

Balance in the medio-lateral direction has been clinically assessed by several tasks involved in Berg balance tests (Berg *et al*., 1989) and Tinetti balance tests (Tinetti *et al*., 1986). Tasks for evaluating the ability to stand in reduced base of support (standing with feet together, tandem stance and one-legged stance) and tasks involving sideward weight shift (alternate stepping onto a stool, turning while standing, one-legged timed score) are used to evaluate postural stability in the frontal plane. Although these tests provide useful information, the scoring systems have been suggested to be very broad and subjective (classification as normal, adaptive or abnormal for the Falls Risk Index) and therefore are not able to detect deficiencies other than major problems in human balance systems (Brauer *et al*., 1999)

The 'lateral reach test' showing the ability of standing subjects to reach directly sideward as far as possible without overbalancing or taking a protective step was used to identify fallers and non-fallers in elderly people (Brauer *et al*., 1999). The test results were found to have high test–to-test repeatability and were symmetrical between the sides and significantly correlated with the measured COP excursions. The Lateral Reach Test has been suggested as a useful tool for investigation of medio-lateral postural stability in the older adult population.

Other than clinical evaluations, subjects' perception of fall risk has been used as a subjective measure of postural stability in standing subjects (Nawayseh and Griffin, 2006). Nawayseh and Griffin (2006) used subjects' estimates of their probability of losing balance as a measure to investigate the postural stability of subjects standing on a floor oscillating in the horizontal directions (fore-and-aft or lateral). Standing subjects exposed to various magnitudes and frequencies of random oscillatory motion were asked to estimate the probability of losing balance if the same exposure were repeated. The estimated probability of losing balance was also compared with COP measurements from a force plate and the actual loss of balance defined by the percentage of people who fell or held a support to prevent falling. Although subjective measures obtained by Nawayseh and Griffin (2006) may not represent the actual probability of losing balance they provide a clear indication of how the perception of the risk of fall depended on various factors and the adopted postural strategies.

2.4. Factors affecting postural stability

There are many factors affecting the performance of human subjects in maintaining their postural stability. In this section, the effects of several characteristics of external perturbations (e.g. magnitude, frequency, waveform), inter-subject variability and supports on postural stability are briefly mentioned. Investigation of these factors is not only important for the design of perturbed balance experiments but also significant for the design of environments (e.g. transport environment) where some of these factors (e.g. magnitude, frequency, support) can be controlled to reduce fall risk.

2.4.1. Effect of perturbation magnitude

Acceleration, velocity or displacement can be used to quantify the magnitude of perturbation. Maki and Ostrovski (1993a and 1993b) suggested that acceleration is a more reasonable parameter for quantifying perturbation amplitudes since joint moments are caused by platform acceleration when balance is perturbed by a moving platform. Acceleration has also been used as a common measure to evaluate external perturbation (e.g. vibration in transport).

The amplitude of the perturbation signal used in balance experiments should not be so large as to cause safety problems, but it should be large enough to provide a reasonable signal-to-noise ratio especially for transient waveforms (Maki, 1986; Maki *et al*., 1987). It should be remembered that low amplitude and high amplitude signals may excite threshold and saturation nonlinearities in the human postural control system. While selecting the perturbation magnitude to be used in human balance experiments, the magnitude range of actual disturbances should also be considered. If the postural stability of train passengers is of interest, the reasonable motion magnitudes are often in the range 0.2 ms^{-2} r.m.s. to 1.0 ms^{-2} r.m.s. (Griffin, 1990).

The effect of the magnitude of perturbation on developed postural strategies has been investigated. Postural strategies developed in response to small gate perturbations have been suggested to be similar to those developed during stance (Nashner, 1980). However, when the perturbation amplitude increased, the dynamic postural control problem during locomotion became more complex (Berger *et al*., 1984). This complexity may be associated with the nonlinearity of the postural control system caused by high amplitude perturbation.

Oddsson *et al.* (2004) observed postural strategies when gait was disturbed by transient perturbations with lateral components. Larger changes in moment arms (step width) and sternum sway occurred in larger perturbation magnitudes and these relations were found to be linear.

Subjects walking on a treadmill were perturbed by lateral acceleration via 6-axis motion platform to assess ride comfort in railway vehicles (Suzuki *et al*., 2006). Subjective ratings of discomfort and centre of gravity displacement were reported to increase with increasing magnitude of lateral acceleration.

The effect of the magnitude of perturbation on the discomfort of seated and standing subjects exposed to whole body vibration has been investigated (e.g., Reiher and Meister 1931, Oborne and Clarke 1974). With whole-body vibration, Morioka and Griffin (2006) found that the frequency-dependence of equivalent comfort contours at magnitudes close to the perception threshold were different from the contours obtained at higher magnitudes.

In terms of the effect of perturbation magnitude on the postural stability of standing subjects, Nawayseh and Griffin (2006) showed that the displacement of the COP, actual loss of balance (falls or grasping to prevent falls), and subject estimates of the probability of losing balance all increased with increasing magnitude of horizontal (either fore-and-aft or lateral) oscillation.

2.4.2. Effect of perturbation frequency

The frequency range to be used in perturbed balance experiments should be selected in accordance to the actual stimuli conditions of interest. “For ride comfort, frequencies of interest in rail vehicles are 0.1 Hz to 2 Hz on curve transitions (roll), 0.5 Hz to 10 Hz in the lateral and longitudinal directions, and 0.5 Hz to 20 Hz in the vertical direction. For ultra-high-speed vehicles (250 km/h and faster) and for tilting trains, vertical accelerations in the low frequency range of 0.1 Hz to 0.5 Hz can occur which may result in motion sickness” (ISO 2631-4, 2001).

The frequency range 0.1 to 2 Hz is reasonable for investigating postural stability. High frequencies should be avoided as they induce fatigue and discomfort rather than postural instability. Caution should also be taken since translational motion in a laboratory environment can create motion sickness in the frequency range between about 0.1 and 0.5 Hz (Griffin, 1990).

The frequency-dependence has been expressed in comfort contours for standing people exposed to whole body vibration (Miwa 1967, Oborne and Boarer, 1982, Thuong and Griffin, 2011). Nawayseh and Griffin (2006) investigated the effects of frequency on the postural stability of standing subjects perturbed by random oscillations of the floor. During fore-and-aft and lateral oscillation with the same velocity at all frequencies from 0.125 to 2 Hz, the displacement of the COP, the loss of balance, and subjective estimates of the probability of losing balance all peaked at around 0.5 Hz. The effect of low frequency whole-body vibration on the postural stability of walking people has not been previously reported.

2.4.3. Effect of perturbation waveform

Postural stability during standing and walking can be disturbed by various types of perturbations with different waveform characteristics. Transient perturbations may characterize slips, trips or short duration oscillations in transport whereas continuous perturbations may represent the continuous destabilizing effect of gravity and may be used to investigate the steady-state characteristics of human postural control system. Some of the studies of human postural control (Nashner, 1980; Horak and Nashner, 1986) have employed transient stimuli (e.g. sudden support surface motions) to evoke typical postural responses. Most of the studies (Peterka, 2002; Mergner *et al.*, 2006) have employed continuously varying stimuli (e.g. sinusoidal or more complex random time series) to evoke steady-state responses that may then be used to obtain transfer function models of human postural control.

The effect of motion waveform on discomfort has been investigated for seated and standing subjects. A common measure of an acceleration waveform is root-mean-square (r.m.s.) value which is a suggested method of predicting discomfort for seated and standing people caused by various types of vibration (ISO 2631-1(1997) and BS6841 (1987)). However, the r.m.s. is not optimum for evaluating all types of waveforms (sinusoidal and octave-bandwidth random waveform with increasing peak levels) in terms of discomfort levels of standing subjects (Thuong and Griffin, 2010b). Oscillations having the same frequency and the same r.m.s. value caused greater discomfort with increasing peak levels for seated subjects exposed to vertical whole-body vibration (Griffin and Whitham, 1980). Howarth and Griffin (1991) also reported an increase in the discomfort of seated people with increasing crest factor of oscillations although the r.m.s. values of the oscillations were kept constant. The effect of waveforms on the postural stability of walking people has not been reported systematically.

Differences in waveforms produce differences in the perception of motion in terms of discomfort, and subjects are more sensitive to random vibration than to sinusoidal vibration (Griffin, 1976) which might be an effect of unpredictability in random motions (Maki, 1986; Maki *et al*., 1987). There is also evidence of nonlinearity in the postural stability of standing people exposed to perturbations with continuous and transient waveforms (Maki and Ostrovski, 1993a). In terms of magnitude-dependence, postural responses to transient stimuli have been found to have a more nonlinear behaviour than responses to continuous perturbations (Maki and Ostrovski, 1993a). Authors (Maki and Ostrovski, 1993a) suggested that predicting responses to transient stimuli from continuous perturbation tests is not reliable. It has been also suggested that the postural control system responds differently to transient and continuous perturbations: feedback control is used for continuous perturbations whereas feed-forward control is utilized for transient recovery (Diener and Dichgans, 1988).

The waveform characteristics of transient platform perturbations experienced by standing subjects have been traditionally reported in terms of peak velocity and displacement (Horak and Nashner,1986; Tang *et al*. 1998). Acceleration has been suggested by Maki and Ostrovski (1993a, 1993b) to quantify the magnitude of external perturbation. Brown *et al*. (2001) also emphasized the necessity of reporting acceleration characteristics of perturbation waveforms. Runge et *al*. (1999) showed that the kinetics of postural recovery is dependent on the velocity of platform translation. There is not a standardized procedure to report the perturbation characteristics of waveforms, which makes it difficult to compare and interpret the results of different perturbed balance experiments conducted in different laboratory environments.

2.4.4. Effect of support

Postural supports, such as mobile assistive devices (e.g. canes and walking aids), can assist the maintenance of stability during quiet standing and when walking. Support may be more beneficial while standing or walking in trains, buses, ships, and aircraft where balance can be disturbed by the oscillatory motion of the floor. There have been many studies regarding the effects of supports in improving postural stability in quiet standing and normal walking but there are no known systematic studies of how the use of a hand support and varying the height of a hand support influence postural stability during perturbed locomotion.

Bateni and Maki (2005) reviewed studies of the benefits and adverse effects of assistive devices on postural stability and mobility. Mobility aids (canes or walkers) are biomechanically advantageous to increase the BOS especially during the single support phase of the gait cycle such that greater range of COM movements can be compensated with an increased BOS. Mobility aids also provide rapid mechanic stabilization by providing stabilizing reaction forces at the hand. Another advantage of using supports is the reduction of loading on the lower limbs which is especially important for patients with injury or pain in the lower limb. Apart from mechanic stabilization, mobility aids provide somatosensory cues which are used as additional spatial sensory information for the central nervous system (CNS). Similar advantages can also be attributed to supports used in transport. Apart from its advantages, mobility aids have adverse effects on balance and mobility due to their demands on attention (e.g. lifting and advancing the device, and contacting the ground in appropriate location). Several other disadvantages have also been summarized by the authors (Bateni and Maki, 2005). Although the type of supports used in transport, being stationary and not required to be lifted or advanced, are quite different from walking aids, investigations of the effect of hand supports on postural stability can provide useful information about to which extent postural stability can be improved via support when balance is perturbed during walking.

Supports used in transport have been shown to improve postural stability for standing subjects. With instantaneous increases in horizontal acceleration, standing people have been reported to maintain balance while exposed to accelerations up to 0.76 ms^{-2} in the backward direction, 0.48 ms^{-2} in the forward direction, and 0.33 ms^{-2} in a sideways direction (Jongkees and Groen, 1942; Graaf and Weperen, 1997). The acceleration in public transport can be greater than these values so standing people cannot maintain stability without holding a support (Jongkees and Groen, 1942), and it has been shown that greater accelerations can be tolerated when using a support (Browning, 1974).

The effect of body support on vibration discomfort has been studied for seated subjects (Wyllie and Griffin, 2007) and for standing subjects (Thuong and Griffin, 2010a). When exposed to horizontal oscillation, whether a support increases or decreases the discomfort of seated people and standing people depends on the frequency and the direction of the oscillation. The discomfort of standing people seems to be increased when a support increases the transmission of high frequency vibration to the upper-body (Thuong and Griffin, 2010a, Figure 2.7), whereas postural instability is caused by low frequency oscillation. When walking, and supported on one leg for 80% of the gait cycle (Woollacott and Tang, 1997), stability may be less than when standing and so supports may be more beneficial.

Figure 2.7: Postures adopted by standing subjects (Thuong and Griffin, 2010a): (i) without support (ii) with bar support (iii) with shoulder support (iv) with back support.

Light touch contact with a surface, even if it does not provide forces sufficient to stabilize the body, has been found to improve standing stability by providing an additional sensory cue to body movement (Jeka and Lackner, 1994 and 1995; Tremblay *et al.*, 2004, Clapp and Wing, 1999; Holden *et al*., 1994). The additional sensory cue is provided by somotasensory information (due to reaction forces) together with the proprioceptive information (via cutaneous stimulation) of the arm-torso configuration (Holden *et al*., 1994).

Touch contact in tandem stance (heel-to-toe) was found to be as effective as force contact (mean vertical force around 5 N, mean horizontal force around 1 N) or vision in reducing postural sway (medio-lateral COP sway) when compared to the no contact, eyes closed condition (Jeka and Lackner, 1994). The forces during light touch were around 40 grams although subjects were allowed to apply up to 100 grams of force. Jeka and Lackner (1994) related this to the possibility that subjects were using a contact force range (30-50 grams) where receptor sensitivity was shown to be greatest (Westling and Johansson, 1987). The correlation between COP sway and contact forces was lower and the time delay between the body sway and fingertip forces was higher in light touch suggesting a sensory cue of the fingertip contact via somatosensory and proprioceptive sensory feedback. If the contact forces were used for mechanic stabilization, they would increase or decrease by body sway such that changes in contact force would follow the changes in body sway.

Standing subjects, eyes closed, in tandem stance position (Figure 2.8) showed a reduction by over 50% in mean sway amplitude (COM sway) both with a light touch (<1 N in medio-lateral or vertical direction) and force touch (~10 N) (Jeka and Lackner, 1995). Although the light touch force levels less than 1 N were far below the levels required for mechanical stabilization (Holden *et al.*, 1994), they resulted in reduction of postural sway by over 50%. During light touch, people were controlled to apply contact forces less than 1 N, which was an additional task added to the main task of keeping postural stability. It is not clear if subjects would really prefer a light touch if they were not restricted to apply forces less than 1 N. The adopted strategy in terms of applied contact forces might also differ if postural stability was threatened to a greater extent by external perturbations. Subjects could apply larger forces if they were able to pull as well as push the support or able to grasp it rather than use fingertip touch, in which case support could be used more as a mechanical tool rather than a sensory cue.

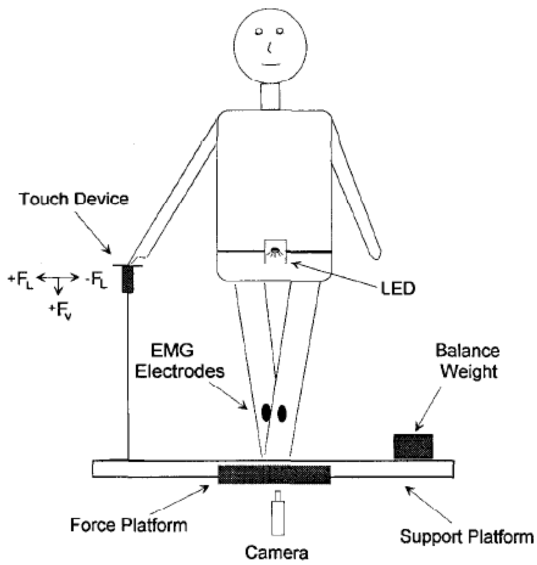

Figure 2.8: Systematic description of a subject in tandem stance on the force platform with the touch device (Jeka and Lackner, 1995).

Tremblay *et al.* (2004) reported that both the young and the elderly showed a significant reduction (40-55 %) in the mean postural sway amplitude in the anterior-posterior direction and smaller but still significant reductions (8-12%) of the mean sway amplitude in the medio-lateral direction with use of support during quiet standing. Clapp and Wing (1999) previously reported a similar reduction of about 40% and about 10% in mean sway amplitude in anterior-posterior

(AP) and medio-lateral (ML) directions for subjects in normal standing position. Sway amplitude was higher in medio-lateral direction in elderly subjects and larger reductions in medio-lateral sway amplitude were observed with touch. This result is consistent with the previous findings of lateral instability problems as an indication of balance problems in the elderly (Maki and McIllroy, 1996). The higher vertical normal forces applied to the touch plate by the elderly (1.21±0.75 N) compared to contact forces applied by young adults (0.32±0.15 N; <0.5 N) were suggested to be compensating for the loss in tactile sensation in the elderly. Fingertip contact forces were not controlled as in previous studies (Dickstein and Laufer, 2004; Jeka and Lackner, 1994 and 1995) such that subjects could apply as much as force they preferred. However, subjects were still instructed that the touch plate was not designed to support heavy forces and therefore could not be used as a cane or a walking aid, which might bias subjects' attempts to minimize contact forces.

Fingertip contact from an external reference (e.g. handrail of height 90 cm) has also been suggested to improve postural stability during treadmill walking (Dickstein and Laufer, 2004). Light fingertip touch (controlled to be less than 200 grams in all three directions) had a similar effect as heavy touch and vision on the centre of mass (COM) sway by decreasing the sway; the effect was more pronounced in the anterior-posterior direction. The side of the rail did not have any significant effect on COM sway, but the force applied to the right handrail was greater than the force applied to the left handrail. Maintaining a better postural control by use of a handrail during locomotion might also be related to the reduction of physiological stress via light touch of the handrail (Manfre *et al.,* 1994).

Assuming an inverted pendulum model of the human body, the stabilizing moment from a hand support will increase as the height of the support increases. Touch bars around waist height (Figure 2.8) have been used to investigate the effects of supports during quiet standing (Clapp and Wing, 1999; Jeka and Lackner, 1994 and 1995), but the effect of supports on stability may depend on their height. The effects of support height on the postural stability during perturbed walking have not been previously reported.

2.4.5. Effect of subject physical characteristics

Inter-subject variability in postural stability may arise from many factors like differences among subjects in terms of age, gender, posture, fitness and prior experience to motion stimuli. Age is one of the most significant subject characteristics the effect of which has been commonly investigated on human balance. Age has been shown to have deteriorating effects on postural stability due to the reduction in ability to sense and actuate movement with ageing. The number of afferent and efferent channels and the quality of the signals transmitted through these channels degrade with age (Horak *et al*., 1999; Alexander, 1996). The elderly (75±2 years) compared to the young (24±5 years) have been found to exhibit slower reaction times and more rigid postures during voluntary movements from quiet stance (Tucker *et al*., 2008). Age-related declines in the speed of postural responses have also been reported during voluntary stepping (Luchies *et al*., 2002; Patla *et al*., 1993) and during sudden turns and termination of gait (Cao *et al*., 1997).

Aging has been associated with increases in mean sway amplitudes and velocities of sway during quiet standing (Tremblay *et al*., 2004, Baloh *et al*., 1998; Prieto *et al*., 1996). Balance performance during one-legged stance was found to be significantly related to age, gender, stature and body weight (Balogun *et al*., 1994). Postural stability parameters evaluated via centre of pressure measurements during quiet standing were also found to be affected by stature, weight, foot width, and base of support area (Chiari *et al*., 2002).

Among the physical characteristics including stature, age, foot length, waist and hip circumference, body weight was found to be the best predictor of postural instability as assessed by the mean COP speed (cumulative distance of COP over sampling period) during quiet standing (Hue *et al,.* 2007). Authors (Hue *et al,.* 2007) suggested that overweight is likely to reduce the sensitivity of mechanoreceptors under the feet which plays an important role in feedback control system to adjust body sway. Another reason for decreased postural stability with increased weight might be related to the extra abdominal mass pushing the centre of mass (COM) to the edges of the base of support (BOS) in which case more corrective action is required to maintain balance. Age contributed to only a small portion of the variance in postural stability which might be caused by the specific age and weight ranges used by the authors (Hue *et al.,* 2007). The age ranged from 24 to 61 years whereas the weight range was 59.2 to 209.5 kg. The mean weight of the group of 59 males was 107.7 kg (±35.6) which indicates that the study was focused on overweight people. Postural instability associated with obesity has also been reported by others (Goulding *et al*., 2003; Chiari *et al*., 2002).

Peak lateral COM-COP inclination angles (Figure 2.6) were found to be significantly greater for elderly patients with balance problems whereas their peak anterior COM-COP inclination angles were significantly smaller than normal elderly (Lee and Chou, 2006). Lee and Chou (2006) suggested that elderly patients might be less stable to disturbances in the lateral direction than to disturbances in the fore-and-aft direction. Elderly people being less stable in the frontal plane may be associated with the tendency of frail elderly people to fall sideways during their daily activities (Greenspan *et al*., 1998).

Most studies of aging and balance were focused on postural stability during quiet standing whereas the ability to keep postural stability requires more rapid and accurate postural reactions to recover from challenges to perturbations in real life like slips, trips, or external disturbances experienced in transport. Stepping is a significant reaction to overcome these perturbations even if the perturbation is relatively small (Maki and McIlroy, 1999). The effect of aging on compensatory stepping reactions to lateral perturbation during standing and walking in place has been previously reported by Maki *et al.* (2000). Studies of the effects of age on gait revealed that older adults have lower gait speeds and step frequencies, higher step width, and increased gait variability (Moe-Nilssen and Helbostad, 2005; Hausdorff *et al*., 2001; Owings and Grabiner, 2004). The effects of a wide range of subject characteristics including age, gender, weight and stature on postural stability during perturbed locomotion have not been previously reported.

2.4.6. Other factors

Although the timing of perturbation is not critical for standing subjects, it is an important consideration in the design of stimuli for walking subjects. Walking subjects can be perturbed randomly at any phase (toe-off, heel strike) of the step cycle, as suggested in Berger *et al*. (1984) and Nashner (1980). Postural stability may vary at different phases of the gait cycle, previously mentioned as “phase dependent modulation of reflexes” (Forssberg *et al*., 1975). Nashner (1980) found that the effects of impulsive perturbation (0.5 cycle sinusoidal displacement of a moveable platform) on walking subjects is strongest at heel strike and the beginning of the single support phase and weaker at the mid-stance and absent at the onset of double-support phase of the gait cycle. During toe-off position (just before heel strike occurs), the effect of perturbation is expected to be less threatening as the subjects are able to compensate the destabilizing effect of perturbation by developing a stepping strategy with their one foot just about to hit the ground. However, it is more difficult to take an appropriate stepping action at heel strike as the stepping action has already been taken. The effect of timing is more emphasized when a shock-type impulsive input is used. The effect may be less during a longer time perturbation which covers the whole gait cycle.

The duration of perturbation may affect subjective and objective measures of postural stability. Griffin and Whitham (1980) showed that with decreasing duration greater levels of acceleration is required to produce the same vibration discomfort in sitting people.

The predictability of perturbations has also been shown to affect postural stability (Maki, 1986; Maki *et al*., 1987). Prediction may result in adaptation to the imposed stimuli and cause a shift to a predictive control strategy such that anticipatory postural adjustments (e.g. leaning) are developed to minimize the effects of perturbation.

The performance of postural tasks has been found to be deteriorated by a secondary cognitive task which requires attention (Teasdale and Simoneau, 2001, Shumway-Cook *et al*., 1997). Attention and motivation of an individual may also influence the subjective judgments as well as task performance (Griffin, 1990). Decrements in postural performance when performing a cognitive task during quiet stance (Stelmach *et al*., 1990; Shumway-Cook and Woolacott, 2000) and during walking (Lundin-Olsson *et al*., 1997 and 1998) have been reported.

2.5. Models of postural stability

A passive rigid body model has been proposed for the prediction of the loss of balance of standing subjects on the decks of ships (Graham, 1990). The human body in a standing posture was represented by a rigid body with a similar shape, size, and mass as the human body (Figure 2.9). The model predicts the number of 'motion-induced interruptions' (MIIs) that are assumed to occur when the centre of pressure threatens to move outside the base of support. MIIs are assumed to interrupt the performance of tasks by postural adjustments made to regain stability by holding on to a fixed structure or by making a significant postural adjustment. A mathematical formula was developed to estimate MIIs based on the root-mean-square acceleration magnitude but not the frequency of motion. Lewis and Griffin (1997) found that the model proposed by Graham (1990) overestimated the number of MIIs in standing subjects on a ship motion simulator.

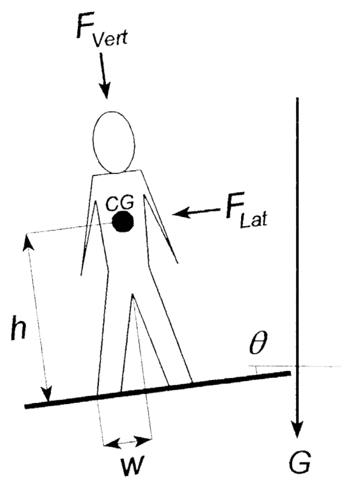

Figure 2.9: Rigid body model of postural stability by Graham (1990).

When the COP is within the limits of the BOS, it has been suggested that the human balance system during standing can be approximated by a linear transfer function model (Maki *et al.*, 1987; Maki and Fernie, 1988). A linear transfer function represents an input-output relation between the acceleration magnitude of the perturbing platform and the COP of the standing subject. The developed model (Maki *et al.*, 1987) represents postural responses to low magnitude perturbations (max 0.15 ms^{-2} r.m.s) where the COP remains within the limits of BOS. Loss of balance is predicted when the stability margin (the distance between the COP and the nearest boundary of the BOS, Figure 2.10) is reduced to zero. The model does not take into account high magnitude perturbations as the human balance system switches to more complex balance strategies (e.g. protective step, grasping a handrail) to avoid falls. Lewis and Griffin (1997) found that the proposed model underestimated the number of MIIs ('motion-induced interruptions') in standing subjects on a ship motion simulator. The model may not represent the actual risk of fall during perturbed standing since subjects make anticipatory postural adjustments while COP is still well within the limits of the base of support area (Lewis and Griffin, 1997).

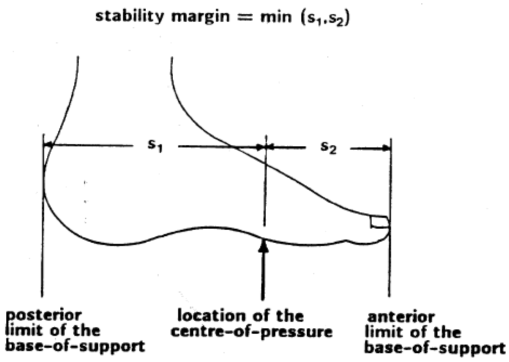

Figure 2.10: Anterior-posterior stability margin (Maki *et al*., 1987).

Passive biomechanical models (Koozekanani *et al*., 1980; Riley *et al*., 1990) have been developed to represent the human body by rigid links connected with hinge joints. These models are useful to develop the kinematic relationships between body segments but do not involve the active elements for controlling human balance. Active models (Mergner *et al.*, 2006; Peterka, 2002 and 2003) of human postural control, as shown in Figure 2.11, represent the effects of sensory systems and the control structure of the central nervous system (CNS). Measuring the human response (e.g. COM sway in the sagittal plane) and perturbation input (e.g. platform displacement), system identification techniques have been used to identify parameters of concern (e.g. sensory contribution, time delay). These active models are promising for identifying balance problems and standardizing perturbation experiments for the purposes of diagnosis and rehabilitation of balance related problems. True selection and interpretation of identified parameters are important and experimental results are required to be repeatable to standardize the experimental procedure.

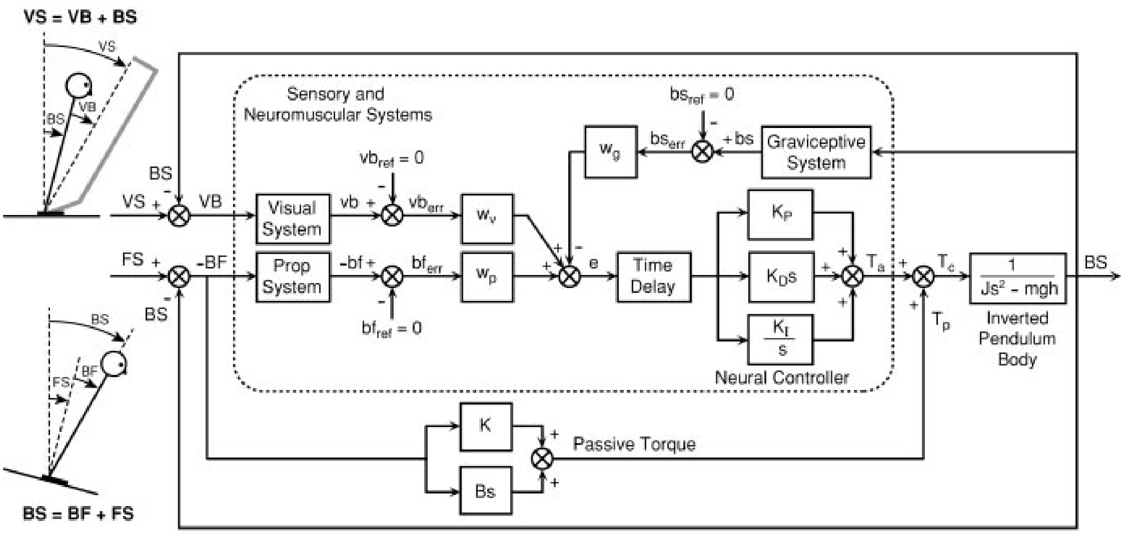

Figure 2.11: Postural control model of postural stability in standing subjects(Peterka, 2002).

The models developed for standing subjects can provide some useful information about the general concept of postural stability in walking subjects. Inverted pendulum models used for standing subjects can resemble partly the single support phase of the gait cycle at an instantaneous time. The differences in the biomechanics of balance between standing and walking, as mentioned in Section 2.2.1, should be taken into account for modelling walking stability.

Two models, as shown in Figure 2.12, have been proposed to represent the postural stability of walking subjects in the frontal plane (MacKinnon and Winter, 1993; Winter 1995). The first model is an inverted pendulum model of the HAT (head, arms and trunk) and swing leg about the hip joint. The second model is an inverted pendulum model of the whole body about the supporting ankle subtalar joint. The models suggest that, apart from the main strategy of stepping, frontal balance is regulated further at two levels: In the first level, the COM of the upper body is regulated about the ankle subtalar joint, and in the second level the COM of the whole body is regulated about the hip joint. The validity of the proposed models (MacKinnon and Winter, 1993) was checked by comparing the net moments about the hip and subtalar joint with the model estimates. The models were verified to be used in the single support phase of the gait cycle as the modelling errors were minimum at this phase. The authors (MacKinnon and Winter, 1993) suggested that once the stepping strategies have been employed, fine tuning strategies are used to compensate the errors in foot placement. The model represents the single support phase of the gait cycle in unperturbed walking, and it was developed based on measurements from four subjects during unperturbed walking. The models do not consider the effect of perturbing forces on walking stability.

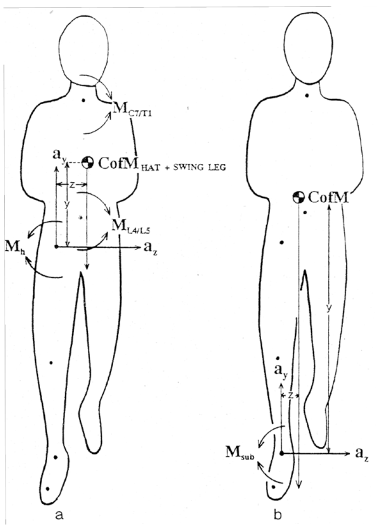

Figure 2.12: Inverted pendulum models of walking stability: (a) HAT (head, arms and trunk) and swing leg about the hip joint and (b) whole body COM around the ankle subtalar joint in the frontal plane (MacKinnon and Winter, 1993).

The necessity for active control of lateral foot placement to maintain frontal balance while walking has been mentioned in Section 2.2.3. As a reactive control strategy to maintain frontal balance developed in response to external perturbation, Oddson *et al.* (2004) suggested that the CNS (central nervous system) adjusts the step width (moment arm) to compensate for the lateral acceleration induced by the perturbation. Consistent with the requirement for active lateral foot placement, gait has been represented by a generalized inverted pendulum model with a moveable support joint to define the lateral foot placement (Townsend, 1985). This inverted pendulum model suggests that upright stability of the human body during locomotion is maintained by controlling the lateral foot placement. Gait stability in the model was provided by discrete foot placements and for the active control of these lateral foot placements, feedback was provided at the onset of each step (Townsend, 1985).

Postural stability models mentioned in this section are useful for understanding the dynamics and control of human balance during standing and walking but they do not predict probability of losing balance when stability is perturbed by external oscillations. The effects of oscillation characteristics (e.g. magnitude, frequency) have not been systematically investigated in these models. Postural stability models have been derived based on the postural responses to either step, sinusoidal, or broad-band random oscillations. These perturbations are not representative of actual disturbances encountered in real life, one example of which is the whole body vibration experienced during a train ride.

2.6. Conclusion

There have been fewer studies of balance during perturbed walking compared to balance during quiet standing, perturbed standing, or normal walking. The complexity of the dynamics of locomotion, the difficulty in applying appropriate stimuli, high inter-subject variability in gait measures, larger laboratory environments required for gait analysis, and restrictions in traditional walkways are several reasons for the comparably few studies.

Walking subjects have been perturbed by means of sudden accelerations or decelerations of a treadmill (Berger *et al*., 1984) or a moveable platform embedded in a walkway (Nashner, 1980; Oddsson *et al*., 2003; Bhatt *et al*., 2005). The effect of support surface perturbations on human walking has been examined by moving platform perturbations (e.g. treadmill embedded on a motion platform) (Brady *et al.*, 2009; O'Connor and Kuo, 2009; McAndrew *et al*., 2010; Suzuki *et al*., 2006). These types of perturbation involve impulsive inputs or longer duration sinusoidal or random inputs but they do not simulate the destabilizing effects of oscillatory motions encountered in transport (e.g. trains and ships).

The effects of perturbation parameters (e.g. magnitude and waveform) have been documented for standing people (Maki, 1986; Nashner, 1986; Horak and Nashner, 1986; Maki and Ostrovski, 1993a). People may be expected to be more stable when standing and supported on two legs than when walking and supported on only one leg for 80% of the gait cycle (Woollacott and

Tang, 1997). Postural stability during locomotion can further be threatened by external disturbances (e.g. slips, oscillatory motions in transport). However, dependence of the postural stability of walking people on the perturbation characteristics (e.g. magnitude, frequency, or direction of motion) has not been systematically investigated. The effects of hand support and subject characteristics on postural stability during perturbed walking are also unknown.

Human beings with feelings are not solely rigid mechanical systems. People feel how much their postural stability is threatened and take appropriate actions. Although subject perception of fall risk may not be the actual probability of falling, it gives an idea about motivations for their short-term reactions (e.g. grasping, or returning to their seat in transport) and their long-term opinion of the environment (e.g. selection of different transport types). The stability thresholds of walking subjects can be determined using the subjective measures of perceived risk of fall. Stability thresholds are useful to determine the tolerance level of walking passengers to lateral oscillations in transport but have not been previously reported for walking subjects.

Previously developed models of postural stability are useful for understanding the dynamics and control of human balance during standing and walking but they do not predict the probability of losing balance when stability is perturbed by external oscillations. For a postural stability model to be applied to train passengers, it is important to understand the point at which passengers believe that they are at risk of falling and so make necessary postural adjustments.

The current research is expected to contribute to the existing knowledge by systematically investigating the effect of lateral oscillations on the perception of fall risk and develop stability thresholds for walking people in transport. Systematic laboratory evaluations of the effects of motion characteristics (e.g. magnitude, frequency, waveform) which are typical of accelerations experienced during a train ride on the perceived risk of fall will be used to develop a subjective model of postural stability. The model is expected to estimate the effects of magnitude, frequency and waveform of lateral oscillatory motion, support and support height, and subject physical characteristics (e.g. age, gender) on the postural stability of walking people exposed to lateral oscillations. Objective measures of the motion of the centre of pressure (COP) will support the subjective model via understanding of the mechanisms of walking stability. The findings of the research are expected to improve understanding of walking stability especially when threatened by external perturbations. The outcome of the research is also expected to improve the postural stability of walking train passengers with a wide range of subject characteristics (e.g. age, gender, weight, stature) in terms of both the optimization of the motions on trains and the design of supports for passengers.

Chapter 3

Methods

3.1. Introduction

The methods used for the assessment of postural stability including the equipment, data processing techniques, and statistical analysis methods are summarized in this chapter.

3.2. Apparatus

3.2.1. Six-axis motion simulator

A 6-axis motion simulator was used to generate lateral oscillatory motions in all four experiments. The vibrator was located in the Human Factors Research Unit of the Institute of Sound and Vibration Research, University of Southampton, Southampton, United Kingdom.

The hydraulic simulator was capable of reproducing multi-axis motions including fore-and-aft, lateral and vertical translation, roll, pitch and yaw (Figure 3.1). The moving platform was approximately 3 meter by 2 meter and can support payloads up to 1000 kg. The maximum stroke is 500 mm in the fore-and-aft and lateral directions, 1000 mm in the vertical direction, and about ±10 degrees in rotational axes. The frequency range of motion is 0 to 50 Hz. The simulator was controlled by a Pulsar Digital Controller provided by Servotest Systems.

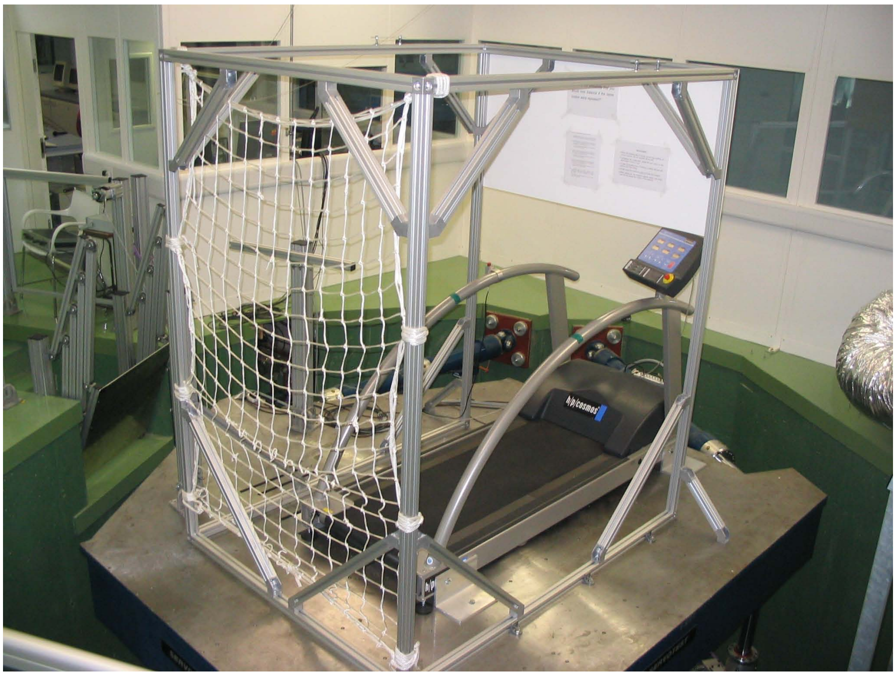

Figure 3.1: Six-axis motion simulator equipped with treadmill, safety frame and safety net.

3.2.1.1. Vibration distortion

In all experiments, the vibration signals were generated and acquired in Pulsar (version 1.4) software, provided by Servotest Testing Systems. The signals were generated and acquired at 256 samples/second. Platform acceleration in the lateral direction was recorded by accelerometers on the simulator platform (FGP model FA101-A2-5G).

Signal distortion was measured for the six-axis simulator. At each frequency of interest, sinusoidal signals were generated at typical magnitudes used in the experiments. The power spectral density of the recorded oscillations was calculated in the frequency band 0 to 128 Hz. The distortion was calculated with Equation (3.1):

$$Distortion = \sqrt{\frac{E_{outside}}{E_{inside}}} \tag{3.1}$$

where E_{outside} is the acceleration power outside an octave band centred on the frequency of the oscillation (Figure 3.2) and E_{inside} is the acceleration power inside that octave band.

An example of a 1-Hz sinusoidal acceleration waveform with distortion 6% is shown in Figure 3.2 and its power spectrum is shown in Figure 3.3.

Distortion was measured in the frequency range of 0.5 Hz to 2 Hz, at magnitudes: the lowest, mid-range and greatest magnitudes used in experiments, as shown in Table 3.1. The distortion values are reported in Table 3.2.

Table 3.1: Magnitude and frequency of lateral oscillations used to measure distortion.

Frequency (Hz)	Low magnitude (ms^{-2} r.m.s.)	Medium magnitude (ms^{-2} r.m.s.)	High magnitude (ms^{-2} r.m.s.)
0.5	0.1	0.25	0.5
0.63	0.125	0.315	0.63
0.8	0.16	0.4	0.8
1	0.2	0.5	1
1.25	0.25	0.63	1.25
1.6	0.315	0.8	1.6
2	0.4	1	2

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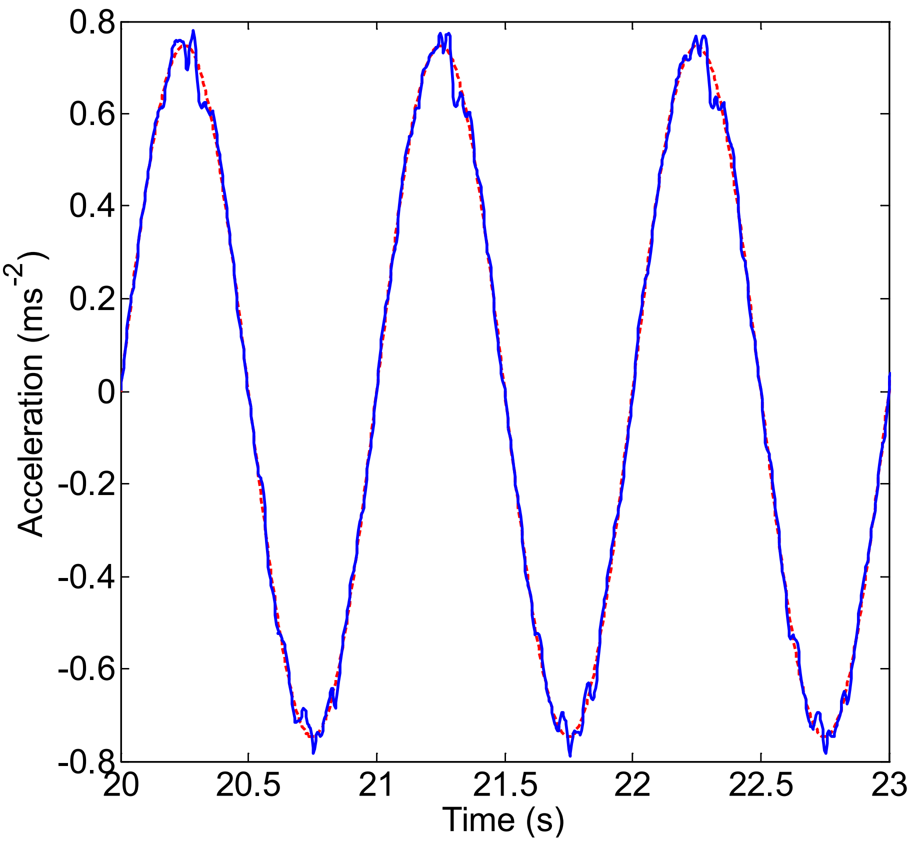


Figure 3.2: An example of lateral 1-Hz oscillation produced by the six-axis simulator (medium magnitude; see Table 3.1). The distortion is 6%: – – –desired acceleration, ——measured acceleration.

Table 3.2: Distortion (%) measured with the six-axis motion simulator.

	Distortion (%)		
Frequency (Hz)	Low magnitude	Medium magnitude	High magnitude
0.5	22.1	10.7	4.5
0.63	18.2	8.0	4.0
0.8	15.6	7.3	4.0
1	13.6	6.0	3.4
1.25	8.7	4.4	2.1
1.6	6.6	3.1	2.0
2	5.9	3.1	1.8

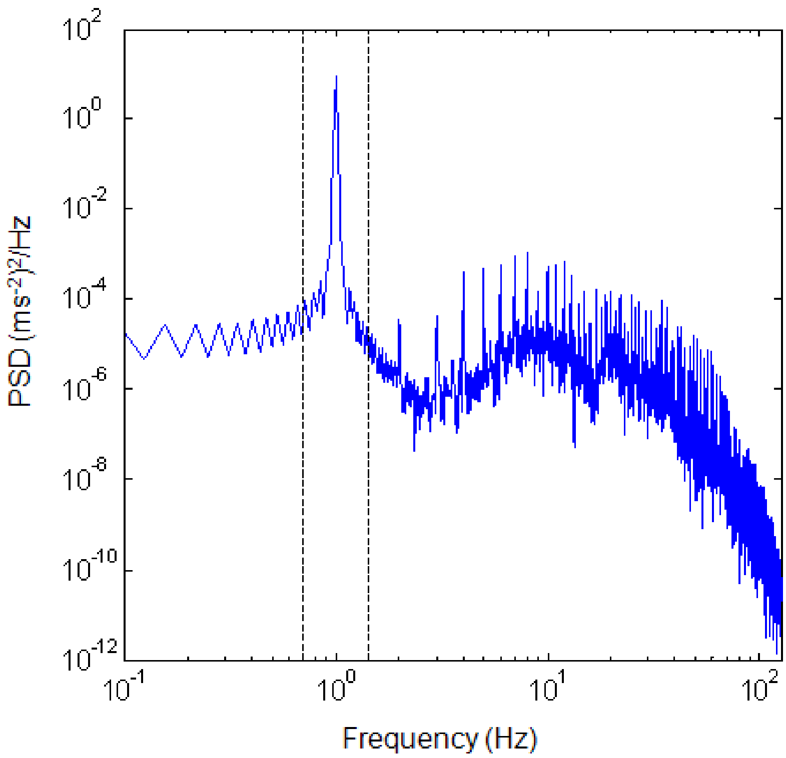


Figure 3.3: Power spectrum of the acceleration shown in Figure 3.2, and octave-band used for the calculation of distortion (6%).

The cross-axis coupling was measured at all frequencies for low, medium, and high magnitudes of oscillation. The cross-axis coupling was calculated as the percentage ratio of the r.m.s. acceleration in non-desired directions to the r.m.s. acceleration in the desired direction of vibration. The cross-axis coupling between the desired direction of vibration (i.e. lateral direction) and other translational directions and rotational axes are reported in Table 3.3.

Table 3.3: Cross-axis coupling between lateral direction of vibration and other translational and rotational axes.

0.5 Hz	Direction of measurement	Low magnitude	Medium magnitude	High magnitude
Coupling with translational axes (%)	Fore-and-aft	6.46	8.99	1.97
	Lateral	100.67	99.77	97.85
	Vertical	1.6	1.7	2.18
Coupling with rotational axes (% rad/m)	Roll	3.19	2.1	1.79
	Pitch	1.86	1.14	1.43
	Yaw	10.53	5.81	2.89
0.63 Hz	Direction of measurement	Low magnitude	Medium magnitude	High magnitude
Coupling with translational axes (%)	Fore-and-aft	5.1	3	1.69
	Lateral	97.54	100.1	98.44
	Vertical	1.83	1.58	1.6
Coupling with rotational axes (% rad/m)	Roll	1.61	1.66	1.38
	Pitch	1.64	1.05	0.94
	Yaw	9.49	4.34	2.27
0.8 Hz	Direction of measurement	Low magnitude	Medium magnitude	High magnitude
Coupling with translational axes (%)	Fore-and-aft	3.8	2.75	1.85
	Lateral	99.33	98.65	99.36
	Vertical	1.77	1.54	1.5
Coupling with rotational axes (% rad/m)	Roll	1.5	1.23	1.43
	Pitch	1.4	0.86	0.85
	Yaw	7.5	3.6	2.11
1 Hz	Direction of measurement	Low magnitude	Medium magnitude	High magnitude
Coupling with translational axes (%)	Fore-and-aft	5.01	3.27	2.11
	Lateral	100.6	99.88	100
	Vertical	1.6	1.53	1.54
Coupling with rotational axes (% rad/m)	Roll	1.4	1.4	1.37
	Pitch	1.4	0.94	0.82
	Yaw	5.41	4.3	2.08
1.25 Hz	Direction of measurement	Low magnitude	Medium magnitude	High magnitude
Coupling with translational axes (%)	Fore-and-aft	3.1	2.11	1.47
	Lateral	99.95	100.03	99.09
	Vertical	1.7	1.57	1.6
Coupling with rotational axes (% rad/m)	Roll	1.95	1.34	1.23
	Pitch	1.37	0.84	0.78
	Yaw	5.5	2.4	1.2
1.6 Hz	Direction of measurement	Low magnitude	Medium magnitude	High magnitude
Coupling with translational axes (%)	Fore-and-aft	2.37	1.8	1.65
	Lateral	99.81	99.64	99.4
	Vertical	1.64	1.52	1.46
Coupling with rotational axes (% rad/m)	Roll	1.47	1.2	1.13
	Pitch	1.08	0.78	0.7
	Yaw	4.07	1.81	1.07
2 Hz	Direction of measurement	Low magnitude	Medium magnitude	High magnitude
Coupling with translational axes (%)	Fore-and-aft	1.56	1.73	0.98
	Lateral	99.47	99.37	99.33
	Vertical	1.12	0.95	0.69
Coupling with rotational axes (% rad/m)	Roll	1.04	0.66	0.65
	Pitch	0.8	0.65	0.48
	Yaw	2.43	1.55	0.88

3.2.2. Kistler treadmill

To provide a walking task, a treadmill (Kistler Gaitway®, Figure 3.4) incorporated with eight force sensors was secured to the 6-axis motion simulator (Figure 3.1). Gaitway® is a complete gait analysis system housed in a commercially manufactured treadmill. It provides the measurement of the vertical ground reaction forces and the centre of pressure (COP) data for complete and consecutive foot strikes during walking. The instrumented treadmill system has been designed using a patented tandem force plate design and includes a patented algorithm which distinguishes left and right foot-strikes. For further technical details the reader may refer to Appendix A.1.

Figure 3.4: Kistler Gaitway® treadmill.

By the measurement of gait data, it was possible to observe stepping strategies developed in response to applied perturbations.

In all experiments, data acquisition via the treadmill software was triggered at the moment the simulator acceleration commenced. The acceleration, vertical ground reaction force, and support contact force data collected by the Gaitway® data acquisition system were sampled at 100 samples per second and stored in a personal computer.

3.2.2.1. Data analysis

There were eight force sensors embedded inside the treadmill to measure the vertical ground reaction force applied by the walking subject. The acquired raw force data (from 8 force sensors, Figure 3.5) was processed to obtain centre of pressure (COP) position. Figure 3.6 shows how the force sensors are arranged inside the treadmill.

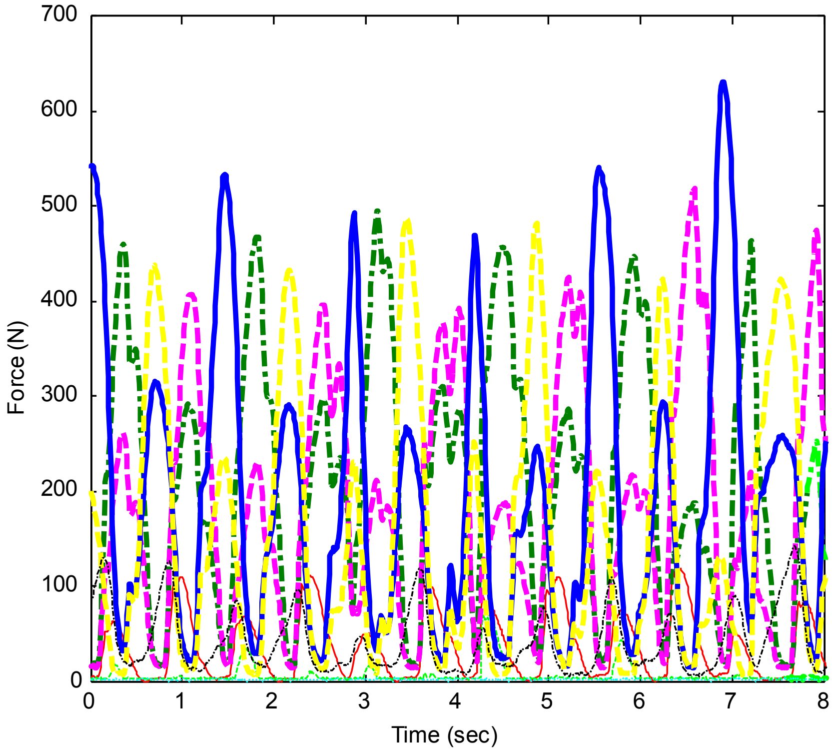


Figure 3.5: Raw force time histories from eight force sensors embedded inside the treadmill (See Figure 3.6). Sensor 1, Sensor 2, Sensor 3, Sensor 4, Sensor 5, Sensor 6, Sensor 7, Sensor 8.

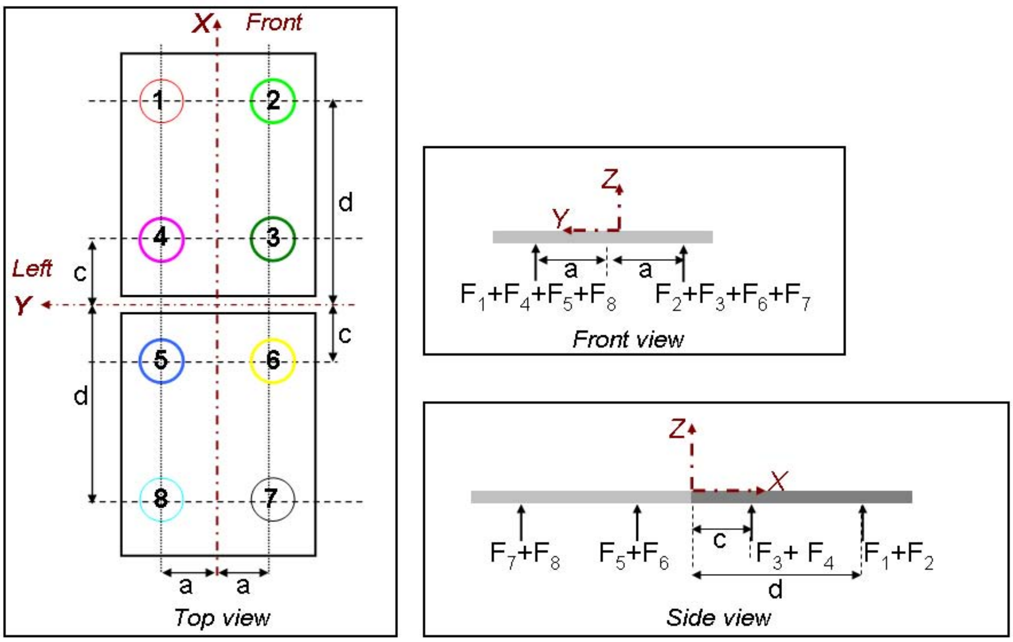


Figure 3.6: Arrangement of force sensors embedded inside the treadmill.

The centre of pressure position (COP) in the lateral direction was obtained by moment equilibrium of the vertical ground reaction forces with respect to longitudinal axis of the treadmill (Figure 3.6).

$$COP_y = \frac{a(F_1 + F_4 + F_5 + F_8) \quad a(F_2 + F_3 + F_6 + F_7)}{F_1 + F_2 + F_3 + F_4 + F_5 + F_6 + F_7 + F_8} \tag{3.2}$$

The COP velocity in the lateral direction was obtained by differentiating the lateral COP position after filtering the centre of pressure position using low-pass Bessel filter at 8 Hz.

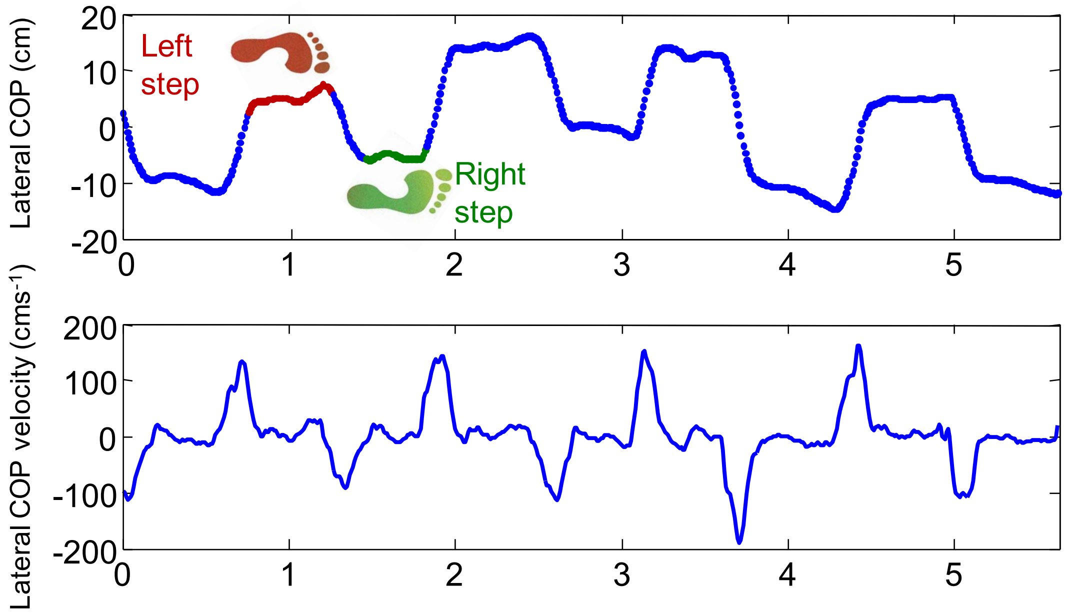


Figure 3.7: COP measurements from experiment data of one subject (a) COP position in the lateral direction (b) COP velocity in the lateral direction.

Figure 3.7 shows the lateral COP position and COP velocity of a subject. The mean is subtracted from the COP position. COP position shows the lateral point location of the resultant of the ground reaction forces and is an indication of lateral foot placement. COP velocity is the rate of change of COP position.

3.2.3. Other transducers

In the second experiment regarding the effect of support on postural stability, contact forces applied to the hand support in the lateral direction were obtained from two single-axis load cells (Tedea Huntleigh Model 1022 1022-20M-C3), attached at the top and bottom end of the vertical handle (Figure 3.8).

The raw force readings from two load cells were amplified (Yokogawa Strain Guage Amplifier, Model 3126) and filtered using a low-pass filter at 10 Hz. The total force applied by the subject on the vertical handle was calculated by summing the force readings from two separate load cells. Detailed technical specifications of the load cells are provided in the Appendix A.2.

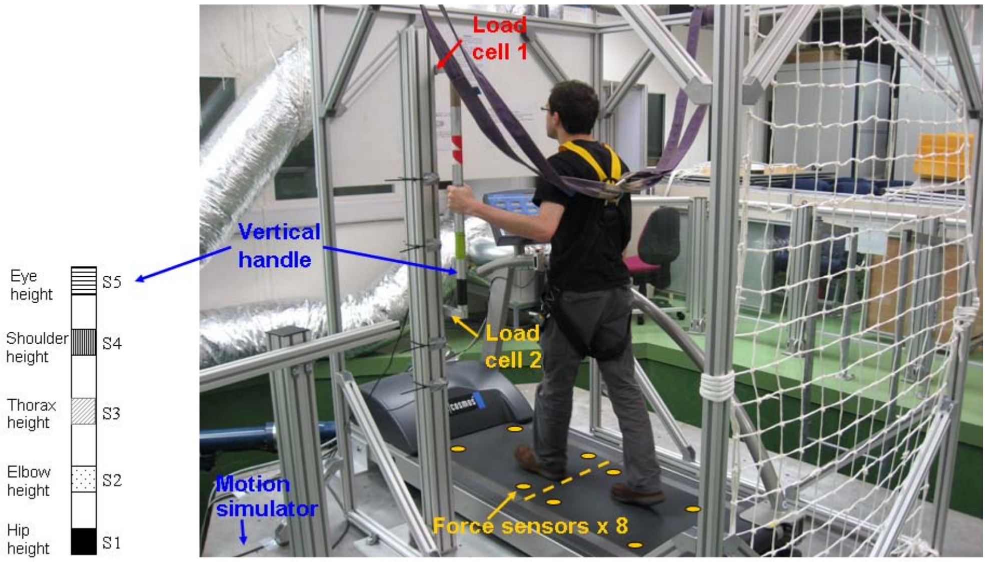


Figure 3.8: A walking subject holding from the hand support while exposed to lateral oscillation.

3.3. Test conditions

3.3.1. Vibration

In all experiments, subjects were exposed to vibration. All experiments were approved by the Human Experimentation Safety and Ethics Committee of the Institute of Sound and Vibration Research, University of Southampton. All subjects were volunteers and could quit the experiment at any time without providing a reason. In all experiments, subjects were provided with instruction sheets which are included in the Appendix B.

3.3.2. Safety frame

The safety frame mounted on the six-axis simulator had dimensions 1900 mm x 1460 mm x 2100 mm (Figure 3.8). Walking subjects were asked to wear a safety harness which was attached to the safety frame via two straps (Figure 3.8). The harness allowed the subjects to move freely in the plane of progression but prevented their knees from contacting the floor if they fell. A safety net was positioned behind the subjects as a precaution in case they slid backwards while walking on the treadmill (Figure 3.8).

3.3.3. Visual field

Walking subjects were asked to fix their vision on the white board in front of them while walking (Figure 3.8). The white board was 1460 mm wide and 750 mm in length and 1150 mm above the treadmill surface. The white board served as a closed visual field to hide visual cues from subjects regarding the lateral movement of the six-axis platform.

3.3.4. Emergency stops

In case subjects felt unsafe or wanted to stop the experiment for some reason, they were supplied with an emergency stop button to automatically stop the motion of the 6-axis simulator. Subjects could also stop the running belt of the treadmill by pressing the red STOP key in the center of the console (Figure 3.9).

Figure 3.9: Control panel of the treadmill.

3.3.5. Acoustic conditions

When the six-axis simulator was running, it produced acoustical noise. The noise level at the location of the subject was less than 51 dB (A).

3.4. Assessment of postural stability

3.4.1. Subjective measure

The discomfort caused by whole-body vibration has been traditionally assessed by subjective methods to obtain discomfort ratings in seated or standing subjects (e.g. Morioka and Griffin, 2006; Thuong and Griffin, 2011). The method of magnitude estimation has been commonly used to obtain discomfort ratings (Morioka and Griffin, 2006; Wyllie and Griffin, 2007). Stevens' power law (Stevens, 1975) was used to relate the magnitude estimates of subject discomfort, ψ, to the physical magnitudes of the motions, ϕ:

$$\Psi = k * (\Phi)^n \tag{3.3}$$

where k (the 'constant' in Stevens' power law) and n (the 'exponent') are assumed to be constant at any frequency. With whole-body vibration of seated persons the exponent depends on the frequency of vibration (Morioka and Griffin, 2006).

The postural stability of stationary standing subjects has been investigated previously using a subjective method in which subjects reported their perceived probability of losing balance (Nawayseh and Griffin, 2006). The postural stability of walking subjects has not been studied systematically using subjective measures.

In the first and fourth experiment, the reported probability of losing balance was used to assess the perceived risk of fall by walking subjects exposed to lateral oscillations. In the second experiment, 'discomfort or difficulty in walking task' was used to assess the effect of hand support in postural stability. In the third experiment, 'discomfort and difficult in walking task' was used together with the 'reported probability of losing balance' to investigate the relative effect of r.m.s. and peak levels of oscillations on discomfort in walking caused by the lateral oscillation.

3.4.2. Objective measure

The postural stability of walking subjects was assessed by the centre of pressure (COP) measurements. The COP was a useful indicator of stepping strategy which is the main strategy to maintain postural stability while walking (Nashner, 1980).

Lateral r.m.s. COP velocity was used as a common measure of stability in all experiments. It is used as an indication of timing and placement of foot placement in the lateral direction.

The peak-to-peak lateral COP position was used as an indication of the range of lateral COP movement.

The mean COP speed and r.m.s. force were used in the fourth experiment while investigating the effects of subject characteristics on the walking stability. Total r.m.s. vertical ground reaction force under the feet was normalized with respect to the weight of each subject and is an indication of loading-unloading strategies employed by the subject. The mean COP speed is defined as the cumulative distance of the COP over the sampling period indicating the amount of physical activity required to maintain stability during quiet standing (Geurts *et al.*, 1993; Hue *et al*, 2007). It is an indication of the walking path taken by the walking subjects (Figure 3.10).

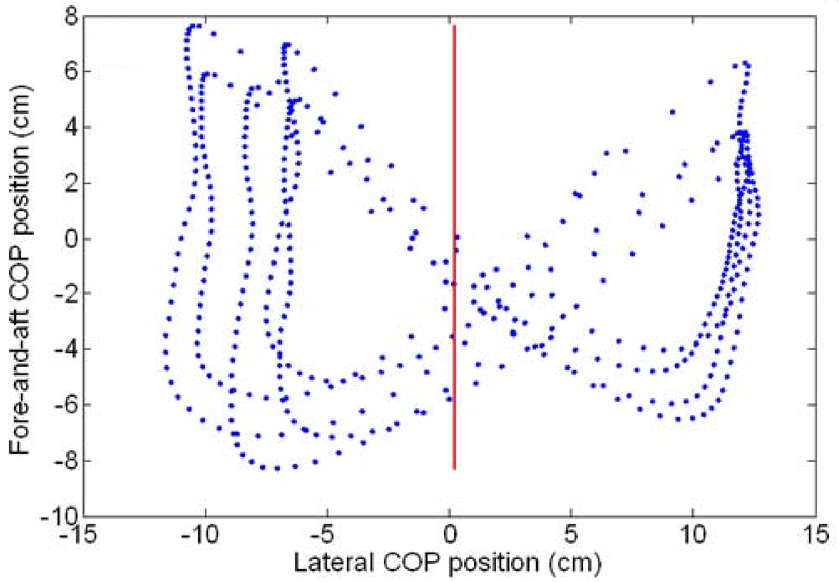


Figure 3.10: Centre of pressure path in the fore-and-aft and lateral direction.

3.5. Statistical methods

SPSS (version 17) was used for statistical analysis. Non-parametric statistical methods (Table 3.4) were used for the data analysis of the results of the first three experiments. The Friedman analysis of variance was used to test for differences between multiple conditions and the Wilcoxon matched-pairs signed ranks test was used to investigate differences between pairs of conditions. Associations between variables were investigated using Spearman's rank correlation.

Parametric statistical methods (Table 3.4) were used for the data analysis of the last experiment which was conducted on 100 subjects. Multiple regression was considered the most appropriate statistical analysis to model the relation between multiple independent variables (i.e. age, weight, height, stature, shoes width) and the subjective and objective measures of walking stability.

The statistical results have not been corrected for multiple comparisons. Although there is a possibility of accumulating Type 1 errors under multiple test conditions, the trends in multiple tests were consistent with each other and were also consistent with the theory and previous work. Because of this consistency, conservative corrections for multiple comparisons have been avoided as they may hide a significant effect when there actually is one.

Table 3.4: Statistical tests used in the analysis of experiment results.

NON-PARAMETRIC STATISTICAL ANALYSIS	
Case	Statistical test used
2 related samples	Wilcoxon signed ranks test
k related samples	Friedman two-way analysis of variance
2 independent samples	Wilcoxon-Mann-Whitney test
k independent samples	Kruskal-Wallis one-way analysis of variance
Correlation between two variables	Spearman rank-order correlation coefficient
2 related samples, binary variable	McNemar change test
k related samples, binary variable	Cochran Q test
PARAMETRIC STATISTICAL ANALYSIS	
Case	Statistical test used
2 related samples	Paired sample t-test
k related samples	Repeated measures ANOVA
2 independent samples	Independent sample t-test
k independent samples	One way ANOVA
Correlation between two variables	Pearson correlation
Relationship between several independent variables and a continuous dependent variable	Multiple regression
Relationship between several independent variables and a binary dependent variable	Logistic regression

Chapter 4

Effect of magnitude and frequency of lateral oscillations on the postural stability of walking people

4.1. Introduction

Standing and walking require continuous postural control to counteract the destabilizing effects of gravity and self-induced movements of the body. Maintaining balance is more challenging when there are external disturbances from motion of the floor, such as when standing or walking in a moving train, bus, aircraft or ship.

In previous studies, walking subjects have been perturbed by sudden accelerations or decelerations of a running belt on a treadmill (Berger *et al*., 1984) or by moveable platforms embedded in a walkway (Nashner, 1980; Oddson *et al*., 2004; Bhatt *et al*., 2005). The perturbations have been impulsive inputs representing slips, trips, or missteps encountered during walking. Longer duration low frequency oscillations (0.2 to 0.5 Hz) were introduced to healthy walking adults via oscillating treadmill embedded on a six-axis motion platform (Brady *et al*., 2009; McAndrew *et al*., 2010). These oscillations were used to investigate dynamic postural responses to perturbations but do not represent typical motions encountered in transport.

How narrow-band random fore-and-aft and lateral oscillations (at frequencies between 0.125 and 2.0 Hz with velocities from 0.04 to 0.16 ms^{-1} r.m.s.) affect the postural stability of standing subjects has been investigated by Nawayseh and Griffin (2006). They found that the displacement of the centre of pressure (COP) and subject estimates of the probability of losing balance increase with increasing magnitude of oscillation and that, with the same velocity at all frequencies, stability problems are greatest around 0.5 Hz. There have been no systematic studies of how the stability of walking persons depends on the magnitude and frequency of oscillations.

With instantaneous increases in horizontal acceleration, standing subjects have been reported to tolerate accelerations up to 0.76 ms^{-2} in the backward direction, 0.48 ms^{-2} in the forward direction, and 0.33 ms^{-2} in a sideways direction (Jongkees and Groen, 1942). Similar thresholds were obtained by Graaf and Weperen (1997), who found that standing subjects were most sensitive to lateral acceleration when standing with their feet almost together. Tolerances of walking subjects to sideward oscillations in transport have not been previously reported.

Understanding of the physiological and biomechanical aspects of balance has been used to develop active models of postural stability when standing (e.g. Mergner *et al*., 2006; Peterka 2003). These models represent the neural, sensory, and biomechanical subsystems involved in human postural control but do not allow the prediction of the probability of falling. People may be expected to be more stable when standing and supported on two legs than when walking and supported on only one leg for 80% of the gait cycle (Woollacott and Tang, 1997), especially when threatened by external perturbations. However, there have been few experimental studies and there are few models of perturbed balance during locomotion, possibly because of difficulty in applying controlled motion stimuli and the complexity of modeling body dynamics during locomotion.

The main strategy used to maintain balance during locomotion is the stepping strategy (Nashner, 1980; Horak and Nashner,1986; Hof *et al*., 2007). Additional strategies (e.g. active hip torque and active ankle subtalar torque) are used for fine tuning (Hof *et al*., 2007; MacKinnon and Winter, 1993) when the foot position is established. Adjusting the step width by varying the foot placement is considered an important strategy for maintaining postural stability in the frontal (i.e. coronal) plane. The step width is used to regulate the trajectory of the centre of mass (COM) so as to maintain balance in the frontal plane (Townsend, 1985) and is considered to have a greater influence on postural control during unperturbed walking than either step length or step time (Owings and Grabiner, 2004). It has also been suggested that step width is adjusted to compensate for lateral acceleration induced by external perturbation (Oddson *et al*., 2004).

The overall aim of the experimental study reported in this chapter was to determine the effects of the magnitude and frequency of lateral oscillation on the postural stability of walking subjects. It was hypothesised that, at each frequency of oscillation, the self-reported probability of losing balance and the movement of the centre of pressure in the lateral direction would increase with increasing magnitude of oscillation. It was expected that the subjective measures of postural stability and some characteristic of the movement of the centre of pressure would have a similar dependence on the frequency of oscillation.

4.2. Method

4.2.1. Subjects

Twenty healthy male subjects with median age 27 years (range 25 to 41), stature 177 cm (range 165 to 192), weight 72.3 kg (48.5 kg to 88.45) participated in the study. Subjects completed a questionnaire to exclude those with relevant disorders or using drugs that might affect postural stability. 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.

4.2.2. Apparatus

A treadmill (Kistler Gaitway®) incorporating eight force sensors was used to provide the walking task and measure the vertical ground reaction forces during walking. Subjects were secured by a safety harness connected via two loose straps to a frame around the treadmill (Figure 4.1). The harness allowed subjects to move freely in the plane of progression but prevented their knees from contacting the floor if they fell. A safety net was positioned behind the subjects as a precaution in case they slid backwards while walking on the treadmill.

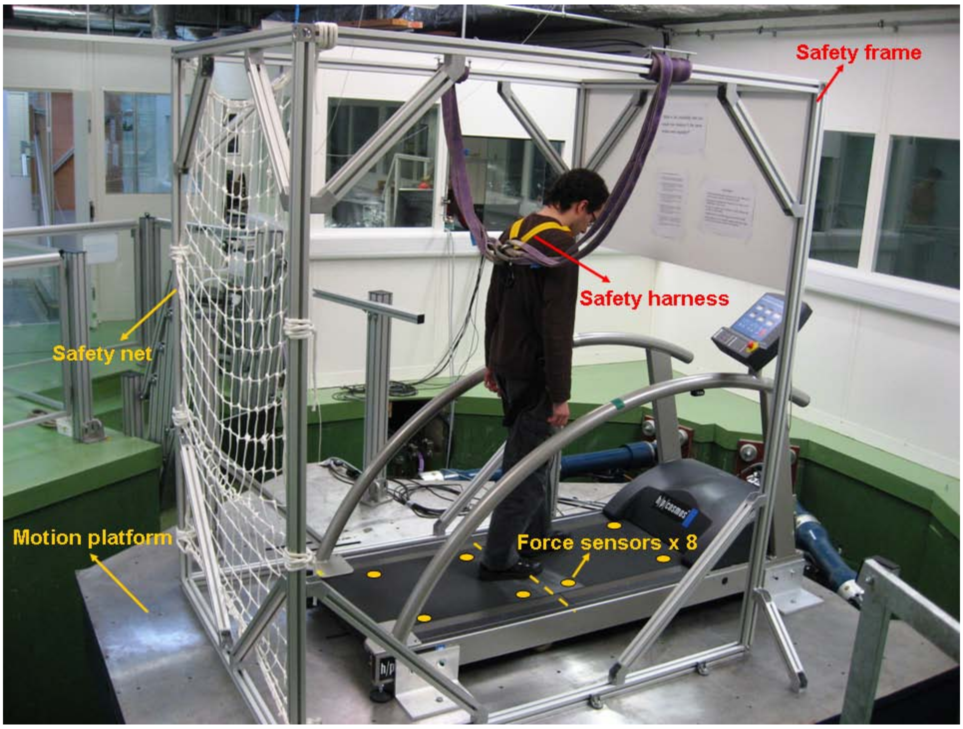


Figure 4.1: Experimental apparatus used in the first experiment.

Lateral oscillatory motion was generated by a six-axis motion simulator in the Human Factors Research Unit at the Institute of Sound and Vibration Research. The simulator is able to provide translational displacements of ±0.25 m in the lateral direction at accelerations up to about ±10 ms^{-2}.

Acceleration in the lateral direction was recorded by accelerometers on the simulator platform (FGP model FA101-A2-5G). Data acquisition via the treadmill software was triggered at the moment the 4½-cycle acceleration commenced. The acceleration and force data collected by the Gaitway® data acquisition system were sampled at 100 samples per second and stored in a personal computer.

4.2.3. Experimental Procedure

While walking on the treadmill, subjects were perturbed by simple transient lateral acceleration stimuli applied at an unpredictable time. The stimuli were 4.5 cycles of sinusoidal motion modulated by a half sine envelope. For these waveforms, the peak acceleration and the peak velocity are, respectively, double the r.m.s. acceleration and r.m.s. velocity. The motions start and end with zero displacement, velocity and acceleration and were chosen as being broadly representative of lateral motions experienced in trains (Figure 4.2a).

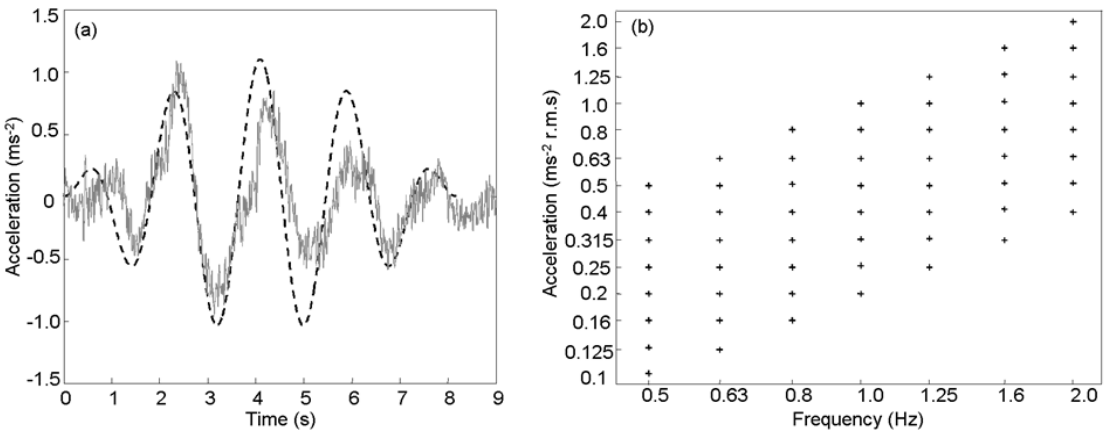


Figure 4.2: A transient lateral acceleration measured on a train compared with a 0.5 Hz 0.5 ms^{-2} r.m.s. 4½-cycle motion stimulus: ——— measured on a train; ----- theoretically generated stimulus. (b) Magnitudes and frequencies investigated in the experiment.

At each of seven frequencies (0.5, 0.63, 0.8, 1.0, 1.25, 1.6, 2.0 Hz), the motions were presented at eight velocities (0.032, 0.04, 0.05, 0.062, 0.08, 0.1, 0.125, 0.16 ms^{-1} r.m.s.). This resulted in accelerations in the range 0.1 to 2.0 ms^{-2} r.m.s. (Figure 4.2b). The frequencies and magnitudes were chosen after preliminary experimentation and so that the effects of stimuli with the same magnitude of acceleration or stimuli with the same magnitude of velocity could be compared across the frequency range. The 56 motions were presented in a random order.

The speed of the treadmill was selected so that subjects walked at 0.7 ms^{-1} throughout the experiment. This was the preferred comfortable walking speed of subjects who participated in preliminary experiments.

The eight channels of force data were acquired throughout each of the 4½-cycle perturbations. After experiencing each motion, subjects were asked to judge their postural stability by answering the following question:

“What is the probability that you would lose balance if the same exposure were repeated?”

Subjects were instructed to grasp the handrails of the treadmill only if it was really necessary. Losing balance was defined as attempting to take protective action not to fall – such as taking a protective step, or grasping an object to regain equilibrium.

4.2.4. Analysis

The raw force time-histories (from 8 force sensors) were processed to determine the centre of pressure (COP) during each motion. The COP in the lateral direction was obtained by moment equilibrium of the vertical ground reaction forces gathered via eight force sensors embedded inside the treadmill with respect to longitudinal axis of the treadmill (Section 3.2.2.1). The COP velocity in the lateral direction was obtained by differentiating the lateral COP position after filtering the centre of pressure position using low-pass Bessel filter at 8 Hz.

An example of the COP position and COP velocity of a subject exposed to 0.8 Hz lateral oscillation at 0.5 ms^{-2} r.m.s. is shown in Figure 4.3. The mean has been subtracted from the COP position, which shows the lateral (y-axis) location of the resultant of the ground reaction forces and is indicative of lateral foot placement. The COP velocity indicates the rate of change of COP position (Figure 4.3b).

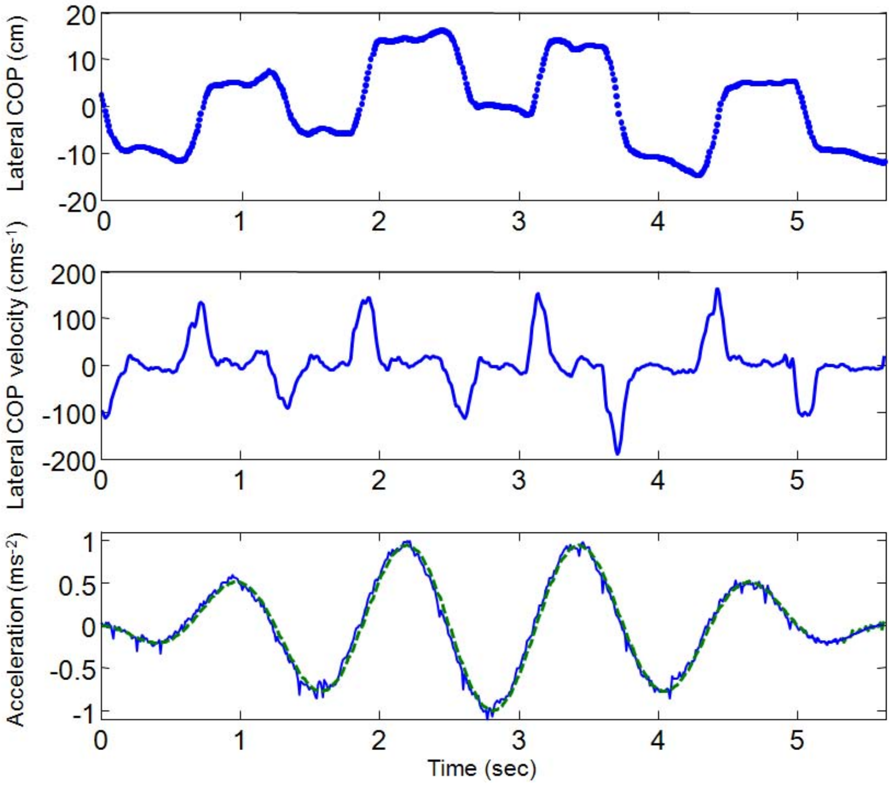


Figure 4.3: Example centre of pressure (COP) and acceleration time histories for a subject walking while exposed to 0.5 ms^{-2} r.m.s. lateral oscillation at 0.8 Hz: (a) COP position in the lateral direction; (b) COP velocity in the lateral direction; (c) lateral acceleration: measured acceleration, desired acceleration.

Non-parametric statistical methods were used for the data analysis using SPSS (version 17). The Friedman analysis of variance was used to test for differences between multiple conditions and the Wilcoxon matched-pairs signed ranks test was used to investigate differences between pairs of conditions. Associations between variables were investigated using Spearman's rank correlation.

4.3. Results

4.3.1. Subjective Data

At a specific frequency, the median reported probability of losing balance increased as the acceleration or velocity magnitude of the lateral motion increased (*p*<0.01 at all seven frequencies; Spearman; Figure 4.4). The increase in the perceived risk of fall with increasing velocity magnitude was broadly similar at all frequencies (Figure 4.4b).

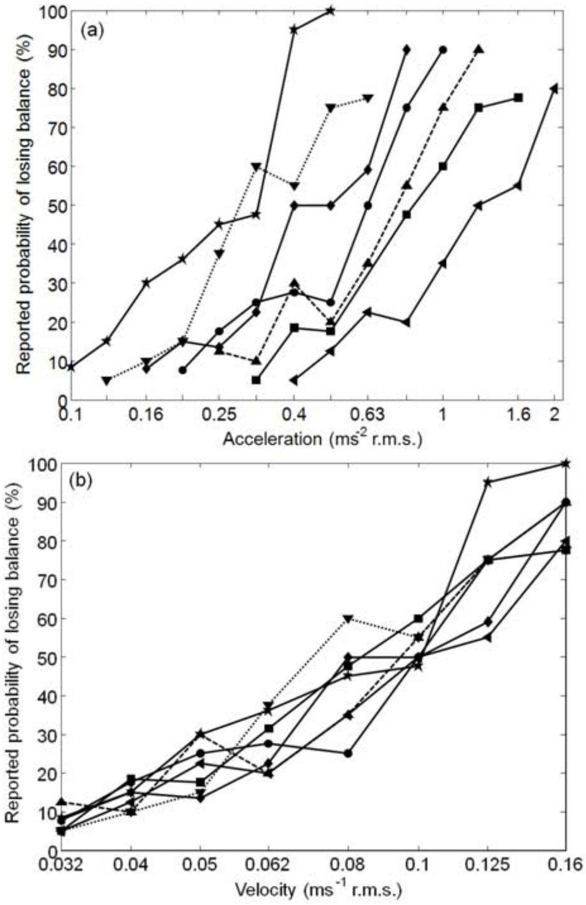


Figure 4.4: (a) Effect of acceleration magnitude on the median reported probability of losing balance at a specific frequency of oscillation (b) Effect of velocity magnitude on the median reported probability of losing balance at each frequency of oscillation: —★—0.5 Hz, —▼— 0.63 Hz, —◆—0.8 Hz, —●—1.0 Hz, --▲-- 1.25 Hz, —■—1.6 Hz, —◀—2.0 Hz.

The effect of the frequency of oscillation on the self-reported probability of losing balance at each magnitude of acceleration is shown in Figure 4.5a. At each acceleration magnitude, the perceived risk of fall decreased as the frequency increased (p<0.01 at 0.125, 0.16, 0.315, 0.4, 0.5, 0.63, 0.8, 1.0, 1.25, and 1.6 ms^{-2} r.m.s.; p<0.05 at 0.2 and 0.25 ms^{-2} r.m.s.; Spearman).

The effect of the frequency of oscillation on the self-reported probability of losing balance at each magnitude of velocity is shown in Figure 4.5b. It can be seen that the median reported probability of losing balance was similar when the velocity was kept constant: there was a significant effect of frequency at only two magnitudes (p<0.05 at 0.08 ms^{-1} r.m.s.; p<0.01 at 0.13 ms^{-1} r.m.s.; Friedman). Within the frequency range 0.63 to 1.6 Hz there was no significant effect of frequency on the probability of losing balance at any magnitude of velocity (p>0.06, Friedman).

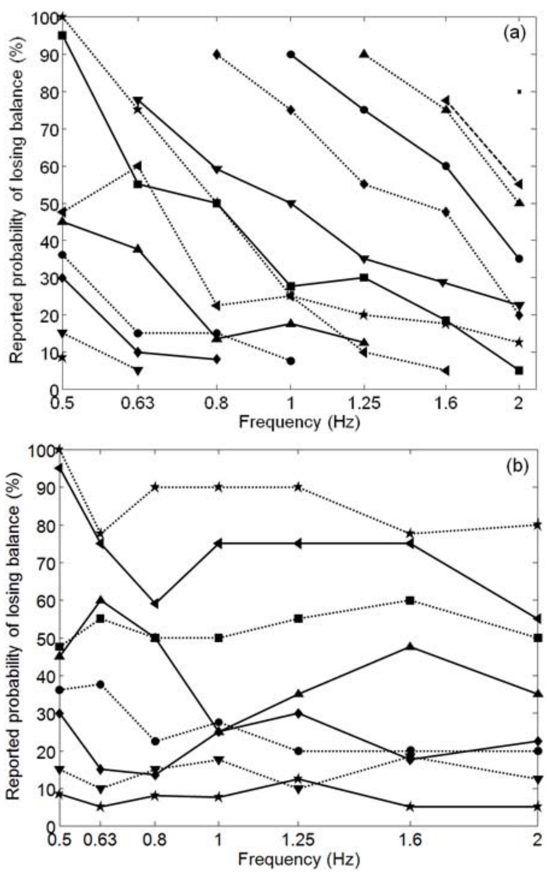


Figure 4.5: Effect of frequency on the median reported probability of losing balance (a) at each magnitude of motion acceleration: 0.1, 0.125, 0.16, 0.2, 0.25, 0.315, 0.4, 0.5, 0.63, 0.8, 1.0, 1.25, 1.6, x 2.0 ms^{-2} r.m.s. (b) at each magnitude of motion velocity: 0.032, 0.04, 0.05, 0.062, 0.08, 0.1, 0.13, 0.16 ms^{-1} r.m.s..

The relation between the number of subjects, N, estimating their probability of losing balance to be 50% or greater was counted and was related to the r.m.s. acceleration, a, at each frequency using linear regression:

$$N = c_1 a + c_2$$

The regression constants, c_1 and c_2, and the correlation coefficients, R^2, are shown in Table 4.1. The accelerations required at each frequency for 25%, 50% and 100% of 20 subjects to report their probability of losing balance to be 50% or greater are shown in Figure 4.6.

Table 4.1: Regression constants and correlation coefficients for the relation between the number of subjects (out of 20) who reported their probability of losing balance to be 50% or greater and the acceleration magnitude at each frequency.

Frequency (Hz)	c_1	c_2	R^2
0.5	38.498	-0.740	0.907
0.63	27.207	-3.945	0.872
0.8	32.495	-1.605	0.855
1	24.585	-4.459	0.936
1.25	19.248	-4.254	0.926
1.6	13.779	-2.062	0.852
2	10.908	-4.403	0.952

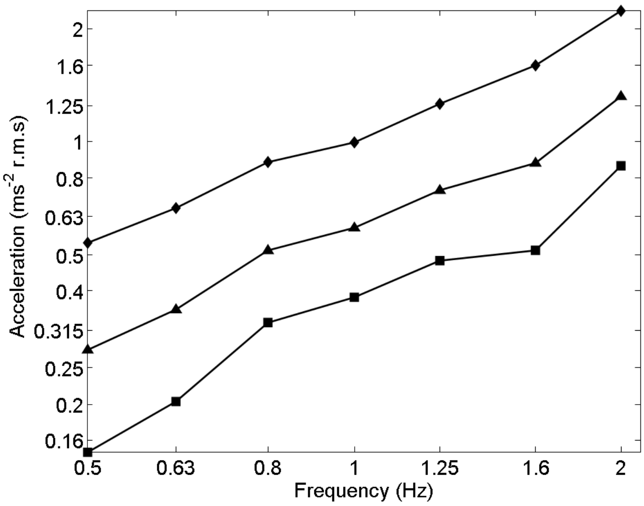


Figure 4.6: Acceleration required at each frequency for 25%, 50% and 100% of 20 subjects to report their probability of losing balance to be 50% or greater: —■— 25% (N=5 subjects), —▲— 50% (N=10 subjects), —◆— 100% (N=20 subjects).

4.3.2. Objective Data

Peak-to-peak lateral COP position and lateral r.m.s. COP velocity were used as objective measures of postural stability. Peak-to-peak COP position is an indication of the range of lateral foot placement and r.m.s. COP velocity is an indication of timing of stepping action.

With each frequency of lateral acceleration, peak-to-peak lateral COP position increased as the magnitude of the motion increased at all frequencies (p<0.01, Spearman; Figure 4.7) except at 0.63 Hz (p=0.091, Spearman). Changes in peak-to-peak COP position was positively correlated with changes in reported probability of losing balance at each frequency of oscillation (p<0.01, Spearman) except at 0.63 Hz (p=0.1, Spearman).

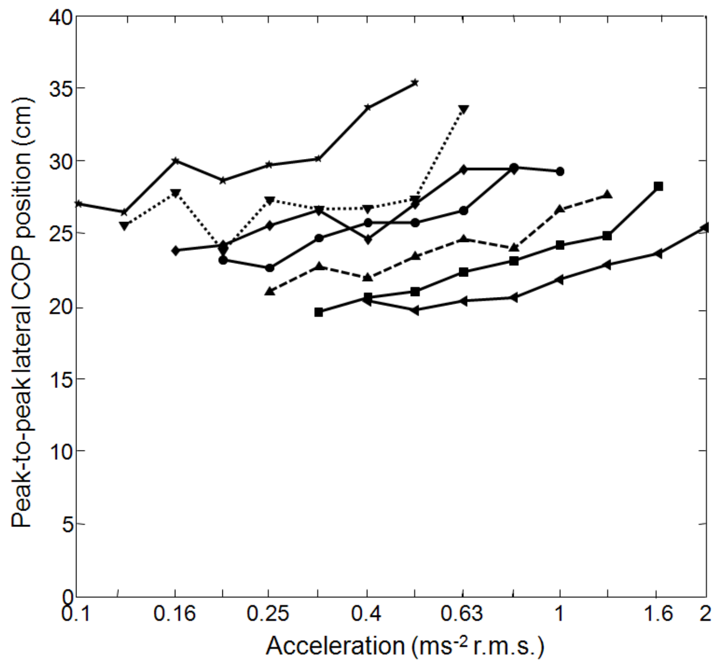


Figure 4.7: Effect of acceleration magnitude on the median peak-to-peak lateral COP position at each frequency of oscillation: ★ 0.5 Hz, ▼ 0.63 Hz, ◆ 0.8 Hz, ● 1.0 Hz, ▲ 1.25 Hz, ■ 1.6 Hz, ◀ 2.0 Hz.

At each acceleration magnitude, peak-to-peak lateral COP position decreased as the frequency increased (p<0.01, Spearman; Figure 4.8) except at 0.2 ms^{-2} r.m.s (p=0.1, Spearman). Peak-to-peak lateral COP position was correlated with the reported probability of losing balance at each acceleration magnitude (p<0.025, Spearman).

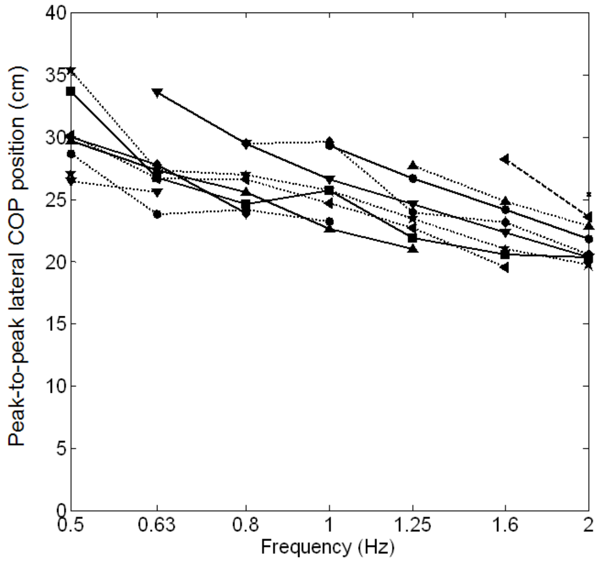


Figure 4.8: Effect of frequency on the median peak-to-peak lateral COP position at each magnitude of motion acceleration: ★ 0.1, —▼— 0.125, —◆— 0.16, —●— 0.2, —▲— 0.25, —◀— 0.315, —■— 0.4, —★— 0.5, —▼— 0.63, —◆— 0.8, —●— 1.0, —▲—1.25, --◀-- 1.6, x 2.0 ms^{-2} r.m.s.

When the motion was applied at the same velocity, peak-to-peak lateral COP position was decreasing with increasing frequency (p<0.05, Spearman) although the reported probability of losing balance was similar (Figure 4.5b). However, r.m.s. COP velocity showed the similar trend with the subjective ratings of postural stability: r.m.s. COP velocity was not correlated with frequency at any magnitudes of lateral velocity (p>0.5, Spearman; Figure 4.9) except the positive correlation with frequency at 0.08 ms^{-1} r.m.s. (p=0.014, Spearman).

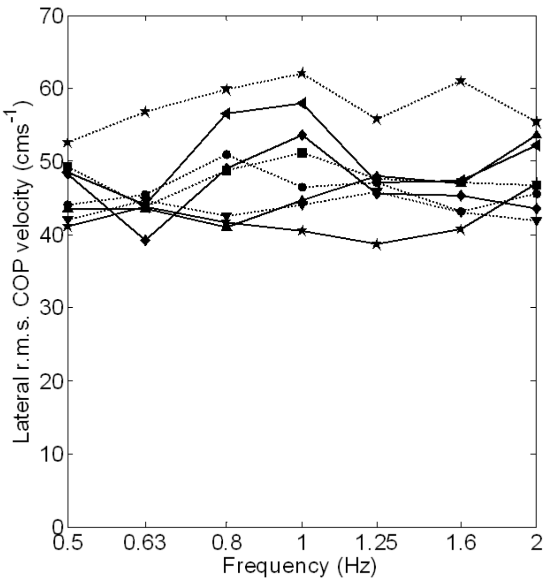


Figure 4.9: Effect of frequency on the median lateral r.m.s. COP velocity at each magnitude of motion velocity: —★— 0.032, —▼— 0.04, —◆— 0.05, —●— 0.062, —▲— 0.08, —■— 0.1, —◀— 0.13, —★— 0.16 ms^{-1} r.m.s.

Lateral r.m.s. COP velocity increased with increasing lateral velocity at all frequencies ($p<0.05$, Spearman; Figure 4.10) except at 0.63 Hz. The increase in lateral r.m.s. COP velocity with increasing velocity magnitude was broadly similar at all frequencies similar to the trend observed in the reported probability of losing balance with increasing velocity of oscillation (Figure 4.4b).

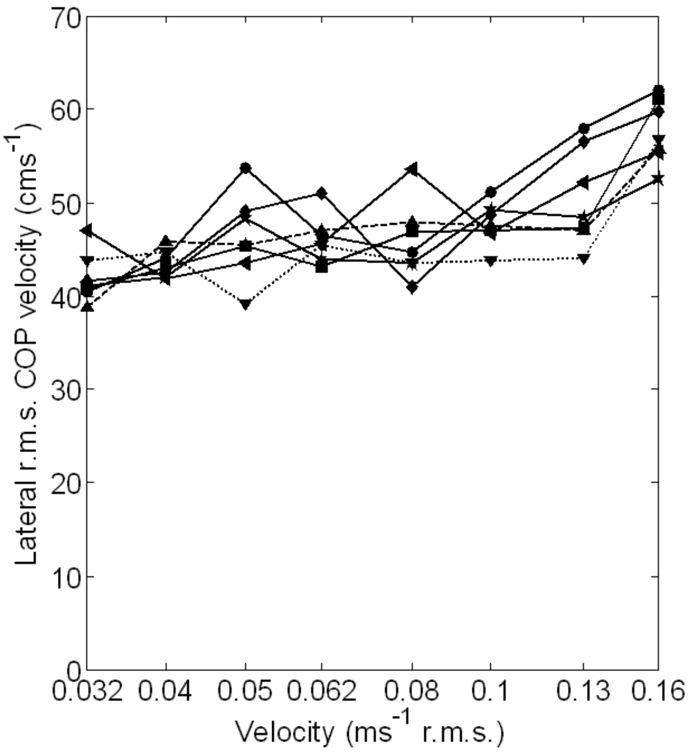


Figure 4.10: Effect of velocity magnitude on the median lateral r.m.s. COP velocity at each frequency of oscillation: ★0.5 Hz, ▼ 0.63 Hz, ◆0.8 Hz, ●1.0 Hz, --▲-- 1.25 Hz, ■1.6 Hz, ◀2.0 Hz.

The median of the lateral peak-to-peak COP position and r.m.s. COP velocity during unperturbed walking were 18.42 cm and 35.79 cms^{-1}, respectively. Peak-to-peak lateral COP position during unperturbed walking was significantly less than when walking and perturbed by lateral oscillation at any frequency and magnitude ($p<0.05$, Wilcoxon), except two conditions with low magnitudes of lateral oscillation at high frequencies (0.032 ms^{-1} r.m.s. with 1.6 Hz oscillation, and 0.04 ms^{-1} r.m.s. with 2-Hz oscillation; $p>0.05$, Wilcoxon). Lateral r.m.s. COP velocity during unperturbed walking was significantly less than during perturbed walking at any frequency and magnitude ($p<0.05$, Wilcoxon), except two conditions with low magnitudes of lateral oscillation (0.032 ms^{-1} r.m.s. with 1.6 Hz oscillation, and 0.05 ms^{-1} r.m.s. with 0.63 Hz oscillation; $p>0.05$, Wilcoxon).

4.4. Discussion

With all frequencies of lateral oscillation, as the magnitude of the perturbation increased the perceived risk of fall increased (Figure 4.4). Peak-to-peak lateral COP position and r.m.s. lateral COP velocity increased during perturbed walking and with increased magnitude of oscillations (Figure 4.7 and Figure 4.10). Increased peak-to-peak COP position is an indication of increased step width which shows the increased effort to maintain stability by compensating the lateral oscillations by means of wider steps. Step width is adjusted to compensate for lateral acceleration induced by external perturbations (Oddson *et al*., 2004; Brady *et al*., 2009). Lateral r.m.s. COP velocity increasing with increasing magnitude of oscillations shows that walking people compensate the lateral oscillations by faster stepping actions. McAndrew *et al*. (2010) also showed that walking people took wider and faster steps during continuous random oscillations than during normal walking without oscillations.

Nawayseh and Griffin (2006) showed that with the same acceleration at all frequencies, the stability of stationary standing people perturbed by lateral one-third octave band random oscillations is less affected by higher frequencies (e.g. 0.5 to 2 Hz) than by lower frequencies (e.g. 0.125 to 0.5 Hz). Over the same frequency range, the current study also found less postural instability (reduced probability of losing balance and reduced peak COP displacement) as the frequency of oscillation increased with constant magnitude acceleration (Figure 4.5a and Figure 4.8). During lateral oscillation of stationary standing people perturbed by oscillations with the same velocity, the displacements of the centre of pressure and subjective estimates of the probability of losing balance were greatest around 0.5 Hz (Nawayseh and Griffin, 2006). In the present study with walking subjects, irrespective of the frequency of oscillation, when the lateral oscillation was applied at the same r.m.s. velocity, the probability of losing balance was broadly similar (Figure 4.5b). Peak COP displacement (i.e. peak-to-peak COP position) was decreasing with increasing frequency but r.m.s. COP velocity was broadly similar at the same r.m.s. velocity (Figure 4.9). Lateral r.m.s. COP velocity may be an indication of the effort to respond to velocity of external perturbation by adjusting the timing of foot placement.

Postural control strategies adjust the centre of pressure in response to movement of the centre of mass (Murray et al 1967; Prieto *et al*. 1993). With the same acceleration at all frequencies, there are greater velocities and greater displacements with lower frequency oscillations, and subjects may have difficulty adjusting their centres of pressure in response to the larger and faster displacements of their centres of mass. Walking people are sensitive to changes in sideward velocity and take corrective actions by stepping (Hof, 2008; Hof *et al*. 2010). Current study also showed that walking people respond to sideward velocity changes by adjusting their lateral COP velocity.

Stability thresholds have not previously been reported for walking subjects. Arbitrarily, the findings of this study have been used to calculate the magnitude of lateral oscillation required at each frequency for 50% of subjects to report at least 50% probability of losing balance (Figure 4.6). Subjects standing with their eyes closed and their feet together have been reported to tolerate 'step' changes in lateral acceleration (sudden constant acceleration followed by a constant deceleration) up to ±0.33 ms^{-2} r.m.s. (Jongkees and Groen, 1942). In the current study, an acceleration of about 0.3 ms^{-2} r.m.s. at about 0.5 Hz resulted in about 50% of subjects reporting at least 50% probability of losing balance, but a much greater acceleration was required to produce the same effect with the higher frequencies of oscillation (Figure 4.6). A stability threshold of ±0.45 ms^{-2} has been reported for subjects standing with their hands free, heels together, and toes 3 to 4 cm apart while exposed to sudden acceleration or deceleration without holding handrails, or taking a protective step, or stabilizing the body by large body sways or arm movements (Graaf and Weperen, 1997). In addition to the use of standing as opposed to walking subjects, and some other important details, these previous studies (Jongkees and Groen, 1942; Graaf and Weperen, 1997) differ in respect of the waveform of the motion stimulus. The present results show that the effects of lateral acceleration on postural stability are highly frequency-dependent and cannot be predicted solely from the peak acceleration although, for the waveforms investigated, stability is well predicted by both the peak velocity and the r.m.s. velocity.

Dynamic balance during normal locomotion is mainly achieved by adjusting the timing and placement of successive steps (Nashner, 1980). To compensate for medio-lateral acceleration induced by perturbations, it has been suggested that the central nervous system adjusts the step width to alter the moment arm (Oddson et al., 2004). Although the main strategy for maintaining balance is the 'stepping strategy', large errors in foot placement are corrected by hip moments (Hof *et al*., 2007; McKinnon and Winter 1993) and fine tuning is achieved by active ankle moments. The overall effects of the magnitude and frequency of oscillation on lateral COP movement in the present study implies that most of the subjects used stepping strategies to counteract the destabilizing effects of lateral motion.

The perceived risk of falling reported in this study may differ from the risk of passengers falling in transport. The subjects were prevented from falling, and so their reported probability of falling was influenced by the extent to which they found it necessary to take protective action, rather than by experiencing a fall. Range of lateral COP movement and r.m.s COP velocity may primarily reflect subject effort to continue walking by compensating with a wider step or a quicker step when the motion threatened their stability. Although the subjective and objective measures of postural stability used in this study reflect threats to subject stability, if the subjects were exposed to the same motions in a transport environment the risk of falling could differ for a variety of reasons (e.g. because in a transport environment the attention of passengers may not be solely focused on developing strategies to prevent falls and the actual risk of fall will be greater without a safety harness and a handrail). However, obtained stability thresholds (Figure 4.6) by simulating typical lateral oscillations experienced in trains provide useful information regarding the tolerance levels of walking subjects to lateral oscillations in transport.

There are differences between walking on a treadmill and walking along a floor. Subjects could not stop walking when their stability was threatened and so, unlike in many forms of transport, remaining stationary for a period of time was not an acceptable response. Otherwise, the biomechanics of walking on a treadmill and walking on a floor may be similar (Wagenaar and Beek, 2000) although the use of a speed preferred by the subject or the same speed controlled for all subjects has been reported to affect postural responses to perturbations during gait (Duysens and Bloem, 2009). The controlled speed of 0.7 ms^{-1} used in the present study was judged to be a comfortable walking speed by subjects in preliminary experiments but stability may differ with faster or slower speeds.

The present study was conducted with fit young male subjects who volunteered to participate in the study. Large differences in postural stability when walking and exposed to perturbations are expected to be associated with differences in age, gender, balance disorders, fitness, clothing, and carrying. The population participating in the study may be assumed to be among those least affected by motion perturbations: greater problems may be expected with some members of the general public, some of whom may be deterred from travelling by the risk of falling when moving around during travel.

4.5. Conclusion

By investigating the effects of systematic variations in the frequency and the magnitude of lateral oscillations it was possible to reveal that stability cannot be predicted solely from either the peak or the r.m.s. value of lateral acceleration although, for the waveforms investigated, stability is reasonably well predicted from both the peak velocity and the r.m.s. velocity of lateral oscillation.

Stability thresholds are obtained for walking people exposed to lateral oscillations that are typical of lateral accelerations experienced in a train ride. The findings may be applicable to passengers walking in moving trains, but further research is required to understand the dependence of postural stability on the motion waveform and variations in individual susceptibility to falling, especially in the elderly.

Chapter 5

Effect of hand support on the postural stability of walking people perturbed by lateral oscillatory motion

5.1. Introduction

Postural supports, such as mobile assistive devices (e.g. canes and walking aids), can assist the maintenance of stability during quiet standing and when walking. Supports may be more beneficial while standing or walking in trains, buses, ships, and aircraft where balance can be disturbed by the oscillatory motion of the floor. There are no known systematic studies of how the use of a hand support and varying the height of a hand support influence postural stability during perturbed locomotion.

Assistive devices may increase the area at the base of support under the feet and reduce the loading on the lower limbs that provide the reaction forces that counteract the destabilizing effects of body movements (Bateni and Maki, 2005). Similar advantages may be expected for hand supports in transport. Additionally, light touch contact with a surface, even if it does not provide force sufficient to stabilize the body may improve standing stability by providing an additional somatosensory cue to body movement (Jeka and Lackner, 1994 and 1995; Tremblay *et al.*, 2004, Clapp and Wing, 1999; Holden *et al.*, 1994). Fingertip contact with a stationary external support (a handrail at a height of 90 cm) has also been suggested to improve stability during treadmill walking (Dickstein and Laufer, 2004).

With instantaneous increases in horizontal acceleration, standing people have been reported to maintain balance while exposed to accelerations up to 0.76 ms^{-2} in the backward direction, 0.48 ms^{-2} in the forward direction, and 0.33 ms^{-2} in a sideways direction (Jongkees and Groen, 1942; Graaf and Weperen, 1997). The acceleration in public transport can be greater than these values so standing people cannot maintain stability without holding support (Jongkees and Groen, 1942), and it has been shown that greater accelerations can be tolerated when using a support (Browning, 1974).

When exposed to horizontal oscillation, whether a support increases or decreases the discomfort of seated people (Wyllie and Griffin, 2007) and standing people (Thuong and Griffin, 2010) depends on the frequency and the direction of the oscillation. The discomfort of standing people seems to be increased when a support increases the transmission of high frequency vibration to the upper-body (Thuong and Griffin, 2010), whereas postural instability is caused by low frequency oscillation. When walking, and supported on one leg for 80% of the gait cycle (Woollacott and Tang, 1997), stability may be less than when standing, and so supports may be more beneficial.

Assuming an inverted pendulum model of the human body, the stabilizing moment from a hand support will increase as the height of the support increases. It may therefore be expected that the effect of supports on stability depend on their height. The effects of support height on the postural stability during perturbed walking have not been previously reported.

When walking along a train, the dominant motions are in the lateral direction, and so in the frontal plane (i.e. coronal plane) of the body. It was shown in Chapter 4 that when exposed to transient lateral oscillations of the same velocity the perceived probability of losing balance and the lateral r.m.s. velocity of the centre of pressure (COP) were approximately constant over the frequency range 0.5 to 2 Hz (Chapter 4). With oscillations of the same frequency, reported probability of losing balance and lateral velocity of the centre of pressure increased with increasing magnitude of oscillations.

The aim of the experimental study reported in this chapter was to investigate the effects of hand support, and the height of hand support, on the postural stability of walking subjects perturbed by lateral oscillations. It was hypothesized that 'discomfort or difficulty' in the walking task, and the r.m.s. velocity of the lateral centre of pressure would decrease when using a hand support and decrease with increasing height of a hand support. It was also hypothesized that the 'discomfort or difficulty' ratings and the r.m.s. COP velocity would not depend on the frequency of oscillation with support and without support. At a specific frequency, it was hypothesized that the 'discomfort or difficulty' ratings, the COP velocity, and the lateral force applied to the hand support would increase with increasing magnitude of oscillation.

5.2. Method

5.2.1. Subjects

Twenty healthy male subjects with median age 28.5 years (range 25 to 40), stature 174 cm (range 166 to 182), weight 70.3 kg (49 kg to 88.7) participated in the study. Subjects completed a questionnaire to exclude those with relevant disorders or using drugs that may affect postural stability. 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.

5.2.2. Apparatus

A treadmill (Kistler Gaitway®) incorporating eight force sensors was used to provide the walking task for subjects and measure the vertical ground reaction forces during walking. Subjects were secured by a safety harness connected via two loose straps to a frame around the treadmill (Figure 5.1). The harness allowed subjects to move freely in the plane of progression but prevented their knees fromcontacting the floor if they fell. A safety net was positioned behind the subjects as a precaution in case they slid backwards while walking on the treadmill.

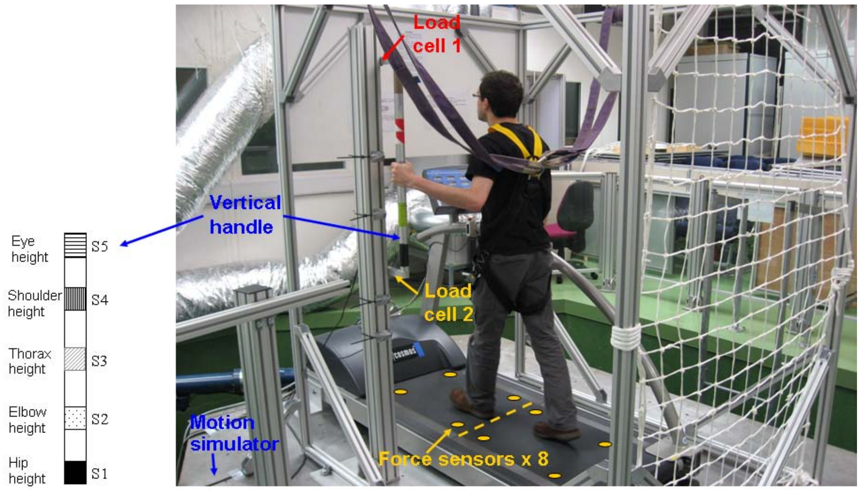


Figure 5.1: Experimental apparatus used in the second experiment.

Lateral oscillatory motion was generated by a six-axis motion simulator in the Human Factors Research Unit at the Institute of Sound and Vibration Research. The simulator is able to provide translational displacements of ±0.25 m in the lateral direction at accelerations up to about ±10 ms^{-2}.

A vertically orientated cylindrical handle, rigidly secured to the platform of the six-axis motion simulator, was placed to the left hand-side of the walking subjects to provide a stationary hand support (Figure 5.1). The handle had five differently coloured sections corresponding to the median values of hip (92 cm), elbow (109 cm), thorax (126 cm), shoulder (143 cm) and eye height (163 cm), respectively (anthropometric data for the British adults aged 19 to 65 years – Pheasant, 1988). Subjects were discouraged from using the handrail of the treadmill on the right hand side.

Contact forces applied to the hand support in the lateral direction were obtained from two single-axis load cells (Tedea Huntleigh Model 1022 1022-20M-C3), attached at the top and bottom end of the vertical handle (Figure 5.1).

Acceleration in the lateral direction was recorded by accelerometers on 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, vertical ground reaction force, and support contact force data collected by the Gaitway® data acquisition system were sampled at 100 samples per second and stored in a personal computer.

5.2.3. Experimental Procedure

While walking on the treadmill, subjects were perturbed by simple transient lateral oscillations. The stimuli – 4.5 cycles of sinusoidal motion modulated by a half sine envelope was the same type of stimuli as used in the first experiment. The motions started and ended with zero displacement, zero velocity, and zero acceleration and were chosen as being broadly representative of the lateral motions experienced in trains (Figure 4.2a).

At each of seven frequencies (0.5, 0.63, 0.8, 1.0, 1.25, 1.6, 2.0 Hz), the motions were presented at the velocity of 0.16 ms^{-1} r.m.s., corresponding to seven acceleration magnitudes (0.5, 0.63, 0.8, 1.0, 1.25, 1.6 and 2 ms^{-2} r.m.s.). At 1 Hz, the motions were also presented at six velocities (0.05, 0.064, 0.08, 0.1, 0.125 and 0.16 ms^{-1} r.m.s.), corresponding to six acceleration magnitudes (0.315, 0.4, 0.5, 0.63, 0.8 and 1.0 ms^{-2} r.m.s).

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. While subjects walked on the treadmill, the lateral oscillatory motions were applied at random unpredictable times.

The experiment involved three parts. In Part A and Part B, subjects were exposed to pairs of motion stimuli. The first stimulus was called the reference motion (1.0 ms^{-2} r.m.s. at 1.0 Hz) and was the same throughout the experiment. During the reference motion, subjects held the vertical handle support at the median thorax height (S3 position, Figure 5.1) continuously throughout the motion. Subjects were asked to report their 'discomfort or difficulty' in walking caused by the second motion (i.e. test motion) relative to the 'discomfort or difficulty' caused by the first motion, assuming the 'discomfort or difficulty' caused by the first motion was 100.

In part A, the test motion was the same as the reference motion. For each test motion, subjects were asked to hold the vertical handle at one of the five vertical positions (Figure 5.1) before the test motion started. In one condition, they were asked not to hold the handle (i.e. without support condition).

In part B, the test motions were applied at seven different frequencies (0.5, 0.63, 0.8, 1, 1.25, 1.6, 2.0 Hz) and at six different magnitudes (0.05, 0.064, 0.08, 0.1, 0.125, and 0.16 ms^{-1} r.m.s.). These test motions were applied in two conditions: with support (subjects held the support at the S3 position throughout the oscillation) and without support.

In Part C, no reference motion was applied. Subjects were exposed to oscillations at seven different frequencies (0.5, 0.63, 0.8, 1, 1.25, 1.6, 2.0 Hz) with a velocity of 0.16 ms^{-1} r.m.s. They were invited to hold the support when required during exposure to the oscillation. They were free to hold the support at whatever position they preferred so as to stabilize their body against the motion. The preferred holding position was recorded by the experimenter. Gait and support contact force data were also gathered.

Parts A, B, and C were applied in sequence but the test motions within each part were presented in random orders for each subject.

Gait measure (i.e. centre of pressure) and lateral force applied to the hand support were also gathered while subject walked normally without oscillation, both with and without support.

5.2.4. Analysis

The change in postural stability when holding the support was quantified by percentage reductions in the subjective measure (i.e. 'discomfort or difficulty' ratings) and the objective measure (i.e. r.m.s. COP velocity). Percentage reduction was calculated as shown by the following equation.

$$\mathrm{Re}\,duction\ (\%) = \frac{\text{measure evaluated with support - measure evaluated without support}}{\text{measure evaluated without support}} * 100$$

The force time-histories (from eight force sensors in the treadmill) were processed to determine centre of pressure (COP) time histories during each motion (Section 3.2.2.1). The COP velocity in the lateral direction was obtained by differentiating the lateral COP position after filtering the centre of pressure position using low-pass Bessel filter at 8 Hz.

Lateral force at the handle was obtained from sum of the forces indicated by the load cells at the top and bottom of the vertical handle. Mass cancelation was performed in the time domain by subtracting the product of the acceleration and the mass of the handle from the total measured force.

Non-parametric statistical tests were performed with SPSS (version 17). The Friedman analysis of variance tested for differences between multiple conditions and the Wilcoxon matched-pairs signed ranks investigated differences between pairs of conditions. Associations between variables were investigated using Spearman's rank correlation.

5.3. Results

An example of the COP position of a subject exposed to 0.8 Hz lateral oscillation at 0.8 ms^{-2} r.m.s. is shown in Figure 5.2a. The COP position shows the lateral location of the resultant of the ground reaction forces and is indicative of lateral foot placement. The COP velocity indicates the rate of change of COP position (Figure 5.2b). An example of the force applied to the vertical handle in the lateral direction is shown in Figure 5.2c.

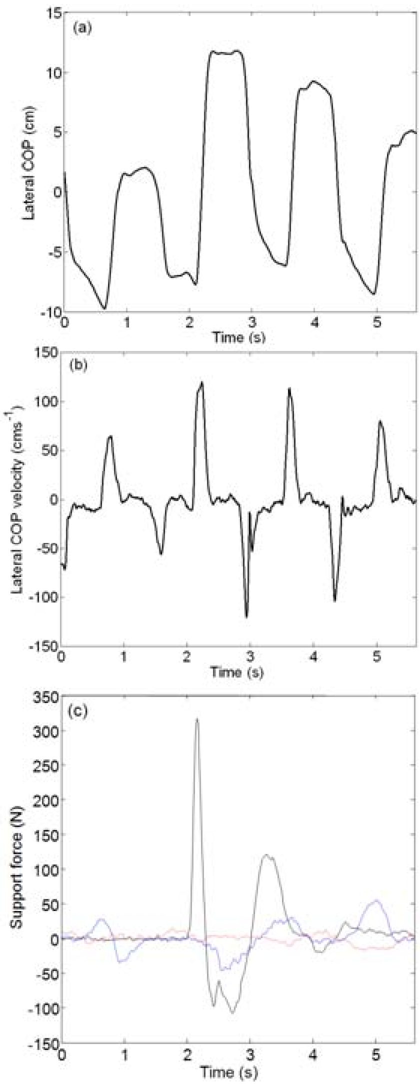


Figure 5.2: Example time histories of the centre of pressure (COP) and lateral force applied to the hand support for a subject walking while exposed to 0.8 ms^{-2} r.m.s. lateral oscillation at 0.8 Hz: (a) COP position in the lateral direction; (b) COP velocity in the lateral direction; (c) lateral force applied to the hand support: ——— support held if required during oscillation, - - - - - support held continuously throughout the oscillation, ········· support held continuously without oscillation.

5.3.1. Effect of height of hand support

During 1-Hz lateral oscillation at 1.0 ms^{-2} r.m.s., the 'discomfort or difficulty' ratings did not depend on the height at which the subjects held the hand support (p=0.224, Friedman; Figure 5.3a). However, as may be expected, the 'discomfort or difficulty' rating was greater without the support than with any of the support heights (p<0.01, Wilcoxon; Figure 5.3a).

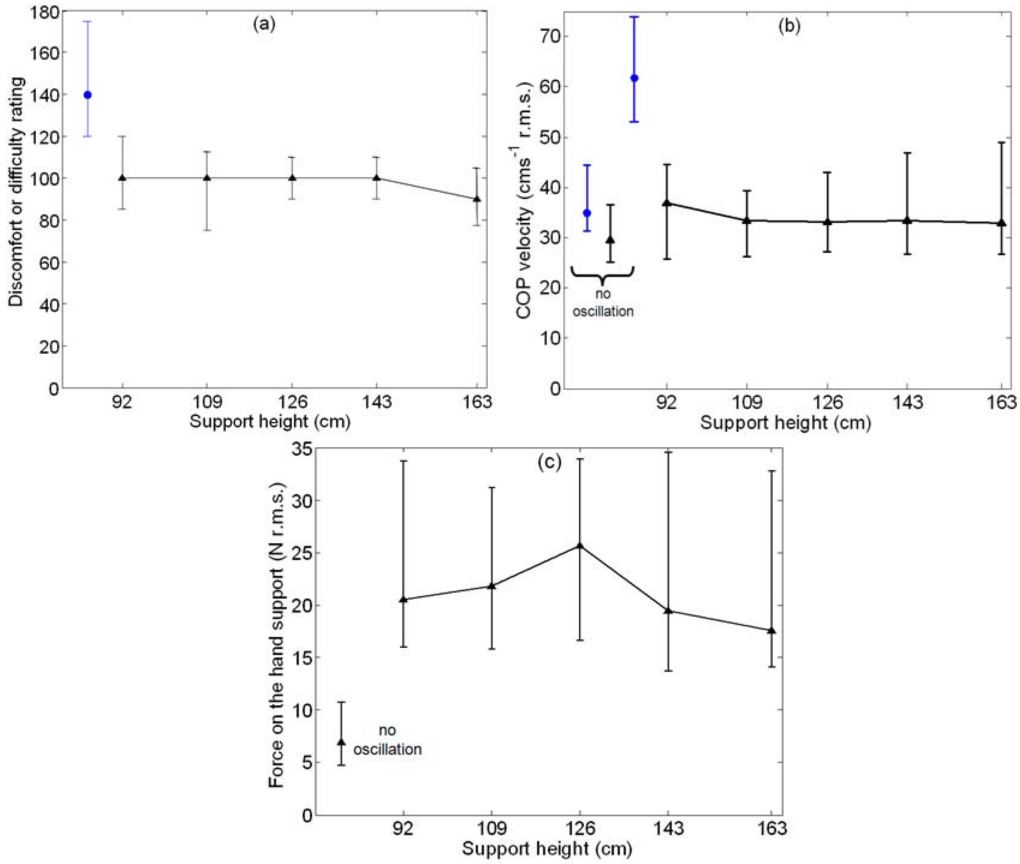


Figure 5.3: Effects of support height while exposed to 1.0 ms^{-2} r.m.s. lateral oscillation at 1.0 Hz (medians and inter-quartile ranges): (a) 'discomfort or difficulty' ratings; (b) r.m.s. COP velocity during oscillation and during normal walking (without oscillation); (c) lateral r.m.s. force on the hand support during oscillation and during normal walking (without oscillation): ▲ support held throughout the oscillation, ● without support.

The r.m.s. COP velocity and the r.m.s. lateral force applied to the hand support were also independent of support height (p=0.78 and p= 0.06, respectively, Friedman; Figure 5.3b and Figure 5.3c), but the r.m.s. COP velocity was greater without support than with any of the five support heights (p<0.01, Wilcoxon; Figure 5.3b).

5.3.2. Effect of frequency of oscillation

With a velocity of 0.16 ms^{-1} r.m.s. at all frequencies, the 'discomfort or difficulty' in walking was independent on the frequency of oscillation when using the hand support (p=0.098, Friedman; Figure 5.4a) but dependent on the frequency of oscillation when not using the support (p<0.01; Friedman, Figure 5.4a). At all frequencies, the 'discomfort or difficulty' was less when the support was held throughout the oscillation than when it was not held (p<0.01, Wilcoxon; Figure 5.4a).

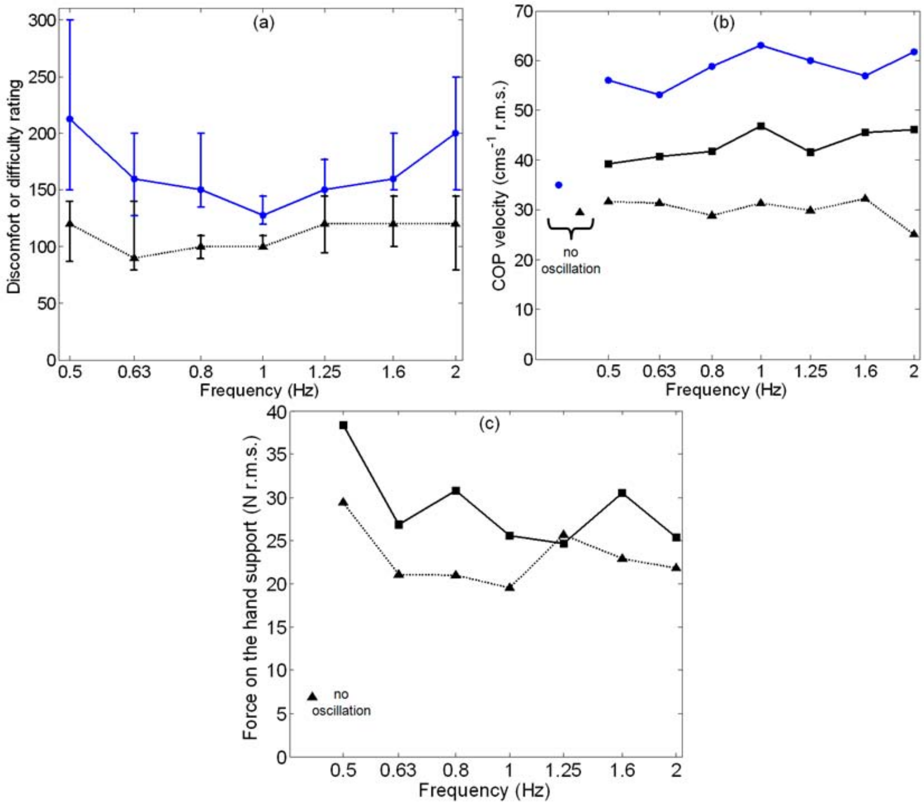


Figure 5.4: Effects of frequency while exposed to 0.16 ms^{-1} r.m.s. lateral oscillation (medians): (a) 'discomfort or difficulty' ratings. (b) r.m.s. COP velocity during oscillation and during normal walking (without oscillation). (c) lateral r.m.s. force applied to the hand support during oscillation and during normal walking (without oscillation): ▲ support held throughout the oscillation ■ support held if required, ● without support.

At all frequencies, the r.m.s. COP velocity was less when the support was held throughout the oscillation than when it was not held (p<0.01, Wilcoxon; Figure 5.4b). Similarly, the r.m.s. COP velocity when the support was used if required was less than when the support was not used (p<0.03, Wilcoxon; Figure 5.4b).

With the same velocity at all frequencies, and no hand support, the r.m.s. COP velocity was independent of the frequency of oscillation (p=0.157, Friedman; Figure 5.4b). Similarly, with use of the support, the r.m.s. COP velocity was independent of the frequency of oscillation both when used throughout the oscillation and when used if required (p=0.284 and p=0.08, respectively, Friedman, Figure 5.4b).

The lateral r.m.s. force applied to the hand support during oscillation at 0.16 ms^{-1} r.m.s. was dependent on the frequency of oscillation, both when used throughout the oscillation and when used if required (p<0.01, Friedman; Figure 5.4c). At all frequencies, when holding the support if required, the force was greater than when holding the support continuously throughout the oscillation (p<0.01, Wilcoxon; Figure 5.4c), except at the three highest frequencies (i.e. 1.25, 1.6, and 2 Hz).

When using the support continuously throughout oscillation, the percentage reduction in the ‘discomfort or difficulty’ ratings depended on the frequency of oscillation ($p<0.01$, Friedman; Table 5.1) The percentage reduction in the r.m.s. COP velocity was not dependent on the frequency of oscillation both when used throughout the oscillation and when used if required ($p=0.089$ and $p=0.922$, respectively, Friedman). The percentage reduction in r.m.s. COP velocity was greater when the support was held throughout the oscillation than when used if required ($p<0.01$, Wilcoxon, Table 5.1).

Table 5.1: Median percentage reductions in ‘discomfort or difficulty’ ratings and r.m.s. COP velocity from holding the hand support as a function of the frequency of 0.16 ms^{-1} r.m.s. lateral oscillation.

	Percentage reductions in ‘discomfort or difficulty’ rating (%)	Percentage reductions in r.m.s. COP velocity (%)	
Frequency (Hz)	Support used throughout oscillation	Support used throughout oscillation	Support used if required
0.5	45.3	43.6	33.0
0.63	44.2	37.7	22.4
0.8	33.3	43.4	28.5
1	22.3	47.2	20.8
1.25	20.0	45.0	27.9
1.6	31.3	34.7	19.5
2	45.3	50.5	22.2
MEDIAN	33.3%	43.6%	22.4%

5.3.3. Effect of magnitude of oscillation

When exposed to lateral oscillation at 1 Hz, the ‘discomfort or difficulty’ ratings increased with increasing magnitude of oscillation, both with and without the hand support ($p<0.01$; Spearman; Figure 5.5a). At all magnitudes of acceleration, the ‘discomfort or difficulty’ ratings were less when a support was used ($p<0.01$, Wilcoxon).

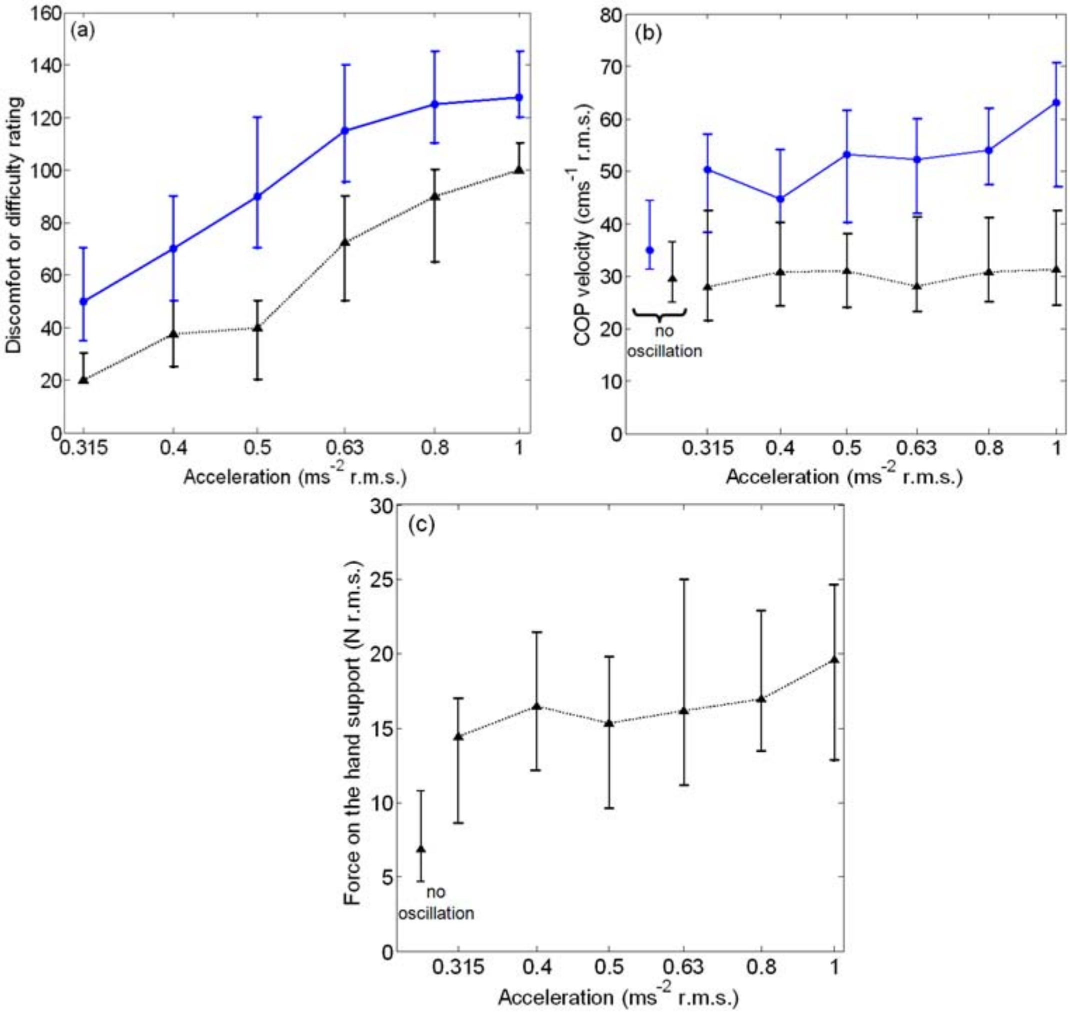


Figure 5.5: Effects of motion magnitude while exposed to lateral oscillation at 1 Hz (medians and inter-quartile ranges): (a) 'discomfort or difficulty' ratings. (b) r.m.s. COP velocity during oscillation and during normal walking (without oscillation). (d) lateral r.m.s. force applied to the hand support during oscillation and during normal walking (without oscillation): ▲ support held continuously throughout oscillation, ■ support used if required, ● without support.

At each magnitude, the r.m.s. COP velocity was less when the support was held throughout the oscillation than without support ($p<0.01$, Wilcoxon; Figure 5.5b).

The r.m.s. COP velocity increased with increasing magnitude of oscillation without support ($p<0.01$, Spearman, Figure 5.5b) but was independent of the magnitude of oscillation when holding the support ($p=0.056$; Friedman, Figure 5.5b). The lateral r.m.s. force applied to the hand support also tended to increase with increasing magnitude of oscillation ($p<0.05$, Spearman, Figure 5.5c).

The percentage reductions in r.m.s. COP velocity increased with increasing magnitude of oscillation ($p=0.019$, Spearman, Table 5.2). The percentage reduction in the 'discomfort or difficulty' ratings arising from holding the hand support decreased as the magnitude of oscillation increased (p<0.01, Spearman, Table 5.2).

Table 5.2: Median percentage reductions in 'discomfort or difficulty' ratings and r.m.s. COP velocity from holding the hand support throughout oscillation as a function of the magnitude of 1-Hz lateral oscillation.

Acceleration (ms^{-2} r.m.s.)	Percentage reductions in 'discomfort or difficulty' rating (%)	Percentage reductions in r.m.s. COP velocity (%)
0.315	50.0	31.7
0.4	50.0	29.6
0.5	49.2	41.5
0.63	33.3	33.3
0.8	31.6	43.0
1	22.3	47.2
MEDIAN	41.25%	37.4%

5.3.4. Effect of support during perturbed walking and normal walking

When walking without perturbation (i.e. no oscillation), holding the support at the median thorax height reduced the r.m.s. COP velocity by 15.6% (Figure 5.4b).

When holding the support at any height (Figure 5.3b), with any frequency (Figure 5.4b), and with any magnitude (Figure 5.5b), the percentage reduction in r.m.s. COP velocity was greater during perturbed walking than during unperturbed walking (p<0.01, Wilcoxon).

The r.m.s. force applied to the support during unperturbed walking was less than the r.m.s. force during lateral oscillation with any height of the hand support (Figure 5.3c), any frequency (Figure 5.4c), and any magnitude (Figure 5.5c) (p<0.01, Wilcoxon, Table 5.3).

To demonstrate the amount of forces applied to the hand support during normal walking and perturbed walking at any support height, frequency and magnitude, peak lateral forces applied to the hand support are provided in Table 5.3.

Table 5.3. Peak lateral forces applied to the hand support during normal walking (without oscillations) and during perturbed walking (with oscillations) at any support height, with any frequency and with any magnitude.

	Peak forces applied to the hand support (N)	
Support height (cm)	support used throughout oscillation	support used if required
92	52.5	
109	55.3	
126	61.4	
143	53.2	
163	46.5	
Frequencies (Hz)		
0.5	83.0	117.7
0.63	59.2	80.6
0.8	55.6	96.2
1	43.7	81.5
1.25	65.7	67.5
1.6	58.7	78.6
2	58.6	81.2
Magnitudes (ms^{-2} r.m.s.)		
0.315	35.0	
0.4	41.4	
0.5	37.3	
0.63	40.6	
0.8	46.3	
1	43.7	
Without oscillations		
	5.7	

5.3.5. Preferred height for hand support

In Part C of the experiment, subjects were invited to hold the support when required and at whatever position they preferred during exposure to 0.16 ms^{-1} r.m.s. at frequencies from 0.5 to 2.0 Hz. About 60% of subjects chose to hold the support at the median thorax height (126 cm above the surface supporting the feet, Figure 5.6). The preferred support height was not affected by the frequency of oscillation (p=0.09, Friedman).

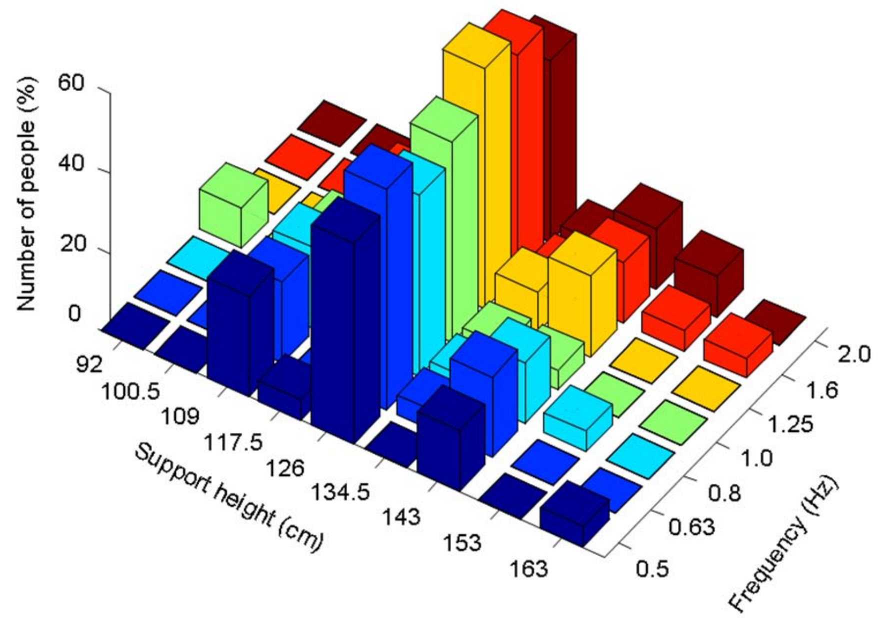


Figure 5.6: Percentage of subjects preferring each support height at each frequency of oscillation with a velocity of 0.16 ms^{-1} r.m.s.

5.4. Discussion

With all magnitudes and frequencies of lateral oscillation, holding the hand support improved postural stability, as indicated by decreased ‘discomfort or difficulty’ ratings and decreased r.m.s. COP velocity (Figure 5.4 and Figure 5.5). Forces less than 1 N applied by a fingertip contact to a stationary support have been reported to improve stability during quiet standing (Jeka and Lackner, 1994, 1995; Clapp and Wing 1999) and during normal (i.e. unperturbed) walking (Dickstein and Laufer, 2004). Lateral forces on the handle support during normal walking without oscillations in the current study were around 6.90 N r.m.s. and are comparable to the mean forces of 5 N applied by standing subjects to a stationary support via fingertip contact (Jeka and Lackner, 1994). Higher forces applied by the walking subjects to the hand support might be caused by a full grasp of the vertical handle rather than a fingertip contact and also from differences in the postural requirements of walking and standing. During oscillation, the forces reached 14.4 to 29.4 N r.m.s. (35.0 to 83 N peak) when the support was used throughout the oscillation and to 24.7 to 38.4 N r.m.s. (67.5. to 117.7 N peak) when the support was used if required (Table 5.3). The current study with perturbation and subjects grasping the support with their chosen force found that the support improved stability (i.e. reduced COP velocity) more during perturbed walking than during normal walking. These findings are consistent with the external perturbation increasing the risk of fall and requiring greater forces, and more rapidly applied forces, to counteract the destabilizing effects of lateral oscillation.

When the walking subjects held the vertical bar support, the 'discomfort or difficulty' was reduced at all frequencies (0.5 to 2 Hz) and with all magnitudes of oscillation (0.315 to 1.0 ms^{-2} r.m.s.) (Figure 5.4a and Figure 5.5a). With a similar support and moderate magnitudes of lateral sinusoidal oscillation (0.04 to 0.25 ms^{-2} r.m.s. at 0.5 Hz, 0.1 to 0.63 ms^{-2} r.m.s at 1 Hz, and 0.16 to 1.0 ms^{-2} r.m.s. at 2 Hz), Thuong and Griffin (2010) found that holding a bar had no significant effect on the comfort of standing people, possibly because postural instability was not the main source of discomfort for the standing subjects. The results of the first experiment as reported in Chapter 4 showed that when walking, the probability of losing balance is about 45% when exposed to lateral oscillation of 0.25 ms^{-2} r.m.s. at 0.5 Hz, 50% with 0.63 ms^{-2} r.m.s at 1 Hz, and 35% with 1.0 ms^{-2} r.m.s at 2 Hz (Sari and Griffin, 2009), with the probability decreasing with decreasing magnitude of oscillation. The postural stability of subjects standing and supported on two legs may be expected to be greater than when supported on only one leg for 80% of the gait cycle during walking (Woollacott and Tang, 1997).

For walking subjects exposed to transient lateral oscillatory motion with a velocity of 0.16 ms^{-1} r.m.s., a 90% probability of losing balance, independent of the frequency of oscillation between 0.5 and 2 Hz, was previously reported in Chapter 4 when not using a support. The motion waveforms used in the first experiment (Chapter 4) were the same as those used in the second experiment reported in this chapter in which a significant effect of the frequency of oscillation was found on discomfort ratings without support when using the same motion velocity (Figure 5.4a). In the second experiment, subjects reported their relative 'discomfort or difficulty' when walking and could use any number for their judgement, whereas in the first experiment subjects were asked to estimate their absolute probability of losing balance using any number between 0 to 100%. Loss of balance is expected to be the main source of 'discomfort or difficulty' during perturbed walking, with zero probability of losing balance when using a support. When the motions became severe, the scale for reporting the probability of losing balance becomes less sensitive due to saturation (towards the maximum value of 100%). The measure of 'discomfort or difficulty' of the test motion relative to the 'discomfort or difficulty' of a reference motion, as used in the current study appears more sensitive to factors influencing walking stability (e.g. the frequency of oscillation and the use of supports).

With the same motion velocity, the r.m.s. COP velocity was independent of frequency of oscillation (Figure 5.4b) similar to the findings in Chapter 4. As suggested by the percentage reduction in the 'discomfort or difficulty' ratings, hand support was most beneficial with the lowest and the highest frequencies of oscillation, where 'discomfort or difficulty' ratings were greatest (Table 5.1). However, the percentage reduction in r.m.s. COP velocity was independent of frequency of oscillations (Table 5.1). Looking at the median values in Table 5.1 and Table 5.2, the percentage improvement in postural stability from holding the hand support can be approximated to 40% when the hand support is used throughout the oscillation and 20% when the hand support is used if required.

Although the ‘discomfort or difficulty’ rating without support was dependent on the frequency of oscillation at the same motion velocity, they were independent of the frequency when the support was held throughout the motion. In part, this may reflect a feeling of being safe with all frequencies when using a support. A reduction in physiological stress may be associated with improved postural control when using a support, as observed here in reductions in the objective measure of postural instability (Figure 5.4b).

With the same frequency of oscillation, ‘discomfort or difficulty’ and r.m.s. COP velocity when walking without support increased with increasing magnitude of the oscillation (Figure 5.5a and Figure 5.5b). When a support was used, a similar increasing trend was observed in the ‘discomfort or difficulty’ ratings, whereas the r.m.s. COP velocity was independent of the magnitude of oscillation (Figure 5.5b). When using the hand support, the percentage reduction in the ‘discomfort or difficulty’ ratings decreased as the magnitude of the oscillation increased, but the percentage reduction in the r.m.s. COP velocity increased with increasing magnitude of oscillation (Table 5.2). The hand support was more beneficial at higher magnitudes of motion, as also suggested by increased lateral force applied to the support (Figure 5.5c).

When subjects held the support continuously throughout lateral oscillation, the ‘discomfort or difficulty’ ratings, r.m.s. COP velocity, and lateral r.ms. force applied to the hand support were similar with all support heights (Figure 5.3). If the support was purely providing a force needed for mechanical stabilization of the body, it would be expected that the subjective and objective evaluations of postural instability would decrease with increasing support height, due to the increased balancing moment provided by support contact forces with a greater moment arm. The absence of an effect of support height suggests the support may not have only provided mechanical stabilization but also sensory cue that assisted spatial orientation (Jeka and Lackner, 1994; Jeka, 1997).

The r.m.s. COP velocity was greater when the support was used if required than when it was used throughout the oscillation (Figure 5.4b). The percentage reduction in r.m.s. COP velocity from using the support throughout the oscillation was also greater than the percentage reduction when only using the support if required (Table 5.1). Forces applied to the support when holding it if required were also greater than when holding it throughout oscillation, except with the higher frequencies (1.25 Hz, 1.6 and 2 Hz). Supports may therefore be useful mechanical aids when they are used only if required and supports may be more required during exposure to low frequency oscillations. When subjects only held the support when it was required, they mostly preferred to hold the vertical handle at the height of 126 cm above the surface supporting the feet which might be ergonomically comfortable for most of the subjects.

5.5. Conclusion

Hand support improves postural stability when walking is perturbed by lateral oscillation at all frequencies (in the range 0.5 to 2 Hz) and at all velocity magnitudes (in the range 0.05 to 0.16 ms^{-1} r.m.s.). The improvement in postural stability is shown by significant reductions in both the subjective ratings of 'discomfort or difficulty' in walking and objective measure of r.m.s. lateral COP velocity when a hand support is used.

The improvement in postural stability from holding the support and the forces applied to the hand support were independent of support height and were greater during perturbed walking than during normal walking, and greater when held throughout the oscillation than when held only if required. Subjects preferred to hold the vertical support at the height of 126 cm above the surface supporting the feet if required during exposure to lateral oscillatory motion. The findings of the study emphasize the importance of supports as mechanical aids in perturbed locomotion and can be used to optimize hand supports in terms of support height in transport.

Chapter 6

Effect of waveform on the postural stability of walking people perturbed by lateral oscillatory motion

6.1. Introduction

Postural stability during standing and walking can be disturbed by various types of perturbations with different waveform characteristics. Transient perturbations may characterize slips, trips or typical short duration oscillations in transport whereas continuous perturbations may represent the continuous destabilizing effect of gravity and may be used to investigate the steady-state characteristics of human postural control system. Sinusoidal perturbations have the disadvantage of predictability whereas random or pseudorandom perturbations are more unpredictable (Maki, 1986; Maki *et al*., 1987).

It has been suggested that differences in waveforms produce differences in perception of motion in terms of discomfort, and subjects can be more sensitive to random vibration than to sinusoidal vibration of the same r.m.s. magnitude (Griffin, 1976). There is evidence of nonlinearity in postural stability of standing people exposed to perturbations with transient waveforms: postural responses to transient stimuli (acceleration pulses) in terms of magnitude-dependence have been shown to have a more nonlinear behavior than responses to continuous pseudo-random perturbations (Maki and Ostrovski, 1993a).

The effect of waveforms on discomfort has been investigated for seated and standing subjects. A common measure of an acceleration waveform is root-mean-square (r.m.s.) value which is a suggested method of predicting discomfort for seated and standing people caused by various types of vibration (ISO 2631-1 (1997) and BS6841 (1987)). However, the r.m.s. is not optimum for evaluating all types of waveforms (sinusoidal and octave-bandwidth random waveform with increasing peak levels) in terms of the discomfort of standing subjects (Thuong and Griffin, 2010b). Oscillations having the same frequency and same r.m.s. value caused greater discomfort with increasing peak levels in seated subjects exposed to vertical whole-body vibration (Griffin and Whitham, 1980). Howarth and Griffin (1991) also reported an increase in the discomfort of seated people with increasing crest factor of oscillations when the r.m.s. values of the oscillations were kept constant. The effect of waveform on the postural stability of walking people has not been reported systematically.

The waveform characteristics of transient platform perturbations applied to standing subjects have been traditionally reported in terms of peak velocity and peak displacement (Horak and Nashner, 1986; Tang *et al*., 1998). Acceleration, velocity, or displacement can be used to quantify the magnitude of a perturbation. Maki and Ostrovski (1993a, 1993b) suggested that acceleration is a more reasonable parameter for quantifying perturbation amplitudes since the acceleration provides the initial destabilizing input to the postural control system and stabilizing joint moments are triggered in response to acceleration. Brown *et al*. (2001) also emphasized the necessity of reporting acceleration characteristics of perturbation waveforms. Sari and Griffin (2009) reported the significance of the velocity of perturbation on the postural stability of walking subjects. Runge et *al*. (1999) showed that kinetics of postural recovery is dependent on the velocity of platform translation. There is no standardized procedure to report perturbation characteristics of waveforms, which makes it difficult to compare and interpret the results of different perturbed balance experiments conducted in different laboratory environments.

In the first and second experiment reported in this thesis, the stimuli were 4.5 cycles of sinusoidal motion modulated by a half sine envelope. For these waveforms, the peak acceleration and the peak velocity were, respectively, double the r.m.s. acceleration and the r.m.s. velocity. The results of the first experiment showed that postural stability is broadly similar when motions are applied at the same velocity magnitude irrespective of changes in frequency. The probability of losing balance increased with increasing magnitude of acceleration and increasing magnitude of velocity. Whether walking people are more sensitive to peak or r.m.s. magnitudes of oscillation is unknown.

The aim of the third experiment that is reported in this chapter was to investigate the effect of waveform on the postural stability of walking subjects in the frontal plane and to determine whether postural stability can be predicted solely from r.m.s. or peak magnitudes of the oscillations. It was hypothesized that within seven acceleration waveforms having the same duration, the same frequency and the same r.m.s. magnitude, the probability of losing balance and the lateral COP velocity would be dependent on the peak magnitude of the oscillations. It was also expected that the reported probability of losing balance and the lateral COP velocity would increase with increasing r.m.s. magnitude of oscillations having the same duration, the same frequency and the same peak magnitude. It was hypothesized that at both frequencies (1 Hz and 2 Hz), the association between ‘discomfort or difficulty’ in the walking task and the r.m.s. magnitude of acceleration would be stronger than the association with the peak magnitude of the acceleration.

6.2. Method

6.2.1. 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 or using drugs that may affect postural stability. 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.

6.2.2. Apparatus

The experimental apparatus was the same as used in the first experiment. For details, please see Section 4.2.2.

6.2.3. Experimental Procedure

While walking on the treadmill, subjects were perturbed by lateral oscillations. Oscillatory motions lasted 8 seconds and were cosine tapered for 1.5 seconds at the beginning and end of the motion. The oscillatory motions were one-third octave band random motions centred at 1 Hz and 2 Hz.

Each subject was exposed to a total of 28 different test motions (Table 6.1) (overall 56 motions including 28 reference motions): seven different waveforms of random vibration having the same r.m.s. magnitude and seven different waveforms of random vibration having the same peak magnitude, both at two frequencies (1 Hz and 2 Hz) (Table 6.1). The seven different random waveforms were selected to have specific values for the crest factor: 1.6, 1.8, 2.0, 2.24, 2.5, 3.8, and 3.15. Figure 6.1 shows some examples of the waveforms centred at 1 Hz with different crest factors. Table 6.1 shows the characteristics of the oscillatory motions used in the experiment.

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. While subjects walked on the treadmill, the lateral oscillatory motions were applied at random unpredictable times.

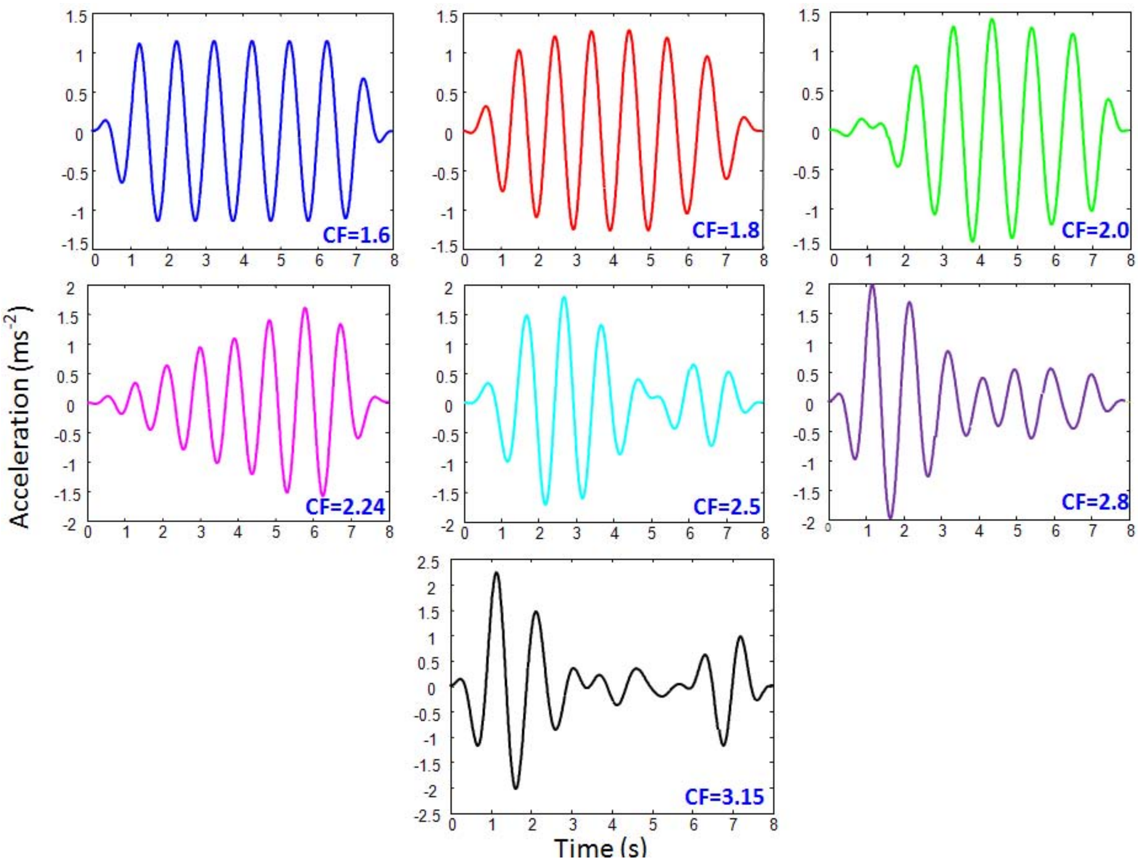


Figure 6.1: One-third octave band random waveforms centred at 1 Hz with different crest factors at a magnitude of 0. 7 ms^{-2} r.m.s.

The experiment was conducted at two frequencies (1 Hz and 2 Hz). Subjects were exposed to pairs of motion stimuli; the first stimulus was called the reference motion. The reference motion at 1 Hz was a sinusoidal waveform (0.7 ms^{-2} r.m.s., 1.14 ms^{-2} peak) with a crest factor of 1.6. The reference motion at 2 Hz was a sinusoidal waveform (1.4 ms^{-2} r.m.s., 2.24 ms^{-2} peak) with a crest factor of 1.6. After the test motion was applied, subjects were invited to answer two questions: For the first question, subjects were asked to report the ‘discomfort or difficulty’ in their walking task caused by the second motion relative to the ‘discomfort or difficulty’ caused by the first motion, assuming the ‘discomfort or difficulty’ caused by the first motion was 100. For the second question, subjects were asked to report their probability of losing balance caused by the test motion by answering the same question asked to the participants of the first experiment reported in Chapter 4:

‘‘What is the probability that you would lose balance if the same test motion were repeated?’’

Table 6.1: Acceleration characteristics of the lateral oscillations used in the experiment.

f=1 Hz PART A					**f=1 Hz PART B**				
waveform	r.m.s. (ms^{-2})	peak (ms^{-2})	CF	taper length (s)	waveform	r.m.s. (ms^{-2})	peak (ms^{-2})	CF	taper length (s)
sinusoidal	**0.7**	1.14	1.60	1.40	random	0.57	**1.8**	3.15	1.50
random	**0.7**	1.28	1.80	1.50	random	0.64	**1.8**	2.80	1.50
random	**0.7**	1.42	2.00	1.50	random	0.72	**1.8**	2.50	1.50
random	**0.7**	1.59	2.24	1.50	random	0.80	**1.8**	2.24	1.50
random	**0.7**	1.77	2.50	1.50	random	0.90	**1.8**	2.00	1.50
random	**0.7**	2.00	2.80	1.50	random	1.00	**1.8**	1.80	1.50
random	**0.7**	2.24	3.15	1.50	sinusoidal	1.12	**1.8**	1.60	1.40
f=2 Hz PART A					**f=2 Hz PART B**				
waveform	r.m.s. (ms^{-2})	peak (ms^{-2})	CF	taper length (s)	waveform	r.m.s. (ms^{-2})	peak (ms^{-2})	CF	taper length (s)
sinusoidal	**1.4**	2.24	1.60	1.40	random	1.11	**3.5**	3.15	1.50
random	**1.4**	2.52	1.80	1.50	random	1.25	**3.5**	2.80	1.50
random	**1.4**	2.80	2.00	1.50	random	1.40	**3.5**	2.50	1.50
random	**1.4**	3.14	2.24	1.50	random	1.56	**3.5**	2.24	1.50
random	**1.4**	3.50	2.50	1.50	random	1.75	**3.5**	2.00	1.50
random	**1.4**	3.92	2.80	1.50	random	1.94	**3.5**	1.80	1.50
random	**1.4**	4.41	3.15	1.50	sinusoidal	2.19	**3.5**	1.60	1.40

At each frequency, the experiment involved two parts (Table 6.1). In Part A, test motions had the same r.m.s. magnitude with increasing peak values and increasing crest factors. In Part B, the test motions having the same peak magnitude were applied at different r.m.s. magnitudes. Test motions within each part were presented in random orders for each subject.

Together with the subjective data, the acceleration of the moving platform and gait data (i.e. ground reaction forces under the feet) were gathered. Gait data were also gathered while subjects walked normally without oscillations.

6.2.4. Analysis

The force time-histories (from 8 force sensors in the treadmill) were processed to determine centre of pressure (COP) time histories during each motion (Section 3.2.2.1). The COP velocity in the lateral direction was obtained by differentiating the lateral COP position after filtering the centre of pressure position using low-pass Bessel filter at 8 Hz.

The association of 'discomfort or difficulty' ratings with the peak and r.m.s. magnitude of the oscillation was quantified by the growth of sensation (*n*) in Stevens' power law (Stevens, 1975) shown by Equation (6.1), where ψ is the sensation magnitude (i.e. reported 'discomfort or difficulty' ratings), ϕ is the stimulus physical magnitude (i.e. r.m.s. and peak magnitudes of oscillations) and *k* is a constant for a given stimulus. Two different *n* values (n_{peak} and n_{rms}) corresponding to the growth of sensation with respect to peak and r.m.s. magnitudes of oscillations were obtained.

$$\Psi = k * (\Phi)^n \quad (6.1)$$

The subjective and objective measures of postural stability were then used to test the hypothesis regarding their dependence on the r.m.s. and peak magnitude of the oscillations.

Non-parametric statistical tests were performed with SPSS (version 17). The Friedman analysis of variance tested for differences between multiple conditions and the Wilcoxon matched-pairs signed ranks investigated differences between pairs of conditions. Associations between variables were investigated using Spearman's rank correlation.

6.3. Results

An example of the COP position of a subject exposed to 1.0 Hz lateral oscillation at 0.7 ms^{-2} r.m.s. is shown in Figure 6.2a. The COP position shows the lateral location of the resultant of the ground reaction forces and is indicative of lateral foot placement. The COP velocity indicates the rate of change of COP position (Figure 6.2b). Figure 6.2c shows the one-third octave band oscillation centred at 1 Hz (0.7 ms^{-2} r.m.s, 2.0 ms^{-2} peak) used to perturb walking subject in the lateral direction.

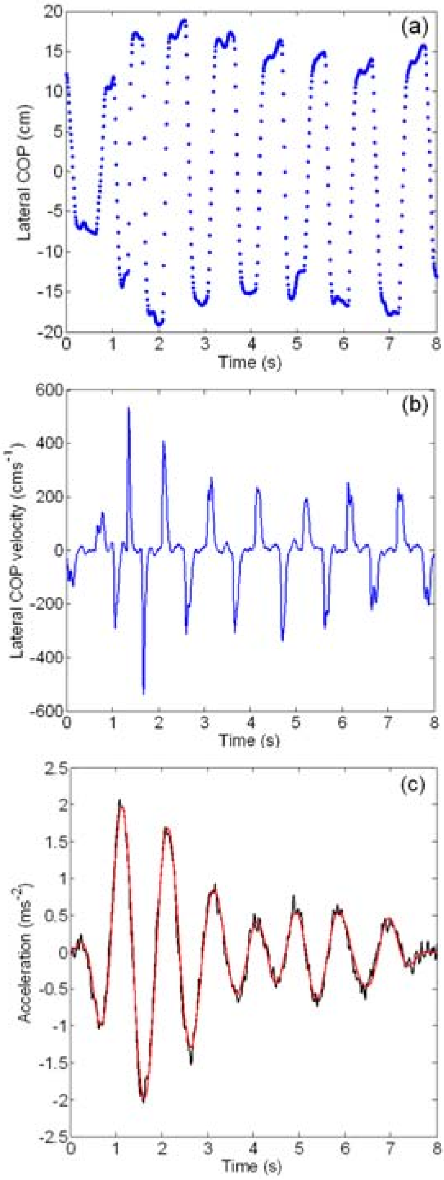


Figure 6.2: Example time histories of the centre of pressure (COP) for a subject walking while exposed to 0.7 ms^{-2} r.m.s. lateral oscillation at 1 Hz: (a) COP position in the lateral direction; (b) COP velocity in the lateral direction; (c) lateral acceleration: ——— desired acceleration, ——— measured acceleration.

6.3.1. Dependence on the peak magnitude of oscillations

The reported probability of losing balance and the 'discomfort or difficulty' ratings were affected by changes in the peak magnitude of acceleration at 1 Hz (p<0.01, Friedman, Figure 6.3a and Figure 6.3b). They increased with increasing peak magnitude of acceleration (p<0.01, Spearman, Figure 6.3a and Figure 6.3b)

The 'discomfort or difficulty' ratings at 2 Hz were also affected by changes in the peak magnitude of acceleration (p<0.01, Friedman, Figure 6.4a), but the self-reported probability of losing balance was not affected by changes in the peak magnitude of acceleration at 2 Hz (p=0.157, Friedman, Figure 6.4b). The 'discomfort or difficulty' ratings increased with increasing peak magnitude of 2-Hz oscillation (p<0.01, Spearman, Figure 6.4a)

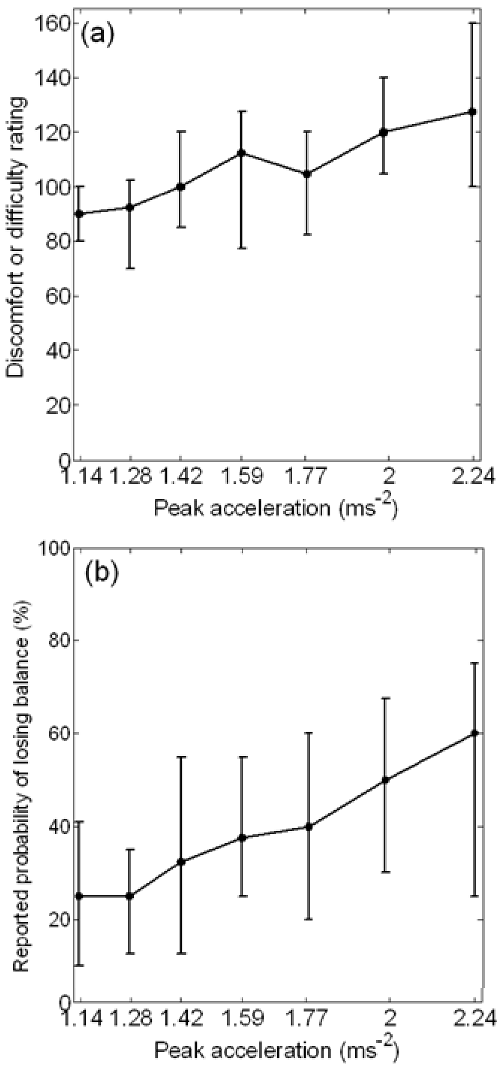


Figure 6.3: (a) 'Discomfort or difficulty' ratings and (b) reported probability of losing balance as a function of peak magnitude of acceleration at 1 Hz.

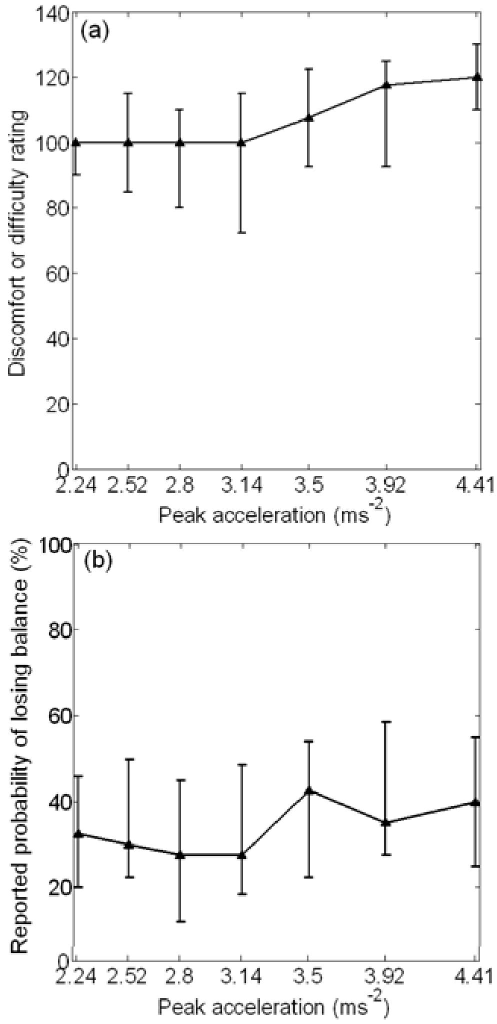


Figure 6.4: (a) 'Discomfort or difficulty' ratings and (b) reported probability of losing balance as a function of peak magnitude of acceleration at 2 Hz.

The peak value of lateral COP velocity was affected by changes in the peak magnitude of acceleration at 1 Hz ($p<0.01$, Friedman, Figure 6.5a). At 1 Hz, the peak value of lateral COP velocity increased as the peak acceleration of the oscillations increased ($p<0.05$, Spearman, Figure 6.5a). Peak COP velocity was not dependent on the peak acceleration of the oscillations at 2 Hz ($p=0.258$, Friedman, Figure 6.5b).

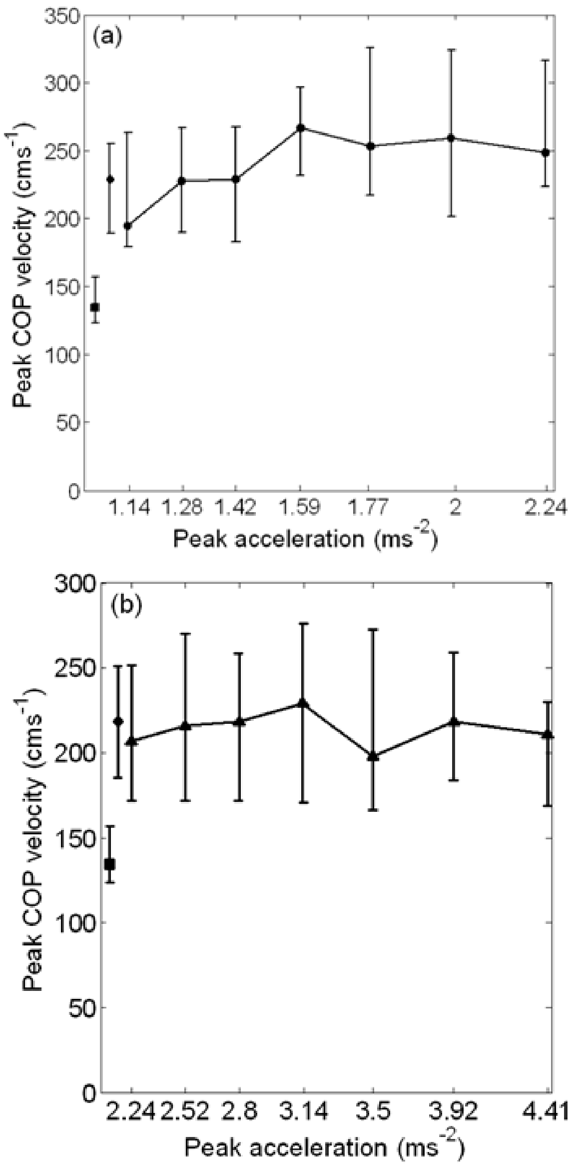


Figure 6.5: (a) Peak lateral COP velocity as a function of peak magnitude of acceleration (a) at 1Hz (b) at 2 Hz:, ■ without oscillation, ◆ during reference oscillation.

The peak value of the lateral COP velocity was less during normal walking without oscillations than during lateral oscillations of 1 Hz and 2 Hz at any peak magnitudes of oscillations (p<0.01, Wilcoxon, Figure 6.5).

6.3.2. Dependence on the r.m.s. magnitude of oscillations

When the oscillations were kept at a constant peak magnitude, the reported probability of losing balance and the 'discomfort or difficulty' ratings were affected by changes in the r.m.s. magnitude of acceleration at 1 Hz and 2 Hz (p<0.01, Friedman, Figure 6.6 and Figure 6.7). The 'discomfort or difficulty' ratings and the reported probability of losing balance increased with increasing r.m.s. magnitude of the acceleration at 1 Hz and 2 Hz (*p*<0.01, Spearman, Figure 6.6 and Figure 6.7)

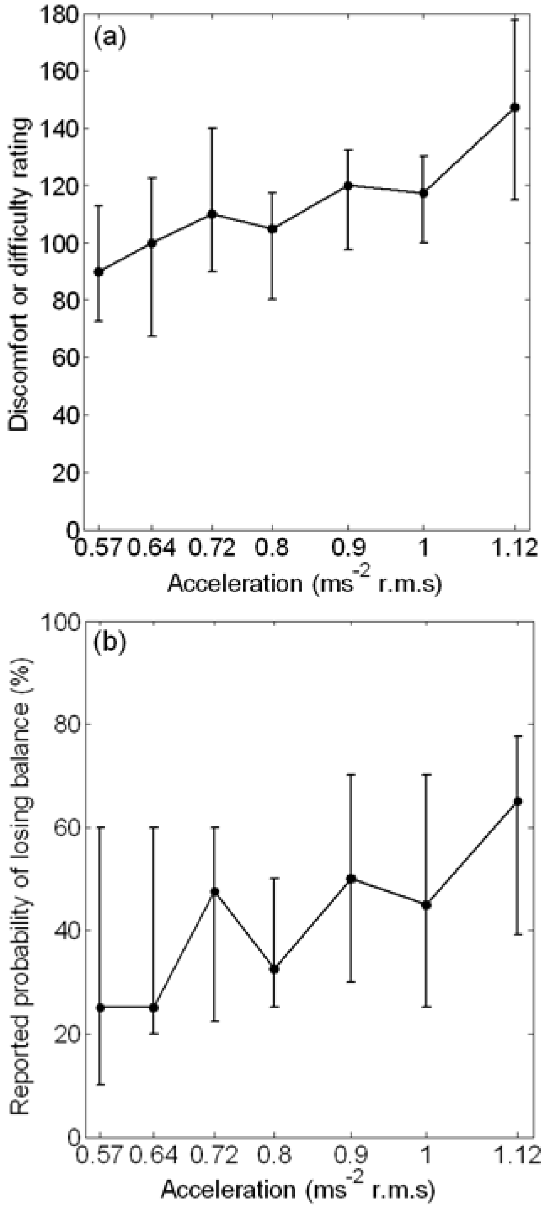


Figure 6.6: (a) ‘Discomfort or difficulty’ ratings and (b) reported probability of losing balance as a function of r.m.s. magnitude of acceleration at 1 Hz.

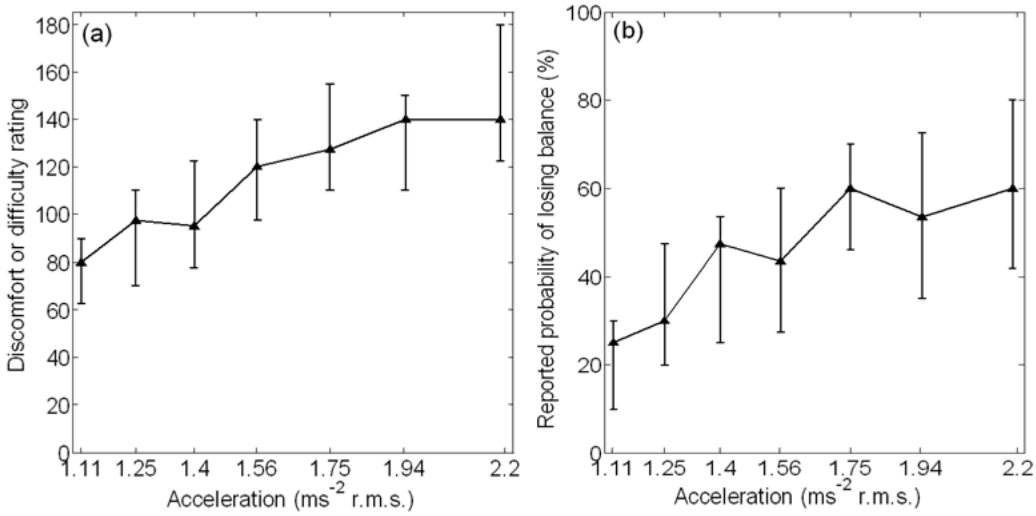


Figure 6.7: (a) 'Discomfort or difficulty' ratings and (b) reported probability of losing balance as a function of r.m.s. magnitude of acceleration at 2 Hz.

The r.m.s. value of lateral COP velocity was dependent on changes in the r.m.s. magnitude of acceleration at 1 Hz and 2 Hz (p<0.01, Friedman). The lateral r.m.s. COP velocity increased with increasing r.m.s. magnitude of acceleration at 1 Hz and 2 Hz (p<0.01, Spearman, Figure 6.8).

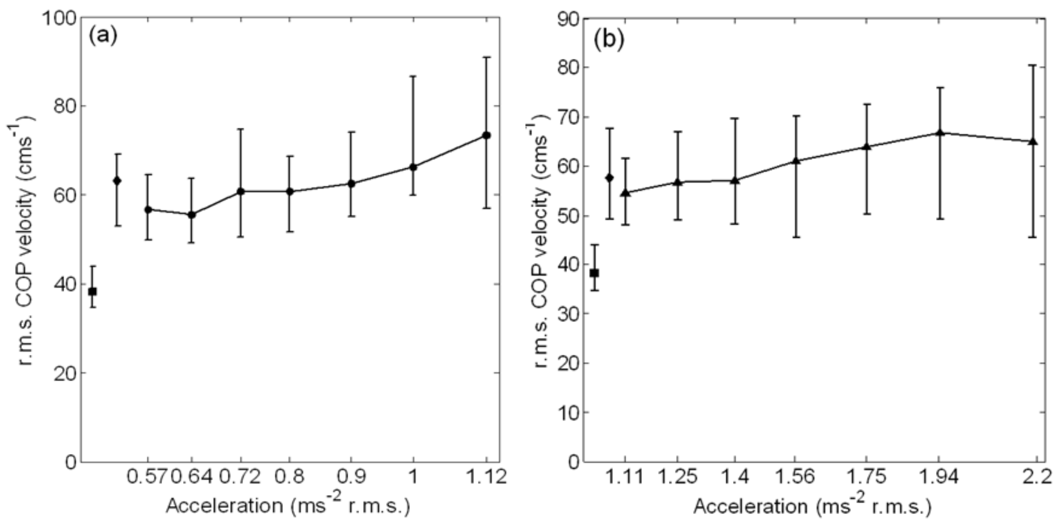


Figure 6.8: Lateral r.m.s. COP velocity as a function of r.m.s. magnitude of acceleration (a) at 1Hz. (b) at 2 Hz: ■ without oscillation ◆ during reference oscillation.

The lateral r.m.s. COP velocity was less during normal walking without oscillation than during lateral oscillations of 1 Hz and 2 Hz at any r.m.s. magnitude of oscillations (p<0.01, Wilcoxon, Figure 6.8).

6.3.3. Association of 'discomfort or difficulty' ratings with the peak and r.m.s. magnitude of oscillation

Associations of 'discomfort or difficulty' ratings with the peak and r.m.s magnitude of oscillation were quantified by the growth of sensation (n_{rms} and n_{peak}) in Steven's power law:

$$\Psi = k * (\Phi_{r.m.s.})^{n_{r.m.s.}} \tag{6.2}$$

$$\Psi = k * (\Phi_{peak})^{n_{peak.}} \tag{6.3}$$

where ψ is the sensation magnitude (i.e. 'discomfort or difficulty' ratings), ϕ_{rms} is the r.m.s. magnitude of the stimulus, ϕ_{peak} is the peak magnitude of the stimulus and k is a constant for a given stimulus. The constants n_{peak} and n_{rms} are the rate of growth of sensation with respect to the peak and r.m.s. magnitude of the stimulus.

The constants n_{peak} and n_{rms} were obtained by linear regression of the individual's reported 'discomfort or difficulty' ratings and the peak and r.m.s. magnitudes of acceleration, respectively. The rate of growth of sensation with respect to the r.m.s. magnitude of 1-Hz oscillation (i.e. n_{rms}=0.597) was not significantly different from the rate of growth of sensation with respect to the peak magnitude of 1-Hz oscillation (i.e. n_{peak}=0.508) (p=0.852, Wilcoxon, Table 6.2).

Table 6.2: Estimated values of growth of sensation at 1 Hz.

	n_{rms}	n_{peak}
Subject 1	0.590	0.801
Subject 2	0.128	0.511
Subject 3	0.800	0.405
Subject 4	0.783	1.076
Subject 5	0.448	1.600
Subject 6	0.438	0.998
Subject 7	0.051	1.226
Subject 8	1.128	-0.642
Subject 9	0.346	0.233
Subject 10	0.638	-0.172
Subject 11	0.265	-0.177
Subject 12	0.541	0.355
Subject 13	0.939	0.505
Subject 14	0.347	0.147
Subject 15	0.604	1.155
Subject 16	0.814	0.147
Subject 17	1.084	0.860
Subject 18	0.128	0.579
Subject 19	1.126	0.486
Subject 20	0.775	0.992
MEDIAN	**0.597**	**0.508**

The rate of growth of sensation with respect to the r.m.s. magnitude of 2 Hz oscillations (i.e. n_{rms}=0.844) was significantly greater than the rate of growth of sensation with respect to the peak magnitude of 2 Hz oscillations (i.e. n_{peak}=0.270) (p<0.001, Wilcoxon, Table 6.3).

Table 6.3: Estimated values of growth of sensation at 2 Hz.

	n_{rms}	n_{peak}
Subject 1	1.032	0.345
Subject 2	0.796	0.470
Subject 3	1.451	0.940
Subject 4	0.751	0.611
Subject 5	0.893	0.251
Subject 6	1.574	-0.128
Subject 7	1.828	0.083
Subject 8	1.571	0.232
Subject 9	0.299	0.261
Subject 10	0.740	0.147
Subject 11	0.466	-0.035
Subject 12	1.137	0.580
Subject 13	2.226	0.839
Subject 14	0.617	0.280
Subject 15	1.109	0.298
Subject 16	0.470	0.615
Subject 17	0.090	-0.220
Subject 18	0.699	0.053
Subject 19	1.008	0.152
Subject 20	0.717	0.396
MEDIAN	**0.844**	**0.270**

6.3.4. Postural stability in response to oscillations at a specific r.m.s. velocity

Crest factors (peak velocity/r.m.s. velocity) based on the velocity of the lateral oscillation were calculated by integrating the acceleration measurements (Table 6.4).

When the peak velocity of 1-Hz and 2-Hz lateral oscillations were kept constant, the reported probability of losing balance decreased with increasing crest factor (p<0.01, Spearman, Figure 6.9a) due to the decreased r.m.s. magnitude of oscillations. At each crest factor, the lateral oscillations of 1 Hz and 2 Hz had similar r.m.s. velocities (Table 6.4).

Table 6.4: Peak and r.m.s. velocity of lateral oscillations and their crest factors (CF).

1 Hz-part A			1 Hz-part B		
r.m.s. velocity [ms^{-1}]	peak velocity [ms^{-1}]	CF	r.m.s. velocity [ms^{-1}]	peak velocity [ms^{-1}]	CF
0.12	0.21	1.7	0.09	0.30	3.08
0.12	0.22	1.91	0.11	0.30	2.77
0.12	0.24	2.03	0.12	0.30	2.48
0.12	0.27	2.26	0.13	0.30	2.34
0.12	0.29	2.48	0.15	0.30	2.05
0.12	0.34	2.84	0.16	0.31	1.86
0.12	0.38	3.14	0.18	0.32	1.74
2 Hz-part A			2 Hz-part B		
r.m.s. velocity [ms^{-1}]	peak velocity [ms^{-1}]	CF	r.m.s. velocity [ms^{-1}]	peak velocity [ms^{-1}]	CF
0.11	0.2	1.8	0.09	0.29	3.18
0.11	0.22	1.93	0.10	0.28	2.84
0.11	0.24	2.13	0.11	0.29	2.57
0.11	0.27	2.36	0.12	0.29	2.32
0.11	0.29	2.6	0.14	0.30	2.14
0.11	0.33	2.86	0.15	0.29	1.90
0.11	0.36	3.26	0.18	0.30	1.72

Probability of losing balance reported at the two frequencies were found to be similar with all crest factors when the oscillations were applied at similar r.m.s. velocities ($p>0.05$, Wilcoxon, Figure 6.9a). The lateral r.m.s. COP velocities measured when exposed to 1-Hz and 2-Hz oscillations of similar r.m.s. velocities were also found to be similar at all crest factors ($p>0.05$, Wilcoxon, Figure 6.9b).

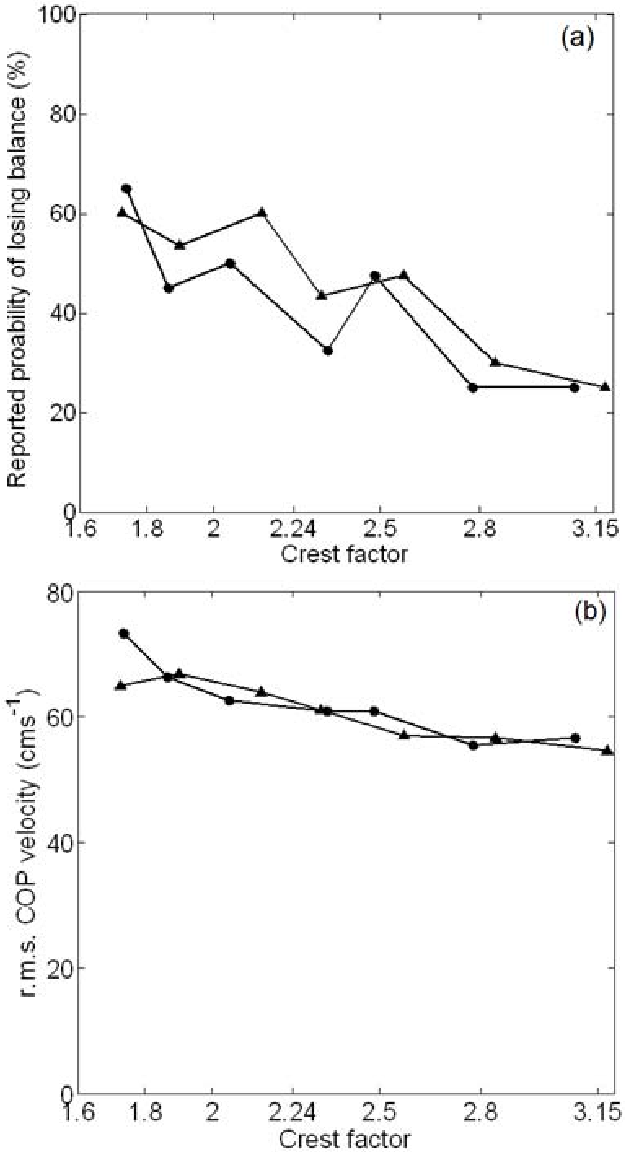


Figure 6.9: (a) Reported probability of losing balance and (b) lateral r.m.s. COP velocity as a function of crest factor when the oscillations were kept at constant peak magnitude: —●— at 1 Hz , —▲—at 2 Hz.

6.4. Discussion

Griffin and Whitham (1980) exposed sitting subjects to 8 Hz vertical complex time varying motions with various crest factors ranging from 2.12 to 8.51. Among the motions having the same r.m.s magnitude, subjects reported greater discomfort for the motions having higher peak levels. Howarth and Griffin (1991) also reported increased discomfort with increasing crest factor of oscillations although the r.m.s value of the oscillations was kept constant. Similar findings are obtained in the current study for walking subjects: when the oscillations were applied at a specific r.m.s. magnitude: ‘discomfort or difficulty’ ratings increased with increased peak value of 1-Hz and 2-Hz oscillations (Figure 6.3 and Figure 6.4). The association of discomfort with the peak magnitude of oscillation (n_{peak}) was similar with the association of discomfort with the r.m.s. magnitude of oscillation (n_{rms}) at 1 Hz (Table 6.2), indicating that both the peak and the r.m.s. have similar effects on 'discomfort or difficulty' ratings at 1 Hz. The association of the discomfort with the r.m.s. magnitude of the oscillation was found to be significantly greater than the association with the peak magnitude when exposed to 2 Hz oscillations (Table 6.3). The r.m.s value may therefore be a better measure of predicting the 'discomfort or difficulty' in a walking task when exposed to 2 Hz oscillations than when exposed to 1 Hz lateral oscillations.

Thuong and Griffin (2010b) suggested that the r.m.s. is not optimum for evaluating all types of waveforms so as to predict the discomfort of standing subjects exposed to random or transient whole-body vibrations at 1 Hz and 8 Hz. They found an exponent greater than 2 was required, although this exponent may depend on the frequency and the direction of vibration. The optimal exponent was about 3 for 1 Hz, and in the range 3 to 4 for 8 Hz vibration. The current study was conducted on walking subjects and was not designed to propose a specific method for evaluating waveforms, but the findings suggest that the r.m.s. by itself is not sufficient to predict the discomfort when walking and exposed to low frequency lateral oscillations.

The reported probability of losing balance and the peak lateral COP velocity increased with increasing peak levels of the oscillations when walking subjects were exposed to 1 Hz oscillations of same r.m.s. velocity (Figure 6.5a). Although there was a slight effect of peak magnitude on the 'discomfort or difficulty' ratings when subjects were exposed to 2 Hz oscillations, the reported probability of losing balance and the lateral r.m.s. COP velocity were not affected by changes in the peak magnitude of 2 Hz oscillations (Figure 6.5b). The current findings suggest that postural stability is sensitive to the peak magnitude of 1-Hz oscillations but not the peak magnitude of 2-Hz oscillations. The peaks in lateral oscillation might create postural stability problems as walking subjects were required to take a fast and sufficient corrective postural action to overcome the destabilizing effect of an unpredictable peak.

When the oscillations were kept at a constant peak magnitude, the reported probability of losing balance and the lateral r.m.s. COP velocity increased with increasing r.m.s. magnitude of oscillations at 1 Hz and 2 Hz (Figure 6.6, Figure 6.7 and Figure 6.8). Increased r.m.s. COP velocity is an indication of wider and faster steps. At higher r.m.s. magnitudes of oscillation, the risk of fall increases and walking subjects adopt stepping strategies by adjusting the placement and timing of successive steps (Nashner, 1980) to overcome the effects of external perturbations. The lateral r.m.s. COP velocity being higher during exposure to lateral oscillations than during normal walking without oscillations is also an indication of stepping strategies being adopted in response to lateral perturbations. McAndrew *et al*. (2010) also showed that walking people took wider and faster steps during continuous random oscillations than during normal walking without oscillations.

The oscillations at two different frequencies but with similar r.m.s. velocity magnitudes caused similar lateral r.m.s. COP velocity and similar probability of losing balance (Figure 6.9). These findings are consistent with the results of the first experiment. The findings of the current study also suggest that probability of losing balance cannot be predicted solely from the r.m.s. velocity of low frequency oscillations and suggest the peak velocity should also be considered.

Seated people exposed to one-third octave random vibration and sinusoidal vibration in the range 3.15 to 20 Hz have been found to be more sensitive to the random vibration at 10 Hz and 12.5 Hz (Griffin, 1976). Seated people exposed to sinusoidal vibration and narrow-band random motion were also found to be more sensitive to random vibration, with the difference decreasing with increasing frequency (Donati *et al.*, 1983). When the oscillations were applied at the same r.m.s. velocity in the current study, walking subjects showed more sensitivity to random vibration than sinusoidal vibration. The sensitivity to random stimuli might be related to the more peaky characteristics and unpredictability associated with random motions having higher crest factors compared to sinusoidal stimulus (Figure 6.1). Prediction may result in adaptation to the imposed stimuli and may cause a shift to a predictive control strategy (Maki, 1986).

6.5. Conclusion

The ‘discomfort or difficulty’ in walking task was affected by the peak magnitude of lateral oscillations with various waveforms, especially with 1-Hz oscillations even when the r.m.s. magnitude of the oscillations was unchanged. When exposed to lateral oscillations of the same r.m.s. magnitude, postural stability assessed by subjectively reported probability of losing balance and lateral COP velocity was found to be dependent on the peak magnitude of the oscillations at 1 Hz but not at 2 Hz. The r.m.s. value may therefore not be the optimum method of predicting the postural stability of walking subjects exposed to lateral oscillations, especially with low frequency motions.

The findings of the study emphasize the importance of considering not only the r.m.s. but also the peak magnitude of oscillations when seeking to eliminate postural stability problems in transport. It is also appropriate that the waveform characteristics, including peak and r.m.s. magnitudes, of platform perturbations should be fully reported in perturbed balance experiments so as to allow comparison and interpretation of experimental findings. If motion velocity is used to quantify the severity of a 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.

Chapter 7

Effect of subject characteristics on the postural stability of walking people perturbed by lateral oscillatory motion

7.1. Introduction

Age is one of the most significant subject characteristics the effect of which has been commonly investigated on human balance. Age has been shown to have deteriorating effects on postural stability due to the reduction in ability to sense and actuate movement with ageing. The number of afferent and efferent channels and the quality of the signals transmitted through these channels degrade with age (Horak *et al*., 1999; Alexander, 1996).

Postural sway amplitude has been found to be higher in the medio-lateral direction in the elderly during quiet stance and larger reductions in medio-lateral sway amplitude were observed when touching a stationary support (Tremblay *et al*., 2004). These findings are consistent with previous findings of lateral instability problems in the elderly (Maki and McIllroy, 1996). Balance performance during one-legged stance was found to be significantly related to age, gender, stature and body weight (Balogun *et al*., 1994). Postural stability parameters evaluated via center of pressure measurements during quiet standing were also found to be affected by stature, weight, foot width, and base of support area (Chiari *et al*., 2002). Among the physical characteristics of the body (including weight, stature, age, foot length, waist and hip circumferences), body weight was found to be the most significant predictor of postural instability during quiet standing (Hue *et al,.* 2007). Postural instability associated with obesity has also been previously reported (Goulding *et al*., 2003; Chiari *et al*., 2002).

Most studies of aging and balance have been focused on postural stability during quiet standing. The ability to keep postural stability requires more rapid and accurate postural reactions to recover from real life challenges to perturbations, such as slips, trips, or external disturbances experienced in transport. Stepping is a significant reaction to overcome these perturbations even if the perturbation is relatively small (Maki and McIlroy, 1999). The effect of aging on compensatory stepping reactions to lateral perturbation during standing and walking in place (i.e., walking on the spot) has been previously reported by Maki *et al.* (2000). Studies of the effect of age on gait revealed that older adults have lower gait speeds and step frequencies, higher step width and increased gait variability (Moe-Nilssen and Helbostad, 2005; Hausdorff *et* *al*., 2001; Owings and Grabiner, 2004). The effects of a wide range of subject characteristics including age, gender, weight, and stature on postural stability during perturbed locomotion have not previously been reported.

The stability thresholds of walking subjects exposed to lateral oscillatory motion, as reported in Chapter 4, were obtained with healthy male subjects aged 25-45 years. Whether these threshold values are applicable to females, or older males, is unknown. The objective of the final experiment reported in this chapter was to investigate the effect of subject characteristics (age, gender, weight, stature, shoes width, walking speed, and fitness level) on the postural stability of walking subjects perturbed by lateral oscillations of the same type used in the first experiment. It was also aimed to investigate short-term learning: the effect of repetitions of the stimulus on subjective and objective measures of postural stability.

It was hypothesized that among all the subject characteristics, age and weight were the most significant predictors of postural stability, with older and heavier people being less stable. It was also expected that the reported probability of losing balance and objective measures of postural stability would be dependent on the number of repetitions of the lateral oscillation.

7.2. Method

7.2.1. Subjects

One hundred healthy adult subjects (50 males, 50 females) aged 18 to 69 years participated in the study (Table 7.1). Subjects completed a questionnaire to exclude those with relevant disorders or using drugs that might affect postural stability. 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.

Table 7.1: Subject characteristics (mean, standard deviation (s.d.) and range values reported).

	All subjects (50 M, 50 F)			Age Group 1 (12 M, 13 F)		
	Mean	s.d.	Range	Mean	s.d.	Range
Age, years	42.62	17.06	18-69	21.16	2.01	18-24
Weight, kg	71.62	14.19	40.85-109.6	69.26	15.07	49.1-107
Stature, cm	170.51	9.17	150-200	172.16	9.53	156-196
Shoes width, cm	10.42	0.92	8-12	10.31	1.03	8-12
Walking speed, ms^{-1}	0.71	0.06	0.53-0.84	0.74	0.05	0.58-0.84
Fitness score	2398.29	2082.25	33-8940	2244.98	1881.69	318-8730

	Age Group 2 (13 M, 12 F)			Age Group 3 (13 M, 13 F)			Age Group 4 (12 M, 12 F)		
	Mean	s.d.	Range	Mean	s.d.	Range	Mean	s.d.	Range
Age, years	33.24	6.19	25-44	52.62	4.08	46-59	63.92	2.70	60-69
Weight, kg	69.56	16.18	40.85-109.6	73.39	12.33	46.7-93.7	74.30	13.00	51.25-95
Stature, cm	170.16	11.91	150-200	170.92	6.69	159-186	168.71	7.99	155-186
Shoes width, cm	10.20	1.15	8.2-12	10.63	0.73	9.2-12	10.53	0.64	9.4-11.5
Walking speed, ms^{-1}	0.71	0.07	0.54-0.84	0.70	0.07	0.54-0.83	0.70	0.07	0.53-0.79
Fitness score	1500.96	1184.97	66-4158	2749.19	2397.32	198-8940	3112.54	2384.00	33-8316

7.2.2. Apparatus

The experimental apparatus was the same as used in the first experiment. For details, see Section 4.2.2.

7.2.3. Experimental Procedure

While walking on the treadmill, subjects were perturbed by lateral oscillations. The stimuli, 4.5 cycles of sinusoidal motion modulated by a half sine envelope, were the same type of stimuli used in the first experiment (Section 4.2.3).

The motions were presented at 0.08 ms^{-1} r.m.s., at each of three frequencies (0.5, 1.0 and 2.0 Hz), which resulted in accelerations of 0.25, 0.5 and 1.0 ms^{-2} r.m.s., respectively. The frequencies and magnitudes were chosen based on the previously reported stability thresholds of walking subjects (Figure 4.5b) and preliminary experimentation considering the safety and comfort of walking subjects especially the older adults.

Subjects were asked to walk on the treadmill at a comfortable constant walking speed throughout the experiment. The speed of the treadmill was selected by the individuals such that they walked at their preferred comfortable walking speed (0.54 ms^{-1} to 0.84 ms^{-1}). Subjects were given 5 minutes of walking prior to the start of the experiment to get accustomed to treadmill walking.

While subjects walked on the treadmill, the lateral oscillatory motions were applied at random unpredictable times. After each oscillatory motion, subjects were asked to report their probability of losing balance caused by the motion by answering the same question asked to the participants of the first experiment reported in Chapter 4:

‘‘What is the probability that you would lose balance if the same test motion were repeated?’’

Subjects were encouraged to grasp the handrails of the treadmill if it was necessary. Losing balance was defined as attempting to take protective action not to fall – such as taking a protective step, or grasping an object to regain equilibrium.

The 1.0-Hz oscillations were repeated seven times to investigate short-term learning. The 0.5-Hz oscillations followed the 1-Hz oscillations and were repeated three times. The 2-Hz oscillations following the 0.5-Hz oscillations were also repeated three times. The experiment together with the body measurements, fitness questionnaire, and walking trial lasted 30 minutes.

Together with the subjective data, the acceleration of the moving platform and gait data (i.e. ground reaction forces under the feet) were gathered. Gait data during walking normally without oscillations were also gathered seven times for 10 seconds after 5 minutes of walking prior to the start of the perturbations to investigate the repeatability of gait measures without oscillations.

7.2.4. Analysis

The dependent variables used to assess postural stability were the reported probability of losing balance, objective gait measures, and grasping strategy, a categorical variable, which was coded as ‘one’ when subjects were observed to grasp the handrail of the treadmill and ‘zero’ otherwise. Peak-to-peak lateral COP position, lateral r.m.s. COP velocity, r.m.s. vertical ground reaction force under the feet, and mean COP speed were used as the objective gait measures of postural stability. Peak-to-peak COP position is an indication of the range of lateral COP displacement, and lateral r.m.s. COP velocity is an indication of the timing of foot placement in the lateral direction. The total r.m.s. vertical ground reaction force under the feet was normalized with respect to the weight of each subject and is an indication of loading-unloading strategies employed by a subject. The mean COP speed is defined as the cumulative distance of the COP over the sampling period, indicating the amount of physical activity required to maintain stability during quiet standing (Geurts *et al*., 1993; Hue *et al*, 2007, Figure 7.1c).

The centre of pressure (COP) time histories was determined by processing the force time-histories (from the eight force sensors in the treadmill) during each motion (Section 3.2.2.1). The COP velocity in the lateral direction was obtained by differentiating the lateral COP position after filtering the centre of pressure position using low-pass Bessel filter at 8 Hz.

Independent variables were the number of repetition of oscillations and subject physical characteristics (age, gender, weight, stature, shoe width, walking speed and fitness score). Fitness score was evaluated by the International Physical Activity Questionnaire (IPAQ, 2002).

The subjective and objective measures of postural stability were then used to test the hypothesis regarding their dependence on the number of repetition of oscillations and subject physical characteristics. Parametric statistical methods were used for the data analysis using SPSS (PASW statistics, version 17.0). Repeated measures ANOVA (RANOVA) was used to test for differences between multiple conditions (i.e. repetitions of the same stimulus). Post hoc analysis with Bonferroni adjustment was carried out for multiple pairwise comparisons. A Greenhouse-Geisser correction was used with RANOVA tests when the assumption of homogeneity of covariances was violated.

The paired samples t-test was used to compare dependent variables between conditions (i.e. frequencies, with and without oscillation). The independent samples t-test was used to compare dependent variables between subjects grouped by their characteristics (i.e. age and gender). The Cohran Q test was used to test for differences in the categorical dependent variables (i.e. number of people who developed grasping strategy) between multiple conditions (i.e. repetitions of the same stimulus) and McNemar change test was used to test for differences between pairs of conditions.

Multiple regression was used to identify significant predictors, drawn from subject characteristics (i.e. age, gender, weight, stature, shoe width, walking speed, fitness level) of postural stability. Logistic regression was used to identify significant predictors of grasping (i.e. whether or not subjects grasped the hand support to maintain balance). For each test condition (i.e. three frequencies of oscillation and without oscillation) all the predictor variables were entered into the multiple regression model using the PASW stepwise procedure (PASW statistics, version 17.0). A significance level of 0.05 was used to enter and retain a variable in the model. Associations between predictor variables were checked by collinearity diagnostics.

Data transformations of dependent variables were used to explore and correct any effects of non-normality in their distributions. Regression analyses using, initially non-transformed data, and subsequently transformed data using Box-Cox transformation, were found to produce almost identical results in terms of the statistical strength of associations. By retaining the variables in their original units the interpretation of the results is made easier.

7.3. Results

An example of the COP position of a subject exposed to 1.0-Hz lateral oscillation at 0.5 ms^{-2} r.m.s. is shown as a function of time in Figure 7.1a. The COP position shows the lateral location of the resultant of the ground reaction forces and is indicative of lateral foot placement. The COP velocity indicates the rate of change of COP position (Figure 7.1b). Figure 7.1c shows the COP in the lateral and fore-and-aft direction as an indication of the walking path followed by the subject.

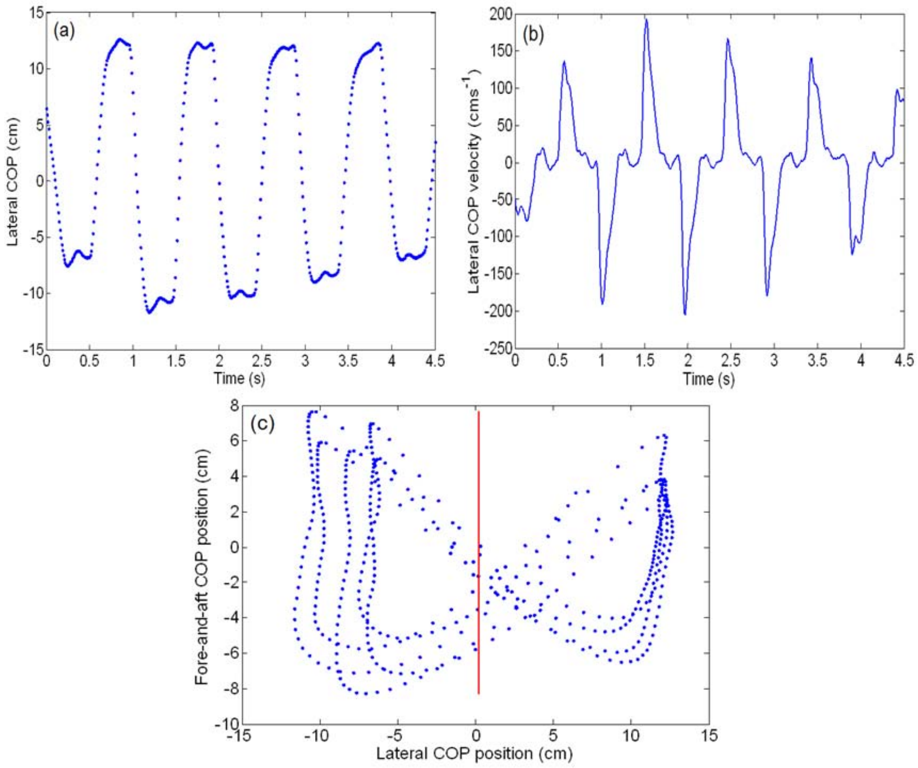


Figure 7.1: Example centre of pressure (COP) for a subject walking while exposed to 0.5 ms^{-2} r.m.s. lateral oscillation at 1 Hz: (a) lateral COP position as a function of time; (b) lateral COP velocity as a function of time; (c) COP path in the fore-and-aft and lateral direction.

7.3.1. Repeated measures of postural stability: short-term learning effect

7.3.1.1. Repeated subjective measures: reported probability of losing balance

The reported probability of losing balance was slightly affected by repeating the 1-Hz stimulus seven times (p=0.024, RANOVA, Figure 7.2a). However, post hoc tests using the Bonferroni correction revealed no significant differences between any of the seven repetitions (p>0.2, t-test).

The reported probability of losing balance did not differ between three repetitions of the 0.5-Hz stimulus (p=0.276, RANOVA, Figure 7.2b).

The reported probability of losing balance differed significantly between three repetitions of the 2-Hz stimulus (p<0.001, RANOVA, Figure 7.2c). The reported probability of losing balance in response to the first stimulus was significantly greater than that in response to the third stimulus (p=0.001, t-test). Subjects reported greater probability of losing balance when exposed to the second stimulus than to the third stimulus (p=0.02, t-test).

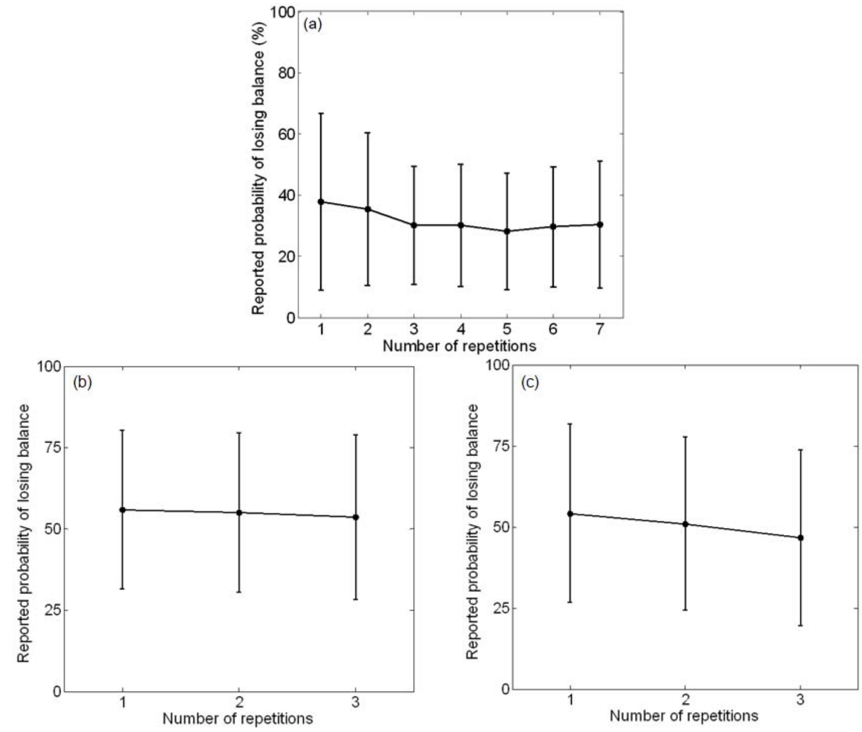


Figure 7.2: Reported probability of losing balance (means and standard deviations) as a function of the number of repetitions of the (a) 1-Hz (b) 0.5-Hz (c) 2-Hz stimuli.

7.3.1.2. Repeated subjective measures: grasping from the hand support

Grasping was significantly affected by repeating the stimulus at each frequency (p<0.01, Cochran Q test, Table 7.2). Grasping was employed by more subjects when exposed to 1-Hz oscillation for the first time than when exposed subsequently (p<0.01, McNemar). There were no significant differences in the percentage of people who held on the hand support between any of the exposures following the first exposure (p>0.05, McNemar).

Table 7.2: Percentage of subjects (%) who held on the hand support during exposure to lateral oscillation as a function of number of repetitions of the stimulus.

	Number of repetitions of stimulus						
Frequency (Hz)	1	2	3	4	5	6	7
1.0	28%	9%	3%	4%	3%	7%	5%
0.5	25%	22%	13%				
2.0	14%	2%	3%				

More subjects grasped the hand support when exposed to 0.5-Hz oscillation for the first and the second time than for the third time (p<0.05, McNemar, Table 7.2), but there were no significant differences in the number of people who held on the support between the first and second exposures (p=0.581, McNemar).

More subjects grasped the hand support during their first exposure to 2-Hz oscillation than during their second and third exposures ($p<0.01$ McNemar, Table 7.2), but there was no significant difference between the second and third exposure ($p=1.00$, McNemar).

7.3.1.3. Repeated gait measures: short-term learning effect

This section concerns the effect of the number of repetitions of lateral oscillations on overall gait measures across the 100 subjects. Figure 7.3, Figure 7.4 and Figure 7.5 also show the trends for different age groups. The effect of age on gait measures will be fully analyzed later in Section 7.3.3.3.

7.3.1.3.1. Repeated gait measures during exposure to 1-Hz stimuli

The peak-to-peak lateral COP position differed between the seven repetitions of the 1-Hz oscillation ($p=0.025$, RANOVA, Figure 7.3a); the peak-to-peak lateral COP during the fifth exposure was significantly less than during the first exposure ($p=0.038$, t-test).

There was a decreasing trend in r.m.s. COP velocity with increasing number of repetitions of 1.0-Hz stimuli, especially for the youngest age group (Figure 7.3b) that showed significantly higher r.m.s. COP velocity during the first exposure than during all other exposures ($p<0.001$, t-test). However, the overall lateral r.m.s. COP velocity was not significantly different between the seven repetitions ($p=0.124$, RANOVA, Figure 7.3b).

The normalized r.m.s. vertical ground reaction force under the feet differed between the seven repetitions ($p<0.001$, RANOVA, Figure 7.3c). The vertical r.m.s. ground reaction force in response to the first stimulus was greater than during all other exposures ($p<0.001$, t-test). The r.m.s. ground reaction force during the second exposure was also greater than during each of the subsequent five exposures ($p<0.02$, t-test). There were no significant differences in ground reaction forces between the third, fourth, fifth, sixth and seventh presentation of the stimulus ($p>0.05$, t-test).

The mean COP speed was also affected by repetitions of the 1-Hz oscillation ($p<0.001$, RANOVA, Figure 7.3d). The mean COP speed during the first exposure did not differ from that during the second exposure but was significantly greater than the mean COP speed during all other exposures ($p<0.05$, t-test).

See Table C.46 in Appendix C.3 for the tabulated values of mean and standard deviations of four objective measures reported for each age group as a function of number of repetitions of 1 Hz stimuli.

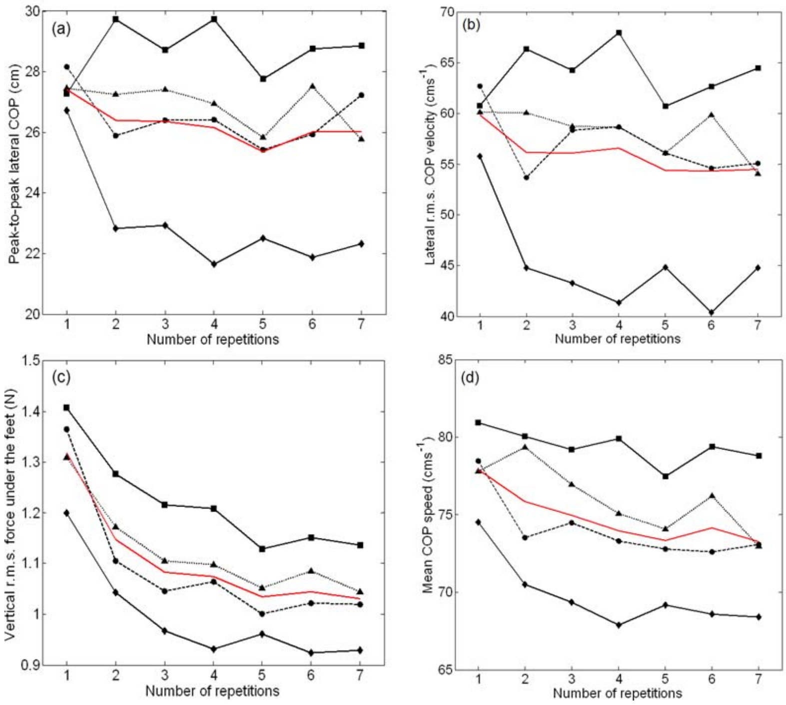


Figure 7.3: Effect of number of repetitions of 1-Hz stimuli on the COP measures: (a) peak-to-peak lateral COP position (b) lateral r.m.s. COP velocity (c) r.m.s. vertical ground reaction force under the feet (d) mean COP speed: age group 1 (18-24), age group 2 (25-45), age group 3 (46-59), age group 4 (60-70), overall.

7.3.1.3.2. Repeated gait measures during exposure to 0.5-Hz stimuli

The peak-to-peak lateral COP position differed between three repetitions of the 0.5-Hz oscillation (p <0.001, RANOVA, Figure 7.4a). The peak-to-peak lateral COP position during the first exposure was greater than during the second and third exposures (p<0.001, t-test) and greater during the second exposure than during the third exposure (p<0.01, t-test).

The lateral r.m.s. COP velocity differed between the three repetitions (p<0.001, RANOVA, Figure 7.4b). The r.m.s. COP velocity in response to the first stimulus was greater than the r.m.s. COP velocities during the second and the third exposures (p<0.01, t-test). However, the r.m.s. COP velocity during the second exposure was not different from that during the third exposure (p=0.942, t-test).

The vertical r.m.s. ground reaction force under the feet differed between the three exposures (p<0.001, RANOVA, Figure 7.4c) showing a similar trend to the r.m.s. COP velocity. The vertical r.m.s. force under the feet during first exposure was greater than during the second and third exposures (p<0.01, t-test). There was no significant difference between the second and third exposures (p=0.423, t-test)

The mean COP speed also differed between the three exposures to the 0.5-Hz stimuli (p<0.001, RANOVA, Figure 7.4d). The mean COP speed during the first presentation of the stimulus was greater than during the second and third presentation of the stimulus ($p<0.01$, t-test). There was no significant difference in the mean COP speed between the second and third exposures ($p=1.0$, t-test)

See Table C.47 in Appendix C.3 for the tabulated values of mean and standard deviations of four objective measures reported for each age group as a function of number of repetitions of 0.5 Hz stimuli.

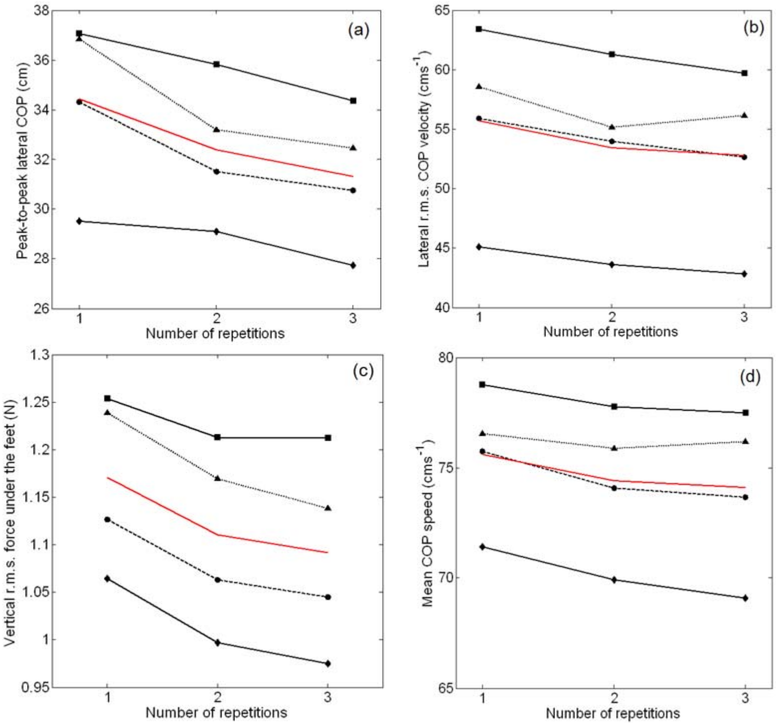


Figure 7.4: Effect of the number of repetitions of 0.5-Hz stimuli on the COP measures: (a) peak-to-peak lateral COP position (b) lateral r.m.s. COP velocity (c) r.m.s. vertical ground reaction force under the feet (d) mean COP speed: age group 1 (18-24), age group 2 (25-45), age group 3 (46-59), age group 4 (60-70), overall.

7.3.1.3.3. Repeated gait measures during exposure to 2-Hz stimuli

The peak-to-peak lateral COP position did not differ between three repetitions of the 2-Hz oscillation ($p=0.386$, RANOVA, Figure 7.5a).

The lateral r.m.s. COP velocity differed between the three exposures to the 2-Hz stimuli ($p=0.013$, RANOVA, Figure 7.5b). r.m.s. The COP velocity during the first exposure was significantly greater than during the third exposure ($p=0.015$, t-test). The r.m.s. COP velocity during the second exposure did not differ from that during the first and third exposures ($p>0.3$, t-test).

The vertical r.m.s. ground reaction force under the feet differed between the three repetitions ($p<0.001$, RANOVA, Figure 7.5c), indicating a decreasing force with increasing repetition of the 1-Hz stimulus. The ground reaction force in response to the first stimulus was greater than the ground reaction forces in response to the second and third stimuli (p<0.001, t-test). The ground reaction force during the second exposure was also greater than during the third exposure (p<0.001, t-test).

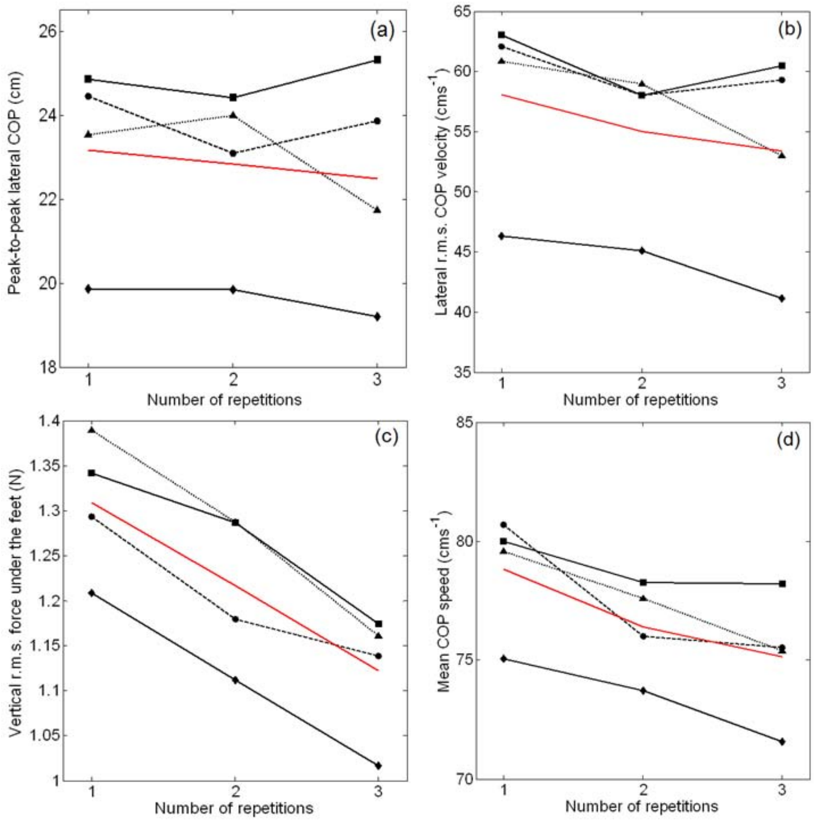


Figure 7.5: Effect of number of repetitions of the 2-Hz stimuli on the COP measures: (a) peak-to-peak lateral COP position (b) lateral r.m.s. COP velocity (c) r.m.s. vertical ground reaction force under the feet (d) mean COP speed: age group 1 (18-24), age group 2 (25-45), age group 3 (46-59), age group 4 (60-70), overall.

The mean COP speed differed between the three exposures to the 2-Hz stimuli (p<0.001, RANOVA, Figure 7.5d). The mean COP speed during the first exposure was greater than during the second and third exposures (p<0.05, t-test), but there was no significant difference in the mean COP speed between the second and third exposures (p=0.209, t-test)

See Table C.48 in Appendix C.3 for the tabulated values of mean and standard deviations of four objective measures reported for each age group as a function of number of repetitions of 2 Hz stimuli.

7.3.1.3.4. Repeated gait measures during normal walking without oscillation

The peak-to-peak lateral COP, r.m.s. COP velocity and mean COP speed did not differ between the seven measurements (p>0.4, RANOVA; Figure 7.6). The vertical r.m.s. ground reaction force under the feet differed between the seven measurements, (p=0.003, RANOVA) but post hoc tests using the Bonferroni correction revealed no significant differences in ground reaction force in pairwise comparisons between the seven measurements (*p*>0.3, t-test, Figure 7.6d). See Table C.49 in Appendix C.3 for the tabulated values of mean and standard deviations of four objective measures reported for each age group.

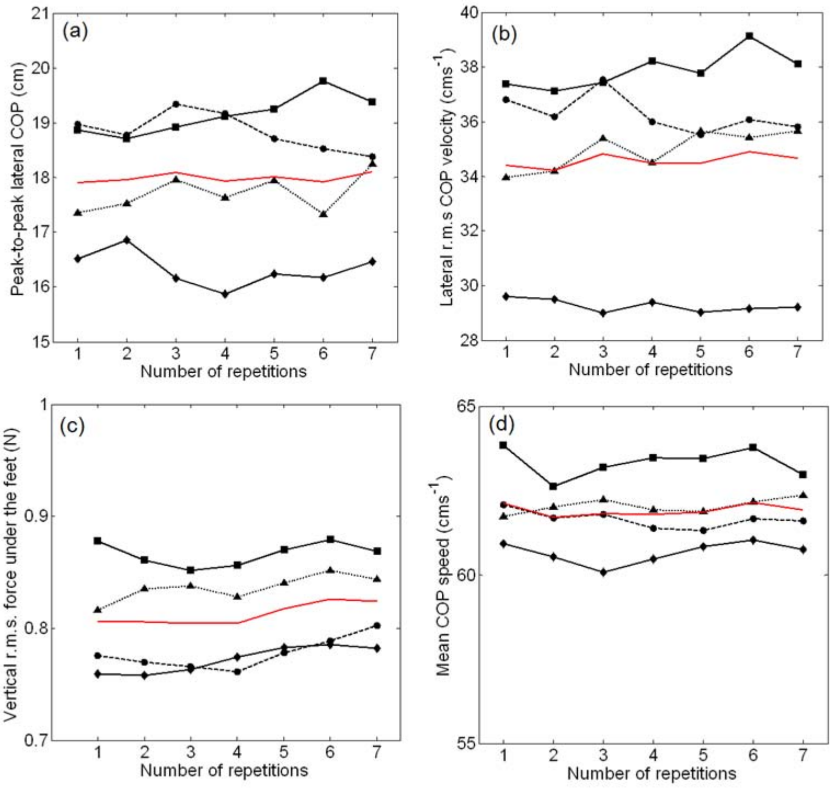


Figure 7.6: Repeated gait measures during normal walking without oscillations: (a) peak-to-peak lateral COP position (b) lateral r.m.s. COP velocity (c) vertical r.m.s. ground reaction force under the feet (d) mean COP speed: age group 1 (18-24), age group 2 (25-45), age group 3 (46-59), age group 4 (60-70), overall.

7.3.3. Effect of subject characteristics: inter-subject variability

7.3.3.1. Effect of subject characteristics on the reported probability of losing balance

The estimated probability of losing balance as a result of 2-Hz oscillation reduced with repetition of the stimulus (Figure 7.2c). A similar decreasing trend, although not statistically significant, was also observed with 1-Hz oscillation (Figure 7.2a). The effects of subject characteristics on the reported probability of losing balance were therefore investigated for subject estimates of the probability of losing balance during the third exposure to the stimuli at each frequency, so that the analyses would be comparable between frequencies.

Stepwise multiple regression analysis revealed that none of the subject characteristics entered into the regression model was a significant predictor of reported probability of losing balance at any frequency. Independent samples t-test also confirmed that there was no significant difference in the reported probability of losing balance by females and males at any frequency (p>0.2, Table 7.3). There was also no significant difference in the reported probability of losing balance between different age groups at any frequency (p>0.05, one way ANOVA, Table 7.3).

Table 7.3: Reported probability of losing balance (mean and standard deviations) for each gender and age groups at each frequency of oscillation.

Mean and (standard deviations)			
Gender	1Hz	0.5 Hz	2 Hz
Female	32.14 (20.81)	48.56 (19.29)	48.56 (25.50)
Male	28.14 (17.83)	45.16 (21.21)	45.16 (27.20)
Age groups	1Hz	0.5 Hz	2 Hz
G1 (18-24)	29.08 (17.01)	43.16 (20.47)	41.8 (25.4)
G2 (25-45)	29.60 (19.29)	41.04 (21.5)	43.84 (26.7)
G3 (46-59)	30.38 (18.70)	46.07 (18.5)	42.5 (24.42)
G4 (60-70)	31.54 (23.30)	45.08 (20.4)	41.5 (28.85)
Overall	30.15 (19.58)	43.84 (20.22)	42.41(26.34)

7.3.3.2. Effect of subject characteristics on grasping from the hand support

Since the number of subjects grasping the hand support to maintain their balance decreased with increasing repetition of the stimulus (Table 7.2), the effects of subject characteristics on grasping were investigated for the first exposure to the stimulus at each frequency. Backward stepwise logistic regression was employed to obtain the subject characteristics having the most significant effect at each frequency on whether a subject grasped the hand support. The significant predictor variables were age and gender at 1 Hz. There was no significant effect of any variables on grasping when exposed to 0.5-Hz and 2-Hz oscillations.

Table 7.4 shows the logistic regression coefficient (β), the Wald test (χ^2), which shows the unique contribution of each predictor while holding the other predictors constant, and the odds ratio (exp(β)) for each of the predictors. The odds ratio is the ratio of probability of grasping to probability of not grasping.

Table 7.4: Logistic regression analysis showing the influence of most significant subject characteristics on whether a subject grasped the hand support during their first exposure to 1-Hz oscillation.

Predictors	*B*	*Wald* χ^2	*p*	Odds ratio EXP(β)
Gender	1.083	5.117	0.024	2.953
Age	0.023	2.775	0.096	1.024
Constant	-2.569			

Employing a 5% significance level, only gender had a significant effect (p=0.024) on predicting the probability that a participant would grasp the hand support. The odds ratio for gender (coded 0 for male, and 1 for female) indicates that when holding all other variables constant, women are about 2.9 times more likely than men to grasp the support. The odds ratio for age reveals that for an increase in age of 1 year the odds that a participant will grasp the support increases by a factor of about 1.024. However, the effect of age on grasping was not statistically significant (p=0.096).

At 0.5 Hz and 2 Hz, there were no significant effects of any of the predictor variables on grasping although there was a tendency for females to use the support more often than males at both frequencies (Table 7.5).

Table 7.5: Effect of age and gender on the number of subjects who grasped the hand support during their first exposure to oscillation at 1 Hz, 0.5 Hz and 2 Hz.

	Age group 1 (18-24)	Age group 2 (25-45)	Age group 3 (46-59)	Age group 4 (60-70)	Overall
1 Hz					
Females	2	5	4	8	19
Males	0	5	0	4	9
Total	2	10	4	12	28
0.5 Hz					
Females	4	3	5	4	16
Males	2	2	2	3	9
Total	6	5	7	7	25
	Age group 1 (18-24)	Age group 2 (25-45)	Age group 3 (46-59)	Age group 4 (60-70)	Overall
2 Hz					
Females	3	1	2	3	9
Males	3	1	0	1	5
Total	6	2	2	4	14

7.3.3.3. Effect of subject characteristics on the gait measures

There was a reduction in the gait measures with repetition of the stimulus at 1 Hz (Figure 7.3). Due to the similar decreasing trend in gait measures with increasing repetitions of the 0.5-Hz and the 2-Hz stimuli (Figure 7.4 and Figure 7.5), multiple regression analyses were conducted on the gait measures obtained during the third exposures to the 0.5-Hz, 1-Hz and 2-Hz stimuli so that the analysis would be comparable between the three frequencies.

Multiple regression analysis for the gait measures without oscillation were conducted on the mean response across seven repetitions as there were no significant differences in gait measures with repeated measurement (Figure 7.6).

Multiple regression analysis for the gait measures during normal walking without oscillation revealed that body weight was the most significant predictor variable of peak-to-peak lateral COP position and that gender contributed 5.3% of the variability in the regression model (Table 7.6). After controlling for the effects of other factors, gender and age were the most significant predictors of lateral r.m.s. COP velocity during normal walking without oscillation. Age and walking speed were identified as significant predictors of normalized vertical r.m.s ground reaction force under the feet. Walking speed was the most significant predictor of mean COP speed during normal walking explaining 36.4% of the variability in the regression model, as expected since the amount of activity increased at greater walking speeds; and adding age and gender contributed to explain an additional 7.2% of the variability in the model.

When walking and exposed to 1-Hz lateral oscillation, age was a significant predictor of all the gait measures (Table 7.6). Adding stature contributed to explain an additional 6.2% of the variability in the peak-to-peak lateral COP position. Walking speed was a significant predictor of mean COP speed, explaining 10.1% of the variability in the regression model; adding age explained a further 12.5% of the variability in the model; 2.7% of the variability in the model was further explained by addition of stature.

During exposure to low frequency lateral oscillation at 0.5 Hz, age was the most significant predictor of postural stability and explained the greatest proportion of the variability in the regression models (Table 7.6). Adding stature explained a further 5.9% of the variability in peak-to-peak lateral COP position, and gender explained a further 3.4% of the variability in lateral r.m.s. COP velocity. After controlling for the effects of other factors, walking speed was the second most significant predictor of r.m.s. ground reaction force under the feet. Walking speed was a significant predictor of mean COP speed, explaining 13.6% of the variability in the regression model, and adding age explained a further 16% of the variability; 3.4% of the variability in the model was further explained by adding gender.

While walking and exposed to 2-Hz lateral oscillation, shoe width and age were identified as the most significant predictors of peak-to-peak lateral COP position (Table 7.6). Age was the most significant predictor of the lateral r.m.s. COP velocity and the r.m.s. ground reaction force under the feet after controlling for the effects of other factors. Walking speed was a significant predictor of mean COP speed explaining 5.9% of the variability in the regression model, and adding age explained an additional 6% of the variability in mean COP speed.

Table 7.6: Multiple regression analyses showing the effects of subject physical characteristics on gait measures in all four conditions (normal walking without oscillation and walking when exposed to 0.5-Hz, 1-Hz and 2-Hz oscillation).

	Normal walking without oscillations					Exposure to 1 Hz oscillation				
Dependent variables	Independent variables	*B*	β	R^2 (%)	*p*	Independent variables	*B*	β	R^2 (%)	*p*
peak-to-peak lateral COP position	Model summary (F=15.298)			22.4	<0.001	Model summary (F=12.094)			18.3	<0.001
	(1) Weight	0.092	0.307	17.1	0.002	(1) Age	0.124	0.384	12.1	<0.001
	(2) Gender	2.313	0.273	5.3	0.006	(2) Stature	0.159	0.266	6.2	0.004
	Constant	10.228				Constant	-6.105			
r.m.s. lateral COP velocity	Model summary (F=8.999)			13.9	<0.001	Model summary (F=11.371)			9.5	0.001
	(1) Gender	6.454	0.308	9.1	0.001	(1) Age	0.374	0.322	9.5	0.001
	(2) Age	0.147	0.238	4.8	0.012					
	Constant	25.08				Constant	40.191			
normalized r.m.s. vertical force under the feet	Model summary (F=5.379)			8.1	0.006	Model summary (F=13.09)			10.9	<0.001
	(1) Age	0.003	0.284	4.9	0.005	(1) Age	0.005	0.343	10.9	<0.001
	(2) Waking speed	0.512	0.206	3.2	0.038					
	Constant	0.334				Constant	0.866			
mean COP speed	Model summary (F=26.46)			43.6	<0.001	Model summary (F=12.151)			25.3	<0.001
	(1) Walking speed	76.629	0.631	36.4	<0.001	(1)Walking speed	63.449	0.363	10.1	<0.001
	(2) Age	0.102	0.218	4.4	0.006	(2) Age	0.252	0.378	12.5	<0.001
	(3) Gender	2.845	0.181	2.8	0.019	(3) Stature	0.232	0.188	2.7	0.039
	Constant	1.673				Constant	-20.531			
	Exposure to 0.5 Hz oscillation					Exposure to 2 Hz oscillation				
Dependent variables	Independent variables	*B*	β	R^2 (%)	*p*	Independent variables	*B*	β	R^2 (%)	*p*
peak-to-peak lateral COP position	Model summary (F=16.014)			23.3	<0.001	Model summary (F=7.741)			12	0.001
	(1) Age	0.15	0.451	17.4	<0.001	(1) Shoes width	1.663	0.26	8	0.008
	(2) Stature	0.16	0.257	5.9	0.004	(2) Age	0.076	0.223	4	0.022
	Constant	-2.299				Constant	1.919	2.374		
r.m.s. lateral COP velocity	Model summary (F=11.978)			18.2	<0.001	Model summary (F=6.552)			5.3	0.012
	(1) Age	0.328	0.389	14.8	<0.001	(1) Age	0.286	0.25	5.3	0.012
	(2) Gender	5.809	0.203	3.4	0.028					
	Constant	35.909				Constant	41.223			
normalized r.m.s. vertical force under the feet	Model summary (F=13.256)			19.8	<0.001	Model summary (F=5.784)			4.6	0.018
	(1) Age	0.006	0.43	13.1	<0.001	(1) Age	0.003	0.236	4.6	0.018
	(2) Walking speed	1.046	0.278	6.7	0.003					
	Constant	0.086				Constant	0.979			
mean COP speed	Model summary (F=17.223)			33	p<0.001	Model summary (F=7.708)			11.9	0.001
	(1) Walking speed	60.444	0.437	13.6	<0.001	(1) Walking speed	48.387	0.315	5.9	0.001
	(2) Age	0.212	0.403	16	<0.001	(2) Age	0.157	0.268	6	0.007
	(3) Gender	3.606	0.202	3.4	0.017					
	Constant	20.264				Constant	34.049	34.049		

B: unstandardized regression coefficient; β: standardized regression coefficient
R^2: percentage of experimental variation accounted for by the model

7.4. Discussion

The reported probability of losing balance was not significantly affected by the repetition of 0.5-Hz and 1-Hz oscillations (Figure 7.2) although there was a trend for reduced imbalance with repeated 1-Hz stimuli. The reported probability of losing balance decreased with increasing repetition of 2-Hz stimuli (Figure 7.2c). The postural strategies developed in response to the higher frequency oscillations might be learned easier with repetition of the stimuli although the learning effect might be better investigated if the stimuli were repeated more than three times.

Predictability may occur due to the repeating of the stimuli and prediction may result in adaptation to the imposed stimuli and cause a shift to a predictive postural strategy (Maki, 1986). Repeated exposure to vibratory proprioceptive stimulation has been found to gradually reduce vibration-induced body sway via adaptation (Fransson *et al*., 2000; Tjernström *et al*., 2002). In the current study, walking subjects were observed to grasp more often during their first exposures to each stimulus (Table 7.2) at all frequencies. Increased number of repetitions of lateral oscillations resulted reductions in the gait measures (Figure 7.3, 7.5 and 7.6). The effect of repeating the stimuli on grasping and the measures of gait suggest that walking subjects could perform better with increased repetition of the motions due to learning although they might have judged the difficulty in maintaining stability caused by the repeated oscillations to be similar.

Age and gender were found to be significant predictors of whether a subject grasped the hand support when exposed to 1 Hz oscillation (Table 7.4). There was not a significant effect of any subject characteristics on grasping during exposure to 0.5 and 2 Hz oscillations, but females were more likely to grasp (Table 7.5). Gender was not the most common predictor variable of gait measures during perturbed walking except during exposure to lateral oscillation at 0.5 Hz where gender was a significant predictor of lateral r.m.s. COP velocity and mean COP speed (Table 7.6), with males having higher r.m.s. COP velocity and mean COP speed. Differences in grasping between genders might not be necessarily caused by less stability in females but might be associated with fear of falling.

Age was the most common significant predictor of all gait measures (peak-to-peak COP position, r.m.s. COP velocity, r.m.s. vertical reaction force under the feet and mean COP speed) in all four conditions (Table 7.6). Elderly were previously reported to have higher magnitude and frequency of centre of mass (COM) and centre of pressure (COP) excursions and higher vertical reaction forces compared to younger subjects while exposed to moderate perturbations of stance via translated platform and during stationary standing on a transversely aligned support beam (Gu *et al*., 1996). Similarly, increased gait measures in elderly as reported in the current study may be an indication of elderly requiring more effort than younger people to maintain postural stability. Older adults were shown to have increased energy expenditure and increased step width during normal walking on a treadmill (Dean *et al*., 2007). Authors (Dean *et al*., 2007) proposed that old subjects having noisier control and sensors compensate their lateral instability by wider steps at the expense of higher metabolic cost. Increased effort in maintaining stability may be an indication of increased risk of fall which might be more critical for frail elderly who were not fit enough to participate in the experiment.

In response to lateral perturbation of stance and walking on the spot, elderly people were more likely to take extra steps and move arms or grasp (Maki *et al*., 2000). In the current study, older people were observed to grasp more often when exposed to 1 Hz lateral oscillations (Table 7.4). Extra steps or arm movements reported by the authors (Maki *et al*., 2000) may also be an indication of greater effort to maintain stability in elderly which is consistent with the increased gait measures in elderly as reported in the current study.

Among other factors like age, stature and foot length, body weight was shown to be a significant predictor of postural stability during quiet standing (Hue *et al*., 2007). When stepwise multiple regression was performed on objective measures of postural stability (range of COP displacement, r.m.s. COP position, r.m.s. COP velocity and mean COP speed in the lateral and back-and-forth direction) body weight was consistently found to be the best predictor explaining the largest variability in the objective measures of stability (Hue *et al*., 2007). According to the results of the current study, there was a significant effect of weight only in the peak-to-peak lateral COP position during normal walking without oscillations but age was generally the most significant predictor of gait measures. Higher range of lateral COP movement in heavier subjects during normal walking might be associated with larger base of support area required to follow the larger lateral movement of the upper body. The significant effect of weight reported by Hue *et al.* (2007) might be associated with rather heavy subjects (ranged 59.2 to 209.5 kg) used in their study and the effect of age would be more clear if the age range (24-61 years) was increased. The differences in findings may also be associated with the differences in standing and walking stability and perturbed and unperturbed stability.

During quiet standing with restricted base of support area in sagittal plane, shorter subjects were shown to have more difficulty in compensating body sway caused by a sudden perturbing torque applied in the sagittal plane at the center of mass level (Berger *et al*., 1992). Longer foot length in taller subjects is advantageous in feet-in-place postural strategies by providing a larger support area. The authors (Berger *et al*., 1992) investigated the bivariate correlations between stature and ankle joint displacement without controlling the effect of other factors like age and foot length. In the current study, stepwise regression analysis showed that after controlling for the effect of other factors, stature was a significant predictor of peak-to-peak COP position during exposure to 0.5 Hz and 1 Hz oscillations. Wider steps adopted by taller subjects might be advantageous to overcome the effects of low frequency oscillations. When exposed to high frequency oscillations shoes width became the most significant predictor of peak-to-peak lateral COP position. Walking subjects having wider shoes were advantageous in terms of larger support area when exposed to high frequency oscillations during which walking subjects had restricted base of support area analogous to quiet standing in sagittal plane.

Although there was a significant effect of subject characteristics especially age on gait measures during normal walking and perturbed walking, reported probability of losing balance was not affected by any subject characteristics. Perceived risk of fall may not be reflecting the actual risk of fall, or the actual risk of fall was not significantly higher in older people as they have managed to recover from lateral perturbations with comparably higher effort than younger people. Older people might also have been more conservative in judging the postural instability caused by the oscillation.

Coefficient of correlation (R^2) values in the multiple regression analysis indicate the proportion of the variability in gait measures accounted for by the predictors in the models. The R^2 values were ranged between 4.6% to 42.6%. Although the R^2 values were considerably low, the effects of predictor variables on gait measures were significant. A great proportion of the variability in gait measures was not explained by the models (Table 7.5 and Table 7.6) suggesting that other postural and anthropometric factors influenced the postural stability of walking subjects. Attention may be an example of these factors which was shown to play an important role in postural stability (Teasdale and Simoneau, 2001, Shumway-Cook *et al.*, 1997). Gait measures used in the current study may not be sufficient to explain all aspects of postural stability during perturbed locomotion, and there may be other measures that are more sensitive to subject characteristics. Measurement uncertainty may also contribute to the unexplained variance.

7.5. Conclusion

Age was the most common significant predictor of postural stability during normal walking and when exposed to lateral oscillations of 0.5 Hz, 1 Hz and 2 Hz, indicating an increase in gait measures with increasing age. Age together with gender was also a significant predictor of whether a subject grasped the hand support to maintain balance when exposed to lateral oscillations of 1 Hz. There was no significant effect of age or any other subject characteristics on the self-reported probability of losing balance. Older adults managed to overcome the destabilizing effects of lateral perturbation with a greater effort and so might have judged the perceived risk of fall to be similar to the judgements of younger adults.

Results of the study indicate that stability thresholds of young male walking subjects (reported in Chapter 4) exposed to lateral oscillations can be applicable to a wider range of subjects including females and older adults. However, increased effort in maintaining stability in older adults may be an indication of increased risk of fall which might be more critical for frail elderly who are not fit enough to participate in the experiment or to use transportation. Further research with older and frail elderly is required to investigate the differences in subjective and objective assessment of postural stability while walking and perturbed by lateral oscillations.

Chapter 8

Discussion

8.1. Introduction

This chapter aims to bring the findings of all four experiments (as reported in Chapters 4, 5, 6 and 7) together to answer the initial research question. The research question which stimulated this study was: ‘Can we develop a model of postural stability from which the effects of lateral oscillation on the postural stability of walking people can be predicted?’. The studies were also designed to determine the stability thresholds of walking subjects when perturbed by lateral oscillation and understand the mechanisms of postural control during perturbed walking.

8.2. Interpretation of the findings

Figure 8.1 shows a proposed model of postural stability during perturbed walking. The model has two outputs. One output is the perceived risk of falling, which was assessed experimentally by asking subjects to indicate their perceived probability of losing balance. Another output of the model is the stepping strategy, which was evaluated from measures of the lateral centre of pressure (COP) during exposure to lateral oscillation.

The postural stability model has additional outputs, including upper-body movement (i.e. of arms and trunk) and grasping from a hand support. Upper-body movements were not measured in the experiments although the relationship between the centre of mass (COM) and the centre of pressure (COP) could be a related measure that also reflects dynamic stability (Kaya *et al.*, 1998; Lee and Chou, 2006). So as to observe the effects of lateral oscillation on the stepping strategy, subjects were discouraged from grasping unless it was really necessary. When they did grasp the hand support, this was recorded by the experimenter.

The proposed model suggests that lateral oscillation is perceived by the vestibular and somatosensory systems. Vision would also contribute to perception of low frequency vibration. However, in our experiments vision is not expected to contribute to perception since subjects were asked to fix their vision on a white board moving with the motion platform (Figure 4.1).

Once the lateral oscillation is sensed by the sensory systems and interpreted in the central nervous system, automated corrective actions (e.g. corrective hip torque or ankle torque) are generated. ‘Body dynamics’ in the model represents the biomechanical structure of the human body (a dynamic model of human body parts, e.g. 2 segment inverted pendulum). As a result of the kinematic and kinetic relations of the human body parts and corrective actions generated by the muscles, appropriate postural strategies (i.e. stepping) are developed to overcome the destabilizing effects of lateral oscillation. The perceived risk of falling may also result in predictive postural strategies (e.g. a more cautious rigid body).

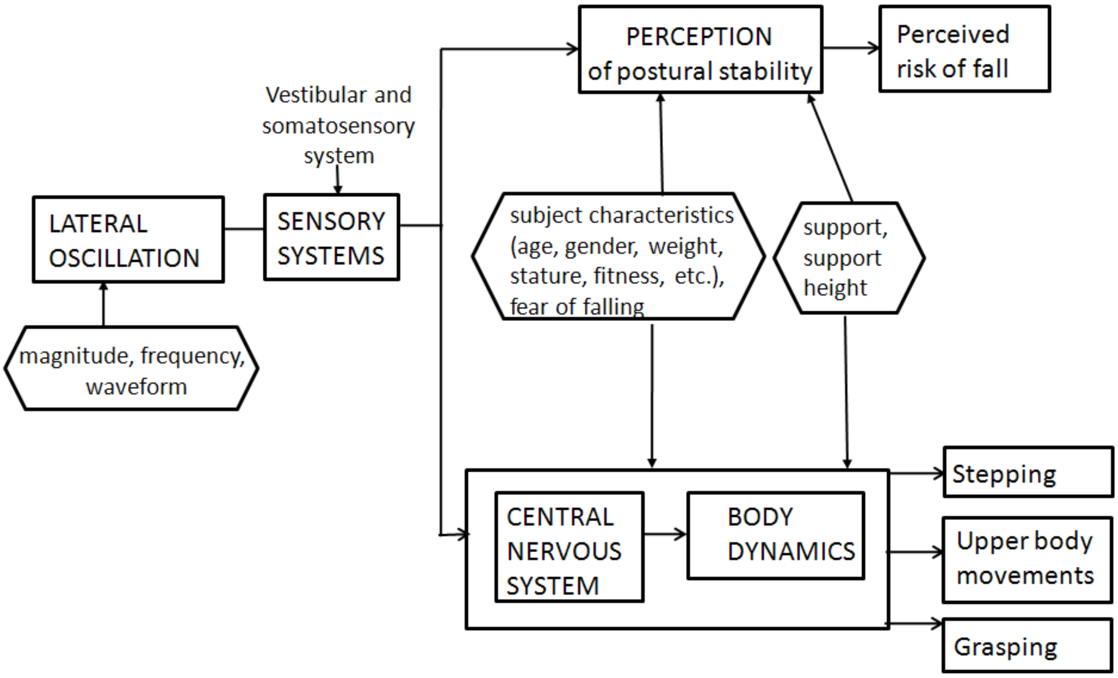


Figure 8.1: Model of postural stability of walking people exposed to lateral oscillations.

In this thesis, various factors have been identified to influence postural stability while walking and perturbed by lateral oscillations. These factors are oscillation characteristics (magnitude, frequency, waveform) which are the input to the postural stability model (Figure 8.1), environmental factors (e.g. support and support height) and physical characteristics of people (e.g. age, gender, weight).

The measured effects of the magnitude and frequency of oscillations on the reported probability of losing balance were used to obtain stability thresholds (Figure 4.6) which have not previously been obtained for walking subjects. Walking people were observed to be sensitive to the velocity of oscillation: when the lateral oscillations were applied at the same velocity the perceived risk of falling was broadly similar irrespective of changes in the frequency oscillation (Figure 4.5b). Sensitivity to the velocity of oscillations suggest that it is not only important to react to perturbations by employing appropriate postural strategies (i.e. wider step) but the timing of reaction (i.e. faster step) is also significant. Altering the stepping strategy, by adjusting the timing and placement of successive steps, is thought to be the principal means of maintaining dynamic balance during locomotion (Nashner, 1980; Oddson *et al.*, 2004). The overall effects of the magnitude and frequency of oscillation on r.m.s. COP velocity (Figure 4.9) confirmed that most of the subjects used stepping strategies to counteract the destabilizing effects of lateral oscillation (Figure 8.1). The lateral r.m.s. COP velocity is an indication of not only the range of base of support in the lateral direction but also the timing of stepping actions.

The stability thresholds obtained for walking passengers are useful for controlling the magnitude and frequency of oscillations to prevent falls related to postural stability problems. Walking passengers are expected to tolerate higher magnitude oscillations when using a hand support. The second study findings showed that postural instability decreased when using a hand support as shown by reductions in subjective ratings of 'discomfort or difficulty' in walking and objective measure of the r.m.s. lateral COP velocity (Figure 5.4a, 5.4b, 5.5a and 5.5b). Reductions in the subjective ratings of 'discomfort or difficulty' when using a hand support might be associated with improved postural control as well as a reduction in the fear of falling. Reduction in r.m.s. COP velocity is an indication of less effort required to maintain stability when using a support. Most studies of hand supports in the literature have focused on the benefits of light touch from a stationary support during quiet standing and normal walking (Jeka and Lackner, 1994; Dickstein and Laufer, 2004). Our study with perturbation and subjects grasping the support with their chosen force found that the reductions in COP velocity from using a hand support were greater during perturbed walking (about 30-50% when the support was held throughout the oscillation, about 20-30% when the support was held only if required) than during normal walking (about 15.6%) (Figure 5.4c and 5.5c). These findings emphasize the importance of supports when walking and perturbed by oscillations that increase the risk of fall and require greater forces, and more rapidly applied forces, to counteract the destabilizing effects of lateral oscillation. If a hand support were used purely as a mechanical aid to maintain balance, it would be expected that walking people would benefit more from a support when the support height is increased since the stabilizing moment from a hand support will increase with increasing height of the support. However, support height did not influence the subjective and objective measures of postural stability when used throughout the oscillations (Figure 5.3). Therefore, a hand support is suggested to not only provide mechanical stabilizing forces but also provide a sensory cue to improve postural stability. However, subjects preferred to hold the vertical support at a height of 126 cm (above the surface supporting the feet) if required during exposure to the lateral oscillatory motion.

For the type of waveform used in the first study, postural stability was well predicted from both the peak velocity and the r.m.s. velocity of the oscillation. Whether the walking subjects were more sensitive to the peak or the r.m.s. magnitude of the oscillations was unknown. A third experiment was conducted to understand the dependence of postural stability on the motion waveform. When exposed to lateral oscillations of the same r.m.s. magnitude, postural stability assessed by the subjectively reported probability of losing balance and the lateral COP velocity was found to be sensitive to the peak magnitude of the oscillations (Figures 6.4b, 6.5b and 6.6). The peaks in the lateral oscillation might create postural stability problems because they require walking subjects to take faster corrective postural action to overcome their unexpected destabilizing effects. The r.m.s. value is therefore not the optimum method of evaluating the postural stability of walking subjects exposed to lateral oscillations and peak value should also be considered. Thuong and Griffin (2010b) have found that the r.m.s. value is not optimum for evaluating the discomfort of standing subjects exposed to random and transient whole-body vibrations. Howarth and Griffin (1991) have also reported increased discomfort in seated subjects exposed to vertical vibration when the crest factor increases even though the r.m.s. value was constant.

Stability thresholds were obtained in the first study for healthy young male subjects aged 25 to 45 years. These thresholds may be applicable to passengers walking in moving trains, but further research was required to understand the dependence of postural stability on variations in individual susceptibility to falling, especially in the elderly. So, a fourth experiment was conducted to investigate the effect of subject characteristics (age, gender, weight, stature, shoe width, fitness level) on walking stability. No significant effect of any subject characteristic was found on the subjectively reported probability of losing balance, suggesting that stability thresholds obtained in the first study may be applicable to females and to older people (45 to70 years). During exposure to lateral oscillations, females were more likely to grasp the handrail of the treadmill than the men (Table 7.3 and 7.4), which does not necessarily mean that females are less stable than men but they may be more cautious due to fear of falling. During exposure to 0.5-Hz, 1-Hz, and 2-Hz oscillations and during normal walking without oscillations, age was the most significant and common predictor of postural stability among all four objective gait measures (i.e., peak-to-peak lateral COP position, r.m.s. lateral COP velocity, r.m.s. vertical ground reaction force under the feet and mean COP speed) (Table 7.5). Increased gait measures in older adults may be an indication of increased effort to maintain stability. The increased effort in the elderly may be an indication of increased risk of falling, although the older adults who participated in the fourth experiment reported a probability of losing balance similar to that of the younger adults. Older adults might be more conservative in judging the perceived risk of fall. Alternatively, the fit elderly who participated in the fourth study managed to overcome the effects of lateral oscillations with greater effort and judged their recovery from perturbation similar to younger subjects. Further research with an older or more frail elderly group is required to understand the effects of age on perturbed stability.

The results of all four experiments were combined to develop a predictive model of postural stability of walking people exposed to lateral oscillations. The details of the predictive model are provided in the next section (Section 8.3).

8.3. Predictive model

The effects of systematic variations in the frequency and the magnitude of lateral oscillations were investigated in the first study (Chapter 4). The findings revealed that postural stability can be reasonably predicted from the r.m.s. velocity of the lateral oscillation. At a specific frequency, the self-reported probability of losing balance increased almost linearly with increasing velocity of perturbation (Figure 4.4b). The increase in the perceived risk of fall with increasing velocity was similar at all frequencies (Figure 4.4b). This suggests a model of postural stability in the form of Equation (8.1):

$$PLB\,(\%) = k_1 * v_{r.m.s.} \tag{8.1}$$

where *PLB* is the probability of losing balance, $v_{r.m.s}$ is the r.m.s. velocity of the lateral oscillation and k_1 is a constant. However, the third study (Chapter 6) suggested that the r.m.s. velocity is not sufficient to evaluate postural instability, and that motions having the same r.m.s. velocity but higher peaks caused greater postural instability (Figure 6.4). This suggests that Equation (8.1) should be modified to take into account the effect of the peaks (or the crest factor of the oscillation). Equation 8.2 shows the proposed model, based on the third study, to predict probability of losing balance from the r.m.s. and peak values of velocity of lateral oscillations:

$$PLB\,(\%) = k_1 * v_{r.m.s.} + k_2 * v_{peak} + c \tag{8.2}$$

k_1 is the average of the slopes obtained by linear regression between r.m.s. velocity of oscillations and estimated probability of losing balance at two frequencies (1 Hz and 2 Hz) when the peak value of oscillations were kept constant.

$$k_1 = \frac{309.7 + 364.1}{2} = 337 \tag{8.3}$$

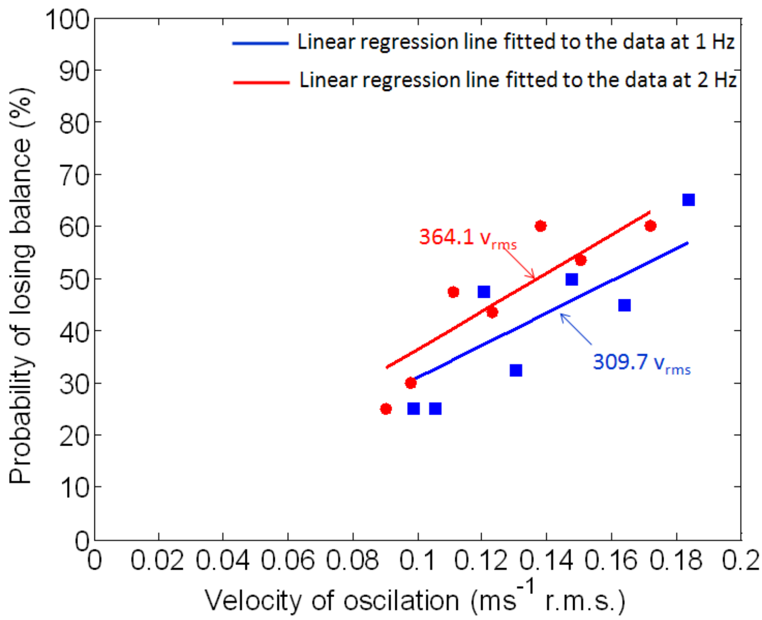


Figure 8.2: Probability of losing balance as a function of r.m.s. velocity of 1-Hz and 2-Hz oscillations: lines fitted to the data points by linear regression with zero intercepts.

Similarly, k_2 is the average of the slopes obtained by linear regression between peak velocity of oscillations and estimated probability of losing balance at two frequencies (1 Hz and 2 Hz) when the r.m.s. value of oscillations were kept constant (Figure 8.3).

$$k_1 = \frac{127.4 + 146.1}{2} = 137 \tag{8.4}$$

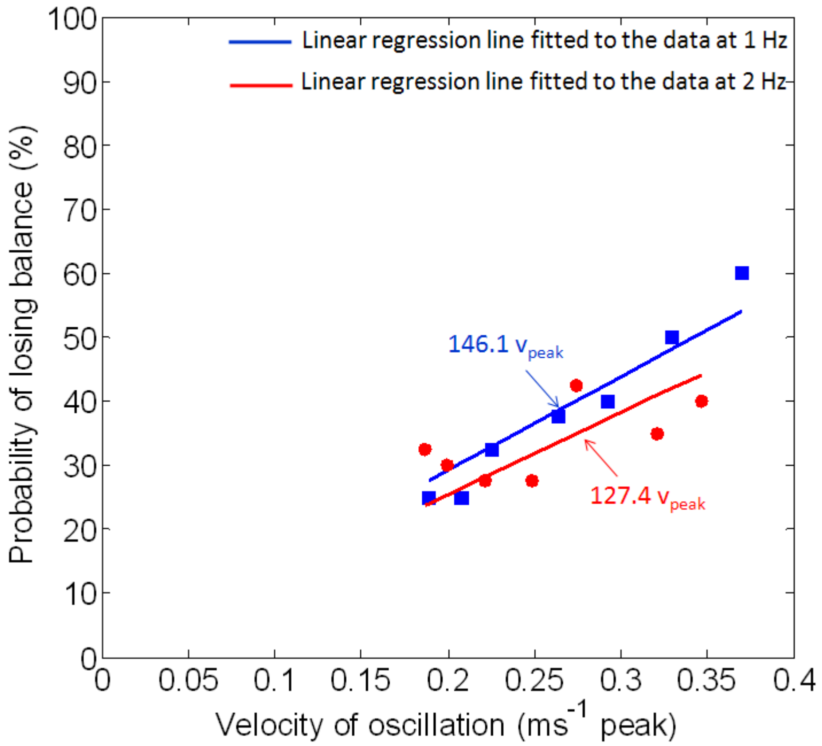


Figure 8.3: Probability of losing balance as a function of peak velocity of 1-Hz and 2-Hz oscillations: lines fitted to the data points by linear regression with zero intercepts.

Having obtained the slopes k_1 and k_2, the constant c in Equation (8.2) is obtained such that the probability of losing balance (*PLB* (%) in Equation (8.2)) satisfies the linear regression lines fitted to the data at both frequencies (1 Hz and 2 Hz) when the peak velocity of oscillations was kept constant (Figure 8.2) and when the r.m.s. value of oscillations was kept constant (Figure 8.3).

When the r.m.s. value of 1-Hz oscillations was kept constant (0.12 ms^{-1}, Table 6.4), Equation (8.2) gives:

$$PLB(\%) = 146.1 * v_{peak} + 309.7 * 0.12 + c \tag{8.5}$$

For this equation (8.5) to satisfy the regression line fitted to 1-Hz data as a function of the peak value of oscillations (Figure 8.3) the intercept (309.7*0.12+*c*) should be equal to zero from which the constant *c* is obtained to be -37.2.

When the peak value of 1-Hz oscillations was kept constant (0.3 ms^{-1}, Table 6.4), Equation (8.2) gives:

$$PLB(\%) = 146.1 * 0.3 + 309.7 * v_{r.m.s.} + c \tag{8.6}$$

For this equation (8.6) to satisfy the regression line fitted to 1-Hz data as a function of ther.m.s. value of oscillations (Figure 8.2), the intercept (146.1*0.3+*c*) should be equal to zero from which the constant *c* is obtained to be -43.8.

If we take the average of two constants obtained when the peak and r.m.s. value of 1-Hz oscillations were kept constant, we can find an approximate value for the constant *c* (c_{1Hz}):

$$c_{1\,Hz} = -\frac{37.2 + 43.8}{2} = -40.5 \tag{8.7}$$

The same procedure can be repeated to find the constant *c* for 2-Hz oscillations. When the r.m.s. value of 2-Hz oscillations was kept constant (0.11 ms^{-1}, Table 6.4), Equation (8.2) gives:

$$PLB(\%) = 127.4 * v_{peak} + 364.1 * 0.12 + c \tag{8.8}$$

For this equation (8.8) to satisfy the regression line fitted to 2-Hz data as a function of peak. value of oscillations (Figure 8.3) the intercept (364.1*0.12+*c*) should be equal to zero from which the constant *c* is obtained to be -43.7.

When the peak value of 2-Hz oscillations was kept constant (0.29 ms^{-1}, Table 6.4), Equation (8.2) gives:

$$PLB(\%) = 127.4 * 0.29 + 364.1 * v_{r.m.s.} + c \tag{8.9}$$

For this equation (8.9) to satisfy the regression line fitted to 2-Hz data as a function of the r.m.s. value of oscillations (Figure 8.2) the intercept (127.4*0.29+*c*) should be equal to zero from which the constant *c* is obtained to be -37.

If we take the average of the two constants obtained when the peak and r.m.s. value of 2-Hz oscillations were kept constant, we can find an approximate value for the constant *c* (c_{2Hz}):

$$c_{2\,Hz} = -\frac{37 + 43.7}{2} = -40.35 \tag{8.10}$$

Taking the average of constant values obtained for 1-Hz and 2-Hz oscillations (c_{1Hz} and c_{2Hz}), the constant *c* value can be obtained as follows:

$$c = -\frac{40.5 + 40.35}{2} \approx -40 \tag{8.11}$$

Substituting for k_1 and k_2 (Equations (8.3) and (8.4)) and the constant c value (Equation (8.11)) into equation (8.2), the general form of the predictive model can be obtained as follows:

$$PLB(\%) = 337 * v_{r.m.s.} + 137 * v_{peak} - 40 \tag{8.12}$$

Figure 8.4 shows the model defined in Equation (8.12). The model fits reasonably well to the experimental data of the third study although it underestimates the values of reported probability of losing balance in the first and fourth experiment. Underestimation of the data from the first and fourth experiments by the model based on the third experiment might be related to differences between the experiments (e.g., waveform and duration of the oscillations, and the range of stimuli). The differences might also be caused by the method applied in the third experiment where subjects were exposed to a reference motion before each test motion, which might have resulted in some adaptation of the walking subjects to the stimuli due to a learning effect.

The second study (Chapter 5) showed that the probability of losing balance decreased when using a hand support (Figures 5.4 and 5.5) but was not dependent on the height of a hand support (Figure 5.3). Looking at the median values in Table 5.1 and Table 5.2, the percentage improvement in postural stability can be approximated to 40% when the hand support is used throughout the oscillation and to 20% when the hand support is used if required. Therefore, the probability of losing balance when using a hand support throughout the oscillation (PLB_{s1}) and when using a hand support if required (PLB_{s2}) is formulated as shown in Equations (8.13) and (8.14):

$$PLB_{s1}(\%) = 337 * v_{r.m.s.} + 137 * v_{peak} - 80 \tag{8.13}$$

$$PLB_{s2}(\%) = 337 * v_{r.m.s.} + 137 * v_{peak} - 60 \tag{8.14}$$

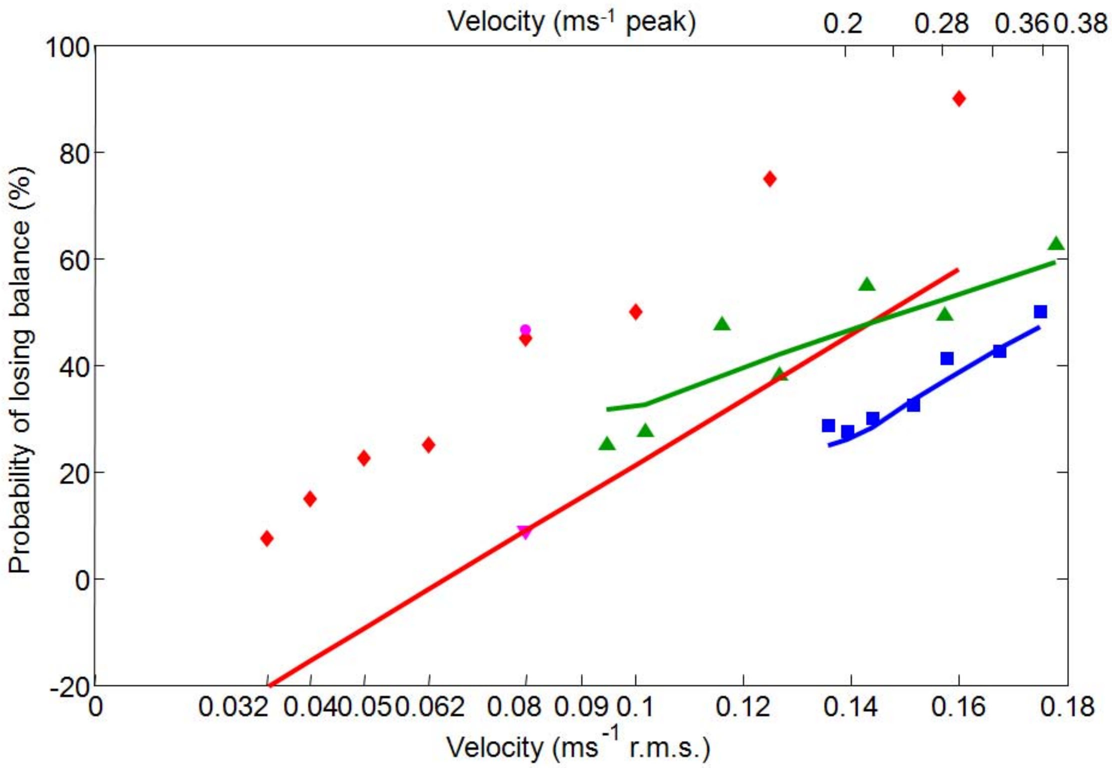


Figure 8. 4: Model of postural stability (Equation 8.12) used to predict probability of losing balance reported in the first, third and fourth experiments. Reported values of probability of losing balance: ◆ in the first experiment, in the third experiment and, ● in the fourth experiment as a function of r.m.s. velocity of oscillations, ■ in the third experiment as a function of peak velocity of oscillation. Predicted values of probability of losing balance in the first experiment, in the third experiment and, ▼ in the fourth experiment as a function of r.m.s. velocity of oscillations in the third experiment as a function of peak velocity of oscillation.

Figure 8.5 shows the linear regression model defined in Equation 8.12 with different constants (*c*). When a constant of 40 is used, the model predicts well the data from the third experiment whereas the data from the first experiment is underestimated. When a constant of 8 is used, the model predicts well the data from the first experiment but overestimates the data from the third experiment. If an average constant (*c*=24) is used, the model predictions seem reasonable for all three experiments. The error in the estimation of the probability of losing balance (0-100%) with the new constant (*c*= 24) is about 16%.

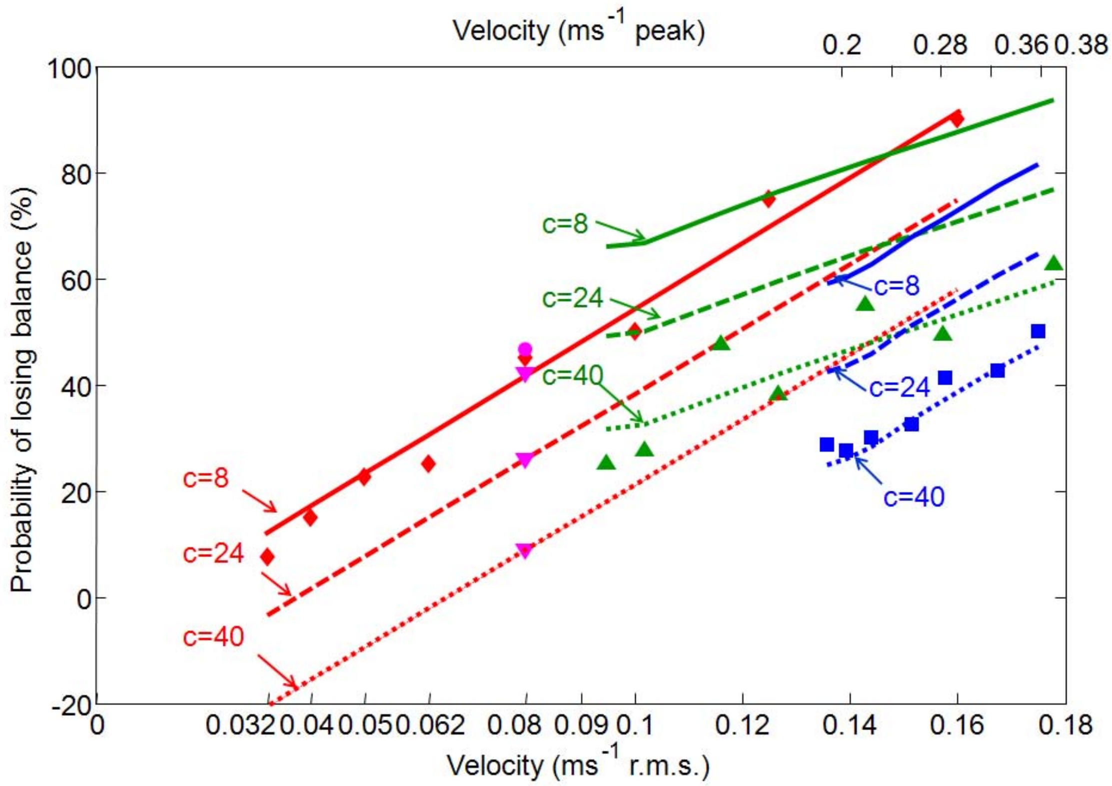


Figure 8.5: Model of postural stability (Equation 8.12) used to predict probability of losing balance reported in the first, third and fourth experiments for different values of constant *c* (*c*=8, *c*=24, *c*=40). Reported values of probability of losing balance: ◆ in the first experiment, in the third experiment and, ● in the fourth experiment as a function of r.m.s. velocity of oscillations, ■ in the third experiment as a function of peak velocity of oscillation. Predicted values of probability of losing balance ——— in the first experiment, ——— in the third experiment and, ▼ in the fourth experiment as a function of r.m.s. velocity of oscillations in the third experiment as a function of peak velocity of oscillation.

With the new constant (*c*=24), the probability of losing balance (with and without hand support) is modeled by the following three equations:

$$PLB\,(\%) = 337 * v_{r.m.s.} + 137 * v_{peak} - 24 \quad (8.15)$$

$$PLB_{s1}\,(\%) = 337 * v_{r.m.s.} + 137 * v_{peak} - 64 \quad (8.16)$$

$$PLB_{s2}(\%) = 337 * v_{r.m.s.} + 137 * v_{peak} - 44 \quad (8.17)$$

The results of the fourth experiment suggested that the reported probability of losing balance was not affected by any subject characteristic (i.e., age, gender, weight, stature, shoe width, and fitness level). Therefore, the same model (Equation 8.15) developed based on experimental results with young (25-45 year old) fit male subjects may be applicable to a wider group of subjects, including females and older adults (aged 45-69 years).

The proposed predictive model (Equation 8.15, 8.16 and 8.17) requires a recorded velocity time history. The model is currently run in Matlab for a randomly generated velocity time history (Figure 8.6). The self-reported probability of losing balance is calculated using Equations 8.15, 8.16 and 8.17 in a finite length running window Figure 8.6 shows an example time history for lateral oscillations of 20 seconds. For a running window of 4 seconds, the probability of losing balance with and without hand support is calculated using Equations 8.15, 8.16, and 8.17. Figure 8.7 shows the calculated probability of losing balance as predicted by the model both when using a hand support and when walking without support. If the predicted probability of losing balance exceeds a previously determined threshold value (e.g. 50%) caution should be taken to prevent falls related to postural instability problems. Note that the probability of losing balance is calculated at each running window of 4 seconds. If the probability of losing balance is observed to be larger than 50 at a certain time (e.g. at 5 s, Figure 8.7), caution should be taken during 4 seconds running window starting from that time (e.g. 5-9 s). Figure 8.8 shows the running r.m.s. and peak levels of oscillations calculated for each running window.

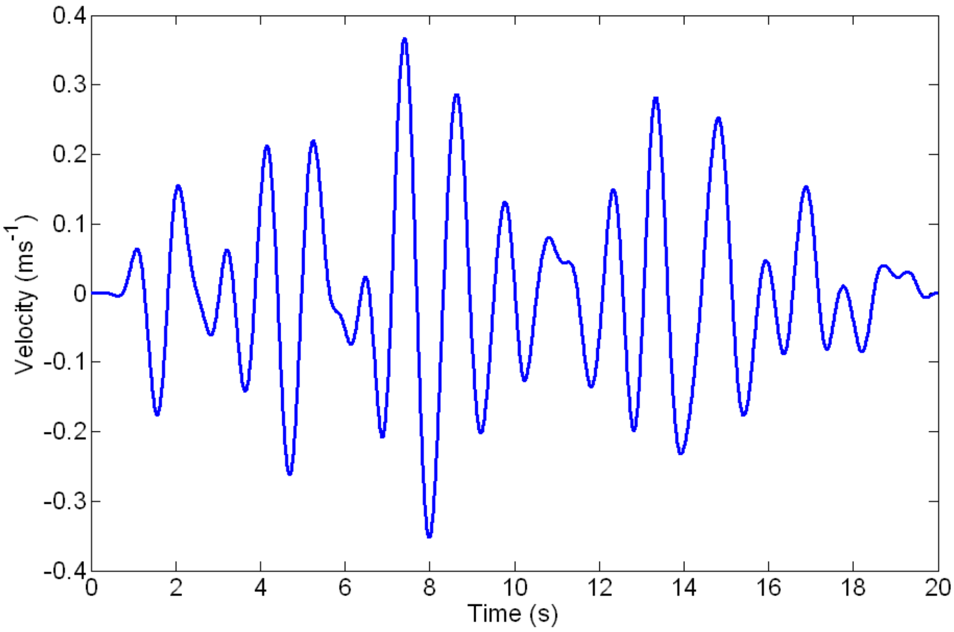


Figure 8.6: Random velocity time history generated in Matlab.

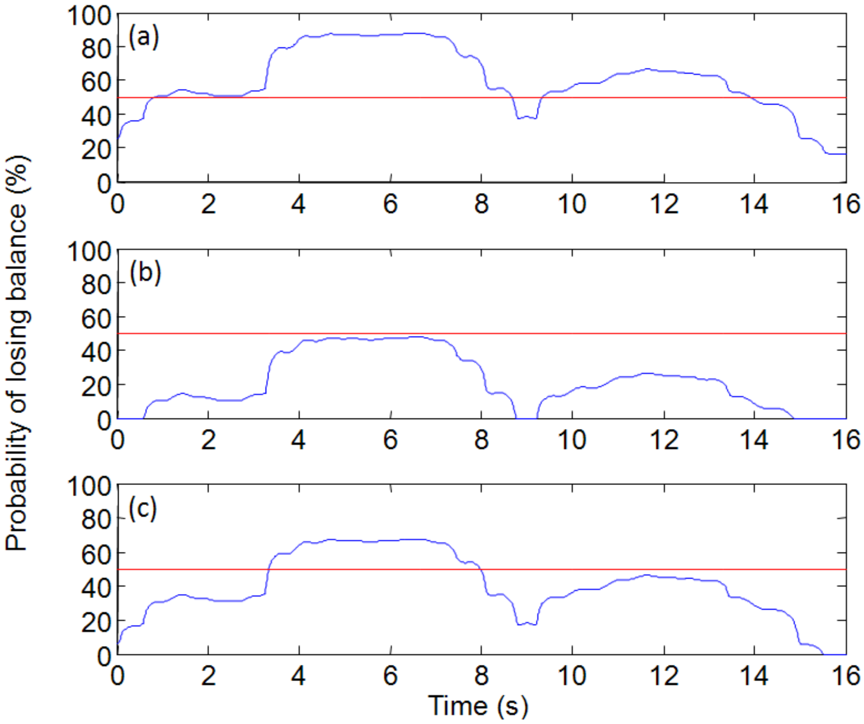


Figure 8.7: Predicted probability of losing balance when exposed to lateral oscillation shown in Figure 8.6 with and without support: (a) when using no hand support (b) when using hand support continuously throughout the oscillation (c) when using hand support if required during the oscillation.

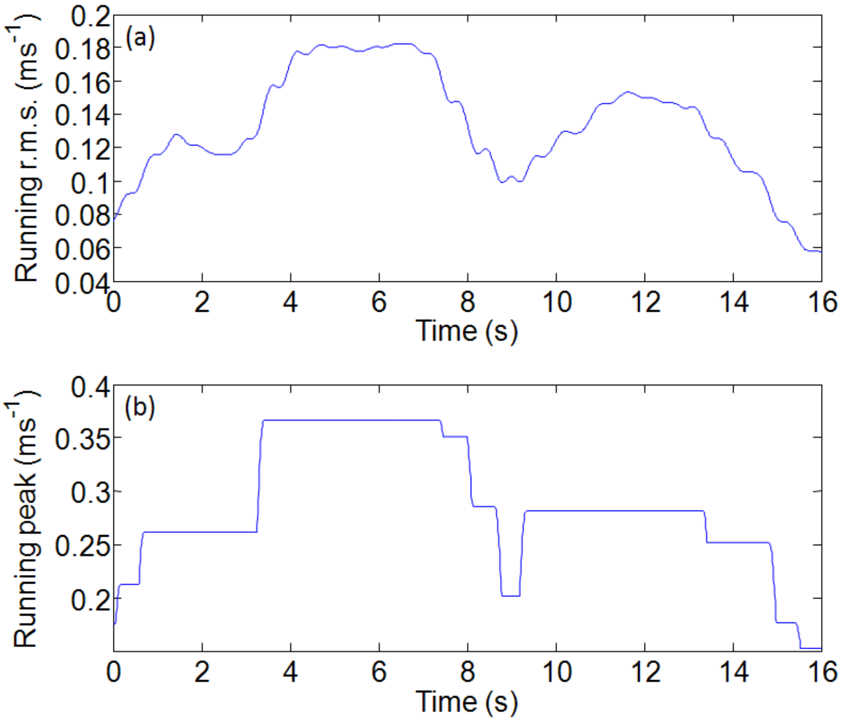


Figure 8.8: (a) Running r.m.s. and (b) running peak values of lateral oscillation (Figure 8.6) calculated for each running window of 4 seconds.

8.4. Discussion of the method

In this study, postural stability when walking and perturbed by lateral oscillations was investigated experimentally using two methods of assessment. One method was a subjective method based on the perceived risk of fall and the other method was an objective method based on the lateral centre of pressure (COP) measurements. The two measures associated with the subjective and objective method were the 'reported probability of losing balance' and the 'lateral r.m.s. COP velocity'.

The reported probability of losing balance may not represent the actual risk of fall, but it is a useful measure that can indicate the short-term and long-term reactions of walking subjects to lateral oscillations. Short-term reactions of walking subjects in transport may be to stop walking and return back to their seat. Long-term reactions may be to prefer a different transport type.

The lateral r.m.s. COP velocity showed consistency with the subjective measure of postural stability regarding the dependency on the frequency, the magnitude, and the waveform of oscillations. The reported probability of losing balance and the r.m.s. COP velocity were not consistent as regards the influence of subject characteristics. The reported probability of losing balance may be an indication of a perceived risk of fall whereas the r.m.s. COP velocity may be an indication of the physical effort to recover from perturbation. Subjects may adopt strategies to overcome the effects of motion and judge what would have happened if they had not applied great effort or skill to minimize the effects of perturbation. The use of an objective measure of stability increased the power to interpret the experimental results and understand mechanisms of walking stability (Figure 8.1).

8.5. Recommendations for future work

This research is an initial framework to develop the tolerances of walking people to lateral oscillations in transport and develop a predictive model of walking stability. The proposed predictive model is based on the subjective measure of postural stability as it aims to predict the probability of losing balance when exposed to lateral oscillations. The reported probability of losing balance was scaled from 0 to 100%. When the motions became severe, the scale for reporting the probability of losing balance becomes less sensitive due to saturation (towards the maximum value of 100%). The predictive model might be further developed to have a greater resolution such that it is more sensitive to factors influencing walking stability (e.g. frequency of oscillations, use of support, subject characteristics). A relative scale can be used to assess perceived risk of fall using the stability thresholds obtained in the current research. The motions shown to result in 0%, 25%, 50% and 100% probability of losing balance can be introduced to the walking subjects as reference motions which will be followed by test motions to be judged by the subjects accordingly.

The overall effects of the lateral oscillation on the r.m.s. COP velocity demonstrated the use of stepping strategies by walking subjects to counteract the destabilizing effects of lateral oscillation. During exposure to lateral oscillations, walking subjects might also switch to different postural strategies (e.g. upper body movement or arm movement) in which case, the centre of pressure (COP) may not be sufficient to show the state of postural instability but the centre of mass (COM) movements or the relative movement of the COM with respect to the COP may provide more reliable information.

8.6. Conclusion

Postural stability while walking and exposed to lateral oscillations has been investigated systematically using two methods of assessment. The changes in centre of pressure measures in response to lateral oscillations show that people respond to lateral oscillations by stepping actions, although different methods of objective measures are recommended to be used in future work to improve the understanding of walking stability.

The subjective measure of reported probability of losing balance has provided a useful measure of tolerances of walking people to lateral oscillations and has been used to develop a predictive model of perturbed walking stability. The model is useful to predict postural stability of wide variety of walking people when exposed to various types of lateral oscillations, although recommendations for future work are made for the improvement of the proposed model.

Chapter 9

Conclusion

Four experiments have been conducted to investigate the effect of lateral oscillations on the postural stability of walking people (Chapters 4 to 7). The results of the four experiments have been combined to develop an understanding of the mechanisms involved in walking stability. Experiment findings have also been used to develop a preliminary predictive model of probability of losing balance from which the effects of oscillation characteristics (e.g. magnitude, frequency, waveform) and support can be predicted.

It appears that postural stability cannot be predicted solely from either the peak or the r.m.s. value of lateral acceleration, but can be reasonably well predicted from both the peak velocity and the r.m.s. velocity of lateral oscillation (0.5-2 Hz). The dependency of centre of pressure on the magnitude and frequency of oscillations reveals that main strategy to overcome the effects of lateral oscillation during walking is stepping. The dependence of the reported probability of losing balance on the magnitude and frequency of oscillations have been used to obtain stability thresholds for walking people exposed to lateral oscillations that are typical of lateral accelerations experienced in a train ride.

Hand support improves postural stability when walking is perturbed by lateral oscillation at all frequencies (in the range 0.5 to 2 Hz) and at all velocity magnitudes (in the range 0.05 to 0.16 ms^{-1} r.m.s.). The improvement in postural stability is shown by significant reductions in both the subjective ratings of 'discomfort or difficulty' in walking and objective measure of r.m.s. lateral COP velocity when a hand support is used. The improvement in postural stability from holding the support and the forces applied to the hand support are independent of support height and are greater during perturbed walking than during normal walking, and greater when held throughout the oscillation than when held only if required. The percentage improvement in postural stability can be approximated to 40% when the hand support is used throughout the oscillation and to 20% when the hand support is used if required. Subjects prefer to hold the vertical support at a height of 126 cm above the surface supporting the feet if required during exposure to the lateral oscillatory motion.

When exposed to lateral oscillations of the same r.m.s. magnitude, postural stability assessed by subjectively reported probability of losing balance and lateral COP velocity has been found to be sensitive to the peak magnitude of oscillations especially at 1 Hz. The r.m.s. value may therefore not be the optimum method of evaluating postural stability of walking subjects exposed to lateral oscillations especially at low frequencies: the peak magnitude of the oscillations should also be considered when minimizing postural stability problems in transport. It is suggested that perturbation characteristics including peak and r.m.s. magnitudes of motion should be fully reported in perturbed balance experiments for an appropriate comparison and interpretation of experimental studies.

Age has been found to be a significant predictor variable of postural instability, with increasing gait measures with increasing age. Age together with gender is also a significant predictor of whether a hand support is grasped when exposed to lateral oscillations of 1 Hz. No significant effect of age or any other subject characteristics has been found on the self-reported probability of losing balance. Therefore, stability thresholds of young (25 to 45 years) male walking subjects (reported in Chapter 4) exposed to lateral oscillations can be applicable to a wider range of subjects including females and older adults (45 to 70 years). Older adults may manage to overcome the destabilizing effects of lateral perturbation with a greater effort and so judge the perceived risk of fall to be similar to the judgements of younger adults. However, increased effort in maintaining stability in older adults as indicated by increased gait measures with age may be an indication of increased risk of fall. A further study is required to investigate postural stability in frail elderly group.

The stability thresholds obtained in these studies may be applicable to passengers walking in moving trains. The findings of the study show the importance of supports as mechanical aids during transport and the findings can be used to optimize the height of hand supports in transport. The findings of the experiments also suggest that not only the r.m.s. but also the peak value of lateral oscillations should be taken into account for preventing falls in transport. Proposed predictive model can be tested and further developed to evaluate lateral oscillations in terms of postural stability of walking people in transport.

Appendix A: Technical specifications of some equipment

A.1. Technical specifications of Kistler Gaitway® treadmill

Force – FSD

KISTLER

Type 9810AS10

GAITWAY® INSTRUMENTED TREADMILL

Gaitway® is a complete gait analysis system housed in a commercially manufactured treadmill.
What makes this treadmill unique is the ability to measure vertical ground reaction forces and center of pressure for complete, consecutive foot strikes during walking and running. The instrumented treadmill system has been designed using a patented tandem force plate design and includes a patented algorithm which distinguishes left and right foot-strikes. Gaitway simplifies the challenges associated with force plate targeting found in conventional walkway systems.

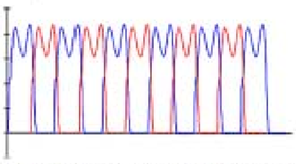
Features

- Gait analysis of consecutive steps in a matter of minutes at controlled speed
- Ideal for both clinical and research applications
- Treadmill system is an approved medical device and includes a POLAR heart rate monitor

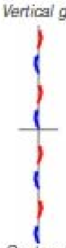
Gaitway® offers the ability to vary cadence, speed and grade. It can be programmed for speeds as low as 0,1 km/h and up to 22,0 km/h (optional 30,0 km/h). The grade can be adjusted from 0 ... 24 % (Optional –24% with reverse belt rotation).
Trials lasting several minutes or more can be easily performed, allowing subtle changes in a gait over time to be quantified. In addition, data from six auxiliary devices can be acquired simultaneously.

Measurements

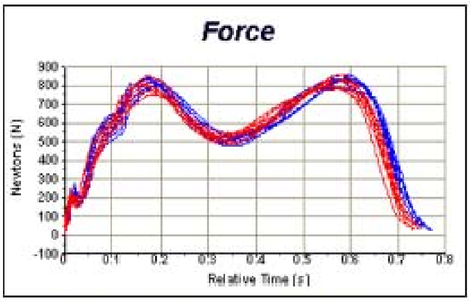
Gaitway measures the vertical ground reaction force and the center of pressure (COP) of consecutive strides in of walking and running for up to 5 minutes.



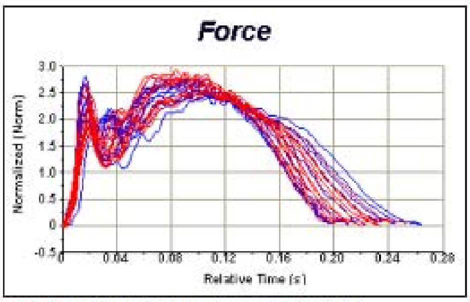
Vertical ground reaction force of consecutive steps



Center of pressure (gait line) path of consecutive steps over a distance of 4 meters



Typical ground reaction force for walking (4 km/h)



Typical ground reaction force for running (12 km/h)

Kistler Instrumente AG, PO Box, CH-8408 Winterthur
Tel + 41 - 52 - 224 11 11, Fax 224 14 14, sales@kistler.ch, www.kistler.com

000-174e-09.01 (DB06.9810Ae)

P. 2

KISTLER

Gaitway® Software

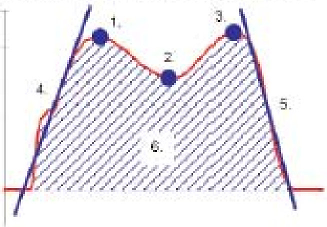
The software is the power behind the Gaitway system. Gaitway collects the data from the force sensors located in the bed of the treadmill, separates the data into left and right foot strikes, plots the results, and quickly produces a report to show everything clearly and in an easy-to-understand format. The user can selectively print and export data to other programs.

A powerful database keeps track of trials by subject name, ID, doctor, therapist, pathology, or any other user-specified classification. Multiple trials can be overlaid on a single graph, and the time base can be varied to view the data in absolute time, relative time (i.e. heel strikes aligned at time=0), percent contact, percent step, and percent gait cycle.

Gait Parameters

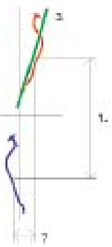
The Gaitway software automatically calculates more than 25 gait parameters. The force measurements allow for calculation of force, center of pressure (COP), and temporal (time-based) gait parameters.

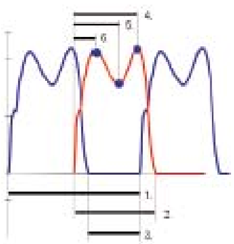
Data from auxiliary devices such as EMG and goniometers can be plotted on the same graphs. Data of multiple trials and multiple subjects can be overlaid for comparison.



Force Parameters

1. First Peak force
2. Mid support force
3. Second Peak force
4. Weight acceptance rate
5. Push-off rate
6. Impulse
- Maximum force
- Peak 1 to Peak 2 ratio





COP parameters / belt speed

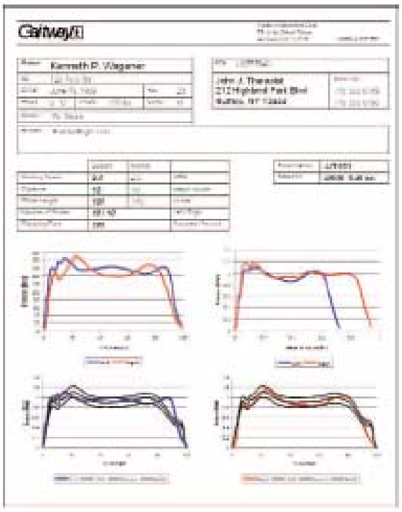
1. Step Length
2. Base of support
3. Angle of progression
- Stride Length (double stride)
- COP coordinates ax and ay versus time
- belt speed

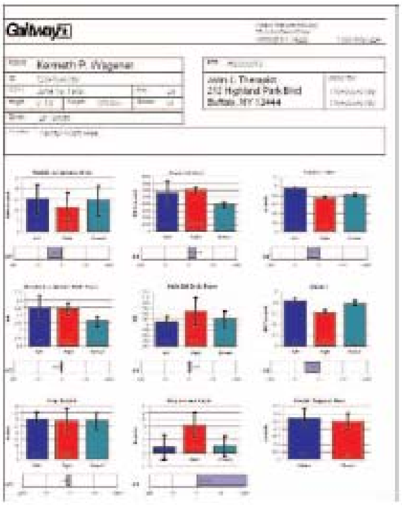
Temporal parameters

1. stride time
2. contact time
3. single limb stance time
4. time to second peak
5. time to mid-support force
6. time to first peak
- cadence

Clinical Report Manager with Normative Data

The Gaitway Clinical Report Manager quickly outputs pertinent data in an easy-to-read format, great for patient education and for physical outcome reporting. The clinical report is 100% user definable. Gait symmetry is a strong focus. All parameters are available as dimensionless symmetry index. Average and standard deviation are shown automatically.





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Tel + 41 - 52 - 224 11 11, Fax 224 14 14, sales@kistler.ch, www.kistler.com

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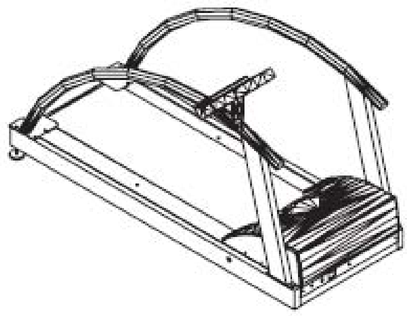
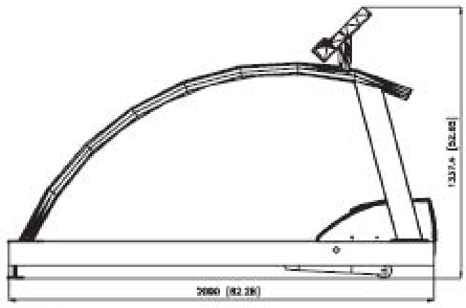
KISTLER

Technical Data

Treadmill

Treadmill Model	h/p cosmos	MERCURY®-med	
Belt Speed	km/h	0,1 ... 22,0	(option 30,0)
	mph	0,1 ... 13,6	(option 18,6)
Grade	%	0 ... 24	(option –24 ... 24)
Dimensions l x w x h	mm	2100 x 820 x 1280	
Transp. box l x w x h	mm	2250 x 1080 x 850	
Bed surface l x w	mm	1500 x 500	
Bed Height (to ground)	mm	180	
Weight	kg	364	
Power supply	VAC	220 ... 240	(options available)
	A	15	
	Hz	50 ... 60	
Max. subject weight	kg	200	
Operating Temp.	°C	–10 ... +40	
Humidity	%RH	30 ... 75	no condensation
Interface		1 x RS-232C CosCom protocol	
Heart rate measurement		POLAR sender and receiver included	
Safety standard		EN 60601-1 (IEC 601-1) isolation transformer	
MDD classification	class	II b	
Color	yellow		





Side View

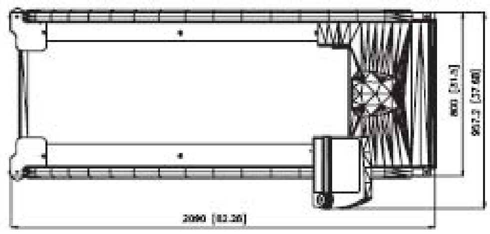
Instrumentation

Range 1(per channel)	N	≈1000	
Range 2 (per channel)	N	≈3000	
Max sampling rate	Hz	≈2500	(typical)
Max sampling time	min.	≈3	(typical)
Auxiliary channels		6	
Range	V	–5 ... +5	
A/D Board (included)		16 bit, 16 channels type PCIM-DAS1602/16	

Recommended Computer

- Windows 95 / 98 / 2000 / NT
- Empty PCI slot for full length A/D board
- CD writer or similar storage media for backup

A medical device approved computer is required in Europe (available from Tulip (http://www.tulip.com/) and other manufacturers).



Top View

Important Note:
For tropical climate (relative humidity > 60%) it is strongly recommended to use Gaitway under air-conditioning or de-humidified conditions at all times.
The patented tandem force plate layout allows measurement only above a certain step length. The subject has to be able to step over an imaginary line. Shuffling gait cannot be measured with Gaitway.

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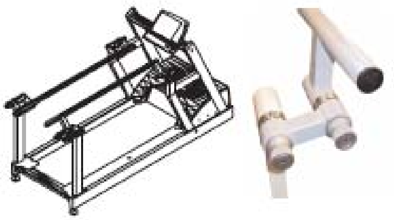
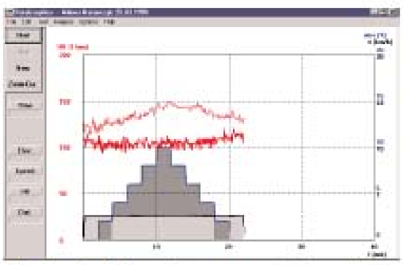
How to Order – Scope of Delivery / Text for Quotation

Item	Type	Description	
1	9810AS10	Gaitway Instrumented Treadmill with built-in force plates for vertical ground reaction force measurement of consecutive foot strikes during walking and running. Dimensions l x w x h = 2100 x 820 x 1280 mm, weight 364 kg. Consisting of: • Gaitway Treadmill with built-in force plates • Cable to A/D board l = 3m D-Sub 37pin male - D-Sub 37pin female, 1:1 • A/D board for data acquisition (PCI type full length) • Gaitway Software for data acquisition, signal processing and reporting, including Clinical Report Manager Not included • Computer Power supply configuration: ☐ 220 ... 240 VAC / 16 A / 50 ... 60 Hz standard ☐ 200 VAC / 20 A / 50 ... 60 Hz typical for Japan, 2 phase 100V ☐ 115 VAC / 20 A / 50 ... 60 Hz Speed limited 0 ... 12,0 km/h ☐ 110 VAC / 20 A / 50 ... 60 Hz Speed limited 0 ... 12,0 km/h ☐ 100 VAC / 20 A / 50 ... 60 Hz Speed limited 0 ... 12,0 km/h Packaging (one type of crate must be selected): ☐ Crate for surface and air transport HPC 001 9701 0001 dimensions 2250 x 1080 x 850 ☐ Crate for sea transport HPC 001 9701 0002	basic system
	HPC 000 9611 0008	☐ Increased speed range 0 ... 30 km/h, Only 220..240V and 200V Version	Option
	HPC 000 9810 0045	☐ Reverse belt rotation (key switch), grade -24 ... 24 %	Option
	HPC COS10030	☐ Adjustable handrails (both sides) see description	Option
	HPC 000 9805 0043	☐ Safety stop magnet switch including harness and waist belt	Option
2	HPC 000 9806 0044	Rehab-support (both sides) see description not compatible with adjustable handrails	Accessory
3	HPC 000 9611 0003	Software ParaGraphics, see description	Accessory
4	HPC 000 9701 0034	Serial cable for ParaGraphics Software, l = 5 m	Accessory

Option = must be included in first purchase Accessory = can be purchased later



Adjustable handrails *for large and small subjects*



The ***ParaGraphics*** *software communicates with Gaitway over the serial interface for registration and control of treadmill parameters and heart-rate measurements. HPC 000 9611 0003*

Rehab-support *HPC 000 9806 0044*

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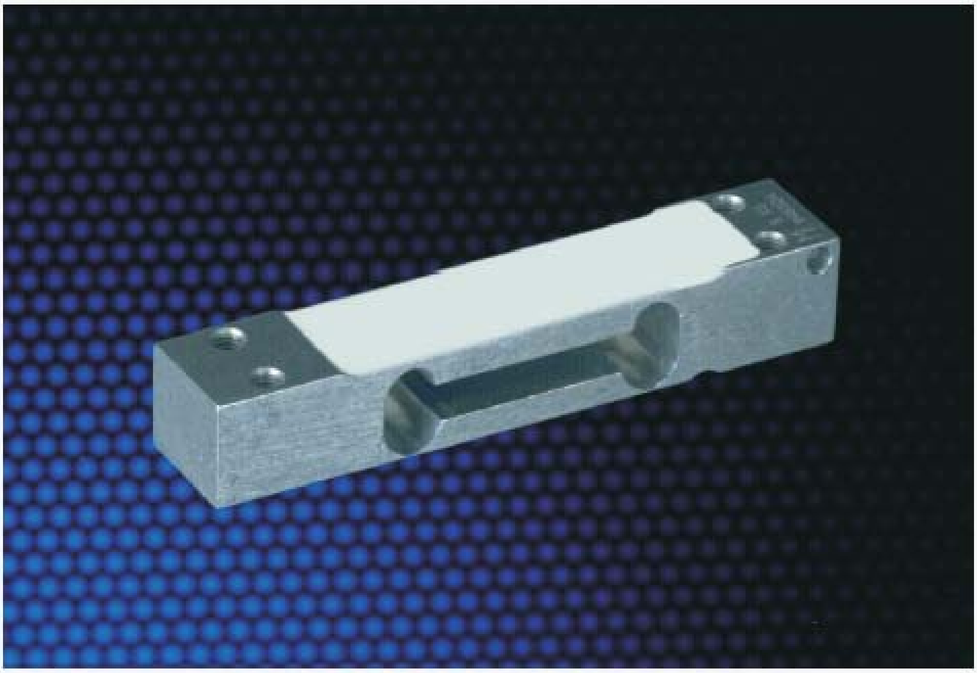
A.2. Technical specifications of Tedea Huntleigh loadcell

1022
Platform Load Cell

SENSOR
TECHNIQUES LIMITED



Platform Load Cell, Model 1022

- Approved to OIML R60 4000D
- Typical platform size 350 x 350 mm
- Available in capacity ranges from 3 kg to 200 kg
- Ideal for retail, bench or counting scales
- Low cost, low profile construction
- Easy to mount and use
- Sealed to IP66
- 4 wire cable with screen
- Optional Hazardous area approval (ATEX) available

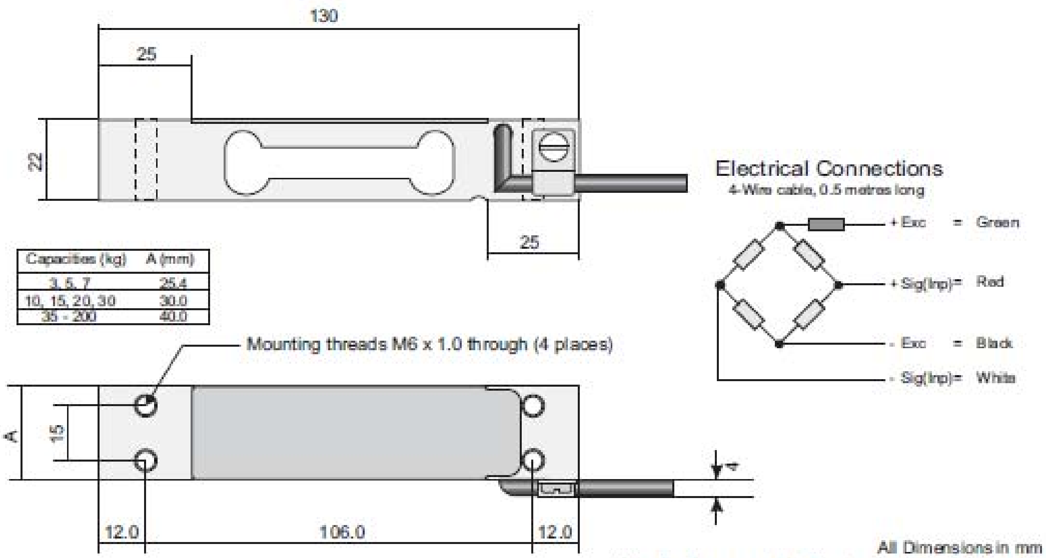
Technical Data

Model 1022

GRADE		E	G	C3	C4
Number of Load Cell Intervals	n (max)			3000	4000
Minimum Utilisation	% Rated Cap.			50	40
Minimum Verification Interval	Vmin=Emax/.			6000	10000
Total Error	% Appl. Load	0.050	0.020	0.020	0.015
Zero Return after 30 mins	% Appl. Load	0.050	0.017	0.017	0.013
Temperature Effect on : Span	%Appl. Ld./10K	0.030	0.010	0.010	0.008
Zero	%ORL/10K	0.100	0.023	0.023	0.014
Output at Rated Load (ORL)	mV/V	2			
Output at Rated Load Tolerance	%	± 10			
Input Impedance	Ohm	415 ± 15			
Output Impedance	Ohm	350 ± 3			
Recommended Supply Voltage	V	10			
Compensated Temp. Range	°C	-10 to +40			
Operating Temperature Range	°C	-20 to +70			
Deflection	mm	< 0.4			
Safe Overload	% Rated Cap.	150			
Maximum Overload	% Rated Cap.	200			
Ultimate Overload	% Rated Cap.	300			
Cable Length	m	0.5			
Typical Platform Size	mm	350 x 350			
Environmental Protection		IP66			
Rated Capacities (Emax)	Kg	3, 5, 7, 10, 15, 20, 30, 35, 50*, 100*, 150* 200*			

NMI-Certificate TC2792. * 50 - 200 kg versions not OIML approved

Dimensions



All Dimensions in mm
Specifications are subject to change without prior notice

DS1022-5, 10/07

SENSOR
TECHNIQUES LIMITED
Tel. +44 (0)1446 771185 Fax +44 (0)1446 771186

Precision Load Cells
Accessories and Mountings
Measuring Instruments and Systems

Appendix B: Instructions to subjects

B.1. Instruction sheet provided to the subjects in the first experiment reported in Chapter 4

GENERAL INFORMATION

Thank you for your participation in this experiment.

The aim of the experiment is to determine the effect of lateral motion on the postural stability of walking subjects.

The experiment has been approved by the Human Experimentation Safety and Ethics Committee of the Institute of Sound and Vibration Research (ISVR), University of Southampton.

During the experiment, you can stop the motion at any time using the emergency stop button (red button) provided. You can stop the treadmill belt using the emergency stop button on the control panel of the treadmill. You can quit the experiment at anytime without providing a reason.

PROCEDURE

You will be exposed to lateral motion while walking on the treadmill. After each motion, you will be asked to rate your ***postural instability*** by answering the question: "*What is the probability that you would lose balance if the same motion were repeated?*". ***Losing balance*** can be associated with attempting to take a protective action not to fall – such as taking a protective step, extending the arms, or grasping an object to regain equilibrium.

You can use any number between 0 and 100.

0 – indicates that if the same motion was repeated many times you are certain you would never lose balance.

25 – indicates that if the same exposure was repeated you think you would lose balance on 25% of occasions. For example, you should say '25 if:

- you think you would lose balance every fourth time the motion was presented.

50 – indicates that if the same exposure was repeated you think you would lose balance on 50% of occasions. For example, you should say '50' if:

- you think you would lose balance every other time the motion was presented.

75 – indicates that if the same exposure was repeated you think you would lose balance on 75% of occasions. For example, you should say '75' if:

- you think you would lose balance three out of four times the motion was presented.

100 – indicates that if the same exposure was repeated many times you are certain that you would lose balance every time. For example, you should say '100' if:

- you think you would lose balance every time the motion was presented.

REMEMBER

While the treadmill belt is moving, do not stop walking, do not turn around, and do not jump off the belt.

Throughout the experiment, please fix your vision on the white board in front of you.

In case of complete loss of balance, a safety harness will prevent you from falling.

Please grasp the handrails *only* if you feel very unsafe.

While walking on the treadmill please centre your body over the line across the middle of the treadmill.

B.2. Instruction sheet provided to the subjects in the second experiment reported in Chapter 5

GENERAL INFORMATION

Thank you for your participation in this experiment.

The aim of the experiment is to determine the effects of supports on the postural stability of walking subjects exposed to lateral motion.

The experiment has been approved by the Human Experimentation Safety and Ethics Committee of the Institute of Sound and Vibration Research.

During the experiment, you can stop the motion at any time using the emergency stop button (red button) provided. You can stop the treadmill belt using the emergency stop button on the control panel of the treadmill. You can quit the experiment at anytime without providing a reason.

PROCEDURE

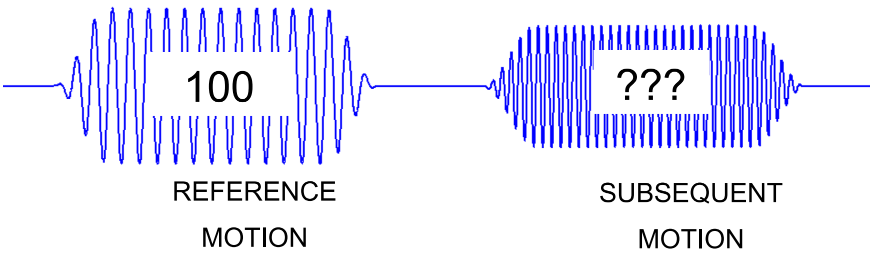
The experiment involves two parts.

In **Part A**, you are asked to judge the **discomfort or difficulty** caused by lateral motion while walking on the treadmill.

While walking, you will be exposed to pairs of motion stimuli. The first stimulus is called the reference motion, and will be the SAME throughout the experiment

Assume the **discomfort or difficulty** caused by the *first motion in each pair* (i.e. the *reference motion*) is 100.

Judge the **discomfort or difficulty** caused by the second motion in each pair relative to the **discomfort or difficulty** caused by the first motion.



For example:

- If the **discomfort or difficulty** is double that caused by the first motion, say '200'
- If the **discomfort or difficulty** is 25% more than that caused by the first motion, say '125'

- If the **discomfort or difficulty** is 25% less than that caused by the first motion, say ‘75’
- If the **discomfort or difficulty** is half that caused by the first motion, say ‘50’

During the experiment, you will be asked to hold the vertical handle support at different positions (indicated by a colour on the handle) or not to hold the support.

You may be asked to change posture (hold or not hold the support, or to hold the support at different positions) between the first motion and the second motion. Please follow the instructions given by the experimenter.

In **PART B**, there will be no reference motion. You will be exposed to various motions at unpredictable times. You will be asked to hold the support when the motion occurs. You can hold the support at whichever position you like during motion so as to stabilize your body against the motion. The position you hold does not need to match the coloured sections on the handle.

REMEMBER

Do not lean or pull on the vertical handle support unnecessarily. Only use the support to help stabilize your body against the motion.

While the treadmill belt is moving, do not stop walking, do not turn around, and do not jump off the belt.

Throughout the experiment, please fix your vision on the white board in front of you.

In case of complete loss of balance, a safety harness will prevent you from falling.

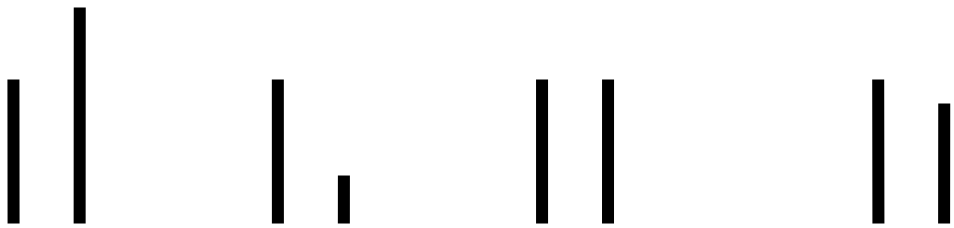
In Part A, please avoid holding any support if you are instructed not to, although you can always use the vertical handle support with your **left hand** *if really necessary*.

Try to avoid holding the treadmill handrail on the right side at any time during the experiment.

While walking on the treadmill please try to centre your body over the line across the middle of the treadmill.

------ Practice -------

To practise your judgements, please rate the length of the second line, assuming that the length of the first line is 100:



The experiment will begin with a short period of practice – so that you feel confident how to judge the **discomfort or difficulty** caused by lateral motion while walking on the treadmill.

B.3. Instruction sheet provided to the subjects in the third experiment reported in Chapter 6

GENERAL INFORMATION

Thank you for your participation in this experiment.

The aim of the experiment is to determine the effects of waveforms on the postural stability of walking subjects exposed to lateral motion.

The experiment has been approved by the Human Experimentation Safety and Ethics Committee of the Institute of Sound and Vibration Research.

During the experiment, you can stop the motion at any time using the emergency stop button (red button) provided. You can stop the treadmill belt using the emergency stop button on the control panel of the treadmill. You can quit the experiment at anytime without providing a reason.

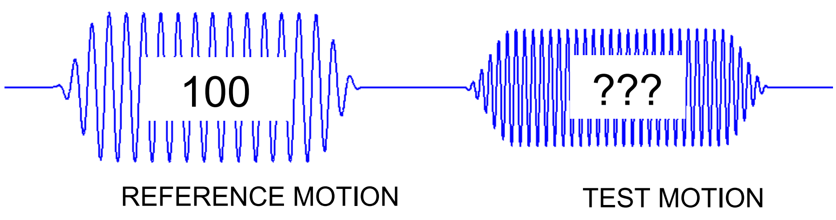
PROCEDURE

While walking, you will be exposed to pairs of motion stimuli. The first stimulus is called the REFERENCE motion.

There will be two parts involved in the experiment. After the first part is finished, a rest break will be provided. The procedure in the second part will be exactly the same as in the first part except that the REFERENCE motion will be different. Please follow the instructions given by the experimenter.

After exposure to each pair of motion stimuli, you will be asked to answer TWO questions.

For the FIRST question, you will be asked to judge the ***relative*** **discomfort or difficulty** caused by lateral motion while walking on the treadmill. Assume the **discomfort or difficulty** caused by the *first motion* (i.e. the *reference motion*) *in each pair* is 100. Judge the **discomfort or difficulty** caused by the second motion (i.e. test motion) in each pair *relative* to the **discomfort or difficulty** caused by the first motion (i.e. reference motion).



For example:

- If the **discomfort or difficulty** is double that caused by the first motion, say '200'

- If the **discomfort or difficulty** is 25% more than that caused by the first motion, say '125'
- If the **discomfort or difficulty** is 25% less than that caused by the first motion, say '75'
- If the **discomfort or difficulty** is half that caused by the first motion, say '50'

For the SECOND question, you will be asked to give an absolute rating for your ***postural instability*** by answering the question: "*What is the probability that you would lose balance if the same **test motion** were repeated?*". Your judgement should be based on the effect of the second motion (i.e. test motion) on your postural instability regardless of the first motion. ***Losing balance*** can be associated with attempting to take a protective action not to fall – such as taking a protective step, extending the arms, or grasping an object to regain equilibrium.

You can use *any number* between 0 and 100.

0 – indicates that if the same motion was repeated many times you are certain you would never lose balance.

25 – indicates that if the same exposure was repeated you think you would lose balance on 25% of occasions. For example, you should say '25 if:

- you think you would lose balance every fourth time the motion was presented.

50 – indicates that if the same exposure was repeated you think you would lose balance on 50% of occasions. For example, you should say '50' if:

- you think you would lose balance every other time the motion was presented.

100 – indicates that if the same exposure was repeated many times you are certain that you would lose balance every time. For example, you should say '100' if:

- you think you would lose balance every time the motion was presented.

REMEMBER

While the treadmill belt is moving, do not stop walking, do not turn around, and do not jump off the belt.

Throughout the experiment, please fix your vision on the white board in front of you.

In case of complete loss of balance, a safety harness will prevent you from falling.

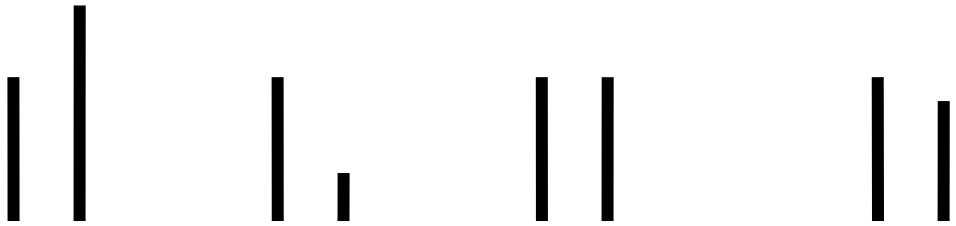
Please grasp the handrails *only* if you feel very unsafe.

While walking on the treadmill please try to centre your body over the line across the middle of the treadmill.

------ Practice -------

To practise your *relative* judgements, please rate the length of the second line, assuming that the length of the first line is 100.

To practise your *absolute* judgement, please rate the length of the second line in terms of centimetres.



Relative:	Relative:	Relative:	Relative:
Absolute:	Absolute:	Absolute:	Absolute:

The experiment will begin with a short period of practice – so that you feel confident about your relative and absolute judgements.

B.4. Instruction sheet provided to the subjects in the fourth experiment reported in Chapter 7

GENERAL INFORMATION

Thank you for your participation in this experiment.

The aim of the experiment is to determine the effect of subject physical characteristics (e.g. age, weight and height) on the postural stability of walking subjects exposed to lateral motion.

The experiment has been approved by the Human Experimentation Safety and Ethics Committee of the Institute of Sound and Vibration Research (ISVR), University of Southampton.

During the experiment, you can stop the motion at any time using the emergency stop button (red button) provided. You can stop the treadmill belt using the emergency stop button on the control panel of the treadmill. You can quit the experiment at anytime without providing a reason.

PROCEDURE

You will be exposed to various lateral motions while walking on the treadmill.

After each motion, you will be asked to rate your ***postural instability*** by answering the question:

> "*What is the probability that you would lose balance if the same motion were repeated?*

Losing balance can be associated with attempting to take a protective action not to fall – such as taking a protective step, extending the arms, or grasping an object to regain equilibrium.

You can use **any number** between 0 and 100.

0 – indicates that if the same motion was repeated many times you are certain you would never lose balance.

25 – indicates that if the same exposure was repeated you think you would lose balance on 25% of occasions. For example, you should say '25 if:

- you think you would lose balance every fourth time the motion was presented.

62 – indicates that if the same exposure was repeated you think you would lose balance on 62% of occasions. For example, you should say '62' if:

- you think you would lose balance in 62 occasions if the motion was presented 100 times.

100 – indicates that if the same exposure was repeated many times you are certain that you would lose balance every time. For example, you should say '100' if:

- you think you would lose balance every time the motion was presented.

REMEMBER

While the treadmill belt is moving, do not stop walking, do not turn around, and do not jump off the belt.

Throughout the experiment, please fix your vision on the white board in front of you.

In case of complete loss of balance, a safety harness will prevent you from falling.

Please grasp the handrails if you feel unsafe.

While walking on the treadmill please centre your body over the line across the middle of the treadmill.

Appendix C: Data

C.1. Data used in the analysis of the first experiment reported in Chapter 4

Table C.1: Reported probability of losing balance as a function of acceleration at each frequency of lateral oscillation.

0.5 Hz	Acceleration (ms^{-2} r.m.s.)							
Subject no	0.1	0.125	0.16	0.2	0.25	0.315	0.4	0.5
1	10	30	60	50	40	50	90	100
2	70	75	10	30	90	60	95	100
3	25	20	80	60	50	80	100	100
4	25	50	25	75	50	100	100	100
5	5	5	15	15	15	10	35	25
6	5	15	5	35	20	35	100	100
7	7	18	95	10	80	95	100	100
8	48	5	35	37	69	70	95	40
9	1	15	2	7	10	33	40	50
10	0	2	5	40	40	5	90	100
11	5	15	5	30	35	45	40	80
12	40	10	40	10	30	40	70	100
13	0	0	10	5	15	75	80	80
14	45	10	20	35	70	30	50	50
15	15	80	60	65	25	35	100	70
16	5	20	60	45	35	25	100	100
17	0	10	50	50	80	90	95	100
18	20	30	75	40	80	100	100	100
19	0	0	0	30	70	0	90	10
20	75	20	65	70	55	60	100	100
MEDIAN	8.50	15.00	30.00	36.00	45.00	47.50	95.00	100.00
0.63 Hz	0.125	0.16	0.20	0.25	0.315	0.40	0.50	0.63
1	5	10	10	70	80	40	90	90
2	50	30	30	80	80	60	85	80
3	30	80	25	50	60	50	100	100
4	25	25	100	100	75	100	100	75
5	0	0	5	10	25	30	25	15
6	5	17	15	35	40	20	30	70
7	0	15	25	45	90	85	95	100
8	1	1	6	55	92	85	82	70
9	5	5	2	2	4	20	10	40
10	0	0	15	20	15	60	30	75
11	0	10	15	10	20	30	40	75
12	0	10	0	0	20	20	70	80
13	0	0	5	30	5	50	45	90
14	2	10	15	40	60	60	75	50
15	5	15	50	30	85	50	75	90
16	10	40	10	50	60	60	80	100
17	0	0	15	50	75	95	60	100

18	20	20	40	30	80	75	60	100
19	5	0	40	10	70	40	80	40
20	60	80	35	65	50	95	100	75
MEDIAN	5.00	10.00	15.00	37.50	60.00	55.00	75.00	77.50
0.8 Hz	0.16	0.20	0.25	0.315	0.40	0.50	0.63	0.80
1	20	30	10	40	40	80	70	90
2	5	20	12	65	45	80	85	95
3	20	20	40	25	40	80	40	80
4	10	25	25	25	50	100	75	100
5	2	10	5	5	15	20	10	35
6	10	0	35	10	45	55	20	90
7	5	4	8	80	55	75	95	100
8	8	38	3	12	92	45	58	97
9	1	5	4	2	3	40	15	80
10	0	5	0	5	10	90	10	40
11	5	0	40	10	40	40	60	55
12	50	30	20	30	60	40	100	90
13	0	15	5	5	25	20	70	20
14	10	15	5	40	50	25	5	80
15	5	35	20	30	50	75	80	100
16	8	5	0	20	85	90	95	40
17	10	0	15	10	65	50	100	95
18	20	60	25	40	70	20	50	60
19	0	5	30	40	80	50	50	100
20	20	50	30	20	65	45	50	100
MEDIAN	8.00	15.00	13.50	22.50	50.00	50.00	59.00	90.00
1.0 Hz	0.20	0.25	0.315	0.40	0.50	0.63	0.80	1.00
1	30	15	25	70	50	70	90	100
2	5	35	45	15	10	20	50	70
3	20	25	40	50	60	25	80	90
4	0	0	25	75	100	75	100	100
5	5	0	0	15	20	20	50	35
6	5	50	37	15	20	35	70	85
7	30	2	15	45	75	75	98	98
8	1	3	16	12	38	62	82	92
9	2	5	15	5	10	3	15	60
10	25	0	5	10	5	30	40	75
11	10	30	20	30	20	50	50	50
12	0	40	60	30	20	70	80	100
13	0	10	0	35	15	20	65	90
14	5	20	60	20	20	40	60	50
15	10	60	90	40	30	50	60	85
16	5	7	10	3	25	35	80	95
17	15	25	80	15	25	50	95	75
18	40	50	75	40	90	100	60	100
19	10	0	0	30	60	50	80	90
20	10	20	35	25	45	75	100	90
MEDIAN	7.50	17.50	25.00	27.50	25.00	50.00	75.00	90.00
1.25 Hz	0.25	0.315	0.40	0.50	0.63	0.80	1.00	1.25
1	30	40	50	60	90	80	100	90
2	20	10	10	20	45	20	80	90

3	25	25	40	50	50	80	100	80
4	0	0	50	50	50	75	75	100
5	0	5	10	20	15	10	20	30
6	5	10	40	65	10	75	65	75
7	8	3	30	17	35	60	80	99
8	15	27	75	18	49	37	88	95
9	2	1	3	10	6	50	15	40
10	0	3	5	15	20	30	70	65
11	15	30	10	25	20	35	45	50
12	10	40	10	20	40	60	60	100
13	0	0	5	0	20	50	85	55
14	15	40	30	10	30	30	60	75
15	40	10	30	50	60	80	100	95
16	0	0	40	50	10	35	90	85
17	25	25	40	15	15	65	75	100
18	50	40	25	75	40	90	80	100
19	20	10	50	20	100	60	60	95
20	10	30	45	20	35	50	65	95
MEDIAN	12.50	10.00	30.00	20.00	35.00	55.00	75.00	90.00
1.6 Hz	0.315	0.40	0.50	0.63	0.80	1.00	1.25	1.60
1	50	70	10	50	60	80	100	100
2	5	10	15	60	30	20	65	90
3	15	25	25	75	40	60	100	50
4	10	10	0	50	100	75	75	100
5	5	20	10	15	20	25	20	15
6	0	5	15	75	55	65	100	55
7	1	35	30	50	25	95	90	95
8	5	17	22	20	89	69	30	50
9	3	3	35	75	10	10	75	75
10	5	2	10	5	25	60	50	70
11	5	25	20	25	35	20	50	60
12	10	10	20	20	80	60	70	100
13	0	5	15	20	30	35	0	60
14	10	5	15	10	60	60	35	75
15	20	70	45	50	60	80	75	80
16	3	0	35	15	0	25	85	85
17	15	65	15	50	90	75	75	100
18	20	30	75	50	75	60	100	100
19	0	20	10	25	40	75	30	25
20	25	35	20	50	90	95	75	80
MEDIAN	5.00	18.50	17.50	50.00	47.50	60.00	75.00	77.50
2.0 Hz	0.40	0.50	0.63	0.80	1.00	1.25	1.60	2.00
1	30	20	50	30	60	70	50	80
2	5	10	15	10	20	20	30	40
3	15	40	40	50	50	50	80	100
4	0	0	25	0	10	50	50	100
5	0	10	5	10	5	15	7	20
6	5	0	20	15	35	45	60	85
7	3	10	40	5	35	85	65	90
8	3	28	48	67	47	72	67	88
9	4	2	3	40	55	30	70	80

10	0	5	10	10	0	20	15	75
11	5	5	10	25	35	25	20	55
12	20	20	10	40	25	60	70	90
13	5	15	0	5	25	30	40	55
14	15	25	10	40	40	30	30	70
15	50	40	40	10	25	70	45	70
16	0	5	0	20	5	40	45	55
17	15	25	50	20	40	65	75	100
18	30	20	30	30	50	50	60	80
19	10	0	50	10	0	60	70	70
20	10	30	75	50	50	75	90	100
MEDIAN	5.00	12.50	22.50	20.00	35.00	50.00	55.00	80.00

Table C.2: Peak to peak lateral COP position as a function of acceleration at each frequency of lateral oscillation.

0.5 Hz	Acceleration (ms^{-2} r.m.s.)							
Subject no	0.1	0.125	0.16	0.2	0.25	0.315	0.4	0.5
1	27.91	25.33	31.63	31.73	29.62	32.90	34.21	35.63
2	34.02	31.88	19.90	20.32	35.62	33.08	34.28	35.10
3	22.70	20.79	23.69	22.83	24.48	26.65	27.61	24.39
4	27.22	29.21	32.73	30.78	28.61	28.74	35.69	29.99
5	23.36	22.30	30.98	29.77	22.53	37.06	33.45	44.89
6	25.53	27.86	25.69	31.37	24.14	30.38	31.30	36.29
7	18.37	21.12	28.37	24.27	23.12	29.80	31.36	46.15
8	25.20	24.54	23.64	23.11	25.60	32.43	30.95	31.96
9	21.65	24.42	27.50	24.95	25.43	28.26	26.75	29.34
10	26.80	24.45	26.18	28.79	33.74	29.43	33.42	50.91
11	34.42	35.69	31.11	35.30	33.54	41.54	35.37	36.40
12	27.46	26.78	30.33	26.33	29.72	29.28	33.83	31.08
13	22.25	20.99	27.30	24.61	29.93	33.31	39.25	49.73
14	35.42	28.12	34.19	27.85	36.90	35.57	32.93	40.18
15	23.68	28.80	31.84	29.42	23.66	27.93	26.30	32.59
16	30.40	28.98	33.74	32.03	35.23	26.77	35.80	32.55
17	27.71	30.63	36.57	34.90	33.02	38.13	35.68	42.16
18	30.96	24.90	29.62	27.14	37.47	45.03	48.17	44.18
19	24.85	26.52	25.04	30.65	34.44	25.99	34.89	31.27
20	31.47	26.36	32.49	28.55	28.52	24.22	27.62	22.18
MEDIAN	27.01	26.44	29.98	28.67	29.67	30.09	33.64	35.37
0.63 Hz	0.125	0.16	0.20	0.25	0.315	0.40	0.50	0.63
1	26.75	30.17	23.44	29.68	28.74	28.83	29.56	38.72
2	32.97	29.76	22.93	32.58	27.83	20.40	23.23	35.87
3	22.96	24.40	20.30	25.34	21.83	26.64	19.39	28.52
4	26.88	28.89	29.74	28.69	30.68	26.80	22.84	26.14
5	22.83	21.25	18.01	26.91	23.73	33.13	31.21	33.64
6	25.10	22.92	29.29	27.72	27.22	25.15	25.63	35.77
7	19.83	23.51	19.00	26.51	22.48	24.58	25.73	33.52
8	27.30	21.81	23.45	24.54	25.70	29.84	25.08	28.67
9	26.27	29.91	22.95	23.12	22.25	24.15	26.07	32.92

10	20.44	21.46	27.36	29.23	25.43	27.50	27.95	40.88
11	30.23	32.32	30.95	31.85	33.77	35.39	31.09	35.52
12	25.94	33.70	29.85	21.89	35.07	29.07	36.12	31.22
13	23.38	26.12	15.11	25.45	25.13	20.68	29.29	37.68
14	28.26	34.39	23.76	33.31	34.12	33.59	26.36	32.53
15	24.06	30.06	28.26	32.65	19.27	21.80	26.73	38.35
16	29.61	27.44	30.19	39.11	27.71	35.90	31.78	27.40
17	24.95	28.11	26.09	28.01	33.78	19.08	34.49	28.18
18	23.98	19.93	23.75	22.50	26.13	24.08	31.46	46.22
19	29.77	27.10	33.22	23.84	37.00	32.46	35.20	42.75
20	24.68	32.55	21.77	20.15	22.57	23.44	16.79	26.21
MEDIAN	25.52	27.78	23.76	27.32	26.67	26.72	27.34	33.58
0.8 Hz	0.16	0.20	0.25	0.315	0.40	0.50	0.63	0.80
1	27.40	24.49	27.47	28.04	23.37	27.99	37.44	30.85
2	15.00	17.55	21.40	20.01	22.49	25.92	28.08	20.83
3	19.13	18.62	23.47	18.91	21.54	26.41	28.12	26.77
4	25.01	24.12	24.42	23.70	20.22	31.05	37.45	26.15
5	24.21	26.55	21.32	24.08	32.50	20.76	20.16	28.94
6	27.06	18.77	30.10	26.90	25.43	24.84	30.15	23.73
7	18.43	16.97	23.65	26.21	22.33	23.85	30.90	24.23
8	23.49	24.03	23.51	25.61	21.55	24.75	26.36	23.72
9	21.57	23.55	26.95	26.18	17.45	25.09	30.38	34.57
10	21.91	25.40	20.87	22.69	26.98	32.67	30.39	29.64
11	32.60	31.82	34.32	31.28	39.10	28.98	26.50	38.51
12	29.17	24.35	30.50	34.52	32.42	27.56	27.01	35.34
13	13.92	21.66	22.40	24.38	23.77	29.51	21.36	29.19
14	25.41	32.18	29.62	32.21	28.99	28.11	33.50	32.93
15	23.10	21.52	20.95	31.26	22.06	22.88	26.99	28.90
16	27.63	33.88	29.34	29.87	23.72	39.32	29.61	34.44
17	28.28	24.18	31.21	30.31	30.22	34.29	27.79	36.60
18	19.76	25.64	25.38	34.42	27.23	28.43	29.22	37.29
19	24.60	26.55	25.71	28.77	30.94	21.80	30.09	32.68
20	20.89	19.83	28.34	26.27	28.46	23.38	32.03	23.05
MEDIAN	23.85	24.15	25.55	26.59	24.60	26.98	29.42	29.42
1.0 Hz	0.20	0.25	0.315	0.40	0.50	0.63	0.80	1.00
1	24.74	28.23	28.29	25.83	26.66	28.42	35.18	36.44
2	13.76	21.03	18.15	19.02	15.95	20.05	23.58	26.38
3	17.32	22.73	21.06	24.62	20.22	19.80	19.96	24.76
4	26.29	20.15	23.50	29.41	30.70	31.69	28.57	29.12
5	15.80	19.91	24.45	23.97	24.95	24.58	29.51	24.16
6	21.27	19.97	27.85	23.66	22.84	27.09	30.56	33.86
7	20.23	21.47	23.41	21.59	22.48	25.26	26.56	26.72
8	23.57	22.54	24.84	23.37	22.26	24.54	25.73	26.21
9	21.92	20.78	25.25	26.74	26.04	21.36	29.65	31.34
10	30.93	19.42	22.56	23.47	27.05	26.56	23.21	25.78
11	36.65	36.57	35.96	34.65	39.32	33.01	33.64	39.63
12	22.82	32.28	28.48	25.60	29.47	34.99	31.89	33.14
13	21.30	20.38	19.04	29.34	25.13	19.94	21.99	32.60
14	25.75	28.33	33.31	28.65	31.15	30.62	32.92	32.71
15	25.03	27.21	32.25	20.48	20.53	26.62	25.36	29.17
16	28.51	27.71	29.95	30.58	30.56	36.17	30.67	29.37

17	26.51	27.33	30.63	27.92	28.56	32.57	32.73	36.49
18	20.10	27.42	22.31	23.43	28.58	26.40	29.86	23.81
19	25.45	33.84	20.42	27.33	24.44	34.98	33.35	38.34
20	22.29	18.67	14.01	27.20	25.42	21.08	18.84	22.49
MEDIAN	23.20	22.63	24.65	25.71	25.73	26.59	29.58	29.27
1.25 Hz	0.25	0.315	0.40	0.50	0.63	0.80	1.00	1.25
1	24.24	25.37	24.00	27.06	29.62	27.60	33.83	23.86
2	18.33	15.46	17.43	23.04	25.07	14.71	23.84	20.17
3	16.45	19.53	18.95	20.06	18.89	21.93	28.62	25.34
4	18.08	21.66	20.89	24.86	24.32	23.01	23.72	29.27
5	20.28	18.71	20.61	23.69	27.12	23.34	19.29	27.94
6	20.00	23.73	21.03	27.58	23.27	24.60	27.87	24.31
7	18.27	17.27	19.14	21.87	21.49	22.81	20.46	22.51
8	22.52	25.16	20.01	20.00	22.40	21.14	18.69	24.87
9	16.71	21.15	23.86	24.59	24.00	25.24	26.69	27.38
10	22.96	24.89	20.12	22.69	26.86	24.58	23.09	31.75
11	34.71	32.83	32.42	30.87	32.26	30.50	37.05	36.83
12	24.65	23.77	23.99	23.15	23.72	29.81	29.64	31.64
13	15.66	16.67	21.77	13.72	21.22	22.32	20.24	28.28
14	24.44	29.64	29.81	26.36	27.95	29.69	28.61	31.22
15	19.27	20.61	22.01	15.04	25.70	19.69	26.55	21.57
16	34.58	25.33	24.23	28.68	24.89	34.17	30.73	29.00
17	27.62	27.58	24.74	28.26	25.74	29.20	29.41	32.49
18	19.97	20.60	26.15	19.97	16.22	16.12	18.62	33.53
19	24.35	23.99	25.63	25.35	36.08	26.32	31.73	26.89
20	21.75	19.81	20.84	18.75	19.16	17.28	21.01	26.12
MEDIAN	21.01	22.70	21.89	23.42	24.61	23.96	26.62	27.66
1.6 Hz	0.315	0.40	0.50	0.63	0.80	1.00	1.25	1.60
1	23.26	23.56	26.88	26.33	25.56	25.59	29.66	32.73
2	12.80	9.15	14.32	13.29	13.58	12.10	13.61	19.17
3	17.87	15.90	17.13	17.99	20.89	19.63	24.89	22.72
4	16.14	20.70	20.53	18.98	24.39	24.67	28.93	37.65
5	16.06	19.88	21.25	25.04	18.94	26.66	30.21	34.03
6	19.11	19.84	22.17	26.84	22.72	23.64	34.17	27.72
7	12.55	19.55	20.04	21.55	15.44	23.03	24.68	22.87
8	21.25	23.17	24.10	20.28	22.92	22.77	23.62	23.24
9	24.15	23.16	22.55	27.88	24.13	30.57	26.71	35.59
10	16.21	17.98	17.45	20.67	22.44	21.99	27.23	32.20
11	26.67	34.19	28.77	32.36	33.51	31.75	24.59	28.74
12	20.01	24.45	25.22	21.95	22.63	25.80	21.49	65.54
13	18.97	13.78	23.26	22.95	23.38	23.34	20.15	24.79
14	24.43	23.86	18.97	18.77	27.04	26.37	33.61	32.79
15	25.26	20.41	23.17	24.89	23.36	26.62	26.38	22.48
16	23.22	23.05	20.76	20.52	26.02	22.47	20.31	24.85
17	20.24	25.42	20.41	24.63	27.57	25.38	24.22	23.35
18	16.87	14.36	14.74	21.30	16.81	22.97	17.09	29.65
19	25.25	21.65	26.81	30.05	25.41	32.08	38.22	36.93
20	14.41	19.73	17.27	22.73	15.21	16.43	14.53	16.90
MEDIAN	19.56	20.55	21.01	22.34	23.14	24.15	24.78	28.23
2.0 Hz	0.40	0.50	0.63	0.80	1.00	1.25	1.60	2.00
1	24.91	22.76	22.43	27.79	26.85	27.82	31.80	29.74

2	13.05	17.38	11.06	11.93	16.96	16.12	23.37	14.02
3	17.04	15.45	21.15	16.65	15.27	18.88	19.27	15.98
4	17.51	18.30	19.82	16.41	23.80	21.85	26.59	26.27
5	19.04	19.40	16.83	17.76	18.29	23.85	19.45	22.50
6	20.67	23.36	21.27	18.49	22.23	20.70	23.11	30.99
7	12.99	16.86	15.55	18.17	20.92	11.10	21.65	19.71
8	22.29	21.48	22.87	24.51	21.14	24.43	22.16	25.23
9	24.41	25.14	17.77	26.90	25.18	26.04	37.09	38.15
10	19.71	17.65	19.08	17.43	25.02	24.46	23.78	26.95
11	29.23	31.58	28.55	29.24	25.71	30.40	32.82	25.06
12	22.87	25.33	23.09	24.36	21.40	24.19	25.73	19.03
13	17.16	20.08	19.43	16.82	22.37	16.71	21.84	23.99
14	19.97	22.53	22.99	22.93	23.97	26.49	26.35	28.58
15	26.57	16.45	14.18	22.03	16.88	20.93	21.51	27.69
16	23.53	17.52	20.84	20.77	19.68	26.91	24.34	25.56
17	23.66	23.18	28.40	23.80	23.51	20.52	27.17	29.34
18	16.02	13.01	22.52	20.36	15.98	17.32	19.71	18.57
19	24.09	24.19	17.89	23.31	24.69	28.24	30.79	28.04
20	16.29	15.49	19.89	13.30	17.44	12.33	13.95	14.79
MEDIAN	20.32	19.74	20.36	20.57	21.82	22.85	23.58	25.40

Table C.3: Lateral r.m.s. COP velocity as a function of acceleration at each frequency of lateral oscillation.

0.5 Hz	Acceleration (ms^{-2} r.m.s.)							
Subject no	0.1	0.125	0.16	0.2	0.25	0.315	0.4	0.5
1	53.70	51.73	61.15	60.28	56.81	65.60	61.36	68.75
2	28.42	25.66	22.39	21.45	30.45	43.19	41.70	37.16
3	40.07	37.08	37.19	40.74	37.82	34.08	37.12	33.62
4	47.62	43.02	56.13	52.59	43.22	48.38	51.03	52.35
5	33.50	33.41	44.48	37.91	34.56	55.95	49.93	71.41
6	36.89	43.22	44.89	45.51	44.25	49.03	33.33	48.67
7	28.52	27.36	36.95	30.77	34.20	54.07	48.33	60.42
8	46.01	44.82	48.00	51.80	49.34	51.25	58.45	50.99
9	41.14	42.85	53.87	52.71	56.13	63.47	60.22	72.87
10	41.16	42.25	38.27	37.80	43.92	46.35	44.82	72.46
11	62.84	60.24	49.83	65.30	63.69	60.16	56.33	54.35
12	42.36	50.82	49.37	52.72	50.50	56.94	69.78	46.32
13	35.86	35.43	37.91	36.32	35.39	47.26	45.88	83.38
14	45.02	39.80	49.35	35.78	39.28	42.80	45.34	52.32
15	36.77	33.75	48.64	42.49	31.40	43.02	32.92	42.71
16	51.91	41.84	48.66	55.89	59.91	43.87	59.26	58.92
17	51.38	46.81	48.97	51.18	44.98	55.06	48.59	52.75
18	31.63	27.30	30.37	33.17	41.03	50.48	44.66	44.97
19	51.95	46.96	48.94	52.04	61.73	49.45	54.46	71.70
20	33.05	33.43	29.96	32.06	37.81	27.67	31.62	28.76
MEDIAN	41.15	42.04	48.32	44.00	43.57	49.24	48.46	52.55

0.63 Hz	0.125	0.16	0.20	0.25	0.315	0.40	0.50	0.63
1	55.44	62.10	47.25	56.28	57.72	49.44	52.15	69.36
2	26.19	35.47	21.95	30.83	32.65	31.36	31.44	56.33
3	37.04	41.32	40.06	42.71	32.65	47.17	29.51	46.75
4	42.09	46.19	43.12	47.17	48.89	46.03	39.20	43.83
5	44.91	40.38	29.08	30.59	30.38	39.57	39.90	47.31
6	42.65	42.16	36.57	48.63	41.99	42.16	51.56	60.15
7	30.63	43.27	32.12	40.12	33.08	39.97	43.92	61.01
8	46.14	51.42	44.99	51.97	53.74	45.45	53.21	53.62
9	56.01	59.29	37.61	47.26	46.10	50.85	62.85	84.67
10	31.18	41.72	40.99	42.13	41.22	42.84	48.08	69.18
11	59.69	65.20	60.75	61.09	61.43	66.44	56.21	54.63
12	49.10	54.47	55.71	37.29	60.41	52.35	59.84	72.09
13	35.84	38.03	24.01	39.39	40.74	28.63	44.19	56.85
14	44.91	50.77	25.30	51.92	51.28	44.67	34.16	56.79
15	28.78	50.52	38.42	43.81	27.52	25.99	35.67	78.81
16	46.44	42.02	44.29	56.47	45.18	69.49	41.73	48.74
17	46.21	56.79	48.29	49.62	56.20	25.26	65.09	38.51
18	32.66	29.22	22.15	26.16	25.17	32.61	42.18	60.74
19	54.41	48.10	65.86	50.92	49.85	58.66	71.43	91.77
20	21.73	38.16	29.48	25.58	26.47	24.78	23.26	32.92
MEDIAN	43.78	44.73	39.24	45.49	43.58	43.75	44.06	56.82
0.8 Hz	0.16	0.20	0.25	0.315	0.40	0.50	0.63	0.80
1	56.08	53.66	58.61	64.68	51.36	54.74	82.46	74.24
2	17.76	27.17	33.17	30.24	37.20	25.29	49.95	23.01
3	37.89	41.85	42.74	31.05	35.73	40.62	53.40	62.82
4	48.37	46.10	48.55	54.85	38.73	52.40	74.62	48.05
5	36.28	42.31	31.27	53.50	48.82	22.49	34.40	53.71
6	38.31	35.68	56.12	49.80	41.07	33.62	68.92	38.48
7	25.81	30.04	36.44	52.20	39.35	47.39	64.55	27.85
8	46.58	50.58	51.22	43.76	30.21	60.24	67.51	59.09
9	45.15	42.80	57.61	59.01	32.99	51.18	64.49	59.54
10	41.46	39.63	33.84	31.45	41.45	42.29	56.50	58.39
11	56.60	55.44	64.44	68.58	70.93	35.44	32.77	78.40
12	55.74	44.42	49.62	66.60	64.42	60.47	45.90	78.95
13	27.91	32.58	44.75	45.59	38.10	50.78	21.47	61.11
14	41.75	40.22	55.21	46.14	43.23	50.07	68.71	64.43
15	33.80	31.57	27.81	47.52	26.57	41.88	55.58	28.45
16	53.99	69.58	52.03	45.57	40.90	58.96	54.74	73.39
17	57.71	49.14	63.25	60.58	62.21	58.73	44.19	79.87
18	29.58	43.06	35.53	53.54	53.53	50.05	56.62	60.12
19	47.88	53.06	57.73	57.21	54.57	28.65	62.12	69.60
20	34.92	24.50	34.51	47.66	33.90	39.79	61.25	38.19
MEDIAN	41.61	42.55	49.08	51.00	40.99	48.72	56.56	59.83
1.0 Hz	0.20	0.25	0.315	0.40	0.50	0.63	0.80	1.00
1	53.08	66.84	67.91	59.29	61.02	51.05	88.97	83.56
2	17.92	26.09	25.34	20.28	21.29	30.10	31.88	36.49
3	37.88	42.48	47.29	44.58	44.65	37.83	33.99	63.67
4	50.95	47.75	54.66	50.06	42.74	58.61	40.90	57.83
5	29.24	37.76	53.52	42.24	42.65	49.00	26.59	32.44
6	35.94	39.90	52.09	52.83	43.35	51.24	54.66	48.57

7	40.65	42.90	39.65	40.16	42.96	53.90	49.71	56.47
8	45.07	47.97	53.83	59.62	63.57	60.48	61.23	73.26
9	42.13	48.72	67.16	61.59	57.10	46.59	79.48	87.80
10	38.24	36.22	39.10	38.36	46.23	39.02	38.56	39.73
11	61.04	83.03	64.17	60.27	61.64	59.49	75.73	77.69
12	48.09	72.32	59.06	57.15	60.64	74.36	86.81	66.61
13	35.34	36.00	38.77	40.21	44.83	33.23	34.31	60.45
14	40.34	54.52	67.64	47.57	52.36	71.63	62.94	88.48
15	39.72	45.14	63.66	32.41	34.29	36.32	45.07	72.07
16	51.48	40.73	62.95	59.71	67.90	58.53	61.86	51.44
17	48.87	45.46	59.63	59.64	54.26	67.39	74.91	89.18
18	32.31	28.26	30.25	31.81	40.74	24.67	69.01	21.18
19	57.02	78.31	41.45	45.41	40.18	85.11	92.05	120.44
20	35.78	30.84	26.37	30.29	34.05	26.48	23.78	24.78
MEDIAN	40.49	44.02	53.67	46.49	44.74	51.14	57.94	62.06
1.25 Hz	0.25	0.315	0.40	0.50	0.63	0.80	1.00	1.25
1	53.32	59.73	56.48	52.95	72.53	57.90	81.95	57.63
2	27.92	22.19	26.51	34.22	26.04	24.34	25.00	28.53
3	31.53	37.89	40.55	44.93	33.75	39.44	38.66	52.65
4	40.79	48.82	42.28	52.02	47.75	46.07	45.42	46.21
5	40.31	26.86	44.19	36.76	43.57	44.23	37.52	44.73
6	37.21	47.76	46.89	46.95	48.86	59.14	63.68	54.47
7	35.13	28.40	29.62	32.14	37.99	34.23	31.55	45.34
8	44.70	58.08	51.41	48.00	58.29	56.21	49.83	68.93
9	35.73	48.29	48.81	62.34	60.26	65.36	83.35	72.26
10	36.52	56.93	37.09	41.10	43.58	47.08	43.12	57.00
11	65.56	59.53	61.52	58.90	64.50	60.40	68.78	82.05
12	50.03	45.30	50.74	57.83	48.16	62.14	76.92	67.47
13	34.51	31.27	36.37	30.45	38.00	30.70	37.88	58.75
14	48.64	50.17	48.68	49.91	56.78	48.06	62.29	71.28
15	28.31	37.76	31.75	24.97	46.45	38.78	45.68	26.09
16	62.36	39.58	50.28	52.89	50.73	69.67	61.66	57.22
17	51.25	46.43	51.92	47.23	53.35	50.63	48.51	54.24
18	27.44	29.56	30.25	32.24	17.77	22.05	25.05	40.49
19	57.27	57.09	59.09	57.48	86.95	57.48	103.21	57.76
20	28.94	37.69	30.21	30.82	25.38	32.45	31.26	32.34
MEDIAN	38.76	45.86	45.54	47.09	47.96	47.57	47.09	55.74
1.6 Hz	0.315	0.40	0.50	0.63	0.80	1.00	1.25	1.60
1	58.81	53.98	59.69	63.78	66.06	62.37	64.72	63.19
2	23.79	16.50	23.04	18.20	23.99	22.76	20.20	26.19
3	40.31	38.45	37.03	26.15	45.67	36.10	39.01	55.36
4	35.75	46.44	46.37	41.94	48.27	50.07	75.61	79.35
5	35.78	35.13	46.48	54.44	39.49	39.70	43.48	52.29
6	41.90	42.93	50.82	33.69	45.49	48.17	78.19	63.75
7	20.32	31.67	38.88	37.58	30.28	43.74	45.61	41.82
8	46.26	45.41	53.42	48.24	57.74	54.84	48.21	58.83
9	55.50	51.76	61.87	62.38	70.38	88.18	74.01	132.72
10	33.21	35.91	25.54	32.15	40.85	45.48	66.99	65.40
11	59.99	61.96	65.60	64.97	67.23	69.04	46.41	65.90
12	35.05	47.77	52.62	41.08	48.20	57.89	41.33	230.82
13	35.56	27.31	44.40	46.05	42.39	41.63	40.06	41.73

14	43.86	45.39	38.15	35.83	62.91	60.47	61.08	68.98
15	48.17	42.94	40.75	57.39	37.01	45.95	52.40	36.61
16	41.25	55.54	46.27	42.46	65.53	45.02	24.79	49.17
17	45.52	48.62	44.20	55.96	53.96	59.38	56.55	40.33
18	29.23	26.05	21.65	39.74	29.04	22.71	35.41	63.33
19	59.33	43.15	51.24	89.76	54.78	94.74	120.16	86.28
20	25.10	34.42	29.58	43.99	29.67	22.93	22.71	29.44
MEDIAN	40.78	43.04	45.33	43.22	46.93	47.06	47.31	61.01
2.0 Hz	0.40	0.50	0.63	0.80	1.00	1.25	1.60	2.00
1	61.46	55.95	59.34	66.40	67.29	72.22	76.53	77.94
2	22.46	27.07	17.93	16.64	28.31	30.65	41.83	31.54
3	47.53	30.24	39.12	38.33	37.32	45.58	48.30	32.05
4	46.99	35.97	50.72	37.63	57.89	36.61	60.90	63.91
5	48.29	43.61	38.10	38.98	46.03	29.55	44.56	49.85
6	47.04	48.53	51.80	53.86	55.22	52.04	44.76	71.32
7	24.52	32.41	31.23	34.97	52.35	22.36	42.82	39.01
8	50.67	45.93	60.53	51.76	54.82	56.69	61.13	59.11
9	51.25	59.55	38.53	67.76	77.55	77.11	126.10	132.29
10	36.72	32.78	41.28	34.48	45.54	47.90	47.30	61.47
11	66.68	62.03	62.64	74.16	73.48	55.98	80.20	54.63
12	42.41	47.35	46.77	54.85	55.96	53.35	56.45	48.56
13	39.17	40.20	34.98	37.92	50.74	37.05	49.31	54.23
14	42.86	47.37	45.87	46.00	56.03	52.67	63.46	61.94
15	47.12	26.44	22.44	45.37	36.39	40.21	55.08	54.66
16	49.20	35.42	53.16	45.76	42.76	59.07	44.49	56.05
17	46.54	51.45	57.07	52.74	56.39	43.16	66.65	72.90
18	28.79	20.68	40.84	33.14	17.08	36.98	40.51	40.44
19	66.15	58.53	48.98	47.96	61.97	70.61	87.82	100.84
20	31.76	31.83	33.70	30.26	31.67	22.46	25.24	26.72
MEDIAN	47.01	41.90	43.58	45.56	53.59	46.74	52.19	55.35

Table C.4: Reported probability of losing balance as a function of frequency at each magnitude of lateral acceleration.

0.125 ms^{-2} r.m.s.	Frequency (Hz)	
Subject no	0.5	0.63
1	30	5
2	75	50
3	20	30
4	50	25
5	5	0
6	15	5

7	18	0
8	5	1
9	15	5
10	2	0
11	15	0
12	10	0
13	0	0
14	10	2
15	80	5
16	20	10
17	10	0
18	30	20
19	0	5
20	20	60
MEDIAN	15	5

0.16 ms^{-2} r.m.s.	0.5	0.63	0.8
1	60	10	20
2	10	30	5
3	80	80	20
4	25	25	10
5	15	0	2
6	5	17	10
7	95	15	5
8	35	1	8
9	2	5	1
10	5	0	0
11	5	10	5
12	40	10	50
13	10	0	0
14	20	10	10
15	60	15	5
16	60	40	8
17	50	0	10
18	75	20	20
19	0	0	0
20	65	80	20
MEDIAN	30	10	8

0.2 ms^{-2} r.m.s.	0.5	0.63	0.8	1
1	50	10	30	30
2	30	30	20	5
3	60	25	20	20
4	75	100	25	0
5	15	5	10	5
6	35	15	0	5
7	10	25	4	30
8	37	6	38	1
9	7	2	5	2
10	40	15	5	25
11	30	15	0	10

12	10	0	30	0		
13	5	5	15	0		
14	35	15	15	5		
15	65	50	35	10		
16	45	10	5	5		
17	50	15	0	15		
18	40	40	60	40		
19	30	40	5	10		
20	70	35	50	10		
MEDIAN	36	15	15	7.5		
0.25 ms^{-2} r.m.s.	0.5	0.63	0.8	1	1.25	
1	40	70	10	15	30	
2	90	80	12	35	20	
3	50	50	40	25	25	
4	50	100	25	0	0	
5	15	10	5	0	0	
6	20	35	35	50	5	
7	80	45	8	2	8	
8	69	55	3	3	15	
9	10	2	4	5	2	
10	40	20	0	0	0	
11	35	10	40	30	15	
12	30	0	20	40	10	
13	15	30	5	10	0	
14	70	40	5	20	15	
15	25	30	20	60	40	
16	35	50	0	7	0	
17	80	50	15	25	25	
18	80	30	25	50	50	
19	70	10	30	0	20	
20	55	65	30	20	10	
MEDIAN	45	37.5	13.5	17.5	12.5	
0.315 ms^{-2} r.m.s.	0.5	0.63	0.8	1	1.25	1.6
1	50	80	40	25	40	50
2	60	80	65	45	10	5
3	80	60	25	40	25	15
4	100	75	25	25	0	10
5	10	25	5	0	5	5
6	35	40	10	37	10	0
7	95	90	80	15	3	1
8	70	92	12	16	27	5
9	33	4	2	15	1	3
10	5	15	5	5	3	5
11	45	20	10	20	30	5
12	40	20	30	60	40	10
13	75	5	5	0	0	0
14	30	60	40	60	40	10
15	35	85	30	90	10	20

16	25	60	20	10	0	3	
17	90	75	10	80	25	15	
18	100	80	40	75	40	20	
19	0	70	40	0	10	0	
20	60	50	20	35	30	25	
MEDIAN	47.5	60	22.5	25	10	5	
0.4 ms^{-2} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	90	40	40	70	50	70	30
2	95	60	45	15	10	10	5
3	100	50	40	50	40	25	15
4	100	100	50	75	50	10	0
5	35	30	15	15	10	20	0
6	100	20	45	15	40	5	5
7	100	85	55	45	30	35	3
8	95	85	92	12	75	17	3
9	40	20	3	5	3	3	4
10	90	60	10	10	5	2	0
11	40	30	40	30	10	25	5
12	70	20	60	30	10	10	20
13	80	50	25	35	5	5	5
14	50	60	50	20	30	5	15
15	100	50	50	40	30	70	50
16	100	60	85	3	40	0	0
17	95	95	65	15	40	65	15
18	100	75	70	40	25	30	30
19	90	40	80	30	50	20	10
20	100	95	65	25	45	35	10
MEDIAN	95	55	50	27.5	30	18.5	5
0.5 ms^{-2} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	100	90	80	50	60	10	20
2	100	85	80	10	20	15	10
3	100	100	80	60	50	25	40
4	100	100	100	100	50	0	0
5	25	25	20	20	20	10	10
6	100	30	55	20	65	15	0
7	100	95	75	75	17	30	10
8	40	82	45	38	18	22	28
9	50	10	40	10	10	35	2
10	100	30	90	5	15	10	5
11	80	40	40	20	25	20	5
12	100	70	40	20	20	20	20
13	80	45	20	15	0	15	15
14	50	75	25	20	10	15	25
15	70	75	75	30	50	45	40
16	100	80	90	25	50	35	5
17	100	60	50	25	15	15	25
18	100	60	20	90	75	75	20
19	10	80	50	60	20	10	0
20	100	100	45	45	20	20	30

MEDIAN	100	75	50	25	20	17.5	12.5
0.63 ms^{-2} r.m.s.	0.63	0.8	1	1.25	1.6	2	
1	90	70	70	90	50	50	
2	80	85	20	45	60	15	
3	100	40	25	50	75	40	
4	75	75	75	50	50	25	
5	15	10	20	15	15	5	
6	70	20	35	10	75	20	
7	100	95	75	35	50	40	
8	70	58	62	49	20	48	
9	40	15	3	6	75	3	
10	75	10	30	20	5	10	
11	75	60	50	20	25	10	
12	80	100	70	40	20	10	
13	90	70	20	20	20	0	
14	50	5	40	30	10	10	
15	90	80	50	60	50	40	
16	100	95	35	10	15	0	
17	100	100	50	15	50	50	
18	100	50	100	40	50	30	
19	40	50	50	100	25	50	
20	75	50	75	35	50	75	
MEDIAN	77.5	59	50	35	50	22.5	
0.8 ms^{-2} r.m.s.	0.8	1	1.25	1.6	2		
1	90	90	80	60	30		
2	95	50	20	30	10		
3	80	80	80	40	50		
4	100	100	75	100	0		
5	35	50	10	20	10		
6	90	70	75	55	15		
7	100	98	60	25	5		
8	97	82	37	89	67		
9	80	15	50	10	40		
10	40	40	30	25	10		
11	55	50	35	35	25		
12	90	80	60	80	40		
13	20	65	50	30	5		
14	80	60	30	60	40		
15	100	60	80	60	10		
16	40	80	35	0	20		
17	95	95	65	90	20		
18	60	60	90	75	30		
19	100	80	60	40	10		
20	100	100	50	90	50		
MEDIAN	90	75	55	47.5	20		
1.0 ms^{-2} r.m.s.	1	1.25	1.6	2			
1	100	100	80	60			

2	70	80	20	20
3	90	100	60	50
4	100	75	75	10
5	35	20	25	5
6	85	65	65	35
7	98	80	95	35
8	92	88	69	47
9	60	15	10	55
10	75	70	60	0
11	50	45	20	35
12	100	60	60	25
13	90	85	35	25
14	50	60	60	40
15	85	100	80	25
16	95	90	25	5
17	75	75	75	40
18	100	80	60	50
19	90	60	75	0
20	90	65	95	50
MEDIAN	90	75	60	35

1.25 ms^{-2} r.m.s.	1.25	1.6	2
1	90	100	70
2	90	65	20
3	80	100	50
4	100	75	50
5	30	20	15
6	75	100	45
7	99	90	85
8	95	30	72
9	40	75	30
10	65	50	20
11	50	50	25
12	100	70	60
13	55	0	30
14	75	35	30
15	95	75	70
16	85	85	40
17	100	75	65
18	100	100	50
19	95	30	60
20	95	75	75
MEDIAN	90	75	50

1.6 ms^{-2} r.m.s.	1.6	2
1	100	50
2	90	30
3	50	80
4	100	50
5	15	7
6	55	60

7	95	65
8	50	67
9	75	70
10	70	15
11	60	20
12	100	70
13	60	40
14	75	30
15	80	45
16	85	45
17	100	75
18	100	60
19	25	70
20	80	90
MEDIAN	77.5	55

Table C.5: Peak to peak lateral COP position as a function of frequency at each magnitude of lateral acceleration.

0.125 ms^{-2} r.m.s.	Frequency (Hz)		
Subject no	0.5	0.63	
1	25.33	26.75	
2	31.88	32.97	
3	20.79	22.96	
4	29.21	26.88	
5	22.30	22.83	
6	27.86	25.10	
7	21.12	19.83	
8	24.54	27.30	
9	24.42	26.27	
10	24.45	20.44	
11	35.69	30.23	
12	26.78	25.94	
13	20.99	23.38	
14	28.12	28.26	
15	28.80	24.06	
16	28.98	29.61	
17	30.63	24.95	
18	24.90	23.98	
19	26.52	29.77	
20	26.36	24.68	
MEDIAN	26.44	25.52	
0.16 ms^{-2} r.m.s.	0.5	0.63	0.8
1	31.63	30.17	27.40
2	19.90	29.76	15.00

3	23.69	24.40	19.13
4	32.73	28.89	25.01
5	30.98	21.25	24.21
6	25.69	22.92	27.06
7	28.37	23.51	18.43
8	23.64	21.81	23.49
9	27.50	29.91	21.57
10	26.18	21.46	21.91
11	31.11	32.32	32.60
12	30.33	33.70	29.17
13	27.30	26.12	13.92
14	34.19	34.39	25.41
15	31.84	30.06	23.10
16	33.74	27.44	27.63
17	36.57	28.11	28.28
18	29.62	19.93	19.76
19	25.04	27.10	24.60
20	32.49	32.55	20.89
MEDIAN	29.98	27.78	23.85

0.2 ms^{-2} r.m.s.	0.5	0.63	0.8	1
1	31.73	23.44	24.49	24.74
2	20.32	22.93	17.55	13.76
3	22.83	20.30	18.62	17.32
4	30.78	29.74	24.12	26.29
5	29.77	18.01	26.55	15.80
6	31.37	29.29	18.77	21.27
7	24.27	19.00	16.97	20.23
8	23.11	23.45	24.03	23.57
9	24.95	22.95	23.55	21.92
10	28.79	27.36	25.40	30.93
11	35.30	30.95	31.82	36.65
12	26.33	29.85	24.35	22.82
13	24.61	15.11	21.66	21.30
14	27.85	23.76	32.18	25.75
15	29.42	28.26	21.52	25.03
16	32.03	30.19	33.88	28.51
17	34.90	26.09	24.18	26.51
18	27.14	23.75	25.64	20.10
19	30.65	33.22	26.55	25.45
20	28.55	21.77	19.83	22.29
MEDIAN	28.67	23.76	24.15	23.20

0.25 ms^{-2} r.m.s.	0.5	0.63	0.8	1	1.25
1	29.62	29.68	27.47	28.23	24.24
2	35.62	32.58	21.40	21.03	18.33
3	24.48	25.34	23.47	22.73	16.45
4	28.61	28.69	24.42	20.15	18.08
5	22.53	26.91	21.32	19.91	20.28
6	24.14	27.72	30.10	19.97	20.00
7	23.12	26.51	23.65	21.47	18.27

8	25.60	24.54	23.51	22.54	22.52		
9	25.43	23.12	26.95	20.78	16.71		
10	33.74	29.23	20.87	19.42	22.96		
11	33.54	31.85	34.32	36.57	34.71		
12	29.72	21.89	30.50	32.28	24.65		
13	29.93	25.45	22.40	20.38	15.66		
14	36.90	33.31	29.62	28.33	24.44		
15	23.66	32.65	20.95	27.21	19.27		
16	35.23	39.11	29.34	27.71	34.58		
17	33.02	28.01	31.21	27.33	27.62		
18	37.47	22.50	25.38	27.42	19.97		
19	34.44	23.84	25.71	33.84	24.35		
20	28.52	20.15	28.34	18.67	21.75		
MEDIAN	29.67	27.32	25.55	22.63	21.01		
0.315 ms^{-2} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	
1	32.90	28.74	28.04	28.29	25.37	23.26	
2	33.08	27.83	20.01	18.15	15.46	12.80	
3	26.65	21.83	18.91	21.06	19.53	17.87	
4	28.74	30.68	23.70	23.50	21.66	16.14	
5	37.06	23.73	24.08	24.45	18.71	16.06	
6	30.38	27.22	26.90	27.85	23.73	19.11	
7	29.80	22.48	26.21	23.41	17.27	12.55	
8	32.43	25.70	25.61	24.84	25.16	21.25	
9	28.26	22.25	26.18	25.25	21.15	24.15	
10	29.43	25.43	22.69	22.56	24.89	16.21	
11	41.54	33.77	31.28	35.96	32.83	26.67	
12	29.28	35.07	34.52	28.48	23.77	20.01	
13	33.31	25.13	24.38	19.04	16.67	18.97	
14	35.57	34.12	32.21	33.31	29.64	24.43	
15	27.93	19.27	31.26	32.25	20.61	25.26	
16	26.77	27.71	29.87	29.95	25.33	23.22	
17	38.13	33.78	30.31	30.63	27.58	20.24	
18	45.03	26.13	34.42	22.31	20.60	16.87	
19	25.99	37.00	28.77	20.42	23.99	25.25	
20	24.22	22.57	26.27	14.01	19.81	14.41	
MEDIAN	30.09	26.67	26.59	24.65	22.70	19.56	
0.4 ms^{-2} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	34.21	28.83	23.37	25.83	24.00	23.56	24.91
2	34.28	20.40	22.49	19.02	17.43	9.15	13.05
3	27.61	26.64	21.54	24.62	18.95	15.90	17.04
4	35.69	26.80	20.22	29.41	20.89	20.70	17.51
5	33.45	33.13	32.50	23.97	20.61	19.88	19.04
6	31.30	25.15	25.43	23.66	21.03	19.84	20.67
7	31.36	24.58	22.33	21.59	19.14	19.55	12.99
8	30.95	29.84	21.55	23.37	20.01	23.17	22.29
9	26.75	24.15	17.45	26.74	23.86	23.16	24.41
10	33.42	27.50	26.98	23.47	20.12	17.98	19.71
11	35.37	35.39	39.10	34.65	32.42	34.19	29.23

12	33.83	29.07	32.42	25.60	23.99	24.45	22.87
13	39.25	20.68	23.77	29.34	21.77	13.78	17.16
14	32.93	33.59	28.99	28.65	29.81	23.86	19.97
15	26.30	21.80	22.06	20.48	22.01	20.41	26.57
16	35.80	35.90	23.72	30.58	24.23	23.05	23.53
17	35.68	19.08	30.22	27.92	24.74	25.42	23.66
18	48.17	24.08	27.23	23.43	26.15	14.36	16.02
19	34.89	32.46	30.94	27.33	25.63	21.65	24.09
20	27.62	23.44	28.46	27.20	20.84	19.73	16.29
MEDIAN	33.64	26.72	24.60	25.71	21.89	20.55	20.32
0.5 ms^{-2} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	35.63	29.56	27.99	26.66	27.06	26.88	22.76
2	35.10	23.23	25.92	15.95	23.04	14.32	17.38
3	24.39	19.39	26.41	20.22	20.06	17.13	15.45
4	29.99	22.84	31.05	30.70	24.86	20.53	18.30
5	44.89	31.21	20.76	24.95	23.69	21.25	19.40
6	36.29	25.63	24.84	22.84	27.58	22.17	23.36
7	46.15	25.73	23.85	22.48	21.87	20.04	16.86
8	31.96	25.08	24.75	22.26	20.00	24.10	21.48
9	29.34	26.07	25.09	26.04	24.59	22.55	25.14
10	50.91	27.95	32.67	27.05	22.69	17.45	17.65
11	36.40	31.09	28.98	39.32	30.87	28.77	31.58
12	31.08	36.12	27.56	29.47	23.15	25.22	25.33
13	49.73	29.29	29.51	25.13	13.72	23.26	20.08
14	40.18	26.36	28.11	31.15	26.36	18.97	22.53
15	32.59	26.73	22.88	20.53	15.04	23.17	16.45
16	32.55	31.78	39.32	30.56	28.68	20.76	17.52
17	42.16	34.49	34.29	28.56	28.26	20.41	23.18
18	44.18	31.46	28.43	28.58	19.97	14.74	13.01
19	31.27	35.20	21.80	24.44	25.35	26.81	24.19
20	22.18	16.79	23.38	25.42	18.75	17.27	15.49
MEDIAN	35.37	27.34	26.98	25.73	23.42	21.01	19.74
0.63 ms^{-2} r.m.s.	0.63	0.8	1	1.25	1.6	2	
1	38.72	37.44	28.42	29.62	26.33	22.43	
2	35.87	28.08	20.05	25.07	13.29	11.06	
3	28.52	28.12	19.80	18.89	17.99	21.15	
4	26.14	37.45	31.69	24.32	18.98	19.82	
5	33.64	20.16	24.58	27.12	25.04	16.83	
6	35.77	30.15	27.09	23.27	26.84	21.27	
7	33.52	30.90	25.26	21.49	21.55	15.55	
8	28.67	26.36	24.54	22.40	20.28	22.87	
9	32.92	30.38	21.36	24.00	27.88	17.77	
10	40.88	30.39	26.56	26.86	20.67	19.08	
11	35.52	26.50	33.01	32.26	32.36	28.55	
12	31.22	27.01	34.99	23.72	21.95	23.09	
13	37.68	21.36	19.94	21.22	22.95	19.43	
14	32.53	33.50	30.62	27.95	18.77	22.99	
15	38.35	26.99	26.62	25.70	24.89	14.18	
16	27.40	29.61	36.17	24.89	20.52	20.84	

17	28.18	27.79	32.57	25.74	24.63	28.40
18	46.22	29.22	26.40	16.22	21.30	22.52
19	42.75	30.09	34.98	36.08	30.05	17.89
20	26.21	32.03	21.08	19.16	22.73	19.89
MEDIAN	33.58	29.42	26.59	24.61	22.34	20.36
0.8 ms^{-2} r.m.s.	0.8	1	1.25	1.6	2	
1	30.85	35.18	27.60	25.56	27.79	
2	20.83	23.58	14.71	13.58	11.93	
3	26.77	19.96	21.93	20.89	16.65	
4	26.15	28.57	23.01	24.39	16.41	
5	28.94	29.51	23.34	18.94	17.76	
6	23.73	30.56	24.60	22.72	18.49	
7	24.23	26.56	22.81	15.44	18.17	
8	23.72	25.73	21.14	22.92	24.51	
9	34.57	29.65	25.24	24.13	26.90	
10	29.64	23.21	24.58	22.44	17.43	
11	38.51	33.64	30.50	33.51	29.24	
12	35.34	31.89	29.81	22.63	24.36	
13	29.19	21.99	22.32	23.38	16.82	
14	32.93	32.92	29.69	27.04	22.93	
15	28.90	25.36	19.69	23.36	22.03	
16	34.44	30.67	34.17	26.02	20.77	
17	36.60	32.73	29.20	27.57	23.80	
18	37.29	29.86	16.12	16.81	20.36	
19	32.68	33.35	26.32	25.41	23.31	
20	23.05	18.84	17.28	15.21	13.30	
MEDIAN	29.42	29.58	23.96	23.14	20.57	
1.0 ms^{-2} r.m.s.	1	1.25	1.6	2		
1	36.44	33.83	25.59	26.85		
2	26.38	23.84	12.10	16.96		
3	24.76	28.62	19.63	15.27		
4	29.12	23.72	24.67	23.80		
5	24.16	19.29	26.66	18.29		
6	33.86	27.87	23.64	22.23		
7	26.72	20.46	23.03	20.92		
8	26.21	18.69	22.77	21.14		
9	31.34	26.69	30.57	25.18		
10	25.78	23.09	21.99	25.02		
11	39.63	37.05	31.75	25.71		
12	33.14	29.64	25.80	21.40		
13	32.60	20.24	23.34	22.37		
14	32.71	28.61	26.37	23.97		
15	29.17	26.55	26.62	16.88		
16	29.37	30.73	22.47	19.68		
17	36.49	29.41	25.38	23.51		
18	23.81	18.62	22.97	15.98		
19	38.34	31.73	32.08	24.69		
20	22.49	21.01	16.43	17.44		
MEDIAN	29.27	26.62	24.15	21.82		

1.25 ms^{-2} r.m.s.	1.25	1.6	2
1	23.86	29.66	27.82
2	20.17	13.61	16.12
3	25.34	24.89	18.88
4	29.27	28.93	21.85
5	27.94	30.21	23.85
6	24.31	34.17	20.70
7	22.51	24.68	11.10
8	24.87	23.62	24.43
9	27.38	26.71	26.04
10	31.75	27.23	24.46
11	36.83	24.59	30.40
12	31.64	21.49	24.19
13	28.28	20.15	16.71
14	31.22	33.61	26.49
15	21.57	26.38	20.93
16	29.00	20.31	26.91
17	32.49	24.22	20.52
18	33.53	17.09	17.32
19	26.89	38.22	28.24
20	26.12	14.53	12.33
MEDIAN	27.66	24.78	22.85

1.6 ms^{-2} r.m.s.	1.6	2
1	32.73	31.80
2	19.17	23.37
3	22.72	19.27
4	37.65	26.59
5	34.03	19.45
6	27.72	23.11
7	22.87	21.65
8	23.24	22.16
9	35.59	37.09
10	32.20	23.78
11	28.74	32.82
12	65.54	25.73
13	24.79	21.84
14	32.79	26.35
15	22.48	21.51
16	24.85	24.34
17	23.35	27.17
18	29.65	19.71
19	36.93	30.79
20	16.90	13.95
MEDIAN	28.23	23.58

Table C.6: Lateral r.ms. COP velocity as a function of frequency at each magnitude of lateral acceleration.

0.125 ms^{-2} r.m.s.	Frequency (Hz)		
Subject no	0.50	0.63	
1	51.73	55.44	
2	25.66	26.19	
3	37.08	37.04	
4	43.02	42.09	
5	33.41	44.91	
6	43.22	42.65	
7	27.36	30.63	
8	44.82	46.14	
9	42.85	56.01	
10	42.25	31.18	
11	60.24	59.69	
12	50.82	49.10	
13	35.43	35.84	
14	39.80	44.91	
15	33.75	28.78	
16	41.84	46.44	
17	46.81	46.21	
18	27.30	32.66	
19	46.96	54.41	
20	33.43	21.73	
MEDIAN	42.04	43.78	
0.16 ms^{-2} r.m.s.	0.50	0.63	0.80
1	61.15	62.10	56.08
2	22.39	35.47	17.76
3	37.19	41.32	37.89
4	56.13	46.19	48.37
5	44.48	40.38	36.28
6	44.89	42.16	38.31
7	36.95	43.27	25.81
8	48.00	51.42	46.58
9	53.87	59.29	45.15
10	38.27	41.72	41.46
11	49.83	65.20	56.60
12	49.37	54.47	55.74
13	37.91	38.03	27.91
14	49.35	50.77	41.75
15	48.64	50.52	33.80
16	48.66	42.02	53.99
17	48.97	56.79	57.71
18	30.37	29.22	29.58
19	48.94	48.10	47.88
20	29.96	38.16	34.92

MEDIAN	48.32	44.73	41.61			
0.2 ms^{-2} r.m.s.	0.50	0.63	0.80	1.00		
1	60.28	47.25	53.66	53.08		
2	21.45	21.95	27.17	17.92		
3	40.74	40.06	41.85	37.88		
4	52.59	43.12	46.10	50.95		
5	37.91	29.08	42.31	29.24		
6	45.51	36.57	35.68	35.94		
7	30.77	32.12	30.04	40.65		
8	51.80	44.99	50.58	45.07		
9	52.71	37.61	42.80	42.13		
10	37.80	40.99	39.63	38.24		
11	65.30	60.75	55.44	61.04		
12	52.72	55.71	44.42	48.09		
13	36.32	24.01	32.58	35.34		
14	35.78	25.30	40.22	40.34		
15	42.49	38.42	31.57	39.72		
16	55.89	44.29	69.58	51.48		
17	51.18	48.29	49.14	48.87		
18	33.17	22.15	43.06	32.31		
19	52.04	65.86	53.06	57.02		
20	32.06	29.48	24.50	35.78		
MEDIAN	44.00	39.24	42.55	40.49		
0.25 ms^{-2} r.m.s.	0.50	0.63	0.80	1.00	1.25	
1	56.81	56.28	58.61	66.84	53.32	
2	30.45	30.83	33.17	26.09	27.92	
3	37.82	42.71	42.74	42.48	31.53	
4	43.22	47.17	48.55	47.75	40.79	
5	34.56	30.59	31.27	37.76	40.31	
6	44.25	48.63	56.12	39.90	37.21	
7	34.20	40.12	36.44	42.90	35.13	
8	49.34	51.97	51.22	47.97	44.70	
9	56.13	47.26	57.61	48.72	35.73	
10	43.92	42.13	33.84	36.22	36.52	
11	63.69	61.09	64.44	83.03	65.56	
12	50.50	37.29	49.62	72.32	50.03	
13	35.39	39.39	44.75	36.00	34.51	
14	39.28	51.92	55.21	54.52	48.64	
15	31.40	43.81	27.81	45.14	28.31	
16	59.91	56.47	52.03	40.73	62.36	
17	44.98	49.62	63.25	45.46	51.25	
18	41.03	26.16	35.53	28.26	27.44	
19	61.73	50.92	57.73	78.31	57.27	
20	37.81	25.58	34.51	30.84	28.94	
MEDIAN	43.57	45.49	49.08	44.02	38.76	
0.315 ms^{-2} r.m.s.	0.50	0.63	0.80	1.00	1.25	1.60
1	65.60	57.72	64.68	67.91	59.73	58.81

2	43.19	32.65	30.24	25.34	22.19	23.79	
3	34.08	32.65	31.05	47.29	37.89	40.31	
4	48.38	48.89	54.85	54.66	48.82	35.75	
5	55.95	30.38	53.50	53.52	26.86	35.78	
6	49.03	41.99	49.80	52.09	47.76	41.90	
7	54.07	33.08	52.20	39.65	28.40	20.32	
8	51.25	53.74	43.76	53.83	58.08	46.26	
9	63.47	46.10	59.01	67.16	48.29	55.50	
10	46.35	41.22	31.45	39.10	56.93	33.21	
11	60.16	61.43	68.58	64.17	59.53	59.99	
12	56.94	60.41	66.60	59.06	45.30	35.05	
13	47.26	40.74	45.59	38.77	31.27	35.56	
14	42.80	51.28	46.14	67.64	50.17	43.86	
15	43.02	27.52	47.52	63.66	37.76	48.17	
16	43.87	45.18	45.57	62.95	39.58	41.25	
17	55.06	56.20	60.58	59.63	46.43	45.52	
18	50.48	25.17	53.54	30.25	29.56	29.23	
19	49.45	49.85	57.21	41.45	57.09	59.33	
20	27.67	26.47	47.66	26.37	37.69	25.10	
MEDIAN	49.24	43.58	51.00	53.67	45.86	40.78	
0.4 ms^{-2} r.m.s.	0.50	0.63	0.80	1.00	1.25	1.60	2.00
1	61.36	49.44	51.36	59.29	56.48	53.98	61.46
2	41.70	31.36	37.20	20.28	26.51	16.50	22.46
3	37.12	47.17	35.73	44.58	40.55	38.45	47.53
4	51.03	46.03	38.73	50.06	42.28	46.44	46.99
5	49.93	39.57	48.82	42.24	44.19	35.13	48.29
6	33.33	42.16	41.07	52.83	46.89	42.93	47.04
7	48.33	39.97	39.35	40.16	29.62	31.67	24.52
8	58.45	45.45	30.21	59.62	51.41	45.41	50.67
9	60.22	50.85	32.99	61.59	48.81	51.76	51.25
10	44.82	42.84	41.45	38.36	37.09	35.91	36.72
11	56.33	66.44	70.93	60.27	61.52	61.96	66.68
12	69.78	52.35	64.42	57.15	50.74	47.77	42.41
13	45.88	28.63	38.10	40.21	36.37	27.31	39.17
14	45.34	44.67	43.23	47.57	48.68	45.39	42.86
15	32.92	25.99	26.57	32.41	31.75	42.94	47.12
16	59.26	69.49	40.90	59.71	50.28	55.54	49.20
17	48.59	25.26	62.21	59.64	51.92	48.62	46.54
18	44.66	32.61	53.53	31.81	30.25	26.05	28.79
19	54.46	58.66	54.57	45.41	59.09	43.15	66.15
20	31.62	24.78	33.90	30.29	30.21	34.42	31.76
MEDIAN	48.46	43.75	40.99	46.49	45.54	43.04	47.01
0.5 ms^{-2} r.m.s.	0.50	0.63	0.80	1.00	1.25	1.60	2.00
1	68.75	52.15	54.74	61.02	52.95	59.69	55.95
2	37.16	31.44	25.29	21.29	34.22	23.04	27.07
3	33.62	29.51	40.62	44.65	44.93	37.03	30.24
4	52.35	39.20	52.40	42.74	52.02	46.37	35.97
5	71.41	39.90	22.49	42.65	36.76	46.48	43.61
6	48.67	51.56	33.62	43.35	46.95	50.82	48.53

7	60.42	43.92	47.39	42.96	32.14	38.88	32.41
8	50.99	53.21	60.24	63.57	48.00	53.42	45.93
9	72.87	62.85	51.18	57.10	62.34	61.87	59.55
10	72.46	48.08	42.29	46.23	41.10	25.54	32.78
11	54.35	56.21	35.44	61.64	58.90	65.60	62.03
12	46.32	59.84	60.47	60.64	57.83	52.62	47.35
13	83.38	44.19	50.78	44.83	30.45	44.40	40.20
14	52.32	34.16	50.07	52.36	49.91	38.15	47.37
15	42.71	35.67	41.88	34.29	24.97	40.75	26.44
16	58.92	41.73	58.96	67.90	52.89	46.27	35.42
17	52.75	65.09	58.73	54.26	47.23	44.20	51.45
18	44.97	42.18	50.05	40.74	32.24	21.65	20.68
19	71.70	71.43	28.65	40.18	57.48	51.24	58.53
20	28.76	23.26	39.79	34.05	30.82	29.58	31.83
MEDIAN	52.55	44.06	48.72	44.74	47.09	45.33	41.90
0.63 ms^{-2} r.m.s.	0.63	0.80	1.00	1.25	1.60	2.00	
1	69.36	82.46	51.05	72.53	63.78	59.34	
2	56.33	49.95	30.10	26.04	18.20	17.93	
3	46.75	53.40	37.83	33.75	26.15	39.12	
4	43.83	74.62	58.61	47.75	41.94	50.72	
5	47.31	34.40	49.00	43.57	54.44	38.10	
6	60.15	68.92	51.24	48.86	33.69	51.80	
7	61.01	64.55	53.90	37.99	37.58	31.23	
8	53.62	67.51	60.48	58.29	48.24	60.53	
9	84.67	64.49	46.59	60.26	62.38	38.53	
10	69.18	56.50	39.02	43.58	32.15	41.28	
11	54.63	32.77	59.49	64.50	64.97	62.64	
12	72.09	45.90	74.36	48.16	41.08	46.77	
13	56.85	21.47	33.23	38.00	46.05	34.98	
14	56.79	68.71	71.63	56.78	35.83	45.87	
15	78.81	55.58	36.32	46.45	57.39	22.44	
16	48.74	54.74	58.53	50.73	42.46	53.16	
17	38.51	44.19	67.39	53.35	55.96	57.07	
18	60.74	56.62	24.67	17.77	39.74	40.84	
19	91.77	62.12	85.11	86.95	89.76	48.98	
20	32.92	61.25	26.48	25.38	43.99	33.70	
MEDIAN	56.82	56.56	51.14	47.96	43.22	43.58	
0.8 ms^{-2} r.m.s.	0.80	1.00	1.25	1.60	2.00		
1	74.24	88.97	57.90	66.06	66.40		
2	23.01	31.88	24.34	23.99	16.64		
3	62.82	33.99	39.44	45.67	38.33		
4	48.05	40.90	46.07	48.27	37.63		
5	53.71	26.59	44.23	39.49	38.98		
6	38.48	54.66	59.14	45.49	53.86		
7	27.85	49.71	34.23	30.28	34.97		
8	59.09	61.23	56.21	57.74	51.76		
9	59.54	79.48	65.36	70.38	67.76		
10	58.39	38.56	47.08	40.85	34.48		
11	78.40	75.73	60.40	67.23	74.16		

12	78.95	86.81	62.14	48.20	54.85
13	61.11	34.31	30.70	42.39	37.92
14	64.43	62.94	48.06	62.91	46.00
15	28.45	45.07	38.78	37.01	45.37
16	73.39	61.86	69.67	65.53	45.76
17	79.87	74.91	50.63	53.96	52.74
18	60.12	69.01	22.05	29.04	33.14
19	69.60	92.05	57.48	54.78	47.96
20	38.19	23.78	32.45	29.67	30.26
MEDIAN	59.83	57.94	47.57	46.93	45.56
1.0 ms^{-2} r.m.s.	1.00	1.25	1.60	2.00	
1	83.56	81.95	62.37	67.29	
2	36.49	25.00	22.76	28.31	
3	63.67	38.66	36.10	37.32	
4	57.83	45.42	50.07	57.89	
5	32.44	37.52	39.70	46.03	
6	48.57	63.68	48.17	55.22	
7	56.47	31.55	43.74	52.35	
8	73.26	49.83	54.84	54.82	
9	87.80	83.35	88.18	77.55	
10	39.73	43.12	45.48	45.54	
11	77.69	68.78	69.04	73.48	
12	66.61	76.92	57.89	55.96	
13	60.45	37.88	41.63	50.74	
14	88.48	62.29	60.47	56.03	
15	72.07	45.68	45.95	36.39	
16	51.44	61.66	45.02	42.76	
17	89.18	48.51	59.38	56.39	
18	21.18	25.05	22.71	17.08	
19	120.44	103.21	94.74	61.97	
20	24.78	31.26	22.93	31.67	
MEDIAN	62.06	47.09	47.06	53.59	
1.25 ms^{-2} r.m.s.	1.25	1.60	2.00		
1	57.63	64.72	72.22		
2	28.53	20.20	30.65		
3	52.65	39.01	45.58		
4	46.21	75.61	36.61		
5	44.73	43.48	29.55		
6	54.47	78.19	52.04		
7	45.34	45.61	22.36		
8	68.93	48.21	56.69		
9	72.26	74.01	77.11		
10	57.00	66.99	47.90		
11	82.05	46.41	55.98		
12	67.47	41.33	53.35		
13	58.75	40.06	37.05		
14	71.28	61.08	52.67		
15	26.09	52.40	40.21		
16	57.22	24.79	59.07		

17	54.24	56.55	43.16
18	40.49	35.41	36.98
19	57.76	120.16	70.61
20	32.34	22.71	22.46
MEDIAN	55.74	47.31	46.74

1.6 ms^{-2} r.m.s.	1.60	2.00
1	63.19	76.53
2	26.19	41.83
3	55.36	48.30
4	79.35	60.90
5	52.29	44.56
6	63.75	44.76
7	41.82	42.82
8	58.83	61.13
9	132.72	126.10
10	65.40	47.30
11	65.90	80.20
12	230.82	56.45
13	41.73	49.31
14	68.98	63.46
15	36.61	55.08
16	49.17	44.49
17	40.33	66.65
18	63.33	40.51
19	86.28	87.82
20	29.44	25.24
MEDIAN	61.01	52.19

Table C.7: Reported probability of losing balance as a function of frequency at each velocity of lateral oscillation.

0.032 ms^{-1} r.m.s.	Frequency (Hz)						
Subject no	0.5	0.63	0.8	1	1.25	1.6	2
1	10	5	20	30	30	50	30
2	70	50	5	5	20	5	5
3	25	30	20	20	25	15	15
4	25	25	10	0	0	10	0
5	5	0	2	5	0	5	0
6	5	5	10	5	5	0	5
7	7	0	5	30	8	1	3
8	48	1	8	1	15	5	3
9	1	5	1	2	2	3	4
10	0	0	0	25	0	5	0
11	5	0	5	10	15	5	5
12	40	0	50	0	10	10	20

13	0	0	0	0	0	0	5
14	45	2	10	5	15	10	15
15	15	5	5	10	40	20	50
16	5	10	8	5	0	3	0
17	0	0	10	15	25	15	15
18	20	20	20	40	50	20	30
19	0	5	0	10	20	0	10
20	75	60	20	10	10	25	10
MEDIAN	8.50	5.00	8.00	7.50	12.50	5.00	5.00
0.04 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	30	10	30	15	40	70	20
2	75	30	20	35	10	10	10
3	20	80	20	25	25	25	40
4	50	25	25	0	0	10	0
5	5	0	10	0	5	20	10
6	15	17	0	50	10	5	0
7	18	15	4	2	3	35	10
8	5	1	38	3	27	17	28
9	15	5	5	5	1	3	2
10	2	0	5	0	3	2	5
11	15	10	0	30	30	25	5
12	10	10	30	40	40	10	20
13	0	0	15	10	0	5	15
14	10	10	15	20	40	5	25
15	80	15	35	60	10	70	40
16	20	40	5	7	0	0	5
17	10	0	0	25	25	65	25
18	30	20	60	50	40	30	20
19	0	0	5	0	10	20	0
20	20	80	50	20	30	35	30
MEDIAN	15.00	10.00	15.00	17.50	10.00	18.50	12.50
0.05 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	60	10	10	25	50	10	50
2	10	30	12	45	10	15	15
3	80	25	40	40	40	25	40
4	25	100	25	25	50	0	25
5	15	5	5	0	10	10	5
6	5	15	35	37	40	15	20
7	95	25	8	15	30	30	40
8	35	6	3	16	75	22	48
9	2	2	4	15	3	35	3
10	5	15	0	5	5	10	10
11	5	15	40	20	10	20	10
12	40	0	20	60	10	20	10
13	10	5	5	0	5	15	0
14	20	15	5	60	30	15	10
15	60	50	20	90	30	45	40
16	60	10	0	10	40	35	0
17	50	15	15	80	40	15	50

18	75	40	25	75	25	75	30
19	0	40	30	0	50	10	50
20	65	35	30	35	45	20	75
MEDIAN	30.00	15.00	13.50	25.00	30.00	17.50	22.50
0.062 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	50	70	40	70	60	50	30
2	30	80	65	15	20	60	10
3	60	50	25	50	50	75	50
4	75	100	25	75	50	50	0
5	15	10	5	15	20	15	10
6	35	35	10	15	65	75	15
7	10	45	80	45	17	50	5
8	37	55	12	12	18	20	67
9	7	2	2	5	10	75	40
10	40	20	5	10	15	5	10
11	30	10	10	30	25	25	25
12	10	0	30	30	20	20	40
13	5	30	5	35	0	20	5
14	35	40	40	20	10	10	40
15	65	30	30	40	50	50	10
16	45	50	20	3	50	15	20
17	50	50	10	15	15	50	20
18	40	30	40	40	75	50	30
19	30	10	40	30	20	25	10
20	70	65	20	25	20	50	50
MEDIAN	36.00	37.50	22.50	27.50	20.00	50.00	20.00
0.08 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	40	80	40	50	90	60	60
2	90	80	45	10	45	30	20
3	50	60	40	60	50	40	50
4	50	75	50	100	50	100	10
5	15	25	15	20	15	20	5
6	20	40	45	20	10	55	35
7	80	90	55	75	35	25	35
8	69	92	92	38	49	89	47
9	10	4	3	10	6	10	55
10	40	15	10	5	20	25	0
11	35	20	40	20	20	35	35
12	30	20	60	20	40	80	25
13	15	5	25	15	20	30	25
14	70	60	50	20	30	60	40
15	25	85	50	30	60	60	25
16	35	60	85	25	10	0	5
17	80	75	65	25	15	90	40
18	80	80	70	90	40	75	50
19	70	70	80	60	100	40	0
20	55	50	65	45	35	90	50
MEDIAN	45.00	60.00	50.00	25.00	35.00	47.50	35.00

0.1 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	50	40	80	70	80	80	70
2	60	60	80	20	20	20	20
3	80	50	80	25	80	60	50
4	100	100	100	75	75	75	50
5	10	30	20	20	10	25	15
6	35	20	55	35	75	65	45
7	95	85	75	75	60	95	85
8	70	85	45	62	37	69	72
9	33	20	40	3	50	10	30
10	5	60	90	30	30	60	20
11	45	30	40	50	35	20	25
12	40	20	40	70	60	60	60
13	75	50	20	20	50	35	30
14	30	60	25	40	30	60	30
15	35	50	75	50	80	80	70
16	25	60	90	35	35	25	40
17	90	95	50	50	65	75	65
18	100	75	20	100	90	60	50
19	0	40	50	50	60	75	60
20	60	95	45	75	50	95	75
MEDIAN	47.50	55.00	50.00	50.00	55.00	60.00	50.00
0.13 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	90	90	70	90	100	100	50
2	95	85	85	50	80	65	30
3	100	100	40	80	100	100	80
4	100	100	75	100	75	75	50
5	35	25	10	50	20	20	7
6	100	30	20	70	65	100	60
7	100	95	95	98	80	90	65
8	95	82	58	82	88	30	67
9	40	10	15	15	15	75	70
10	90	30	10	40	70	50	15
11	40	40	60	50	45	50	20
12	70	70	100	80	60	70	70
13	80	45	70	65	85	0	40
14	50	75	5	60	60	35	30
15	100	75	80	60	100	75	45
16	100	80	95	80	90	85	45
17	95	60	100	95	75	75	75
18	100	60	50	60	80	100	60
19	90	80	50	80	60	30	70
20	100	100	50	100	65	75	90
MEDIAN	95.00	75.00	59.00	75.00	75.00	75.00	55.00
0.16 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	100	90	90	100	90	100	80
2	100	80	95	70	90	90	40
3	100	100	80	90	80	50	100

4	100	75	100	100	100	100	100
5	25	15	35	35	30	15	20
6	100	70	90	85	75	55	85
7	100	100	100	98	99	95	90
8	40	70	97	92	95	50	88
9	50	40	80	60	40	75	80
10	100	75	40	75	65	70	75
11	80	75	55	50	50	60	55
12	100	80	90	100	100	100	90
13	80	90	20	90	55	60	55
14	50	50	80	50	75	75	70
15	70	90	100	85	95	80	70
16	100	100	40	95	85	85	55
17	100	100	95	75	100	100	100
18	100	100	60	100	100	100	80
19	10	40	100	90	95	25	70
20	100	75	100	90	95	80	100
MEDIAN	100.00	77.50	90.00	90.00	90.00	77.50	80.00

Table C.8: Peak to peak lateral COP position as a function of frequency at each velocity of lateral oscillation.

0.032 ms^{-1} r.m.s.	Frequency (Hz)						
Subject no	0.5	0.63	0.8	1	1.25	1.6	2
1	27.91	26.75	27.40	24.74	24.24	23.26	24.91
2	34.02	32.97	15.00	13.76	18.33	12.80	13.05
3	22.70	22.96	19.13	17.32	16.45	17.87	17.04
4	27.22	26.88	25.01	26.29	18.08	16.14	17.51
5	23.36	22.83	24.21	15.80	20.28	16.06	19.04
6	25.53	25.10	27.06	21.27	20.00	19.11	20.67
7	18.37	19.83	18.43	20.23	18.27	12.55	12.99
8	25.20	27.30	23.49	23.57	22.52	21.25	22.29
9	21.65	26.27	21.57	21.92	16.71	24.15	24.41
10	26.80	20.44	21.91	30.93	22.96	16.21	19.71
11	34.42	30.23	32.60	36.65	34.71	26.67	29.23
12	27.46	25.94	29.17	22.82	24.65	20.01	22.87
13	22.25	23.38	13.92	21.30	15.66	18.97	17.16
14	35.42	28.26	25.41	25.75	24.44	24.43	19.97
15	23.68	24.06	23.10	25.03	19.27	25.26	26.57
16	30.40	29.61	27.63	28.51	34.58	23.22	23.53
17	27.71	24.95	28.28	26.51	27.62	20.24	23.66
18	30.96	23.98	19.76	20.10	19.97	16.87	16.02
19	24.85	29.77	24.60	25.45	24.35	25.25	24.09
20	31.47	24.68	20.89	22.29	21.75	14.41	16.29

MEDIAN	27.01	25.52	23.85	23.20	21.01	19.56	20.32
0.04 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	25.33	30.17	24.49	28.23	25.37	23.56	22.76
2	31.88	29.76	17.55	21.03	15.46	9.15	17.38
3	20.79	24.40	18.62	22.73	19.53	15.90	15.45
4	29.21	28.89	24.12	20.15	21.66	20.70	18.30
5	22.30	21.25	26.55	19.91	18.71	19.88	19.40
6	27.86	22.92	18.77	19.97	23.73	19.84	23.36
7	21.12	23.51	16.97	21.47	17.27	19.55	16.86
8	24.54	21.81	24.03	22.54	25.16	23.17	21.48
9	24.42	29.91	23.55	20.78	21.15	23.16	25.14
10	24.45	21.46	25.40	19.42	24.89	17.98	17.65
11	35.69	32.32	31.82	36.57	32.83	34.19	31.58
12	26.78	33.70	24.35	32.28	23.77	24.45	25.33
13	20.99	26.12	21.66	20.38	16.67	13.78	20.08
14	28.12	34.39	32.18	28.33	29.64	23.86	22.53
15	28.80	30.06	21.52	27.21	20.61	20.41	16.45
16	28.98	27.44	33.88	27.71	25.33	23.05	17.52
17	30.63	28.11	24.18	27.33	27.58	25.42	23.18
18	24.90	19.93	25.64	27.42	20.60	14.36	13.01
19	26.52	27.10	26.55	33.84	23.99	21.65	24.19
20	26.36	32.55	19.83	18.67	19.81	19.73	15.49
MEDIAN	26.44	27.78	24.15	22.63	22.70	20.55	19.74
0.05 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	31.63	23.44	27.47	28.29	24.00	26.88	22.43
2	19.90	22.93	21.40	18.15	17.43	14.32	11.06
3	23.69	20.30	23.47	21.06	18.95	17.13	21.15
4	32.73	29.74	24.42	23.50	20.89	20.53	19.82
5	30.98	18.01	21.32	24.45	20.61	21.25	16.83
6	25.69	29.29	30.10	27.85	21.03	22.17	21.27
7	28.37	19.00	23.65	23.41	19.14	20.04	15.55
8	23.64	23.45	23.51	24.84	20.01	24.10	22.87
9	27.50	22.95	26.95	25.25	23.86	22.55	17.77
10	26.18	27.36	20.87	22.56	20.12	17.45	19.08
11	31.11	30.95	34.32	35.96	32.42	28.77	28.55
12	30.33	29.85	30.50	28.48	23.99	25.22	23.09
13	27.30	15.11	22.40	19.04	21.77	23.26	19.43
14	34.19	23.76	29.62	33.31	29.81	18.97	22.99
15	31.84	28.26	20.95	32.25	22.01	23.17	14.18
16	33.74	30.19	29.34	29.95	24.23	20.76	20.84
17	36.57	26.09	31.21	30.63	24.74	20.41	28.40
18	29.62	23.75	25.38	22.31	26.15	14.74	22.52
19	25.04	33.22	25.71	20.42	25.63	26.81	17.89
20	32.49	21.77	28.34	14.01	20.84	17.27	19.89
MEDIAN	29.98	23.76	25.55	24.65	21.89	21.01	20.36
0.062 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	31.73	29.68	28.04	25.83	27.06	26.33	27.79

2	20.32	32.58	20.01	19.02	23.04	13.29	11.93
3	22.83	25.34	18.91	24.62	20.06	17.99	16.65
4	30.78	28.69	23.70	29.41	24.86	18.98	16.41
5	29.77	26.91	24.08	23.97	23.69	25.04	17.76
6	31.37	27.72	26.90	23.66	27.58	26.84	18.49
7	24.27	26.51	26.21	21.59	21.87	21.55	18.17
8	23.11	24.54	25.61	23.37	20.00	20.28	24.51
9	24.95	23.12	26.18	26.74	24.59	27.88	26.90
10	28.79	29.23	22.69	23.47	22.69	20.67	17.43
11	35.30	31.85	31.28	34.65	30.87	32.36	29.24
12	26.33	21.89	34.52	25.60	23.15	21.95	24.36
13	24.61	25.45	24.38	29.34	13.72	22.95	16.82
14	27.85	33.31	32.21	28.65	26.36	18.77	22.93
15	29.42	32.65	31.26	20.48	15.04	24.89	22.03
16	32.03	39.11	29.87	30.58	28.68	20.52	20.77
17	34.90	28.01	30.31	27.92	28.26	24.63	23.80
18	27.14	22.50	34.42	23.43	19.97	21.30	20.36
19	30.65	23.84	28.77	27.33	25.35	30.05	23.31
20	28.55	20.15	26.27	27.20	18.75	22.73	13.30
MEDIAN	28.67	27.32	26.59	25.71	23.42	22.34	20.57
0.08 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	29.62	28.74	23.37	26.66	29.62	25.56	26.85
2	35.62	27.83	22.49	15.95	25.07	13.58	16.96
3	24.48	21.83	21.54	20.22	18.89	20.89	15.27
4	28.61	30.68	20.22	30.70	24.32	24.39	23.80
5	22.53	23.73	32.50	24.95	27.12	18.94	18.29
6	24.14	27.22	25.43	22.84	23.27	22.72	22.23
7	23.12	22.48	22.33	22.48	21.49	15.44	20.92
8	25.60	25.70	21.55	22.26	22.40	22.92	21.14
9	25.43	22.25	17.45	26.04	24.00	24.13	25.18
10	33.74	25.43	26.98	27.05	26.86	22.44	25.02
11	33.54	33.77	39.10	39.32	32.26	33.51	25.71
12	29.72	35.07	32.42	29.47	23.72	22.63	21.40
13	29.93	25.13	23.77	25.13	21.22	23.38	22.37
14	36.90	34.12	28.99	31.15	27.95	27.04	23.97
15	23.66	19.27	22.06	20.53	25.70	23.36	16.88
16	35.23	27.71	23.72	30.56	24.89	26.02	19.68
17	33.02	33.78	30.22	28.56	25.74	27.57	23.51
18	37.47	26.13	27.23	28.58	16.22	16.81	15.98
19	34.44	37.00	30.94	24.44	36.08	25.41	24.69
20	28.52	22.57	28.46	25.42	19.16	15.21	17.44
MEDIAN	29.67	26.67	24.60	25.73	24.61	23.14	21.82
0.1 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	32.90	28.83	27.99	28.42	27.60	25.59	27.82
2	33.08	20.40	25.92	20.05	14.71	12.10	16.12
3	26.65	26.64	26.41	19.80	21.93	19.63	18.88
4	28.74	26.80	31.05	31.69	23.01	24.67	21.85
5	37.06	33.13	20.76	24.58	23.34	26.66	23.85
6	30.38	25.15	24.84	27.09	24.60	23.64	20.70

7	29.80	24.58	23.85	25.26	22.81	23.03	11.10
8	32.43	29.84	24.75	24.54	21.14	22.77	24.43
9	28.26	24.15	25.09	21.36	25.24	30.57	26.04
10	29.43	27.50	32.67	26.56	24.58	21.99	24.46
11	41.54	35.39	28.98	33.01	30.50	31.75	30.40
12	29.28	29.07	27.56	34.99	29.81	25.80	24.19
13	33.31	20.68	29.51	19.94	22.32	23.34	16.71
14	35.57	33.59	28.11	30.62	29.69	26.37	26.49
15	27.93	21.80	22.88	26.62	19.69	26.62	20.93
16	26.77	35.90	39.32	36.17	34.17	22.47	26.91
17	38.13	19.08	34.29	32.57	29.20	25.38	20.52
18	45.03	24.08	28.43	26.40	16.12	22.97	17.32
19	25.99	32.46	21.80	34.98	26.32	32.08	28.24
20	24.22	23.44	23.38	21.08	17.28	16.43	12.33
MEDIAN	30.09	26.72	26.98	26.59	23.96	24.15	22.85
0.13 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	34.21	29.56	37.44	35.18	33.83	29.66	31.80
2	34.28	23.23	28.08	23.58	23.84	13.61	23.37
3	27.61	19.39	28.12	19.96	28.62	24.89	19.27
4	35.69	22.84	37.45	28.57	23.72	28.93	26.59
5	33.45	31.21	20.16	29.51	19.29	30.21	19.45
6	31.30	25.63	30.15	30.56	27.87	34.17	23.11
7	31.36	25.73	30.90	26.56	20.46	24.68	21.65
8	30.95	25.08	26.36	25.73	18.69	23.62	22.16
9	26.75	26.07	30.38	29.65	26.69	26.71	37.09
10	33.42	27.95	30.39	23.21	23.09	27.23	23.78
11	35.37	31.09	26.50	33.64	37.05	24.59	32.82
12	33.83	36.12	27.01	31.89	29.64	21.49	25.73
13	39.25	29.29	21.36	21.99	20.24	20.15	21.84
14	32.93	26.36	33.50	32.92	28.61	33.61	26.35
15	26.30	26.73	26.99	25.36	26.55	26.38	21.51
16	35.80	31.78	29.61	30.67	30.73	20.31	24.34
17	35.68	34.49	27.79	32.73	29.41	24.22	27.17
18	48.17	31.46	29.22	29.86	18.62	17.09	19.71
19	34.89	35.20	30.09	33.35	31.73	38.22	30.79
20	27.62	16.79	32.03	18.84	21.01	14.53	13.95
MEDIAN	33.64	27.34	29.42	29.58	26.62	24.78	23.58
0.16 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	35.63	38.72	30.85	36.44	23.86	32.73	29.74
2	35.10	35.87	20.83	26.38	20.17	19.17	14.02
3	24.39	28.52	26.77	24.76	25.34	22.72	15.98
4	29.99	26.14	26.15	29.12	29.27	37.65	26.27
5	44.89	33.64	28.94	24.16	27.94	34.03	22.50
6	36.29	35.77	23.73	33.86	24.31	27.72	30.99
7	46.15	33.52	24.23	26.72	22.51	22.87	19.71
8	31.96	28.67	23.72	26.21	24.87	23.24	25.23
9	29.34	32.92	34.57	31.34	27.38	35.59	38.15
10	50.91	40.88	29.64	25.78	31.75	32.20	26.95
11	36.40	35.52	38.51	39.63	36.83	28.74	25.06

12	31.08	31.22	35.34	33.14	31.64	65.54	19.03
13	49.73	37.68	29.19	32.60	28.28	24.79	23.99
14	40.18	32.53	32.93	32.71	31.22	32.79	28.58
15	32.59	38.35	28.90	29.17	21.57	22.48	27.69
16	32.55	27.40	34.44	29.37	29.00	24.85	25.56
17	42.16	28.18	36.60	36.49	32.49	23.35	29.34
18	44.18	46.22	37.29	23.81	33.53	29.65	18.57
19	31.27	42.75	32.68	38.34	26.89	36.93	28.04
20	22.18	26.21	23.05	22.49	26.12	16.90	14.79
MEDIAN	35.37	33.58	29.42	29.27	27.66	28.23	25.40

Table C.9: Lateral r.m.s. COP velocity as a function of frequency at each velocity of lateral oscillation.

0.032 ms^{-1} r.m.s.	Frequency (Hz)						
Subject no	0.5	0.63	0.8	1	1.25	1.6	2
1	53.70	55.44	56.08	53.08	53.32	58.81	61.46
2	28.42	26.19	17.76	17.92	27.92	23.79	22.46
3	40.07	37.04	37.89	37.88	31.53	40.31	47.53
4	47.62	42.09	48.37	50.95	40.79	35.75	46.99
5	33.50	44.91	36.28	29.24	40.31	35.78	48.29
6	36.89	42.65	38.31	35.94	37.21	41.90	47.04
7	28.52	30.63	25.81	40.65	35.13	20.32	24.52
8	46.01	46.14	46.58	45.07	44.70	46.26	50.67
9	41.14	56.01	45.15	42.13	35.73	55.50	51.25
10	41.16	31.18	41.46	38.24	36.52	33.21	36.72
11	62.84	59.69	56.60	61.04	65.56	59.99	66.68
12	42.36	49.10	55.74	48.09	50.03	35.05	42.41
13	35.86	35.84	27.91	35.34	34.51	35.56	39.17
14	45.02	44.91	41.75	40.34	48.64	43.86	42.86
15	36.77	28.78	33.80	39.72	28.31	48.17	47.12
16	51.91	46.44	53.99	51.48	62.36	41.25	49.20
17	51.38	46.21	57.71	48.87	51.25	45.52	46.54
18	31.63	32.66	29.58	32.31	27.44	29.23	28.79
19	51.95	54.41	47.88	57.02	57.27	59.33	66.15
20	33.05	21.73	34.92	35.78	28.94	25.10	31.76
MEDIAN	41.15	43.78	41.61	40.49	38.76	40.78	47.01
0.04 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	51.73	62.10	53.66	66.84	59.73	53.98	55.95
2	25.66	35.47	27.17	26.09	22.19	16.50	27.07
3	37.08	41.32	41.85	42.48	37.89	38.45	30.24
4	43.02	46.19	46.10	47.75	48.82	46.44	35.97
5	33.41	40.38	42.31	37.76	26.86	35.13	43.61
6	43.22	42.16	35.68	39.90	47.76	42.93	48.53
7	27.36	43.27	30.04	42.90	28.40	31.67	32.41

8	44.82	51.42	50.58	47.97	58.08	45.41	45.93
9	42.85	59.29	42.80	48.72	48.29	51.76	59.55
10	42.25	41.72	39.63	36.22	56.93	35.91	32.78
11	60.24	65.20	55.44	83.03	59.53	61.96	62.03
12	50.82	54.47	44.42	72.32	45.30	47.77	47.35
13	35.43	38.03	32.58	36.00	31.27	27.31	40.20
14	39.80	50.77	40.22	54.52	50.17	45.39	47.37
15	33.75	50.52	31.57	45.14	37.76	42.94	26.44
16	41.84	42.02	69.58	40.73	39.58	55.54	35.42
17	46.81	56.79	49.14	45.46	46.43	48.62	51.45
18	27.30	29.22	43.06	28.26	29.56	26.05	20.68
19	46.96	48.10	53.06	78.31	57.09	43.15	58.53
20	33.43	38.16	24.50	30.84	37.69	34.42	31.83
MEDIAN	42.04	44.73	42.55	44.02	45.86	43.04	41.90
0.05 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	61.15	47.25	58.61	67.91	56.48	59.69	59.34
2	22.39	21.95	33.17	25.34	26.51	23.04	17.93
3	37.19	40.06	42.74	47.29	40.55	37.03	39.12
4	56.13	43.12	48.55	54.66	42.28	46.37	50.72
5	44.48	29.08	31.27	53.52	44.19	46.48	38.10
6	44.89	36.57	56.12	52.09	46.89	50.82	51.80
7	36.95	32.12	36.44	39.65	29.62	38.88	31.23
8	48.00	44.99	51.22	53.83	51.41	53.42	60.53
9	53.87	37.61	57.61	67.16	48.81	61.87	38.53
10	38.27	40.99	33.84	39.10	37.09	25.54	41.28
11	49.83	60.75	64.44	64.17	61.52	65.60	62.64
12	49.37	55.71	49.62	59.06	50.74	52.62	46.77
13	37.91	24.01	44.75	38.77	36.37	44.40	34.98
14	49.35	25.30	55.21	67.64	48.68	38.15	45.87
15	48.64	38.42	27.81	63.66	31.75	40.75	22.44
16	48.66	44.29	52.03	62.95	50.28	46.27	53.16
17	48.97	48.29	63.25	59.63	51.92	44.20	57.07
18	30.37	22.15	35.53	30.25	30.25	21.65	40.84
19	48.94	65.86	57.73	41.45	59.09	51.24	48.98
20	29.96	29.48	34.51	26.37	30.21	29.58	33.70
MEDIAN	48.32	39.24	49.08	53.67	45.54	45.33	43.58
0.062 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	60.28	56.28	64.68	59.29	52.95	63.78	66.40
2	21.45	30.83	30.24	20.28	34.22	18.20	16.64
3	40.74	42.71	31.05	44.58	44.93	26.15	38.33
4	52.59	47.17	54.85	50.06	52.02	41.94	37.63
5	37.91	30.59	53.50	42.24	36.76	54.44	38.98
6	45.51	48.63	49.80	52.83	46.95	33.69	53.86
7	30.77	40.12	52.20	40.16	32.14	37.58	34.97
8	51.80	51.97	43.76	59.62	48.00	48.24	51.76
9	52.71	47.26	59.01	61.59	62.34	62.38	67.76
10	37.80	42.13	31.45	38.36	41.10	32.15	34.48
11	65.30	61.09	68.58	60.27	58.90	64.97	74.16

12	52.72	37.29	66.60	57.15	57.83	41.08	54.85
13	36.32	39.39	45.59	40.21	30.45	46.05	37.92
14	35.78	51.92	46.14	47.57	49.91	35.83	46.00
15	42.49	43.81	47.52	32.41	24.97	57.39	45.37
16	55.89	56.47	45.57	59.71	52.89	42.46	45.76
17	51.18	49.62	60.58	59.64	47.23	55.96	52.74
18	33.17	26.16	53.54	31.81	32.24	39.74	33.14
19	52.04	50.92	57.21	45.41	57.48	89.76	47.96
20	32.06	25.58	47.66	30.29	30.82	43.99	30.26
MEDIAN	44.00	45.49	51.00	46.49	47.09	43.22	45.56
0.08 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	56.81	57.72	51.36	61.02	72.53	66.06	67.29
2	30.45	32.65	37.20	21.29	26.04	23.99	28.31
3	37.82	32.65	35.73	44.65	33.75	45.67	37.32
4	43.22	48.89	38.73	42.74	47.75	48.27	57.89
5	34.56	30.38	48.82	42.65	43.57	39.49	46.03
6	44.25	41.99	41.07	43.35	48.86	45.49	55.22
7	34.20	33.08	39.35	42.96	37.99	30.28	52.35
8	49.34	53.74	30.21	63.57	58.29	57.74	54.82
9	56.13	46.10	32.99	57.10	60.26	70.38	77.55
10	43.92	41.22	41.45	46.23	43.58	40.85	45.54
11	63.69	61.43	70.93	61.64	64.50	67.23	73.48
12	50.50	60.41	64.42	60.64	48.16	48.20	55.96
13	35.39	40.74	38.10	44.83	38.00	42.39	50.74
14	39.28	51.28	43.23	52.36	56.78	62.91	56.03
15	31.40	27.52	26.57	34.29	46.45	37.01	36.39
16	59.91	45.18	40.90	67.90	50.73	65.53	42.76
17	44.98	56.20	62.21	54.26	53.35	53.96	56.39
18	41.03	25.17	53.53	40.74	17.77	29.04	17.08
19	61.73	49.85	54.57	40.18	86.95	54.78	61.97
20	37.81	26.47	33.90	34.05	25.38	29.67	31.67
MEDIAN	43.57	43.58	40.99	44.74	47.96	46.93	53.59
0.1 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	65.60	49.44	54.74	51.05	57.90	62.37	72.22
2	43.19	31.36	25.29	30.10	24.34	22.76	30.65
3	34.08	47.17	40.62	37.83	39.44	36.10	45.58
4	48.38	46.03	52.40	58.61	46.07	50.07	36.61
5	55.95	39.57	22.49	49.00	44.23	39.70	29.55
6	49.03	42.16	33.62	51.24	59.14	48.17	52.04
7	54.07	39.97	47.39	53.90	34.23	43.74	22.36
8	51.25	45.45	60.24	60.48	56.21	54.84	56.69
9	63.47	50.85	51.18	46.59	65.36	88.18	77.11
10	46.35	42.84	42.29	39.02	47.08	45.48	47.90
11	60.16	66.44	35.44	59.49	60.40	69.04	55.98
12	56.94	52.35	60.47	74.36	62.14	57.89	53.35
13	47.26	28.63	50.78	33.23	30.70	41.63	37.05
14	42.80	44.67	50.07	71.63	48.06	60.47	52.67
15	43.02	25.99	41.88	36.32	38.78	45.95	40.21
16	43.87	69.49	58.96	58.53	69.67	45.02	59.07

17	55.06	25.26	58.73	67.39	50.63	59.38	43.16
18	50.48	32.61	50.05	24.67	22.05	22.71	36.98
19	49.45	58.66	28.65	85.11	57.48	94.74	70.61
20	27.67	24.78	39.79	26.48	32.45	22.93	22.46
MEDIAN	49.24	43.75	48.72	51.14	47.57	47.06	46.74
0.13 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	61.36	52.15	82.46	88.97	81.95	64.72	76.53
2	41.70	31.44	49.95	31.88	25.00	20.20	41.83
3	37.12	29.51	53.40	33.99	38.66	39.01	48.30
4	51.03	39.20	74.62	40.90	45.42	75.61	60.90
5	49.93	39.90	34.40	26.59	37.52	43.48	44.56
6	33.33	51.56	68.92	54.66	63.68	78.19	44.76
7	48.33	43.92	64.55	49.71	31.55	45.61	42.82
8	58.45	53.21	67.51	61.23	49.83	48.21	61.13
9	60.22	62.85	64.49	79.48	83.35	74.01	126.10
10	44.82	48.08	56.50	38.56	43.12	66.99	47.30
11	56.33	56.21	32.77	75.73	68.78	46.41	80.20
12	69.78	59.84	45.90	86.81	76.92	41.33	56.45
13	45.88	44.19	21.47	34.31	37.88	40.06	49.31
14	45.34	34.16	68.71	62.94	62.29	61.08	63.46
15	32.92	35.67	55.58	45.07	45.68	52.40	55.08
16	59.26	41.73	54.74	61.86	61.66	24.79	44.49
17	48.59	65.09	44.19	74.91	48.51	56.55	66.65
18	44.66	42.18	56.62	69.01	25.05	35.41	40.51
19	54.46	71.43	62.12	92.05	103.21	120.16	87.82
20	31.62	23.26	61.25	23.78	31.26	22.71	25.24
MEDIAN	48.46	44.06	56.56	57.94	47.09	47.31	52.19
0.16 ms^{-1} r.m.s.	0.5	0.63	0.8	1	1.25	1.6	2
1	68.75	69.36	74.24	83.56	57.63	63.19	77.94
2	37.16	56.33	23.01	36.49	28.53	26.19	31.54
3	33.62	46.75	62.82	63.67	52.65	55.36	32.05
4	52.35	43.83	48.05	57.83	46.21	79.35	63.91
5	71.41	47.31	53.71	32.44	44.73	52.29	49.85
6	48.67	60.15	38.48	48.57	54.47	63.75	71.32
7	60.42	61.01	27.85	56.47	45.34	41.82	39.01
8	50.99	53.62	59.09	73.26	68.93	58.83	59.11
9	72.87	84.67	59.54	87.80	72.26	132.72	132.29
10	72.46	69.18	58.39	39.73	57.00	65.40	61.47
11	54.35	54.63	78.40	77.69	82.05	65.90	54.63
12	46.32	72.09	78.95	66.61	67.47	230.82	48.56
13	83.38	56.85	61.11	60.45	58.75	41.73	54.23
14	52.32	56.79	64.43	88.48	71.28	68.98	61.94
15	42.71	78.81	28.45	72.07	26.09	36.61	54.66
16	58.92	48.74	73.39	51.44	57.22	49.17	56.05
17	52.75	38.51	79.87	89.18	54.24	40.33	72.90
18	44.97	60.74	60.12	21.18	40.49	63.33	40.44
19	71.70	91.77	69.60	120.44	57.76	86.28	100.84
20	28.76	32.92	38.19	24.78	32.34	29.44	26.72
MEDIAN	52.55	56.82	59.83	62.06	55.74	61.01	55.35

C.2. Data used in the analysis of the second experiment reported in Chapter 5

Table C.10: 'Discomfort or difficulty' ratings as a function of acceleration with and without hand support.

WITHOUT SUPPORT							WITH SUPPORT					
	Acceleration (ms^{-2} r.m.s.)						Acceleration (ms^{-2} r.m.s.)					
Subject no	0.315	0.4	0.5	0.63	0.8	1	0.315	0.4	0.5	0.63	0.8	1
1	5	20	120	170	130	200	10	10	50	70	100	100
2	20	80	80	120	120	120	20	50	50	80	100	100
3	100	110	105	110	120	130	25	20	15	90	90	80
4	60	150	110	180	130	150	20	50	20	40	50	100
5	10	60	150	150	170	120	25	20	10	70	40	100
6	70	60	70	100	125	140	30	40	65	90	85	110
7	60	80	90	90	150	100	20	40	40	60	80	80
8	50	75	70	190	130	120	50	20	70	110	100	105
9	70	50	200	150	150	200	20	30	20	60	100	150
10	30	50	50	80	90	120	20	60	40	30	90	110
11	40	30	70	70	90	120	20	30	40	50	60	90
12	50	50	50	100	125	125	50	50	50	100	100	150
13	40	70	100	110	100	110	20	60	40	50	90	100
14	20	70	70	60	90	100	10	30	40	80	70	95
15	45	100	145	110	160	175	20	35	25	75	110	115
16	120	120	120	130	150	130	70	50	100	90	100	100
17	110	110	90	120	140	140	20	20	10	50	80	100
18	40	10	40	60	120	50	30	50	30	80	60	150
19	50	80	65	120	100	130	50	50	50	100	100	100
20	100	70	120	130	120	150	20	30	20	30	50	100
MEDIAN	50	70	90	115	125	127.5	20	37.5	40	72.5	90	100

Table C.11: Lateral r.m.s. COP velocity as a function of acceleration with and without hand support.

WITHOUT SUPPORT						
	Acceleration (ms^{-2} r.m.s.)					
Subject no	0.315	0.4	0.5	0.63	0.8	1
1	36.71	34.38	51.22	48.30	52.00	39.28
2	34.36	33.96	35.66	44.27	37.98	55.41
3	52.92	44.83	38.30	50.54	58.57	63.43
4	57.01	54.12	59.59	49.34	51.11	53.00
5	50.72	41.36	68.48	61.27	81.22	72.90
6	56.57	44.66	54.88	35.49	61.04	85.93
7	59.77	72.21	91.21	58.48	95.47	85.41
8	61.30	59.46	69.13	86.06	67.65	63.33

9	57.65	58.99	64.69	60.05	62.77	67.21
10	42.18	43.37	45.82	37.79	55.34	66.32
11	54.04	54.49	54.45	59.94	69.62	79.92
12	57.75	58.12	63.76	65.46	47.01	48.45
13	39.97	42.44	53.70	31.23	38.05	46.25
14	35.36	52.57	29.50	40.22	48.07	52.00
15	34.31	31.31	42.11	32.45	26.13	36.34
16	43.80	44.26	52.81	63.56	52.65	77.43
17	57.64	50.26	34.58	54.44	55.78	41.82
18	43.11	39.11	50.75	54.08	48.38	68.72
19	50.18	50.91	57.65	57.18	56.66	63.27
20	31.91	27.66	35.46	48.36	43.79	43.26
MEDIAN	50.30	44.74	53.24	52.24	53.97	63.13
	WITH SUPPORT					
	Acceleration (ms^{-2} r.m.s.)					
Subject no	0.315	0.4	0.5	0.63	0.8	1
1	19.02	23.69	17.20	12.59	12.39	12.04
2	14.59	7.86	8.87	14.23	9.71	13.57
3	14.28	13.42	13.07	15.58	15.00	20.23
4	18.26	21.81	14.19	24.88	25.53	26.15
5	15.78	11.86	21.46	28.61	17.12	17.20
6	8.32	10.71	10.42	10.17	12.97	8.42
7	14.85	19.84	16.84	14.80	24.51	19.46
8	13.42	12.49	7.45	23.16	16.81	15.49
9	30.61	23.90	23.68	50.56	32.09	19.68
10	11.44	19.00	12.83	9.61	13.96	7.96
11	8.83	12.47	8.29	11.38	15.11	15.64
12	7.17	12.96	11.49	16.80	15.74	21.47
13	8.34	11.78	8.83	5.80	11.39	8.36
14	14.70	21.09	16.42	28.76	21.44	21.88
15	13.62	16.94	16.93	25.04	17.41	23.29
16	25.07	17.65	27.28	24.75	28.70	34.04
17	27.91	22.91	22.11	26.47	23.02	37.49
18	14.95	16.05	18.26	22.13	19.67	35.09
19	5.17	9.41	5.30	5.43	6.00	8.01
20	7.46	21.98	25.61	10.85	22.77	26.48
MEDIAN	14.43	16.49	15.30	16.19	16.96	19.57

Table C.12: Lateral force applied to the hand support as a function of acceleration.

	Acceleration (ms^{-2} r.m.s.)					
Subject no	0.315	0.4	0.5	0.63	0.8	1
1	19.02	23.69	17.20	12.59	12.39	12.04
2	14.59	7.86	8.87	14.23	9.71	13.57
3	14.28	13.42	13.07	15.58	15.00	20.23
4	18.26	21.81	14.19	24.88	25.53	26.15
5	15.78	11.86	21.46	28.61	17.12	17.20
6	8.32	10.71	10.42	10.17	12.97	8.42

7	14.85	19.84	16.84	14.80	24.51	19.46
8	13.42	12.49	7.45	23.16	16.81	15.49
9	30.61	23.90	23.68	50.56	32.09	19.68
10	11.44	19.00	12.83	9.61	13.96	7.96
11	8.83	12.47	8.29	11.38	15.11	15.64
12	7.17	12.96	11.49	16.80	15.74	21.47
13	8.34	11.78	8.83	5.80	11.39	8.36
14	14.70	21.09	16.42	28.76	21.44	21.88
15	13.62	16.94	16.93	25.04	17.41	23.29
16	25.07	17.65	27.28	24.75	28.70	34.04
17	27.91	22.91	22.11	26.47	23.02	37.49
18	14.95	16.05	18.26	22.13	19.67	35.09
19	5.17	9.41	5.30	5.43	6.00	8.01
20	7.46	21.98	25.61	10.85	22.77	26.48
MEDIAN	14.43	16.49	15.30	16.19	16.96	19.57

Table C.13: 'Discomfort or difficulty' ratings as a function of frequency with and without hand support..

WITHOUT SUPPORT							
	Frequency (Hz)						
Subject no	0.315	0.4	0.5	0.63	0.8	1	2
1	500	700	350	200	200	400	500
2	250	200	150	120	150	200	150
3	200	175	145	130	150	225	250
4	250	130	120	150	150	160	200
5	350	350	250	120	250	300	500
6	300	120	150	140	95	140	200
7	150	200	200	100	120	150	150
8	150	150	125	120	125	150	170
9	300	300	200	200	300	300	250
10	150	110	130	120	130	160	200
11	130	120	80	120	110	130	150
12	150	125	150	125	150	150	150
13	250	120	150	110	180	150	100
14	150	130	125	100	100	120	160
15	380	450	250	175	190	200	200
16	160	170	140	130	150	200	200
17	200	150	150	140	175	160	250
18	300	180	350	50	150	90	500
19	225	200	150	130	150	175	150
20	150	130	150	150	120	150	200
MEDIAN	212.5	160	150	127.5	150	160	200
WHEN SUPPORT USED THROUGHOUT OSCILLATION							
	Frequency (Hz)						
Subject no	0.315	0.4	0.5	0.63	0.8	1	2
1	170	80	150	100	150	200	300
2	200	120	100	100	120	100	180
3	135	90	95	80	120	160	140
4	120	150	120	100	80	140	120

5	120	150	70	100	170	70	70
6	95	70	110	110	80	135	90
7	120	90	90	80	115	90	80
8	140	150	150	105	90	125	125
9	250	250	100	150	150	350	120
10	90	90	100	110	130	120	150
11	70	50	50	90	90	110	130
12	125	125	150	150	150	150	50
13	140	130	80	100	140	100	175
14	80	50	95	95	90	60	50
15	150	130	90	115	130	125	80
16	80	90	110	100	110	120	120
17	120	80	100	100	100	110	110
18	85	70	80	150	130	150	180
19	125	150	100	100	170	120	100
20	30	80	100	100	120	100	80
MEDIAN	120	90	100	100	120	120	120

Table C.14: Lateral r.m.s. COP velocity as a function of frequency with and without hand support.

WITHOUT SUPPORT							
	Frequency (Hz)						
Subject no	0.315	0.4	0.5	0.63	0.8	1	2
1	59.62	53.68	57.30	39.13	57.93	50.34	38.80
2	56.08	36.07	49.88	55.36	57.59	40.51	52.84
3	67.36	68.72	63.94	63.41	73.38	70.82	88.95
4	61.42	52.54	60.37	52.98	61.40	56.18	61.83
5	56.10	49.12	46.46	72.87	65.63	73.70	90.13
6	59.62	60.70	60.37	85.91	64.89	56.57	69.89
7	68.28	65.62	79.04	85.36	72.21	57.42	69.05
8	65.63	63.89	72.82	63.06	91.66	98.65	77.78
9	73.32	63.32	62.39	67.20	62.90	75.70	83.46
10	59.62	44.78	45.74	66.31	40.04	40.32	42.05
11	67.77	55.81	71.60	79.92	65.82	71.29	84.65
12	56.83	58.37	52.06	48.34	73.75	79.89	81.56
13	59.62	47.81	48.25	46.21	29.37	33.45	17.20
14	51.96	48.09	28.19	52.00	40.52	35.66	65.99
15	55.11	49.05	50.01	36.24	33.95	33.32	26.40
16	59.62	56.83	61.89	77.43	52.51	102.68	65.99
17	58.39	51.92	70.97	41.75	78.79	33.44	49.07
18	40.43	50.28	36.03	68.72	53.35	62.96	128.01
19	55.94	59.06	61.64	63.19	58.69	57.38	60.13
20	59.62	47.20	45.58	43.26	45.42	50.84	65.99
MEDIAN	59.62	53.11	58.83	63.13	60.05	56.98	65.99
WHEN SUPPORT USED THROUGHOUT OSCILLATION							
	Frequency (Hz)						
Subject no	0.315	0.4	0.5	0.63	0.8	1	2

1	26.23	19.41	35.01	27.87	30.54	32.08	19.63
2	31.55	19.06	20.43	20.79	20.16	38.60	18.04
3	28.90	19.63	21.61	27.23	35.68	27.73	42.22
4	40.20	48.32	43.60	39.52	49.59	58.73	50.56
5	36.42	41.87	35.99	43.74	36.26	34.37	27.98
6	38.85	45.15	35.07	39.38	34.73	44.89	23.11
7	40.34	41.02	45.94	42.49	43.82	44.69	53.04
8	45.46	48.32	25.13	44.77	31.46	59.91	23.43
9	52.26	46.47	52.08	46.93	47.63	63.09	36.64
10	24.51	22.64	27.32	25.55	21.91	27.80	17.03
11	43.60	41.55	51.87	42.43	44.01	52.78	52.20
12	48.62	54.84	50.30	49.15	45.19	50.55	51.21
13	28.55	30.26	21.78	27.41	22.57	22.37	25.34
14	21.93	20.33	19.37	23.07	24.00	20.60	20.39
15	21.79	28.00	22.79	23.31	18.03	24.81	18.59
16	27.43	31.27	34.05	33.30	29.29	32.32	37.26
17	25.93	17.30	26.30	22.67	19.75	23.22	18.28
18	25.82	19.45	19.50	29.55	24.39	29.80	35.96
19	31.66	36.61	30.36	33.17	29.03	26.78	24.66
20	32.92	31.55	23.52	16.41	28.93	17.75	22.12
MEDIAN	31.61	31.41	28.84	31.36	29.92	32.20	25.00

WHEN SUPPORT USED IF REQUIRED

	Frequency (Hz)						
Subject no	0.315	0.4	0.5	0.63	0.8	1	2
1	31.64	33.72	40.74	43.90	43.28	31.88	38.99
2	33.17	29.55	49.09	47.52	33.19	32.12	48.50
3	29.18	30.23	50.13	74.64	44.10	46.89	29.02
4	38.90	41.85	38.70	39.57	46.75	55.72	42.56
5	46.42	42.54	30.76	43.69	46.42	40.34	53.49
6	63.63	41.91	53.18	57.25	46.22	47.04	57.68
7	51.30	47.14	57.68	61.02	47.19	49.60	56.91
8	55.53	53.61	37.38	100.95	39.48	72.08	52.57
9	59.04	63.16	58.54	56.93	59.99	67.35	63.27
10	34.06	34.67	20.31	34.97	35.01	26.83	34.86
11	47.84	50.93	62.47	60.23	48.10	49.82	63.68
12	62.93	64.65	74.08	71.25	67.56	69.11	64.66
13	35.34	31.23	26.00	32.54	28.78	28.82	42.91
14	39.56	37.42	48.46	26.20	27.57	30.25	41.18
15	31.67	31.05	26.00	30.08	32.97	27.22	25.41
16	31.98	37.59	36.89	41.53	35.75	40.12	56.59
17	36.10	45.85	50.99	46.21	40.05	48.81	37.42
18	44.73	39.46	42.61	58.23	42.52	44.06	111.88
19	44.23	43.80	40.12	63.78	39.38	60.91	43.60
20	24.19	11.50	25.43	23.00	40.74	13.63	19.99
MEDIAN	39.23	40.66	41.67	46.86	41.63	45.48	46.05

Table C.15: Lateral force applied to the hand support as a function of frequency when the hand support is used throughout the oscillation and when the hand support is used if required.

WHEN SUPPORT USED THROUGHOUT OSCILLATION							
	Frequency (Hz)						
Subject no	0.315	0.4	0.5	0.63	0.8	1	2
1	25.10	24.68	24.81	12.04	19.81	22.14	17.35
2	30.06	16.20	13.94	13.57	18.89	17.97	20.68
3	25.07	11.86	13.65	20.23	27.21	39.26	24.47
4	27.73	30.32	32.72	26.15	24.23	16.61	19.82
5	30.15	22.24	19.87	17.20	26.42	20.67	25.25
6	20.97	15.01	15.65	8.42	17.94	7.89	10.62
7	25.53	12.51	27.77	19.46	31.64	25.70	29.99
8	49.08	33.31	28.25	15.49	16.15	16.96	17.76
9	32.07	46.19	50.87	19.68	33.23	65.60	40.79
10	26.63	18.24	16.91	7.96	15.40	28.73	23.02
11	28.75	20.56	20.33	15.64	15.85	12.64	18.17
12	18.68	20.46	22.61	21.47	34.49	40.20	31.08
13	33.16	19.67	17.65	8.36	17.87	12.50	11.41
14	32.37	21.65	17.84	21.88	26.63	22.13	15.49
15	32.62	20.00	15.22	23.29	24.92	23.78	19.24
16	25.80	25.71	38.00	34.04	39.04	60.58	51.70
17	42.79	32.69	28.38	37.49	37.60	44.62	45.61
18	39.70	21.88	21.69	35.09	51.95	49.74	55.55
19	9.64	10.38	5.53	8.01	10.62	9.24	8.25
20	44.53	42.53	34.98	26.48	36.06	26.00	33.24
MEDIAN	29.40	21.11	21.01	19.57	25.67	22.96	21.85
WHEN SUPPORT USED IF REQUIRED							
	Frequency (Hz)						
Subject no	0.315	0.4	0.5	0.63	0.8	1	2
1	20.57	37.11	17.95	29.99	27.35	18.59	23.60
2	32.36	18.81	7.48	1.99	4.60	18.94	15.56
3	40.48	23.91	24.17	1.31	19.44	31.92	33.72
4	47.22	42.04	33.37	24.30	41.22	29.76	59.88
5	41.29	23.45	58.32	27.68	20.50	31.29	44.89
6	32.16	32.41	16.37	26.82	12.44	31.58	28.50
7	29.29	26.22	35.78	34.23	20.99	31.93	9.65
8	53.31	54.92	29.36	23.88	30.59	10.94	18.30
9	52.78	32.19	53.43	56.22	73.25	72.68	44.25
10	22.28	17.78	29.83	17.28	23.46	16.30	16.84
11	26.25	13.42	13.61	14.47	20.19	14.27	11.25
12	36.27	38.91	22.85	30.99	22.21	14.11	27.19
13	41.22	25.37	52.93	7.93	26.03	36.78	11.02
14	46.52	20.66	14.33	35.83	19.26	17.23	3.42
15	63.44	58.94	76.78	62.51	32.43	36.26	53.35
16	36.03	27.37	49.45	18.21	35.69	33.29	54.47
17	67.25	43.46	49.65	42.35	73.87	52.99	52.20
18	34.08	26.35	31.74	17.79	25.93	49.08	4.71
19	13.38	16.39	8.95	5.65	11.28	2.85	11.72

20	56.50	32.58	35.27	43.60	28.03	28.12	30.76
MEDIAN	38.38	26.86	30.79	25.56	24.69	30.52	25.40

Table C.16: ‘Discomfort or difficulty’ ratings as a function of support height.

	Support height (cm)					
Subject no	No support	92	109	126	143	163
1	250	90	110	100	130	150
2	150	80	100	120	100	150
3	180	110	100	100	115	120
4	175	80	110	80	120	80
5	200	70	50	110	40	110
6	200	130	115	100	90	85
7	120	90	50	50	80	100
8	130	110	110	125	90	120
9	80	120	140	100	60	100
10	120	100	90	120	70	100
11	120	50	90	90	70	80
12	100	100	150	150	100	150
13	120	70	90	90	100	70
14	80	90	90	105	100	85
15	175	125	110	110	105	115
16	175	90	90	100	130	130
17	175	85	110	100	110	100
18	80	80	100	80	90	90
19	150	75	125	100	125	75
20	120	50	100	80	50	100
MEDIAN	140	90	100	100	100	100

Table C.17: Lateral r.m.s. COP velocity as a function of support height.

	Support height (cm)					
Subject no	No support	92	109	126	143	163
1	56.11	30.83	32.82	28.17	31.84	34.74
2	50.93	27.62	24.37	31.67	18.74	16.24
3	66.78	32.78	42.03	29.27	30.00	32.81
4	66.23	49.04	58.41	51.96	57.95	56.14
5	76.07	36.36	36.39	40.62	30.91	42.44
6	71.16	78.09	39.43	39.27	40.85	44.61
7	77.41	48.88	48.56	40.61	35.66	41.24
8	94.91	62.95	49.49	49.86	61.40	59.61
9	100.11	53.38	61.90	66.48	56.43	63.03
10	53.08	26.02	28.36	23.43	25.95	28.02
11	52.84	45.90	45.14	45.23	37.58	42.98
12	81.53	53.79	51.84	46.71	43.29	56.27
13	54.53	26.78	27.55	37.34	24.33	24.38
14	44.73	21.74	22.43	26.38	19.12	19.33
15	61.27	26.62	21.47	26.20	27.70	27.40
16	48.21	33.11	33.92	34.63	37.69	44.58
17	71.73	16.28	23.78	22.84	22.79	21.87

18	62.10	47.28	28.62	30.04	35.15	27.23
19	54.18	31.07	30.44	31.04	36.13	39.00
20	41.91	25.59	25.98	26.35	26.52	22.90
MEDIAN	61.68	32.94	33.37	33.15	33.49	36.87

Table C.18: Lateral force applied to the hand support as a function of support height.

	Support height (cm)				
Subject no	92	109	126	143	163
1	14.98	29.05	25.27	33.06	39.15
2	15.13	13.77	12.15	13.33	19.73
3	17.66	18.28	23.06	20.17	18.91
4	35.81	45.83	27.72	20.32	52.22
5	39.11	55.98	26.71	14.20	30.57
6	27.16	16.22	19.53	21.53	16.41
7	16.19	18.99	22.75	28.42	21.59
8	22.34	7.20	26.02	14.54	10.86
9	13.67	11.33	13.87	22.04	15.85
10	12.94	14.56	13.80	17.12	16.19
11	12.79	13.66	20.78	17.20	18.61
12	17.52	25.20	27.65	25.85	12.18
13	11.21	8.66	7.66	12.02	14.54
14	18.31	19.88	33.69	25.06	21.25
15	63.14	62.84	35.52	29.37	34.98
16	29.77	32.44	47.33	57.69	40.36
17	40.07	37.39	43.20	42.77	35.16
18	14.52	27.22	34.11	34.92	32.52
19	5.82	9.25	7.69	7.76	7.97
20	38.24	36.79	39.06	39.08	32.10
MEDIAN	17.59	19.43	25.64	21.79	20.49

C.3. Data used in the analysis of the third experiment reported in Chapter 6

Table C.19: 'Discomfort or difficulty' ratings as a function of peak and r.m.s. magnitudes of 1 Hz lateral acceleration.

	Peak acceleration (ms^{-2})						
Subject no	1.28	1.39	1.51	1.65	1.86	2.04	2.28
1	100	60	75	75	75	300	80
2	100	100	100	60	100	150	130
3	210	100	160	85	114	130	300
4	90	80	140	150	130	160	180
5	55	60	65	30	110	140	140
6	50	50	95	50	85	90	100
7	80	85	90	80	80	110	250
8	120	90	70	80	70	60	80
9	100	110	100	130	120	110	120
10	110	105	110	105	130	120	80
11	80	95	90	125	90	60	90
12	100	100	50	160	120	80	125
13	90	120	100	120	70	120	170
14	110	90	115	140	110	135	100
15	60	60	120	110	120	110	135
16	80	125	80	125	100	100	100
17	80	100	105	115	95	130	170
18	100	120	120	150	140	150	150
19	80	90	120	120	110	110	120
20	80	40	180	70	80	140	130
MEDIAN	90	92.5	100	112.5	105	120	127.5
	r.m.s. acceleration (ms^{-2})						
Subject no	0.58	0.65	0.73	0.82	0.92	1.02	1.14
1	80	150	200	100	400	100	150
2	125	115	180	90	120	130	150
3	90	110	220	105	280	100	200
4	90	160	140	120	200	160	180
5	115	195	150	80	200	105	250
6	75	65	75	60	60	82	110
7	100	100	80	115	110	130	80
8	70	60	70	80	90	140	120
9	100	120	120	110	100	130	145
10	90	80	100	120	110	120	130
11	60	125	90	125	95	100	90
12	70	100	140	75	130	60	175
13	140	80	50	140	100	150	200
14	115	95	120	145	135	115	140
15	90	50	135	80	120	100	110

16	80	100	100	105	125	125	150
17	180	30	100	110	130	125	200
18	110	140	120	100	120	100	120
19	70	70	90	110	80	140	150
20	40	60	90	60	50	80	90
MEDIAN	90	100	110	105	120	117.5	147.5

Table C.20: Reported probability of losing balance as a function of peak and r.m.s. magnitudes of 1 Hz lateral acceleration.

	Peak acceleration (ms^{-2})						
Subject no	1.28	1.39	1.51	1.65	1.86	2.04	2.28
1	25	25	25	25	25	100	25
2	40	40	35	20	40	70	45
3	42	10	15	12	14	24	67
4	20	0	30	60	40	50	70
5	5	10	8	5	20	30	35
6	45	30	42	25	85	65	82
7	25	40	60	45	20	65	95
8	70	60	40	60	50	50	60
9	10	20	10	35	30	15	23
10	40	45	40	40	80	40	30
11	5	15	10	25	10	5	10
12	50	25	25	60	60	40	60
13	25	25	25	50	25	75	25
14	10	5	10	40	15	30	5
15	30	20	90	75	100	80	100
16	0	25	0	25	0	0	25
17	25	30	80	30	60	50	80
18	70	75	80	90	70	70	90
19	40	30	50	50	50	50	70
20	10	5	90	30	40	50	60
MEDIAN	25	25	32.5	37.5	40	50	60
	r.m.s. acceleration (ms^{-2})						
Subject no	0.58	0.65	0.73	0.82	0.92	1.02	1.14
1	25	100	100	25	100	50	50
2	45	40	85	35	40	45	60
3	15	20	45	6	52	10	32
4	10	70	60	30	80	70	70
5	20	70	50	15	70	15	90
6	75	80	70	27	35	45	85
7	45	50	10	60	55	75	45
8	50	40	60	50	70	90	70
9	10	25	30	20	10	35	43
10	70	25	40	50	60	60	70
11	0	20	10	25	15	20	5
12	25	25	50	25	50	25	75

13	25	10	10	50	25	25	50
14	10	10	20	40	35	10	35
15	70	0	100	50	80	70	75
16	0	25	25	25	0	25	25
17	100	25	30	60	50	45	100
18	80	80	60	65	70	80	85
19	30	20	50	50	50	70	80
20	5	10	20	10	5	30	20
MEDIAN	25	25	47.5	32.5	50	45	65

Table C.21: Peak lateral COP velocity as a function of peak magnitude of 1 Hz lateral acceleration.

	Peak acceleration (ms^{-2})						
Subject no	1.28	1.39	1.51	1.65	1.86	2.04	2.28
1	168.86	264.56	227.34	237.31	335.30	129.62	258.18
2	220.33	209.83	282.12	245.78	211.81	276.42	298.73
3	434.09	426.58	368.95	466.05	486.05	537.12	481.29
4	244.81	199.18	245.01	228.87	324.28	256.54	216.41
5	187.60	193.50	211.58	139.16	231.83	187.20	149.65
6	189.64	242.47	133.62	285.81	189.04	282.65	272.15
7	192.58	179.28	220.30	222.58	191.62	372.47	145.34
8	169.99	221.45	230.22	258.40	233.49	201.56	333.29
9	352.73	359.68	405.53	356.44	365.95	462.63	401.94
10	300.07	229.63	236.05	301.92	325.84	262.38	233.27
11	207.95	181.72	182.07	234.45	258.05	200.62	245.06
12	174.52	268.50	182.28	291.72	264.29	223.95	270.99
13	151.13	166.83	160.89	219.63	155.46	263.39	251.32
14	195.94	265.50	251.74	270.78	248.93	150.08	337.74
15	156.93	182.00	161.45	187.19	221.24	213.56	151.14
16	188.40	267.45	222.58	300.83	315.73	255.66	219.41
17	183.03	226.43	166.09	270.42	201.60	171.56	239.17
18	270.32	279.79	256.63	309.35	263.91	377.65	227.94
19	271.55	228.19	277.49	263.10	325.94	332.94	350.30
20	256.74	186.22	303.02	280.39	247.90	315.73	231.22
MEDIAN	194.26	227.31	228.78	266.76	253.49	259.46	248.19

Table C.22: r.m.s. lateral COP velocity as a function of r.m.s. magnitude of 1 Hz lateral acceleration.

	r.m.s. acceleration (ms^{-2})						
Subject no	0.58	0.65	0.73	0.82	0.92	1.02	1.14
1	66.37	44.08	49.54	63.59	61.44	66.29	106.48
2	52.11	56.15	56.41	54.26	52.23	63.88	69.58
3	116.52	111.67	115.25	106.94	92.56	142.41	134.67
4	61.40	68.25	68.25	60.00	71.28	65.72	67.70
5	48.08	56.79	58.21	57.42	57.91	55.71	53.75

6	34.45	42.53	39.37	49.99	62.06	74.37	77.16
7	35.00	41.80	51.45	68.86	73.55	66.28	49.00
8	57.32	51.53	63.47	50.75	58.84	63.82	54.64
9	116.81	123.01	104.32	133.18	128.79	148.49	147.64
10	50.71	59.51	55.30	64.70	80.47	93.87	87.96
11	49.10	46.92	45.40	42.64	48.51	46.20	57.74
12	61.13	56.23	72.95	61.70	68.56	65.15	65.32
13	54.36	49.85	54.18	44.82	45.48	53.45	48.30
14	56.69	54.86	69.35	56.79	74.67	74.39	77.21
15	53.41	48.24	42.56	48.72	46.18	46.35	56.18
16	62.65	67.73	70.42	62.30	58.57	75.55	83.77
17	40.51	49.92	49.26	52.47	43.42	55.99	64.30
18	71.82	59.78	76.73	68.13	69.84	88.33	78.73
19	56.68	71.98	82.37	79.15	63.04	89.39	94.08
20	70.25	54.34	83.87	78.14	82.12	85.19	96.32
MEDIAN	56.68	55.50	60.84	60.85	62.55	66.29	73.37

Table C.23: 'Discomfort or difficulty' ratings as a function of peak and r.m.s. magnitudes of 2 Hz lateral acceleration.

	Peak acceleration (ms^{-2})						
Subject no	2.38	2.57	2.85	3.16	3.51	3.92	4.40
1	100	100	150	140	120	130	130
2	100	100	80	110	110	120	130
3	90	100	100	100	105	150	180
4	90	90	80	130	130	120	120
5	95	120	110	105	150	95	130
6	100	120	100	130	60	95	120
7	100	90	180	70	100	120	110
8	100	60	60	70	110	90	80
9	132	135	120	140	150	150	150
10	110	60	110	120	100	80	110
11	90	100	95	105	85	100	90
12	80	75	60	50	125	80	110
13	100	120	80	100	120	150	180
14	90	110	100	110	105	115	115
15	70	80	60	50	110	120	60
16	80	125	100	100	125	150	125
17	100	100	100	75	95	85	90
18	120	90	120	90	90	100	130
19	80	90	70	80	70	70	110
20	120	80	90	60	70	120	150
MEDIAN	100	100	100	100	107.5	117.5	120
	r.m.s. acceleration (ms^{-2})						
Subject no	1.12	1.26	1.41	1.57	1.76	1.96	2.21
1	90	110	130	200	160	150	200
2	80	90	90	90	110	100	160
3	90	120	120	180	130	205	280

4	80	100	120	140	120	140	140
5	75	90	230	155	190	185	125
6	55	125	75	95	150	150	200
7	75	100	100	80	120	225	280
8	50	60	70	120	110	120	140
9	120	150	160	140	160	155	160
10	80	90	75	120	75	140	130
11	80	100	80	100	110	110	110
12	80	60	75	80	100	110	160
13	50	50	80	120	150	150	200
14	90	95	110	115	130	125	135
15	50	50	100	90	90	80	120
16	100	100	125	100	125	150	125
17	105	110	170	115	160	140	100
18	90	110	80	150	130	120	150
19	50	60	70	120	120	110	80
20	70	80	90	140	170	90	110
MEDIAN	80	97.5	95	120	127.5	140	140

Table C.24: Reported probability of losing balance as a function of peak and r.m.s. magnitudes of 2 Hz lateral acceleration.

	Peak acceleration (ms^{-2})						
Subject no	2.38	2.57	2.85	3.16	3.51	3.92	4.40
1	50	50	75	50	75	75	75
2	60	60	40	55	50	50	55
3	12	30	2	15	17	20	5
4	10	10	0	20	40	30	30
5	15	30	25	15	45	35	25
6	92	95	50	92	60	85	87
7	40	30	90	25	40	75	40
8	30	30	30	40	70	30	40
9	42	45	35	47	58	52	56
10	50	20	40	40	50	25	20
11	13	15	14	17	10	15	13
12	25	50	25	60	40	65	55
13	25	25	10	25	25	35	75
14	5	4	5	6	5	7	5
15	40	50	50	25	50	70	30
16	25	25	25	25	25	50	25
17	35	45	40	45	45	33	35
18	60	50	60	50	60	50	50
19	30	40	20	30	20	30	40
20	40	15	10	5	10	10	50
MEDIAN	32.5	30	27.5	27.5	42.5	35	40
	r.m.s. acceleration (ms^{-2})						

Subject no	1.12	1.26	1.41	1.57	1.76	1.96	2.21
1	25	50	50	100	75	100	100
2	30	40	40	40	60	60	80
3	6	20	45	16	42	10	34
4	10	20	30	50	30	30	60
5	10	20	70	40	60	45	60
6	45	75	90	65	100	100	100
7	30	45	50	30	60	100	100
8	30	40	50	60	70	60	80
9	35	60	57	42	67	57	68
10	25	30	20	40	25	60	50
11	10	15	10	15	20	20	20
12	25	40	50	60	75	50	60
13	0	10	25	25	50	75	100
14	5	5	5	8	10	8	15
15	0	20	50	70	70	50	80
16	25	25	25	25	50	50	25
17	40	50	80	60	75	75	60
18	45	60	60	75	70	70	75
19	20	30	30	60	70	40	50
20	20	20	15	45	60	10	20
MEDIAN	25	30	47.5	43.5	60	53.5	60

Table C.25: Peak lateral COP velocity as a function of peak magnitude of 2 Hz lateral acceleration.

	Peak acceleration (ms^{-2})						
Subject no	2.38	2.57	2.85	3.16	3.51	3.92	4.40
1	234.44	255.38	318.84	352.16	270.00	248.54	261.52
2	255.47	275.45	268.88	283.08	232.01	192.72	226.36
3	338.90	411.03	336.69	447.34	409.03	340.35	347.86
4	201.15	175.59	205.60	252.61	212.56	207.66	232.04
5	159.36	119.58	220.74	171.86	131.63	241.85	152.48
6	142.65	163.37	165.17	169.83	171.72	161.64	164.94
7	102.25	194.04	222.29	112.40	162.87	143.21	133.24
8	212.14	207.42	222.86	154.48	161.04	175.54	181.85
9	438.53	384.03	469.38	449.81	520.49	411.65	464.88
10	245.22	222.96	248.24	270.85	202.06	231.66	190.31
11	166.51	257.63	191.97	172.32	169.35	200.52	172.04
12	222.01	280.79	163.28	260.38	309.68	222.59	228.08
13	268.67	157.36	178.12	144.72	193.46	246.80	220.34
14	193.71	209.74	183.92	218.09	156.98	197.68	194.74
15	199.96	168.36	156.80	150.72	178.26	214.18	145.00
16	183.58	216.55	226.79	239.91	274.49	269.66	210.98
17	136.70	214.92	164.35	185.22	185.06	153.74	137.55
18	263.19	332.11	271.63	252.34	282.92	281.47	218.34
19	247.59	264.70	215.60	281.94	232.13	311.17	336.77
20	177.82	164.34	133.20	204.40	144.78	170.97	210.92
MEDIAN	206.65	215.74	218.17	229.00	197.76	218.39	210.95

Table C.26: r.m.s. lateral COP velocity as a function of r.m.s. magnitude of 2 Hz lateral acceleration.

	r.m.s. acceleration (ms^{-2})						
Subject no	1.12	1.26	1.41	1.57	1.76	1.96	2.21
1	74.15	78.39	81.07	67.14	83.84	78.42	84.50
2	57.50	60.64	53.38	63.77	65.87	66.04	68.71
3	96.03	104.78	133.76	105.14	128.60	112.58	126.25
4	53.91	57.35	61.24	58.39	66.00	67.59	64.04
5	48.64	53.34	49.57	53.77	51.55	38.78	54.57
6	50.48	51.91	40.97	47.73	57.07	52.75	41.57
7	43.70	38.49	42.18	33.49	42.56	41.01	20.17
8	48.07	50.64	48.10	43.73	29.52	42.55	48.50
9	112.74	118.31	134.35	155.80	143.08	148.21	163.88
10	50.81	56.05	57.52	71.72	63.60	70.33	77.68
11	36.40	35.54	35.52	34.72	49.54	53.39	40.23
12	65.10	81.68	82.77	87.32	95.13	81.74	87.36
13	48.09	45.40	48.48	47.46	41.60	60.38	50.25
14	57.40	59.41	65.70	55.55	51.12	62.69	61.02
15	37.17	47.30	47.00	40.33	51.27	45.79	41.19
16	61.67	65.34	53.54	69.89	73.43	83.15	78.49
17	45.01	46.34	57.11	35.58	47.66	31.63	42.70
18	61.60	64.44	68.05	63.96	71.44	71.18	72.62
19	55.35	68.70	71.38	65.73	70.74	73.54	82.86
20	55.38	54.67	56.94	70.36	64.38	68.56	65.98
MEDIAN	54.63	56.70	57.02	61.08	63.99	66.82	65.01

C.4. Data used in the analysis of the fourth experiment reported in Chapter 7

Table C.27: Reported probability of losing balance as a function of number of repetition of 1 Hz oscillation.

1 Hz	Number of repetition of stimuli						
Subject no	1	2	3	4	5	6	7
1	10	5	10	0	15	30	5
2	20	50	30	50	30	50	20
3	10	12	10	12	8	9	5
4	95	80	60	50	50	30	20
5	80	70	75	75	70	50	60
6	40	35	30	20	5	3	10
7	0	10	0	30	30	20	25
8	10	50	40	30	25	25	30
9	26	20	20	23	25	23	20
10	80	30	30	15	25	30	40
11	100	100	73	73	25	73	25
12	100	75	80	65	50	65	50
13	0	10	10	10	10	0	10
14	20	60	10	75	80	50	90
15	50	30	42	37	30	30	25
16	100	95	60	25	10	5	2
17	50	65	55	40	53	55	60
18	25	25	20	10	10	5	10
19	25	40	40	30	45	40	45
20	10	5	10	5	0	0	25
21	10	10	20	20	25	20	20
22	60	20	5	10	5	7	15
23	10	15	10	30	15	25	35
24	20	30	40	40	45	45	40
25	0	20	45	10	15	35	12
26	100	20	20	5	5	5	5
27	30	90	20	22	10	10	5
28	50	25	35	25	35	50	60
29	55	50	40	25	24	24	17
30	15	15	10	10	10	6	10
31	20	65	50	80	85	40	50
32	0	5	15	10	10	10	20
33	30	25	40	50	50	50	60
34	20	21	15	20	20	25	20
35	40	30	25	35	30	35	45
36	20	15	30	20	35	15	15
37	40	25	30	40	55	45	25
38	65	45	40	30	35	20	40
39	80	100	50	30	20	20	25

40	30	30	25	25	25	20	15
41	30	40	55	25	25	40	65
42	70	80	85	75	65	50	45
43	100	25	10	10	0	20	20
44	80	40	50	40	25	20	30
45	20	10	30	20	25	30	40
46	5	5	5	5	10	5	5
47	60	50	30	30	20	20	15
48	20	15	30	18	10	25	5
49	20	25	35	40	40	45	48
50	40	40	30	30	25	20	10
51	10	15	15	30	40	42	45
52	30	20	35	35	20	15	50
53	10	15	15	15	10	15	18
54	80	80	60	70	30	50	90
55	10	20	20	30	30	30	35
56	30	10	40	30	60	55	50
57	25	40	20	40	15	35	40
58	20	25	25	25	10	30	15
59	40	30	40	50	50	50	60
60	40	50	70	65	80	85	70
61	30	20	15	15	30	35	30
62	5	20	20	21	25	23	28
63	70	55	55	60	55	60	50
64	75	50	25	40	10	10	40
65	15	50	50	60	60	55	65
66	40	50	25	30	25	25	30
67	10	5	15	12	20	17	10
68	10	20	10	20	20	15	10
69	5	25	25	25	50	50	50
70	0	10	0	0	10	0	0
71	5	10	10	10	10	20	20
72	20	15	0	10	5	5	5
73	75	75	60	60	30	20	15
74	25	18	22	25	30	28	50
75	20	10	10	5	5	5	10
76	30	34	20	15	15	20	35
77	40	45	35	25	20	22	20
78	25	25	25	25	25	25	25
79	40	30	30	70	40	30	30
80	40	45	30	25	15	15	15
81	10	5	5	8	10	8	15
82	20	20	15	15	25	23	20
83	40	40	40	30	42	25	25
84	45	60	50	75	60	60	75
85	30	35	30	30	45	50	45
86	90	80	50	50	40	40	30
87	20	0	0	0	0	0	5
88	70	50	5	0	0	0	0
89	60	45	45	30	25	50	20
90	70	70	20	20	20	10	15

91	80	20	20	20	35	55	50
92	50	15	20	20	12	25	10
93	30	35	50	50	60	65	70
94	4	25	12	25	35	35	50
95	20	25	25	30	25	28	35
96	80	70	40	30	20	30	15
97	5	5	20	15	25	25	20
98	20	15	15	20	20	10	10
99	80	70	75	70	60	70	50
100	90	95	25	25	25	90	70
MEAN	37.8	35.5	30.14	30.11	28.24	29.61	30.35

Table C.28: Peak-to-peak lateral COP position as a function of number of repetition of 1 Hz oscillation.

1 Hz	Number of repetition of stimuli						
Subject no	1	2	3	4	5	6	7
1	25.09	19.00	21.01	16.84	14.30	17.29	17.53
2	42.30	26.47	29.38	33.92	33.79	27.76	32.02
3	28.70	19.27	30.86	37.90	28.69	29.41	28.20
4	30.50	33.35	27.21	28.28	24.61	28.44	24.77
5	29.81	22.24	26.76	23.19	28.26	22.81	24.87
6	32.39	29.17	21.79	24.58	24.46	32.04	28.64
7	29.48	29.18	24.32	24.36	33.33	24.28	27.61
8	25.03	25.78	20.94	13.43	16.08	22.27	26.06
9	24.20	22.03	30.56	29.52	27.63	31.78	27.85
10	22.14	26.22	24.23	24.62	24.81	24.77	25.02
11	21.51	19.26	29.43	32.00	28.30	28.79	31.45
12	23.54	28.16	32.33	28.77	25.73	22.14	28.36
13	23.10	22.28	21.11	21.22	29.54	27.91	22.26
14	23.83	21.52	16.26	19.15	19.59	16.14	16.68
15	13.98	19.55	18.70	18.72	15.36	16.28	13.22
16	24.05	33.48	26.77	28.79	25.64	28.25	25.08
17	31.09	27.93	31.80	27.19	24.34	29.86	21.70
18	26.13	24.38	22.37	20.93	19.69	20.97	19.91
19	39.64	27.81	34.45	26.81	22.62	34.74	28.12
20	24.87	27.50	20.68	22.87	25.49	18.25	33.63
21	26.31	24.20	23.77	23.30	31.03	18.74	20.25
22	37.64	34.20	33.80	28.49	27.18	32.70	30.99
23	26.48	32.27	28.35	32.73	23.59	25.91	32.90
24	21.34	22.30	22.03	22.37	23.22	28.02	23.96
25	25.00	22.26	31.80	25.49	30.26	32.90	24.14
26	25.03	26.55	33.52	23.71	26.66	21.02	33.26
27	22.37	23.49	25.84	22.80	24.15	22.81	21.23
28	24.49	25.64	23.91	18.70	24.36	20.48	19.72
29	28.10	33.18	18.09	22.66	27.84	22.69	26.21
30	20.80	22.87	21.78	22.43	21.79	22.57	16.44

31	18.26	17.51	19.47	12.35	11.38	17.74	20.06
32	26.12	23.78	19.86	22.49	24.65	22.53	23.68
33	34.58	24.53	26.52	26.27	23.37	21.02	28.32
34	23.25	22.40	16.74	21.05	18.80	22.30	21.75
35	30.67	27.43	30.09	27.77	24.84	27.22	27.70
36	31.81	24.35	28.13	27.23	26.06	28.69	24.89
37	40.26	36.22	29.96	33.30	26.56	31.36	30.02
38	29.79	25.03	25.18	23.00	19.99	18.86	20.73
39	21.31	18.21	32.57	33.58	35.19	37.88	18.72
40	22.14	37.69	37.52	39.50	20.87	37.07	27.84
41	21.25	14.74	16.74	23.22	20.71	22.29	19.80
42	30.29	41.72	33.85	35.82	19.81	31.31	31.32
43	35.99	37.44	34.31	35.18	30.87	32.27	33.97
44	32.58	30.80	26.57	33.23	29.68	32.58	33.10
45	32.80	27.09	28.12	24.45	28.73	26.45	24.98
46	21.96	27.64	26.40	24.12	22.96	22.45	19.40
47	36.49	26.25	32.72	31.18	31.71	33.64	29.46
48	22.59	13.50	22.53	18.04	14.49	14.01	15.13
49	23.05	17.94	22.95	24.06	23.46	19.92	22.79
50	23.91	28.30	30.61	30.25	27.79	28.06	27.43
51	32.17	28.53	29.74	29.68	24.26	24.96	27.36
52	35.26	30.76	30.87	35.47	34.34	31.65	45.47
53	29.08	29.14	26.13	28.26	22.01	24.16	23.73
54	24.90	20.76	20.88	19.95	17.55	22.76	20.81
55	21.66	19.18	20.68	21.43	19.31	17.85	19.38
56	32.34	27.96	28.40	28.05	29.37	24.81	26.76
57	26.94	34.05	26.86	26.42	21.32	21.84	29.14
58	29.91	27.82	14.76	23.12	23.72	16.16	20.90
59	22.36	23.02	26.75	23.51	22.39	26.60	26.05
60	30.31	32.35	33.32	27.35	29.40	36.27	32.01
61	30.29	37.15	31.58	25.89	35.73	24.10	32.86
62	25.13	20.48	17.81	25.23	24.21	17.51	30.32
63	24.47	29.65	30.45	29.36	27.92	28.64	31.55
64	30.26	18.35	28.10	20.87	23.52	24.49	24.03
65	31.56	23.61	20.50	33.72	27.69	22.50	26.79
66	34.21	25.49	22.51	17.32	25.41	31.52	31.50
67	22.81	18.10	17.69	17.32	20.28	15.16	15.43
68	24.53	21.71	20.29	25.72	23.98	23.82	18.76
69	8.11	23.20	20.63	19.40	26.13	20.89	21.41
70	19.50	12.68	25.19	19.78	18.45	19.53	22.56
71	25.27	31.96	30.27	29.83	28.08	34.46	28.63
72	26.12	25.01	24.00	21.97	18.64	18.59	23.07
73	20.42	29.31	35.05	41.29	39.98	35.76	32.04
74	26.78	24.06	26.86	25.15	26.66	22.57	27.30
75	13.87	21.60	14.89	18.63	25.08	15.94	19.07
76	31.26	20.43	18.25	14.42	19.93	20.03	19.70
77	23.63	17.77	25.48	24.65	13.62	23.99	22.17
78	34.33	32.81	33.49	28.48	30.65	29.57	25.32
79	36.81	31.74	30.97	31.82	31.52	31.26	31.80
80	30.91	32.80	34.00	25.40	26.11	33.67	31.32
81	26.73	21.01	21.49	20.04	26.76	23.09	24.28

82	35.52	36.95	35.55	40.11	29.11	39.19	39.00
83	29.09	23.14	20.52	20.94	22.50	23.51	24.71
84	28.68	31.01	32.59	31.23	32.58	30.33	31.35
85	34.62	20.82	27.69	29.51	26.74	34.18	36.23
86	26.19	33.03	30.19	26.47	27.30	29.01	29.60
87	29.49	20.24	28.78	30.28	25.38	28.86	26.66
88	24.50	28.65	24.55	24.78	24.44	22.33	22.89
89	26.86	27.45	25.76	25.44	23.74	18.13	22.30
90	29.23	22.08	18.25	21.98	26.01	21.05	25.24
91	21.97	35.97	28.88	26.19	29.91	27.17	30.52
92	32.53	24.05	25.88	26.71	28.06	32.09	23.48
93	27.52	24.72	27.56	24.90	25.49	28.91	24.55
94	34.39	40.23	35.20	33.93	31.85	32.69	35.12
95	27.57	34.51	25.53	32.33	28.32	34.56	28.66
96	37.18	34.19	36.03	35.19	36.03	35.10	34.91
97	27.82	22.29	23.61	26.23	22.29	22.26	26.56
98	27.46	33.76	26.21	30.11	21.08	31.02	19.26
99	36.73	34.19	33.87	34.24	30.95	35.22	30.53
100	13.30	26.19	25.80	24.38	26.11	31.46	25.06
MEAN	27.16	26.40	26.11	26.16	25.36	26.01	26.01

Table C.29: r.m.s. lateral COP velocity as a function of number of repetition of 1 Hz oscillation.

1 Hz	Number of repetition of stimuli						
Subject no	1	2	3	4	5	6	7
1	34.84	32.11	40.43	31.77	25.15	30.08	18.77
2	101.72	47.62	63.71	87.87	74.02	49.88	73.88
3	82.50	36.63	62.86	78.26	72.38	69.72	59.24
4	50.31	67.71	53.69	62.91	59.26	60.60	52.11
5	42.47	36.34	42.18	35.14	40.62	44.56	32.60
6	88.07	80.99	54.78	68.12	62.63	88.56	82.44
7	50.71	87.99	67.78	55.77	88.02	69.49	77.88
8	74.78	40.93	23.39	24.23	21.60	30.05	32.25
9	36.14	37.67	57.57	58.54	65.53	58.24	49.43
10	47.01	63.96	54.29	59.90	63.26	62.69	59.37
11	54.54	42.60	81.22	84.30	82.41	53.47	80.97
12	35.62	77.77	50.67	80.58	69.98	58.45	69.25
13	41.34	29.77	35.86	31.46	55.04	46.26	41.19
14	48.74	26.88	24.86	29.75	24.42	23.25	28.63
15	22.32	48.26	39.31	41.55	33.86	35.05	26.76
16	55.63	94.47	62.33	85.73	57.88	83.52	71.60
17	68.00	46.19	51.71	57.52	49.12	53.60	39.00
18	67.88	48.67	41.96	41.43	40.00	37.57	38.10
19	121.94	47.25	106.09	73.28	48.06	91.47	66.53
20	42.66	47.54	41.95	39.25	48.08	31.73	64.84
21	73.10	32.34	41.67	33.86	87.16	29.28	33.00
22	94.28	70.94	69.51	62.29	56.04	58.38	60.47
23	43.56	68.59	61.87	56.35	58.66	60.04	70.17
24	48.92	52.46	34.60	35.01	50.13	64.46	62.30

25	45.00	41.14	40.24	48.07	64.36	55.50	56.58
26	46.98	45.37	72.45	50.53	60.77	37.50	64.63
27	44.21	30.48	54.71	30.96	52.95	44.16	35.48
28	44.94	67.90	60.01	39.41	56.17	47.90	41.53
29	59.05	55.08	28.96	41.66	38.44	33.13	35.73
30	28.11	61.60	44.67	56.02	57.49	54.43	32.11
31	42.93	38.76	36.52	19.13	18.54	27.23	39.66
32	61.81	62.15	35.17	57.19	54.35	55.25	53.02
33	68.73	57.82	40.93	48.29	44.65	33.18	55.03
34	55.34	42.70	40.20	40.73	37.82	34.42	46.99
35	50.20	44.12	51.32	59.76	39.60	55.68	48.58
36	92.63	60.14	61.63	78.25	50.88	75.59	63.85
37	108.83	87.96	59.19	58.23	51.24	69.80	76.17
38	79.23	53.61	60.72	58.60	50.92	53.83	53.93
39	38.79	29.89	93.92	84.57	100.88	104.05	35.79
40	40.19	99.00	76.59	108.41	39.74	81.37	49.45
41	39.16	26.76	30.91	53.79	47.63	49.44	37.84
42	59.93	97.47	84.71	71.81	40.94	77.20	72.85
43	65.08	88.36	85.61	71.95	66.35	55.73	67.91
44	66.09	75.38	66.82	59.51	57.97	76.86	53.78
45	54.14	53.33	38.83	43.26	49.89	44.75	42.02
46	35.50	55.00	45.10	45.77	34.70	40.80	37.71
47	76.25	48.44	49.41	66.36	79.46	67.97	71.56
48	38.14	21.03	38.70	30.60	23.89	23.83	26.97
49	56.67	44.53	66.05	60.01	53.44	35.74	47.95
50	41.00	40.76	65.32	62.30	40.61	60.33	57.94
51	59.79	60.82	58.76	53.81	58.44	46.03	58.53
52	62.66	60.14	47.15	77.00	52.39	59.94	83.73
53	70.54	61.64	45.20	49.50	37.60	40.45	40.65
54	61.08	44.73	43.96	32.31	30.70	41.85	30.27
55	54.78	36.70	45.03	47.21	42.21	45.08	39.04
56	81.49	60.21	54.80	55.74	62.07	39.60	42.17
57	58.32	68.13	59.52	35.60	44.84	44.83	46.36
58	74.13	61.26	26.23	59.07	61.47	31.28	57.03
59	58.48	60.93	75.36	58.59	69.00	64.93	68.22
60	52.00	56.56	49.99	47.78	41.77	66.69	61.48
61	65.79	78.15	88.18	65.16	78.71	51.28	65.14
62	59.32	34.19	29.09	56.98	44.97	34.05	53.42
63	38.39	65.40	61.52	68.39	58.90	53.76	62.15
64	78.94	27.92	54.52	42.72	50.41	40.98	57.67
65	76.50	41.69	27.21	56.17	60.65	49.38	48.95
66	58.75	39.86	33.54	24.71	35.04	51.34	66.04
67	50.94	37.22	33.08	23.72	41.42	33.15	31.12
68	52.04	36.55	49.86	41.94	49.35	54.53	35.91
69	16.71	73.99	43.37	41.65	60.96	54.62	64.37
70	40.21	32.31	66.23	34.77	26.91	42.03	51.17
71	46.76	52.33	52.08	46.06	58.33	62.90	47.88
72	79.66	55.97	55.00	53.88	36.56	34.31	44.65
73	36.86	54.99	102.16	93.22	116.14	78.80	72.31
74	69.03	51.38	80.87	73.42	66.13	41.61	46.63
75	24.16	58.77	26.77	40.25	76.83	32.86	29.07

76	55.29	31.56	45.93	30.08	45.06	33.14	49.38
77	61.58	37.83	75.86	73.31	21.89	42.59	64.94
78	79.75	86.66	81.25	53.56	82.48	70.74	69.91
79	84.59	79.69	76.32	69.47	80.33	83.56	80.22
80	51.44	63.37	60.61	47.14	40.16	63.00	44.53
81	52.92	48.29	32.60	27.65	43.98	36.62	34.13
82	88.47	82.42	90.56	80.07	49.27	95.33	94.39
83	85.27	60.94	53.58	58.78	42.22	62.41	62.37
84	59.59	63.75	89.90	89.50	61.50	75.06	68.86
85	92.91	36.16	51.96	59.01	50.74	78.32	65.76
86	64.88	68.67	63.01	45.72	53.75	68.19	65.19
87	66.69	54.01	82.88	70.72	51.42	66.05	43.15
88	54.62	67.01	68.77	64.75	59.67	54.82	61.43
89	66.43	64.66	59.94	55.38	61.13	26.28	38.46
90	68.30	46.75	23.50	59.92	60.69	52.68	55.14
91	27.27	61.05	43.13	49.17	47.17	41.50	52.50
92	41.94	44.47	38.61	54.60	62.42	71.90	41.31
93	72.89	72.57	52.13	73.70	57.64	50.75	71.72
94	106.96	95.77	100.64	92.92	92.18	77.21	64.83
95	47.18	56.81	50.53	56.19	50.79	49.36	63.38
96	102.96	102.27	113.29	109.34	100.23	82.92	103.12
97	74.19	56.87	39.92	70.24	37.05	45.89	63.76
98	73.39	87.70	75.96	71.36	39.69	79.30	46.18
99	86.11	79.30	75.37	87.43	82.63	80.11	89.03
100	24.51	60.98	62.34	66.61	56.71	63.90	55.64
MEAN	59.38	56.16	55.56	56.56	54.40	54.34	54.49

Table C.30: Normalized r.m.s. vertical force applied by the feet as a function of number of repetition of 1 Hz oscillation.

1 Hz	Number of repetition of stimuli						
Subject no	1	2	3	4	5	6	7
1	0.83	0.79	0.83	0.68	1.07	0.86	0.77
2	1.62	1.05	0.79	1.37	1.06	0.78	1.08
3	1.79	1.21	1.26	1.35	1.30	1.40	1.21
4	1.47	0.97	0.96	1.05	1.13	1.04	0.93
5	0.86	0.74	0.79	0.72	0.76	0.74	0.65
6	1.29	1.17	0.98	1.05	0.99	1.41	1.21
7	0.82	1.10	0.96	0.98	0.96	0.94	0.94
8	1.54	0.94	0.75	0.81	0.67	0.71	0.71
9	0.92	0.81	0.95	0.89	1.06	0.88	0.75
10	0.99	0.92	0.84	0.89	0.91	0.88	0.89
11	1.01	0.92	1.17	1.36	1.20	1.07	1.13
12	1.49	1.40	1.33	1.31	1.27	1.65	1.17
13	0.86	0.77	0.79	0.71	0.77	0.72	0.62
14	0.79	0.81	0.80	0.71	0.75	0.73	0.72
15	1.02	1.10	0.78	0.72	0.60	0.63	0.63
16	2.25	1.63	0.85	1.24	0.98	1.19	1.09
17	1.62	1.66	1.14	1.23	1.03	1.19	1.13

18	1.38	1.34	1.04	0.99	1.07	0.94	1.01
19	2.00	1.43	1.73	1.43	1.30	1.60	1.43
20	1.10	1.07	1.06	1.03	1.02	1.04	1.21
21	1.24	0.89	1.33	0.98	1.39	1.06	1.09
22	2.25	1.20	1.05	0.90	0.75	0.75	0.86
23	0.96	0.97	1.11	1.12	0.99	1.03	1.01
24	1.08	1.00	0.88	0.92	1.03	1.22	1.05
25	1.04	0.86	1.10	1.04	1.14	1.31	1.10
26	0.97	0.97	1.07	0.86	1.00	0.84	0.97
27	0.71	1.13	0.91	0.87	0.87	0.78	0.67
28	0.84	1.01	0.92	0.86	0.91	0.98	0.97
29	1.05	1.23	1.01	0.95	0.94	0.93	0.88
30	1.06	1.04	1.00	0.98	1.03	0.98	0.88
31	0.90	1.01	0.92	0.74	0.91	0.85	0.90
32	1.28	1.35	0.99	1.28	1.14	1.19	1.26
33	1.05	0.73	0.76	0.76	0.64	0.69	0.71
34	1.19	0.98	0.98	0.87	0.90	0.85	0.90
35	0.95	0.61	0.67	0.79	0.78	0.78	0.89
36	1.58	1.46	1.50	1.40	1.54	1.53	1.50
37	1.95	1.43	1.29	1.17	1.18	1.39	1.17
38	1.83	1.16	1.13	1.42	1.16	1.14	1.12
39	1.28	1.32	1.31	1.10	1.18	1.24	0.92
40	1.35	1.53	1.08	1.39	0.93	1.07	0.90
41	1.01	0.78	0.90	0.96	0.87	1.09	0.90
42	1.02	1.45	1.01	1.07	1.03	0.88	0.93
43	1.18	0.96	1.07	0.76	0.74	0.64	0.83
44	1.70	1.46	1.27	1.01	1.00	1.28	0.89
45	1.14	0.96	0.94	0.92	0.92	0.90	0.87
46	0.71	0.75	0.85	0.78	0.68	0.82	0.72
47	1.16	1.20	1.06	1.07	1.19	1.14	1.14
48	1.00	0.75	0.81	0.83	0.79	0.81	0.76
49	1.13	0.98	1.20	0.95	1.01	0.70	0.86
50	1.54	1.63	1.55	1.54	1.32	1.30	1.33
51	1.39	1.00	0.94	1.14	1.04	0.81	1.12
52	1.58	1.24	1.39	1.52	1.17	1.12	1.63
53	1.03	1.02	0.79	0.88	0.70	0.70	0.90
54	1.42	1.37	1.23	1.23	1.09	1.22	1.19
55	1.01	0.83	1.01	1.03	0.96	0.93	0.79
56	1.62	1.15	1.19	1.16	1.15	1.17	0.97
57	1.04	1.24	0.91	0.88	0.83	0.93	0.77
58	1.49	1.33	1.01	1.14	1.21	1.02	1.17
59	1.36	1.12	1.24	1.05	1.00	1.10	1.07
60	1.05	1.01	1.05	0.93	0.93	1.12	1.09
61	1.05	0.87	0.97	0.72	0.93	0.78	0.82
62	1.15	0.86	0.86	1.16	1.05	0.88	1.11
63	1.01	1.00	0.98	1.10	0.99	0.96	0.93
64	1.61	0.88	0.96	0.98	0.99	0.90	1.14
65	1.83	1.33	0.99	1.38	1.31	1.21	1.30
66	1.42	0.97	0.80	0.78	0.78	0.88	1.04
67	1.12	1.03	0.93	0.85	0.93	0.91	0.86
68	1.20	1.21	1.09	1.03	1.06	1.15	1.11

69	1.58	2.29	1.99	1.68	1.84	1.83	1.75
70	1.25	1.46	1.44	1.21	1.29	1.18	1.11
71	0.90	0.86	0.98	0.87	0.88	0.88	0.81
72	2.08	1.44	1.22	1.18	0.98	0.91	1.04
73	1.24	1.12	1.25	1.32	1.06	1.08	1.01
74	1.70	1.74	1.75	1.54	1.49	1.47	1.34
75	1.51	1.49	1.15	1.30	1.46	1.16	1.15
76	1.40	0.77	0.82	0.60	0.77	0.63	0.69
77	1.49	1.62	1.55	1.54	1.00	1.08	1.26
78	1.42	1.30	1.41	1.16	1.27	1.18	1.23
79	1.33	1.13	1.12	1.34	1.26	1.22	1.27
80	1.49	1.21	1.11	1.14	0.90	1.13	1.02
81	1.18	0.94	0.70	0.75	0.84	0.71	0.90
82	1.62	1.32	1.42	1.19	1.07	1.32	1.31
83	1.43	1.24	1.19	1.30	1.16	1.24	1.19
84	1.13	0.83	1.01	1.19	0.94	0.96	0.87
85	1.73	1.28	0.97	0.94	1.02	1.22	1.37
86	1.21	1.12	1.00	0.84	0.87	1.06	1.01
87	1.85	1.17	1.38	1.22	1.09	1.24	1.16
88	1.79	1.23	0.96	0.98	0.93	0.70	0.76
89	1.46	1.22	1.20	1.13	1.07	0.92	0.95
90	1.69	1.10	0.87	1.27	1.28	1.21	1.18
91	1.49	1.29	1.19	1.21	1.14	1.24	1.11
92	1.27	0.89	0.88	0.85	0.85	1.02	0.77
93	1.31	1.36	0.93	1.11	0.98	1.09	1.15
94	1.33	1.14	1.28	1.20	1.28	1.09	1.16
95	0.86	0.89	0.91	0.90	0.81	0.76	1.06
96	2.00	1.71	1.60	1.33	1.31	1.51	1.34
97	1.39	1.26	0.97	1.24	0.95	0.86	1.23
98	1.78	1.73	1.76	1.68	1.43	1.62	1.38
99	1.82	1.35	1.40	1.46	1.33	1.38	1.23
100	1.09	1.24	1.08	1.01	1.03	1.26	1.10
MEAN	1.32	1.15	1.08	1.07	1.04	1.05	1.03

Table C.31: Mean COP speed as a function of number of repetition of 1 Hz oscillation.

1 Hz	Number of repetition of stimuli						
Subject no	1	2	3	4	5	6	7
1	68.70	68.09	68.24	64.93	72.25	68.80	70.84
2	97.86	72.39	77.33	88.10	83.46	70.14	81.90
3	81.01	75.42	80.04	83.49	82.31	84.01	80.18
4	68.08	72.65	59.90	62.51	64.57	61.38	58.05
5	72.49	73.54	68.58	68.88	67.71	73.86	64.96
6	83.35	74.20	62.86	70.18	65.42	84.63	78.65

7	59.50	79.14	68.27	59.72	79.84	71.44	75.42
8	78.49	66.61	60.53	60.56	58.25	59.66	57.67
9	68.31	69.48	69.71	73.66	75.84	72.69	69.08
10	59.82	61.14	58.91	64.36	68.15	68.91	65.43
11	62.55	58.96	77.02	81.78	76.43	66.56	74.30
12	89.98	84.94	90.00	85.05	82.56	91.73	80.75
13	67.45	61.43	61.85	59.11	69.41	65.06	63.05
14	65.21	59.37	57.13	57.52	57.04	59.53	58.41
15	67.29	65.89	66.63	62.82	59.03	60.26	61.20
16	74.15	80.70	63.24	78.36	61.46	76.29	75.04
17	75.77	72.83	71.43	66.80	64.55	64.18	58.14
18	74.25	70.47	71.01	67.36	69.05	65.58	68.70
19	112.44	84.59	110.42	89.40	88.25	97.15	91.15
20	70.69	73.35	72.90	67.92	70.94	62.38	75.71
21	92.25	74.34	80.78	75.97	91.51	75.17	74.88
22	94.96	80.06	80.16	74.69	65.41	67.40	72.22
23	74.76	78.03	80.24	76.51	74.41	76.32	80.58
24	66.42	66.94	63.44	61.71	66.50	71.24	68.82
25	73.00	73.61	71.02	66.27	78.64	82.50	66.27
26	71.01	71.84	83.66	74.27	79.31	70.63	77.64
27	67.50	65.70	72.38	68.12	71.77	68.29	67.73
28	64.31	73.64	73.49	66.73	67.79	68.54	68.70
29	66.22	63.08	57.88	52.55	58.07	57.14	60.26
30	69.54	82.44	76.44	80.07	77.53	80.26	74.04
31	63.97	64.30	62.27	55.43	61.43	58.22	59.65
32	85.31	85.04	70.45	81.61	83.92	80.45	79.66
33	75.20	66.69	64.87	69.62	65.97	63.74	69.46
34	74.82	70.72	70.33	65.78	69.45	63.66	71.79
35	67.85	63.20	62.98	65.61	59.13	63.49	66.83
36	95.65	98.19	102.70	92.09	97.48	96.60	93.76
37	104.75	101.19	87.31	82.83	87.15	95.18	95.29
38	87.98	75.10	83.49	81.82	69.47	70.93	72.06
39	88.91	90.00	97.42	94.32	98.47	101.00	83.55
40	81.89	100.49	93.58	107.17	79.89	95.31	81.10
41	63.01	61.28	65.60	67.24	64.77	75.58	68.66
42	60.12	82.65	79.26	70.97	61.05	72.60	71.23
43	78.38	86.87	86.13	78.48	80.45	72.70	75.60
44	81.76	85.11	84.45	78.88	83.56	86.53	76.16
45	85.27	82.70	76.83	74.76	74.48	72.73	71.48
46	62.62	66.82	63.42	62.02	63.78	63.60	59.62
47	84.73	75.69	68.27	74.48	83.64	80.87	82.58
48	64.94	55.60	60.97	60.13	59.06	57.44	56.20
49	71.12	59.86	81.14	75.31	69.73	57.81	66.05
50	73.24	76.13	72.10	80.63	61.02	75.64	71.68
51	73.87	75.38	72.03	73.90	75.17	66.95	73.25
52	86.60	93.59	82.58	88.96	79.78	89.36	99.01
53	79.41	82.58	76.19	77.38	74.13	72.52	75.66
54	76.25	73.32	74.06	68.84	70.65	74.27	71.96
55	64.12	59.24	59.29	58.78	60.46	61.19	56.21
56	85.48	79.38	74.56	76.51	78.49	71.90	67.51
57	81.91	80.56	76.77	71.26	74.49	70.08	72.12

58	80.54	81.18	72.95	75.55	78.35	74.79	77.02
59	78.97	63.92	68.41	60.53	68.14	63.20	66.24
60	71.92	74.42	66.82	66.13	66.21	74.33	71.01
61	76.72	79.17	78.50	69.15	77.81	72.48	74.79
62	68.09	66.61	66.69	66.92	67.34	67.26	66.49
63	66.37	75.41	71.90	77.99	74.26	67.49	72.45
64	82.75	68.70	79.26	71.73	74.16	73.77	75.91
65	91.12	93.97	73.76	80.07	80.30	82.40	82.85
66	77.71	70.15	61.34	67.12	62.64	69.88	76.64
67	62.66	60.05	60.54	60.81	61.23	59.43	59.13
68	79.34	77.78	73.05	72.12	69.37	76.05	65.83
69	66.61	78.49	74.02	75.66	81.76	76.55	75.86
70	70.17	74.29	87.94	71.83	67.15	77.61	68.67
71	66.53	73.08	68.75	67.75	70.11	68.29	70.03
72	83.52	68.38	68.00	65.27	61.89	60.73	68.28
73	75.86	76.99	91.00	97.90	106.51	97.76	90.97
74	91.79	88.62	88.88	80.53	89.08	86.98	85.17
75	72.03	68.35	59.88	69.10	69.27	68.97	63.34
76	77.92	64.46	54.95	56.78	55.74	61.63	63.70
77	81.46	83.38	91.36	97.18	70.03	73.49	84.83
78	80.01	86.02	82.26	70.57	83.77	71.36	73.07
79	92.47	91.23	90.07	85.35	90.50	91.56	89.31
80	72.89	81.19	77.04	67.09	60.29	76.56	66.57
81	70.95	63.66	61.67	60.33	63.46	64.03	62.32
82	90.27	85.20	86.35	83.28	81.62	91.01	92.55
83	86.00	82.58	79.68	79.97	72.44	83.57	79.84
84	70.74	73.05	80.71	78.47	73.00	73.70	69.77
85	85.27	79.33	73.46	80.35	80.28	90.96	88.17
86	92.56	79.20	80.09	75.58	74.91	80.54	76.51
87	94.83	78.73	85.10	79.01	77.38	72.39	64.15
88	85.06	74.92	74.63	73.16	74.70	66.82	72.74
89	76.77	73.56	68.10	69.26	68.07	66.45	62.64
90	95.34	71.43	66.65	68.32	67.81	64.25	68.50
91	84.08	94.93	82.21	89.59	89.96	87.23	88.54
92	80.98	79.44	78.91	79.01	79.71	82.90	73.55
93	71.18	69.23	57.39	67.15	62.57	65.88	66.34
94	101.06	99.44	94.67	98.19	95.81	91.98	77.70
95	76.49	76.86	75.54	75.65	72.23	71.08	86.57
96	90.54	88.27	95.66	90.36	86.79	87.87	88.87
97	75.92	61.74	71.55	67.41	60.66	59.25	67.83
98	94.39	108.39	106.89	93.54	75.74	102.12	73.75
99	91.85	79.45	87.71	93.30	88.26	89.37	91.85
100	71.21	72.90	78.21	79.30	69.33	80.24	78.35
MEAN	77.16	75.85	74.29	73.99	73.35	74.16	73.27

Table C.32: Reported probability of losing balance as a function of number of repetition of 0.5 Hz oscillation.

0.5 Hz	Number of repetition of stimuli		
Subject no	1	2	3
1	80	50	50
2	50	60	60
3	14	16	11
4	40	30	25
5	100	100	90
6	70	45	20
7	60	70	70
8	70	60	50
9	30	35	36
10	70	50	50
11	25	25	0
12	50	50	40
13	20	10	20
14	100	100	100
15	77	80	55
16	30	40	30
17	90	98	70
18	75	50	50
19	60	60	80
20	20	20	50
21	40	15	15
22	50	60	55
23	50	60	75
24	60	60	50
25	15	8	20
26	5	10	10
27	100	60	15
28	75	100	100
29	47	50	45
30	25	30	25
31	90	70	85
32	50	40	30
33	73	80	60
34	90	100	90
35	70	60	60
36	45	25	45
37	60	75	75
38	55	60	60
39	70	80	65
40	75	75	90
41	70	70	75
42	45	40	40
43	100	70	50
44	50	50	60

45	50	60	50
46	50	50	75
47	20	30	25
48	55	60	64
49	55	50	60
50	20	20	10
51	70	80	85
52	55	50	65
53	20	20	20
54	20	22	15
55	45	50	55
56	70	60	75
57	30	15	25
58	70	75	75
59	70	80	80
60	90	85	90
61	40	75	75
62	40	40	50
63	70	70	70
64	55	80	70
65	75	50	70
66	90	100	70
67	35	40	20
68	20	25	20
69	70	70	70
70	50	65	70
71	35	35	40
72	8	15	10
73	60	60	50
74	68	75	86
75	15	20	20
76	60	65	70
77	60	60	58
78	70	70	70
79	80	90	70
80	60	35	50
81	30	50	45
82	40	50	45
83	50	56	56
84	85	85	90
85	70	65	70
86	60	40	40
87	10	0	5
88	85	50	50
89	65	72	50
90	85	85	100
91	70	95	95
92	30	30	10
93	78	80	80
94	100	85	85
95	50	55	75

96	40	40	50
97	60	60	55
98	20	20	20
99	80	75	70
100	95	70	40
MEAN	55.95	55.07	53.61

Table C.33: Peak to peak lateral COP position as a function of number of repetition of 0.5 Hz oscillation.

0.5 Hz	Number of repetition of stimuli		
Subject no	1	2	3
1	33.40	25.12	29.62
2	41.40	35.93	35.07
3	52.32	37.91	29.50
4	29.02	27.29	26.93
5	25.85	32.40	28.78
6	35.69	33.03	28.68
7	35.74	31.04	28.64
8	33.30	27.80	24.27
9	34.96	30.62	30.93
10	29.19	29.87	28.15
11	33.41	34.27	31.35
12	32.62	28.46	27.15
13	31.99	30.97	32.81
14	25.10	17.46	16.25
15	22.07	21.25	22.89
16	36.54	33.16	33.18
17	27.84	38.32	35.65
18	29.94	18.17	27.43
19	42.15	38.94	41.87
20	33.72	31.04	29.80
21	31.83	38.68	35.21
22	37.70	29.91	32.28
23	37.94	32.36	35.39
24	30.27	30.56	26.93
25	31.68	27.48	32.23
26	31.54	30.21	28.51
27	21.87	29.19	26.94
28	35.37	27.07	27.41
29	29.24	31.21	24.82
30	31.88	34.59	31.42
31	22.81	22.97	24.66
32	35.79	30.77	29.01
33	34.40	30.02	27.34
34	30.16	36.16	29.47
35	36.79	36.44	31.90
36	39.04	32.04	31.06
37	31.68	34.00	35.77

38	23.15	30.94	23.14
39	40.24	36.71	34.38
40	50.78	45.16	37.90
41	28.51	29.96	25.21
42	43.71	37.08	41.12
43	40.10	36.19	35.50
44	36.78	36.86	35.61
45	32.03	42.43	38.69
46	36.48	28.64	32.86
47	35.31	34.92	30.08
48	22.89	20.66	18.30
49	28.08	20.18	22.56
50	39.55	32.92	25.79
51	32.24	32.90	27.37
52	36.27	44.61	41.62
53	31.38	28.39	28.07
54	20.73	16.93	23.07
55	20.89	23.05	23.63
56	38.99	32.17	30.41
57	29.16	30.12	25.19
58	32.21	28.20	24.69
59	32.45	27.65	28.80
60	32.31	32.46	35.90
61	37.91	39.90	39.43
62	37.28	29.70	35.46
63	34.00	32.73	31.82
64	28.24	30.56	25.95
65	32.99	24.09	31.58
66	36.90	33.02	35.04
67	21.70	19.18	21.70
68	35.80	30.99	31.75
69	35.01	30.57	27.09
70	33.69	33.67	32.09
71	39.98	36.01	37.01
72	27.92	24.86	24.83
73	39.51	38.53	31.70
74	39.78	38.81	28.34
75	29.12	29.24	28.37
76	30.67	29.59	24.84
77	33.89	26.71	26.49
78	46.31	44.44	43.33
79	42.88	40.82	41.09
80	41.29	36.40	37.07
81	41.51	34.68	30.71
82	47.44	42.36	37.04
83	38.11	38.84	37.78
84	35.66	34.83	36.35
85	40.85	33.13	30.81
86	33.33	25.26	30.21
87	40.27	36.25	35.16
88	39.67	31.93	29.87

89	34.52	31.02	33.19
90	27.35	29.70	30.62
91	36.28	47.42	42.49
92	30.71	35.49	29.14
93	33.48	28.45	31.30
94	45.11	40.94	38.37
95	36.68	34.73	35.02
96	37.89	40.36	42.83
97	44.76	41.77	37.18
98	41.87	42.59	43.63
99	40.25	41.05	40.28
100	32.56	31.71	34.76
MEAN	34.44	32.38	31.31

Table C.34: Lateral r.m.s. COP velocity as a function of number of repetition of 0.5 Hz oscillation.

0.5 Hz	Number of repetition of stimuli		
Subject no	1	2	3
1	52.76	36.59	38.87
2	68.85	59.55	57.57
3	81.43	67.17	61.91
4	47.93	41.06	45.62
5	35.82	38.83	45.75
6	82.98	73.72	68.93
7	71.82	69.84	66.77
8	36.36	35.12	38.24
9	58.07	50.64	56.34
10	74.15	66.82	63.29
11	65.28	67.35	71.60
12	39.38	45.69	46.21
13	54.45	46.96	46.91
14	24.05	22.21	21.06
15	35.34	31.78	36.40
16	80.67	78.82	71.91
17	37.00	43.96	53.04
18	43.30	24.23	35.28
19	86.69	73.49	86.36
20	51.46	55.90	48.63
21	60.19	58.57	66.83
22	56.92	52.03	51.28
23	57.73	53.23	61.89
24	54.97	58.15	58.68
25	50.25	40.43	50.85
26	49.87	49.37	49.41
27	33.87	43.06	37.70
28	51.34	51.23	45.62
29	47.81	40.07	34.31
30	49.33	51.09	45.27

31	33.31	33.37	33.97
32	54.63	47.24	46.42
33	48.17	50.28	46.91
34	57.30	58.38	49.29
35	48.41	40.85	41.35
36	63.82	56.72	61.11
37	45.04	52.23	53.85
38	36.97	45.14	40.32
39	75.53	75.79	63.74
40	55.64	53.64	45.16
41	35.13	50.82	52.17
42	74.66	70.28	68.40
43	69.21	60.46	61.39
44	49.46	49.48	44.85
45	55.10	65.71	65.35
46	47.98	42.49	51.03
47	51.59	45.88	45.56
48	31.32	30.97	22.14
49	42.42	37.56	39.86
50	42.61	37.73	38.17
51	37.67	43.40	34.21
52	57.77	59.63	55.38
53	47.64	40.81	34.58
54	24.05	25.70	30.15
55	34.02	33.53	36.73
56	51.13	49.88	47.21
57	46.39	47.78	47.50
58	51.29	41.51	39.64
59	72.24	70.60	71.07
60	43.61	44.60	46.09
61	76.82	75.30	80.00
62	56.07	51.22	48.72
63	64.65	61.27	57.92
64	47.25	50.77	53.57
65	49.88	35.19	46.03
66	48.35	54.91	49.92
67	28.49	31.95	30.14
68	55.30	55.18	39.63
69	65.30	51.54	57.48
70	50.43	50.24	45.52
71	65.45	62.37	61.81
72	30.35	37.65	34.99
73	87.70	77.87	60.86
74	60.80	46.62	39.95
75	51.19	59.88	53.00
76	45.44	41.79	38.65
77	54.81	47.28	55.36
78	51.22	69.34	60.67
79	78.30	81.50	85.73
80	68.75	53.98	55.17
81	46.97	35.27	32.36

82	81.77	82.32	69.55
83	88.98	93.22	88.74
84	73.67	68.22	76.12
85	51.85	58.24	51.52
86	52.29	47.47	56.20
87	77.80	71.70	66.89
88	65.84	53.41	57.50
89	61.72	50.17	58.12
90	34.66	41.67	38.27
91	52.28	48.83	54.13
92	51.06	52.83	48.73
93	69.48	60.14	63.49
94	82.88	75.07	68.35
95	58.14	63.62	62.38
96	79.50	85.82	85.16
97	70.62	61.20	61.55
98	81.64	80.88	79.83
99	80.93	80.36	81.78
100	46.84	48.88	48.12
MEAN	55.70	53.45	52.80

Table C.35: Normalized vertical r.m.s. force applied by the feet as a function of number of repetitions of 0.5 Hz oscillation,

0.5 Hz	Number of repetition of stimuli		
Subject no	1	2	3
1	1.17	1.04	1.09
2	1.14	1.09	0.98
3	1.70	1.53	1.26
4	0.82	0.76	0.77
5	0.74	0.71	0.78
6	1.36	1.37	1.06
7	1.03	1.07	0.97
8	1.22	1.12	1.13
9	1.14	1.06	0.86
10	1.24	0.89	0.96
11	1.10	1.04	0.85
12	0.91	1.30	1.07
13	0.98	0.74	0.81
14	0.77	0.77	0.81
15	0.91	0.96	0.89
16	1.35	1.25	1.15
17	1.63	1.59	1.08
18	1.20	0.92	1.02
19	1.67	1.62	1.95
20	1.00	1.06	1.12
21	1.61	1.00	1.11

22	0.93	1.01	0.96
23	0.97	1.01	1.02
24	1.19	1.07	1.16
25	0.95	0.83	1.13
26	0.85	0.90	0.92
27	0.65	0.85	0.79
28	1.15	1.01	1.07
29	1.43	1.06	1.03
30	1.06	1.08	0.92
31	0.82	0.80	0.77
32	1.43	1.16	1.14
33	0.80	0.90	0.78
34	1.15	1.26	1.08
35	1.12	0.81	0.80
36	2.08	1.66	2.06
37	1.23	1.17	1.38
38	0.95	1.07	1.02
39	1.16	1.19	1.13
40	1.14	1.34	1.31
41	0.91	0.99	0.94
42	1.01	0.99	0.86
43	1.41	1.00	0.89
44	1.07	0.99	0.89
45	0.99	1.25	1.00
46	0.84	0.80	0.92
47	1.15	1.19	1.06
48	0.86	0.83	0.77
49	0.91	0.85	0.88
50	1.19	1.23	1.16
51	1.32	1.10	1.25
52	1.41	1.70	1.46
53	1.14	0.90	0.91
54	0.93	1.06	1.11
55	0.87	0.80	0.86
56	1.43	1.49	1.21
57	0.84	0.78	0.87
58	1.49	1.36	1.50
59	1.23	1.07	1.14
60	1.28	1.09	1.21
61	0.96	0.93	0.98
62	1.04	1.00	1.06
63	1.00	0.96	0.93
64	1.01	1.07	0.95
65	1.32	1.01	1.14
66	1.15	1.09	0.94
67	0.83	0.90	0.69
68	1.16	1.17	1.16
69	1.82	1.66	1.69
70	1.30	1.18	1.09
71	0.96	1.00	0.98
72	0.93	0.98	0.95

73	1.28	1.14	1.29
74	1.62	1.50	1.27
75	1.39	1.39	1.33
76	0.86	0.77	0.85
77	1.25	1.24	1.34
78	1.24	1.28	1.22
79	1.56	1.43	1.30
80	1.50	1.28	1.29
81	0.86	0.84	0.86
82	1.37	1.47	1.41
83	1.42	1.62	1.53
84	0.95	0.94	1.02
85	1.23	1.20	1.11
86	0.89	0.87	0.90
87	1.42	1.27	1.26
88	1.75	1.25	1.50
89	1.40	1.18	1.11
90	0.94	1.08	1.16
91	1.37	1.24	1.41
92	0.99	0.94	0.85
93	1.14	1.07	1.00
94	1.37	1.11	1.04
95	0.98	0.90	1.00
96	1.31	1.48	1.40
97	1.41	1.06	1.15
98	1.68	1.70	1.61
99	1.36	1.24	1.29
100	1.00	1.08	1.02
MEAN	1.17	1.11	1.09

Table C.36: Mean COP speed as a function of number of repetitions of 0.5 Hz oscillation.

0.5 Hz	Number of repetition of stimuli		
Subject no	1	2	3
1	81.86	75.21	73.10
2	82.45	76.09	77.58
3	88.62	81.25	79.79
4	59.98	52.17	55.55
5	66.75	67.86	69.34
6	80.38	74.85	72.59
7	68.35	69.97	71.84
8	66.32	65.58	64.24
9	72.13	75.26	72.10
10	71.94	68.55	65.24
11	68.40	70.38	73.96
12	75.48	81.39	76.66

13	71.72	66.83	68.70
14	58.95	57.66	54.76
15	63.36	61.46	60.82
16	74.38	74.00	71.53
17	65.19	66.79	68.64
18	72.14	63.75	65.72
19	100.06	96.36	97.92
20	73.92	74.59	73.91
21	86.34	80.62	80.03
22	73.55	76.66	72.30
23	77.46	75.22	78.40
24	67.57	70.93	69.02
25	74.57	64.96	75.55
26	71.48	71.62	74.55
27	62.81	70.02	65.92
28	73.04	67.94	69.49
29	65.70	58.78	57.24
30	79.70	82.73	76.35
31	60.27	58.78	60.01
32	86.79	81.19	80.60
33	69.13	70.53	66.26
34	75.27	75.40	73.58
35	73.69	67.03	71.02
36	101.17	95.27	100.97
37	81.17	81.17	80.93
38	66.90	74.37	71.17
39	94.90	95.42	92.33
40	81.65	84.73	79.82
41	71.79	80.35	75.84
42	69.71	65.64	67.23
43	83.88	84.78	80.83
44	81.31	80.66	80.11
45	82.61	86.32	79.06
46	70.80	68.13	76.42
47	72.52	77.73	70.45
48	64.39	60.14	60.80
49	64.06	62.62	62.85
50	68.08	69.34	67.43
51	68.96	66.84	65.95
52	85.18	83.54	80.21
53	80.98	78.55	77.62
54	65.82	65.52	68.98
55	58.15	52.97	54.26
56	78.50	76.62	75.89
57	71.18	68.76	69.03
58	82.37	77.78	79.47
59	70.79	69.63	67.42
60	68.69	70.54	70.48
61	80.37	79.18	81.31
62	71.89	69.76	69.37
63	74.40	72.50	70.01

64	70.07	74.83	75.35
65	79.75	77.21	79.42
66	74.84	77.05	70.68
67	59.49	62.58	60.13
68	79.76	79.24	73.81
69	76.78	75.27	78.55
70	71.03	70.75	65.98
71	71.50	72.04	70.86
72	65.61	61.92	62.97
73	95.02	86.24	79.96
74	86.74	74.16	75.66
75	70.00	71.16	73.04
76	66.83	65.91	66.28
77	77.88	78.91	81.70
78	73.49	76.95	73.78
79	89.73	92.52	94.56
80	83.42	74.28	78.36
81	64.08	65.49	66.32
82	87.70	87.37	84.47
83	94.42	97.99	98.27
84	77.94	72.88	78.25
85	86.35	89.91	85.14
86	73.83	71.36	73.65
87	79.80	76.68	76.23
88	88.97	78.90	78.24
89	70.18	69.86	74.78
90	69.87	71.80	69.59
91	88.50	85.84	90.21
92	79.07	76.44	76.60
93	72.65	64.86	69.17
94	83.60	84.61	80.42
95	77.66	79.43	79.97
96	78.81	79.96	88.93
97	73.17	69.83	69.10
98	94.22	96.33	90.12
99	81.72	82.63	82.41
100	73.47	75.57	76.26
MEAN	75.60	74.40	74.10

Table C.37: Reported probability of losing balance as a function of number of repetitions of 2 Hz oscillation.

2 Hz	Number of repetition of stimuli		
Subject no	1	2	3
1	80	25	20
2	70	60	60
3	18	16	7
4	70	70	80
5	60	65	50

6	60	15	10
7	75	75	75
8	30	35	20
9	40	40	35
10	95	90	60
11	25	25	0
12	35	30	25
13	10	10	20
14	100	20	60
15	85	85	87
16	90	80	80
17	55	63	30
18	80	80	85
19	70	65	45
20	75	30	30
21	25	25	25
22	50	30	25
23	35	40	50
24	60	62	50
25	70	65	70
26	5	5	5
27	100	80	5
28	100	100	75
29	25	47	35
30	25	10	10
31	30	25	25
32	50	30	30
33	80	85	88
34	70	60	70
35	65	60	45
36	10	20	25
37	20	20	15
38	45	50	30
39	60	60	50
40	30	35	30
41	85	85	90
42	35	30	25
43	50	50	20
44	40	45	55
45	50	20	20
46	50	50	15
47	15	15	15
48	67	68	70
49	60	62	60
50	10	10	10
51	70	75	78
52	15	40	35
53	30	35	30
54	100	90	55
55	60	60	65
56	80	50	50

57	60	50	70
58	78	78	50
59	90	90	90
60	45	40	35
61	60	50	40
62	40	50	50
63	80	75	75
64	90	90	85
65	75	75	65
66	100	90	70
67	10	10	15
68	20	15	20
69	80	80	90
70	30	50	30
71	5	5	10
72	35	35	25
73	60	65	60
74	94	100	36
75	20	15	20
76	85	90	100
77	65	65	68
78	100	100	100
79	90	90	70
80	60	60	60
81	20	30	30
82	35	30	25
83	56	50	50
84	75	80	95
85	50	45	40
86	60	50	50
87	5	10	0
88	30	15	5
89	50	35	45
90	20	25	90
91	90	95	85
92	25	25	20
93	90	90	92
94	25	50	50
95	75	80	70
96	55	50	45
97	75	70	80
98	25	15	15
99	70	60	70
100	20	50	40
MEAN	54.18	50.96	46.61

Table C.38: Peak to peak lateral COP position as a function of number of repetitions of 2 Hz oscillation.

2 Hz	Number of repetition of stimuli		
Subject no	1	2	3
1	18.42	16.55	13.73
2	25.07	25.99	25.57
3	26.47	26.42	25.40
4	24.81	25.84	22.19
5	20.59	20.79	18.58
6	33.09	26.85	26.04
7	31.81	28.28	27.02
8	11.20	16.86	13.94
9	20.49	16.67	21.06
10	28.74	26.14	24.24
11	22.31	30.95	29.61
12	20.36	19.30	20.50
13	25.05	17.94	27.24
14	16.46	17.24	17.18
15	14.30	14.32	14.00
16	34.92	31.31	33.36
17	23.33	15.80	19.43
18	13.88	15.34	12.37
19	36.05	21.89	22.09
20	21.75	24.45	19.86
21	28.71	22.17	25.38
22	26.26	21.20	24.80
23	23.97	26.96	26.06
24	27.43	23.02	25.09
25	29.31	28.10	24.67
26	21.02	18.96	16.20
27	16.63	21.17	19.49
28	17.09	20.24	16.67
29	11.74	18.83	6.91
30	20.56	26.03	21.87
31	15.53	13.83	14.31
32	20.38	19.91	17.43
33	22.96	15.15	24.84
34	24.76	22.45	19.15
35	20.80	23.94	19.31
36	16.01	19.33	16.90
37	23.86	23.69	22.69
38	21.03	18.94	16.25
39	22.59	32.58	31.50
40	30.74	28.54	25.05
41	21.19	16.30	20.06
42	23.61	29.80	27.12
43	26.23	29.84	33.59
44	19.85	25.25	30.89
45	21.29	29.52	25.72

46	25.44	23.13	17.82
47	21.17	12.95	14.88
48	14.65	10.77	17.32
49	16.51	16.73	17.72
50	26.88	22.42	22.76
51	19.87	21.73	26.01
52	16.13	30.21	19.94
53	19.99	20.81	15.91
54	20.21	13.65	16.39
55	20.30	12.68	20.09
56	18.26	22.10	26.24
57	29.86	25.07	19.71
58	14.45	10.26	16.30
59	31.68	33.06	32.26
60	20.21	24.45	23.05
61	31.20	29.68	27.64
62	18.87	26.60	19.94
63	25.76	24.39	29.24
64	25.34	22.35	24.87
65	24.77	22.75	18.82
66	34.55	23.78	27.70
67	17.36	20.72	13.56
68	20.15	20.21	22.09
69	24.60	14.86	23.09
70	12.63	19.19	15.04
71	15.15	13.74	25.35
72	20.76	23.39	16.23
73	30.92	26.98	35.27
74	18.17	12.04	24.10
75	20.45	19.84	25.29
76	20.34	23.35	17.45
77	18.69	18.23	13.18
78	18.69	28.79	18.27
79	32.05	27.27	31.57
80	24.54	28.55	28.02
81	23.28	17.90	16.70
82	27.61	22.85	32.27
83	28.21	31.73	28.15
84	29.14	37.24	34.80
85	32.04	26.79	30.22
86	18.35	29.56	22.99
87	20.68	32.72	18.79
88	20.97	20.03	21.58
89	19.25	24.30	14.12
90	25.41	16.11	18.79
91	30.43	28.00	28.02
92	21.59	22.30	22.60
93	21.70	24.54	20.88
94	29.86	24.65	24.64
95	33.07	31.36	28.30
96	31.61	29.85	30.43

97	23.76	26.32	28.97
98	28.42	21.88	19.94
99	35.27	34.09	31.20
100	22.59	19.91	27.71
MEAN	23.16	22.84	22.50

Table C.39:Lateral r.m.s. COP velocity as a function of number of repetition of 2 Hz oscillation.

2 Hz	Number of repetition of stimuli		
Subject no	1	2	3
1	30.34	38.70	22.26
2	57.06	70.39	64.29
3	72.17	65.41	72.82
4	61.02	49.11	46.33
5	46.11	44.62	42.17
6	120.38	76.57	83.93
7	66.97	84.10	64.26
8	22.11	29.91	26.45
9	57.96	42.27	60.32
10	115.30	84.38	63.72
11	48.31	92.49	84.48
12	51.27	43.44	44.10
13	57.75	45.89	59.11
14	23.95	32.61	23.30
15	40.44	34.82	33.47
16	109.70	93.60	95.40
17	40.69	36.85	37.73
18	33.24	36.22	20.19
19	94.79	65.99	64.58
20	39.67	53.66	50.43
21	67.99	60.54	45.36
22	62.60	50.32	57.73
23	54.61	67.65	65.38
24	102.61	67.26	63.53
25	63.13	60.21	49.36
26	46.79	43.39	36.90
27	41.01	45.89	39.47
28	39.12	51.17	44.46
29	15.21	22.77	14.38
30	45.31	52.84	49.86
31	38.29	32.97	33.58
32	43.33	58.32	41.77
33	42.90	36.65	55.93
34	83.46	42.82	45.06
35	44.56	43.22	33.27
36	41.97	51.70	36.81
37	46.96	42.75	39.22

38	52.29	45.62	35.25
39	77.86	73.23	88.26
40	63.13	64.54	53.94
41	55.41	42.42	52.58
42	63.13	70.94	59.28
43	62.85	67.85	77.81
44	48.48	66.30	61.90
45	36.72	59.13	58.24
46	60.50	48.82	39.69
47	54.21	27.75	26.12
48	26.93	19.03	37.81
49	41.25	37.52	36.93
50	63.37	47.77	47.12
51	42.68	64.28	58.96
52	33.29	66.82	38.50
53	29.33	40.22	35.18
54	37.12	28.88	29.57
55	46.95	31.89	42.98
56	47.42	54.93	59.83
57	70.89	52.32	49.71
58	28.53	22.29	39.41
59	120.07	107.79	107.13
60	44.45	53.80	50.26
61	86.29	77.33	61.03
62	45.31	53.63	47.56
63	66.78	63.83	70.27
64	66.67	50.99	54.57
65	76.60	68.25	45.51
66	97.93	57.32	56.89
67	42.77	33.42	21.52
68	37.65	42.39	47.25
69	58.97	41.01	55.19
70	25.73	39.54	33.35
71	27.09	23.73	51.79
72	44.26	46.74	35.97
73	89.01	81.40	129.31
74	45.82	31.07	57.00
75	45.14	55.39	66.59
76	55.84	46.03	48.99
77	58.11	46.88	37.86
78	44.80	71.83	49.44
79	97.01	80.45	87.45
80	52.26	60.04	57.66
81	43.27	34.38	33.09
82	67.88	49.74	70.99
83	85.73	86.10	84.94
84	75.21	87.36	85.51
85	70.58	64.12	83.78
86	50.26	67.14	43.48
87	66.82	61.93	55.32
88	56.36	50.76	60.98

89	40.65	64.79	43.91
90	63.69	34.43	34.97
91	67.27	53.51	52.44
92	53.74	51.96	60.22
93	63.63	76.50	55.83
94	76.74	60.16	55.58
95	72.51	75.73	75.07
96	101.90	81.91	80.72
97	67.05	58.74	57.93
98	75.18	73.92	57.24
99	102.40	102.79	87.17
100	63.69	50.46	49.75
MEAN	23.16	22.84	22.50

Table C.40: Normalized vertical r.m.s. force applied by the feet as a function of number of repetitions of 2 Hz oscillation.

2 Hz	Number of repetition of stimuli		
Subject no	1	2	3
1	1.10	0.94	0.86
2	1.93	1.13	1.09
3	2.24	1.69	1.48
4	0.99	0.99	0.75
5	0.80	0.80	0.74
6	1.57	1.35	1.61
7	1.20	1.23	1.22
8	0.90	1.18	0.87
9	1.34	0.89	1.09
10	1.53	1.34	1.04
11	1.40	1.59	1.30
12	1.43	1.29	0.99
13	0.92	0.90	0.96
14	0.92	0.84	0.71
15	1.75	1.08	1.22
16	1.85	1.91	1.44
17	1.34	1.28	1.13
18	1.44	1.24	1.01
19	1.58	1.27	1.00
20	1.36	1.08	1.13
21	1.51	1.20	0.82
22	1.22	1.03	1.14
23	1.01	1.19	1.14
24	1.98	1.32	1.27
25	1.41	1.57	1.39
26	0.93	0.94	0.93
27	0.86	0.93	0.83
28	1.12	1.18	1.13
29	0.97	0.99	1.03

30	1.12	1.03	0.95
31	0.93	0.91	0.80
32	1.25	1.26	1.03
33	0.98	0.88	0.90
34	1.67	1.14	1.23
35	0.82	0.77	0.73
36	1.74	2.14	1.49
37	1.35	1.07	0.90
38	1.40	1.21	0.88
39	1.27	1.33	0.99
40	0.96	0.98	0.94
41	1.18	0.89	1.05
42	1.01	1.02	0.93
43	0.97	1.07	1.09
44	0.85	0.98	1.09
45	1.03	0.94	1.01
46	0.86	0.94	0.78
47	0.86	1.05	0.92
48	1.02	0.85	0.98
49	1.06	0.95	0.74
50	1.41	1.41	1.22
51	1.15	1.54	1.40
52	1.07	1.21	1.19
53	1.03	1.24	0.93
54	1.51	1.16	1.29
55	1.33	1.07	1.12
56	1.31	1.14	1.31
57	1.00	0.90	0.90
58	1.54	1.42	1.30
59	1.73	1.63	1.51
60	1.10	1.30	1.19
61	1.23	1.07	1.11
62	1.21	1.18	1.19
63	0.92	0.96	1.04
64	1.30	1.08	1.06
65	2.40	2.22	1.90
66	1.25	1.03	1.07
67	1.05	0.92	0.73
68	1.05	1.32	1.00
69	2.15	1.82	1.38
70	0.99	1.26	1.10
71	0.78	0.80	0.84
72	1.41	1.34	0.99
73	1.36	1.58	1.35
74	1.60	1.46	1.56
75	1.54	1.51	1.56
76	1.35	1.10	1.00
77	1.46	1.35	1.10
78	1.20	1.43	1.16
79	1.69	1.36	1.42
80	1.54	1.50	1.33

81	1.07	0.76	0.86
82	1.43	1.22	1.41
83	1.69	1.53	1.55
84	0.88	1.09	1.35
85	1.53	1.34	1.69
86	1.19	1.12	1.02
87	1.54	1.38	1.18
88	1.41	1.29	1.17
89	1.59	1.25	1.18
90	1.60	1.01	1.02
91	1.83	1.52	1.11
92	1.00	0.86	0.93
93	1.67	1.69	1.25
94	1.03	0.88	0.83
95	1.33	1.44	1.06
96	1.36	1.38	1.66
97	1.39	1.20	1.21
98	1.72	1.73	1.50
99	1.72	1.51	1.33
100	1.26	1.39	0.99
MEAN	1.31	1.22	1.12

Table C.41: Mean COP speed as a function of number of repetitions of 2 Hz oscillation.

2 Hz	Number of repetition of stimuli		
Subject no	1	2	3
1	78.18	71.41	73.34
2	83.67	81.68	86.72
3	85.12	81.60	84.32
4	65.96	70.60	58.61
5	77.09	72.34	67.50
6	107.77	86.99	88.23
7	70.46	80.62	67.16
8	61.15	60.31	59.45
9	77.08	66.31	70.36
10	99.85	81.87	65.22
11	63.49	82.13	94.81
12	82.53	81.11	73.66
13	68.28	62.33	66.20
14	64.98	55.02	58.15
15	65.40	68.85	68.36
16	103.78	79.67	87.09
17	63.65	58.32	56.34
18	81.52	74.58	73.10
19	93.50	73.81	72.15
20	69.36	72.43	70.93

21	84.63	81.64	73.18
22	79.93	71.80	78.72
23	75.41	82.95	84.79
24	94.02	76.44	77.24
25	84.16	83.27	84.16
26	76.54	70.93	69.39
27	66.35	69.50	63.10
28	64.33	69.91	71.91
29	55.08	63.80	58.68
30	77.05	74.77	77.53
31	67.40	60.75	60.48
32	80.31	85.11	77.46
33	70.12	72.99	75.60
34	90.70	76.01	76.58
35	69.90	71.33	63.45
36	92.27	102.15	89.32
37	80.34	77.03	74.35
38	77.49	68.41	70.71
39	94.09	88.91	98.49
40	81.17	80.27	81.48
41	77.17	68.85	74.65
42	69.45	75.78	68.52
43	93.34	80.16	79.94
44	81.74	87.57	76.70
45	79.14	82.28	80.64
46	74.99	74.68	68.97
47	67.90	74.87	69.49
48	74.67	63.58	67.91
49	67.95	58.09	67.83
50	74.51	61.01	72.02
51	67.12	95.95	76.35
52	78.57	90.12	80.25
53	73.51	70.46	72.89
54	76.51	77.51	69.24
55	59.70	54.49	57.17
56	74.39	83.42	73.26
57	72.20	68.89	72.64
58	78.37	77.52	79.67
59	104.57	88.76	86.66
60	69.41	72.14	64.92
61	85.10	76.02	71.00
62	67.45	71.90	66.06
63	71.06	77.73	76.04
64	74.59	73.24	72.19
65	101.89	88.43	79.94
66	87.74	83.07	77.64
67	70.79	58.66	60.80
68	73.79	74.63	72.96
69	79.78	71.68	70.05
70	61.10	66.52	65.57
71	66.17	64.34	63.54

72	83.05	68.54	62.50
73	96.98	87.00	113.81
74	90.14	88.16	80.30
75	68.60	74.60	75.33
76	74.97	70.22	77.28
77	77.92	69.69	81.91
78	72.41	79.33	78.66
79	99.22	91.63	90.35
80	79.18	80.47	80.19
81	63.56	70.15	57.85
82	83.85	79.63	84.34
83	98.79	93.37	97.48
84	78.91	73.25	85.80
85	84.51	83.08	83.86
86	77.03	78.89	68.18
87	82.85	75.28	73.52
88	78.00	73.68	78.37
89	75.44	81.57	87.78
90	85.34	68.22	70.50
91	88.39	99.14	81.60
92	85.01	71.77	79.58
93	82.69	81.03	74.10
94	83.04	79.40	75.11
95	81.93	93.48	88.81
96	90.51	75.84	86.04
97	72.83	69.45	67.51
98	85.59	96.96	82.94
99	98.96	91.93	88.51
100	83.62	91.72	78.78
MEAN	78.82	76.40	75.15

Table C.42: Repeated measures during normal walking without oscillation: peak to peak lateral COP position.

Subject no	1	2	3	4	5	6	7
1	17.56	18.65	16.35	13.52	15.37	14.61	13.83
2	22.61	23.74	22.67	23.80	19.93	18.64	19.04
3	22.69	21.19	23.07	24.22	19.33	22.21	22.85
4	15.05	14.69	14.28	15.53	14.05	15.46	13.96
5	15.90	15.90	15.76	16.76	17.15	16.87	15.21
6	19.92	19.80	20.25	20.04	18.97	23.13	22.75
7	18.16	17.20	23.49	17.35	15.69	17.14	19.19
8	13.38	22.79	15.22	16.64	19.27	16.62	21.85
9	17.00	18.05	18.25	14.69	20.29	15.47	19.70
10	21.42	19.78	22.46	22.61	22.48	22.30	18.13
11	14.90	14.03	19.96	15.34	15.08	14.03	15.08
12	20.84	18.98	19.25	19.51	19.82	26.74	21.56
13	14.56	11.76	13.44	15.76	14.29	13.60	13.28
14	11.97	13.17	10.97	10.37	12.69	11.96	13.83

15	12.86	12.17	11.76	12.07	13.76	11.20	12.04
16	23.49	27.88	24.97	27.78	22.83	25.99	26.34
17	19.54	17.16	17.72	17.30	14.37	15.90	17.90
18	9.81	10.78	10.46	8.16	8.64	9.30	8.15
19	23.28	19.97	22.56	23.01	21.74	24.06	24.57
20	20.95	18.45	19.49	17.72	19.29	18.26	20.45
21	18.30	12.74	14.83	13.94	16.67	13.61	14.14
22	24.18	26.87	27.35	26.16	24.51	23.28	25.53
23	18.47	18.90	18.76	16.43	19.79	20.14	17.63
24	18.48	20.81	19.46	21.83	17.71	17.87	19.09
25	13.80	16.47	17.44	14.43	11.69	11.27	12.34
26	14.14	14.44	18.82	18.90	18.00	14.92	16.11
27	13.01	15.12	14.01	15.31	16.64	15.62	16.89
28	17.38	20.28	15.77	17.96	16.34	18.51	18.04
29	19.77	18.01	17.84	19.98	20.15	18.97	20.40
30	16.44	14.56	14.94	13.90	16.35	15.51	16.00
31	11.61	12.84	11.92	9.56	12.43	12.40	12.00
32	20.51	17.24	17.51	18.57	16.92	17.05	18.22
33	15.37	17.32	16.33	18.83	16.03	17.72	19.30
34	17.59	16.91	17.96	18.00	17.79	18.94	18.54
35	24.36	20.74	22.90	26.79	24.03	21.56	22.95
36	17.52	17.99	18.94	18.69	21.81	19.39	17.48
37	24.12	23.34	24.11	23.64	24.80	26.44	24.86
38	14.43	17.38	14.48	14.53	14.95	18.05	14.79
39	20.40	22.34	19.67	22.21	19.11	19.75	19.69
40	16.94	18.31	19.67	18.13	19.63	15.45	18.07
41	15.43	14.25	16.62	14.02	14.48	16.15	16.16
42	17.66	18.21	16.75	18.34	18.31	16.74	19.99
43	28.39	26.65	25.97	30.12	28.75	27.12	25.84
44	26.46	26.84	26.09	23.09	19.40	18.70	18.85
45	18.61	21.32	20.41	19.56	19.76	18.39	22.18
46	14.97	14.95	16.93	16.04	16.06	14.48	13.96
47	25.60	24.49	25.35	23.36	21.00	24.51	25.81
48	10.30	13.04	14.17	13.82	12.64	14.54	13.96
49	16.42	16.43	15.10	14.31	18.31	16.79	16.82
50	19.14	19.20	19.38	16.19	20.25	21.24	20.03
51	23.55	25.03	21.87	21.20	24.69	24.13	22.45
52	25.73	21.90	22.99	29.43	36.53	19.81	27.19
53	12.25	11.29	9.61	11.51	10.09	12.95	11.51
54	8.29	9.55	10.73	9.79	10.16	10.69	11.70
55	15.65	14.85	13.38	13.82	15.14	15.48	14.44
56	23.08	21.05	20.62	20.69	17.43	18.55	16.89
57	11.84	13.83	13.76	13.12	11.96	12.29	12.33
58	20.20	20.05	16.10	17.48	18.75	16.69	16.84
59	17.84	18.92	20.03	17.40	19.76	17.18	18.10
60	14.61	14.93	14.31	16.19	16.52	13.78	17.10
61	17.27	13.51	16.59	16.83	15.79	15.77	14.24
62	20.00	19.70	20.10	17.43	15.98	16.50	16.95
63	22.26	19.33	22.40	18.32	22.86	20.25	19.44
64	15.96	15.37	15.04	16.81	15.55	17.75	17.86
65	9.61	11.73	11.47	11.60	10.02	10.62	11.90

66	18.16	18.48	18.86	22.00	22.14	22.18	23.51
67	13.50	12.81	15.24	10.14	10.73	13.22	11.64
68	22.58	22.48	20.89	21.25	21.37	20.33	23.01
69	14.01	16.29	12.55	13.37	12.97	13.68	15.75
70	15.39	14.64	14.12	10.36	13.56	14.53	14.54
71	15.90	14.92	16.92	18.06	15.02	16.22	16.13
72	15.51	12.97	13.44	12.32	12.66	13.67	11.98
73	17.35	20.77	19.15	15.67	16.87	18.59	18.19
74	12.19	10.15	15.28	15.57	14.52	15.58	13.23
75	13.63	11.90	11.42	10.59	11.61	10.80	10.79
76	16.22	16.22	15.43	18.13	16.60	18.09	21.19
77	13.76	13.69	14.44	12.95	14.84	12.25	15.79
78	13.90	16.67	17.10	15.68	20.08	18.57	18.55
79	30.19	27.93	28.96	30.57	31.50	30.04	29.53
80	17.33	18.68	22.20	17.87	17.00	20.10	20.02
81	10.17	8.62	8.77	8.48	9.55	8.59	8.92
82	17.12	17.50	19.33	17.98	17.31	17.94	19.37
83	23.68	23.70	24.10	22.63	22.26	21.42	19.36
84	20.61	22.35	24.86	25.13	22.82	24.44	22.21
85	22.09	20.49	19.56	17.97	17.96	20.16	22.00
86	23.87	21.31	23.05	25.44	26.37	22.67	23.87
87	14.79	16.34	14.58	14.39	15.34	16.44	13.23
88	13.39	12.27	12.45	14.35	13.30	14.01	14.87
89	11.36	11.99	11.43	10.90	10.73	13.18	11.72
90	16.72	19.11	16.23	15.51	16.56	16.47	16.82
91	19.40	17.62	17.07	17.57	17.44	17.82	18.35
92	19.42	22.50	18.33	21.04	24.10	21.13	18.38
93	21.92	22.85	19.11	20.14	21.16	18.59	18.80
94	21.41	19.23	23.82	22.06	21.71	23.22	21.65
95	23.38	22.95	23.55	23.27	23.09	23.70	25.06
96	17.62	17.66	18.86	19.86	17.34	19.91	18.66
97	18.09	17.81	17.33	18.53	21.77	20.54	20.09
98	21.49	22.02	22.27	23.24	23.93	21.62	23.09
99	24.11	26.08	27.20	25.90	28.16	30.50	28.92
100	17.33	19.46	19.98	23.71	18.91	21.19	17.06
MEAN	17.91	17.96	18.09	17.93	18.02	17.92	18.11

Table C.43: Repeated measures during normal walking without oscillation: lateral r.m.s. COP velocity.

Subject no	1	2	3	4	5	6	7
1	31.10	31.01	26.36	27.13	28.14	23.55	26.62
2	44.92	43.37	45.82	43.51	38.81	38.41	36.64
3	39.92	39.49	41.63	44.15	35.92	41.04	43.37
4	24.76	24.43	23.93	25.70	24.14	24.72	23.49
5	29.07	28.12	28.93	29.19	27.82	29.59	27.74
6	41.55	42.72	45.81	40.17	40.70	48.05	42.92
7	42.77	40.94	42.76	41.90	40.66	41.12	43.07
8	20.86	36.06	24.66	26.50	32.84	31.90	32.35

9	31.39	29.53	34.23	26.54	30.25	25.93	35.53
10	41.84	40.54	44.90	47.35	45.06	43.40	35.57
11	33.62	28.47	36.94	31.32	31.02	29.48	31.19
12	39.12	34.57	35.71	34.99	38.67	39.66	40.38
13	20.73	20.53	19.88	22.72	20.69	21.91	19.71
14	20.87	17.86	17.44	18.95	18.29	19.54	19.07
15	21.81	20.89	20.45	21.20	22.38	22.36	22.56
16	50.84	60.46	62.33	59.50	49.86	58.07	56.91
17	39.91	37.02	38.91	32.40	31.88	31.29	34.66
18	15.26	14.18	15.87	10.34	11.73	14.08	11.62
19	59.83	51.86	52.31	53.12	50.84	55.46	59.05
20	40.51	37.97	40.54	36.15	43.63	39.74	41.49
21	27.28	22.33	24.08	27.22	29.55	23.89	26.38
22	38.01	49.15	47.53	43.23	41.97	39.26	39.46
23	36.50	34.52	36.47	31.32	37.36	36.33	34.70
24	36.35	38.89	40.82	38.42	38.05	35.31	37.44
25	24.58	28.94	24.43	22.59	18.51	16.51	17.29
26	27.92	27.55	34.27	33.04	31.07	31.76	30.01
27	25.31	24.67	23.89	26.66	27.32	28.77	27.74
28	38.62	39.98	34.67	36.44	34.99	38.19	38.08
29	28.18	27.12	28.85	30.88	29.83	27.37	30.71
30	29.75	29.13	30.85	29.52	32.26	33.72	31.16
31	24.02	24.45	23.44	21.87	23.10	24.26	24.44
32	41.58	35.37	35.40	36.87	34.38	34.29	35.38
33	31.87	32.59	34.37	33.30	31.48	32.55	35.37
34	31.25	34.60	34.20	34.21	36.41	34.34	33.18
35	49.14	40.13	42.02	43.20	40.87	40.84	39.11
36	40.12	43.52	44.75	46.25	47.37	47.36	44.42
37	45.15	42.49	44.82	42.15	47.22	50.12	47.34
38	24.40	29.98	27.15	30.65	28.78	35.53	29.98
39	41.01	43.25	41.70	41.46	40.98	42.99	42.27
40	26.50	27.55	27.81	26.37	30.79	29.12	29.43
41	27.70	26.00	27.85	26.18	25.83	31.75	32.03
42	31.96	34.20	29.78	31.69	32.79	32.33	34.29
43	57.30	57.65	58.17	57.64	52.76	44.14	48.45
44	41.77	44.73	41.03	39.11	33.41	34.81	29.08
45	37.06	34.76	36.87	33.02	33.46	34.75	39.01
46	28.45	27.01	31.20	27.80	29.15	26.37	28.45
47	41.40	42.65	39.26	39.69	34.57	46.00	33.63
48	16.56	17.26	21.10	22.55	20.34	25.12	22.42
49	31.93	35.32	33.35	33.19	34.49	38.33	37.77
50	39.12	37.20	37.94	32.44	37.02	37.04	34.56
51	35.38	37.93	36.18	38.02	38.49	38.93	35.92
52	40.18	38.67	36.75	46.05	36.29	31.73	41.62
53	20.74	19.47	18.52	19.96	20.07	20.54	23.42
54	16.76	19.01	19.91	18.38	20.90	19.60	23.19
55	30.70	28.61	28.85	29.17	29.66	29.04	29.25
56	35.66	34.89	34.54	33.13	30.97	32.56	29.81
57	22.58	23.79	22.79	24.77	25.07	25.03	24.24
58	42.32	38.41	37.91	40.53	39.36	36.69	34.62
59	46.87	45.04	45.74	42.37	44.71	42.49	39.87

60	26.19	26.29	26.66	26.47	30.80	27.86	28.88
61	28.26	26.66	32.79	32.50	32.65	31.95	27.33
62	37.86	35.02	36.35	33.33	31.31	33.79	31.99
63	41.12	40.33	45.39	40.31	43.91	43.25	42.22
64	32.15	31.24	28.19	35.22	31.37	34.25	37.13
65	16.67	17.01	16.22	17.47	16.60	15.72	19.07
66	29.12	31.03	31.58	36.68	33.38	32.94	33.94
67	23.18	23.62	24.60	18.46	20.77	21.90	19.74
68	47.76	43.53	40.60	42.38	41.40	42.33	43.58
69	28.28	29.53	26.51	28.83	26.99	25.96	31.12
70	24.64	29.47	26.51	20.12	24.04	25.10	28.03
71	30.02	26.20	30.02	28.63	26.95	31.25	29.75
72	22.42	21.43	22.46	22.58	20.56	26.12	22.74
73	29.79	32.26	38.90	33.43	31.65	35.12	35.07
74	17.97	14.35	21.64	20.42	23.53	24.18	22.94
75	23.79	24.33	24.76	25.11	23.89	26.12	25.55
76	31.28	28.57	30.43	34.35	38.53	36.27	39.40
77	29.09	28.00	27.29	27.55	26.62	26.09	25.15
78	29.33	33.70	36.50	32.96	42.73	41.69	38.04
79	64.77	60.70	59.79	63.68	61.94	63.44	63.95
80	34.28	32.43	37.20	34.16	34.93	33.34	34.39
81	16.73	13.33	12.54	13.42	15.74	15.39	14.10
82	32.00	32.03	31.17	30.31	30.18	33.31	32.02
83	59.23	58.28	59.87	56.81	61.01	57.67	51.25
84	49.02	50.91	54.09	55.93	58.09	62.48	46.31
85	33.42	33.27	33.43	26.27	31.16	33.40	43.46
86	46.64	45.43	50.69	50.44	50.77	48.23	50.28
87	31.46	31.37	31.18	29.53	32.49	32.02	32.50
88	26.50	26.13	27.75	30.76	27.90	28.21	28.65
89	19.33	18.59	17.42	17.99	19.02	19.79	18.70
90	33.73	33.57	36.15	32.14	34.23	34.02	35.79
91	30.24	30.87	28.12	29.07	28.27	27.15	27.56
92	36.83	39.99	35.94	34.98	38.98	36.03	34.87
93	50.06	48.66	46.07	45.48	45.02	41.52	42.97
94	50.80	47.42	50.72	54.16	49.60	48.82	54.74
95	46.06	46.68	45.04	48.30	48.52	48.53	47.96
96	42.66	42.80	46.17	45.63	39.86	45.65	43.76
97	37.97	38.52	40.87	40.67	43.75	46.40	39.39
98	56.10	53.78	60.55	63.73	62.30	60.69	59.56
99	57.89	60.15	59.20	62.31	66.16	67.58	68.76
100	32.75	35.44	32.52	41.58	34.85	36.29	35.54
MEAN	34.40	34.22	34.83	34.48	34.47	34.91	34.68

Table C.44: Repeated measures during normal walking without oscillation: normalized vertical force applied by the feet.

Subject no	1	2	3	4	5	6	7
1	0.76	0.77	0.86	0.87	0.85	0.75	0.80
2	0.58	0.53	0.57	0.53	0.58	0.52	0.56

3	0.92	0.91	0.86	0.93	0.83	0.85	0.94
4	0.53	0.60	0.56	0.64	0.60	0.56	0.57
5	0.64	0.65	0.68	0.68	0.68	0.68	0.67
6	0.79	0.83	0.84	0.79	0.75	0.89	0.78
7	0.67	0.63	0.75	0.68	0.68	0.66	0.70
8	0.71	0.84	0.78	0.79	0.84	0.72	0.73
9	0.57	0.60	0.65	0.56	0.60	0.57	0.61
10	0.66	0.64	0.61	0.65	0.65	0.63	0.65
11	0.85	0.85	0.81	0.83	0.83	0.88	0.79
12	0.75	0.71	0.72	0.75	0.81	0.74	0.74
13	0.63	0.64	0.59	0.61	0.61	0.62	0.63
14	0.74	0.77	0.68	0.74	0.73	0.79	0.73
15	0.56	0.61	0.67	0.65	0.64	0.65	0.71
16	0.77	0.82	0.79	0.81	0.84	0.79	0.77
17	0.91	0.94	0.91	0.87	0.92	0.92	0.99
18	0.65	0.69	0.70	0.66	0.70	0.73	0.72
19	0.91	0.93	1.05	0.91	1.03	0.91	1.00
20	0.95	0.95	1.00	0.90	0.95	0.92	0.96
21	0.64	0.64	0.63	0.66	0.68	0.75	0.75
22	0.75	0.82	0.80	0.75	0.82	0.83	0.85
23	0.77	0.75	0.77	0.72	0.84	0.87	0.84
24	0.66	0.72	0.74	0.71	0.73	0.72	0.71
25	0.80	0.77	0.70	0.67	0.75	0.66	0.71
26	0.77	0.75	0.82	0.77	0.75	0.80	0.79
27	0.59	0.50	0.56	0.58	0.64	0.69	0.66
28	0.91	0.85	0.78	0.79	0.75	0.86	0.76
29	0.82	0.84	0.85	0.83	0.88	0.82	0.83
30	0.80	0.76	0.77	0.79	0.77	0.87	0.85
31	0.74	0.76	0.77	0.81	0.79	0.82	0.76
32	0.92	0.84	0.87	0.93	0.93	0.85	0.90
33	0.74	0.72	0.68	0.74	0.72	0.73	0.79
34	0.65	0.66	0.69	0.75	0.76	0.81	0.81
35	0.61	0.51	0.58	0.58	0.55	0.62	0.65
36	0.94	0.96	0.99	1.02	1.03	1.12	1.00
37	1.04	1.05	1.02	1.02	1.08	1.08	1.09
38	0.87	0.87	0.89	0.82	0.88	0.77	0.70
39	0.81	0.94	0.82	0.77	0.82	0.83	0.81
40	0.93	0.93	0.88	0.89	0.93	0.93	0.91
41	0.70	0.75	0.71	0.76	0.75	0.68	0.75
42	0.75	0.85	0.82	0.87	0.91	0.89	0.90
43	0.62	0.58	0.62	0.64	0.64	0.53	0.56
44	0.71	0.74	0.78	0.68	0.72	0.65	0.64
45	0.92	0.81	0.88	0.82	0.81	0.82	0.87
46	0.67	0.67	0.74	0.68	0.67	0.69	0.69
47	0.90	0.86	0.82	0.80	0.74	0.82	0.73
48	0.66	0.71	0.71	0.73	0.72	0.81	0.84
49	0.76	0.81	0.74	0.74	0.77	0.78	0.71
50	0.98	0.98	0.99	0.95	0.97	0.96	1.04
51	0.76	0.77	0.78	0.74	0.82	0.84	0.83
52	0.94	0.78	0.73	0.82	0.84	0.85	0.90
53	0.74	0.71	0.68	0.73	0.77	0.70	0.74

54	0.94	0.94	0.98	0.94	0.96	0.99	1.00
55	0.69	0.58	0.57	0.59	0.65	0.63	0.71
56	0.60	0.61	0.62	0.64	0.57	0.61	0.59
57	0.88	0.88	0.89	0.88	0.94	0.92	0.94
58	1.01	1.01	0.93	1.06	0.98	1.00	0.98
59	0.98	0.95	0.85	0.80	0.93	0.96	0.96
60	0.86	0.85	0.85	0.92	0.98	0.93	0.93
61	0.66	0.66	0.75	0.62	0.73	0.85	0.78
62	0.90	0.87	0.85	0.87	0.85	0.90	0.91
63	0.68	0.62	0.69	0.67	0.74	0.71	0.66
64	0.78	0.84	0.84	0.95	0.89	0.88	0.92
65	0.81	0.84	0.84	0.86	0.86	0.85	0.90
66	0.93	0.83	0.80	0.82	0.77	0.85	0.82
67	0.63	0.68	0.69	0.60	0.62	0.69	0.61
68	0.97	0.94	0.87	0.91	0.91	0.93	1.01
69	1.39	1.05	1.17	1.11	1.24	1.45	1.42
70	1.11	1.09	1.09	1.20	1.12	1.15	1.04
71	0.78	0.68	0.67	0.68	0.67	0.72	0.66
72	0.73	0.70	0.72	0.74	0.73	0.80	0.80
73	1.12	1.31	1.10	1.04	1.10	1.23	1.10
74	0.76	0.85	0.79	0.79	0.88	0.94	1.01
75	0.89	0.90	0.96	0.92	0.91	0.92	0.86
76	0.62	0.67	0.64	0.66	0.71	0.74	0.75
77	0.99	0.98	0.94	0.91	0.90	0.93	0.91
78	0.71	0.75	0.72	0.75	0.76	0.81	0.79
79	0.82	0.80	0.84	0.83	0.85	0.86	0.95
80	1.01	1.02	1.01	1.01	1.06	1.05	1.01
81	0.52	0.55	0.52	0.55	0.54	0.56	0.54
82	0.99	0.91	0.87	0.84	0.90	0.91	0.84
83	1.18	1.21	1.22	1.21	1.29	1.25	1.13
84	0.65	0.65	0.63	0.71	0.68	0.72	0.67
85	0.62	0.66	0.58	0.67	0.62	0.78	0.85
86	0.93	0.90	0.91	0.93	0.88	0.93	0.92
87	0.93	0.92	0.93	0.90	0.92	0.94	1.01
88	0.51	0.53	0.52	0.54	0.54	0.54	0.55
89	0.85	0.84	0.83	0.84	0.85	0.83	0.79
90	1.09	1.10	1.15	1.12	0.93	1.05	1.04
91	0.66	0.74	0.73	0.78	0.63	0.69	0.68
92	0.70	0.70	0.68	0.66	0.76	0.61	0.72
93	0.73	0.72	0.72	0.70	0.78	0.69	0.71
94	0.87	0.92	0.85	0.88	0.84	0.74	0.78
95	0.75	0.74	0.74	0.73	0.70	0.71	0.77
96	0.98	0.97	0.98	0.99	0.98	1.00	1.00
97	0.83	0.80	0.86	0.83	0.80	0.68	0.67
98	1.43	1.42	1.54	1.56	1.53	1.51	1.46
99	1.01	1.04	1.06	1.02	1.04	1.05	1.03
100	0.72	0.77	0.80	0.85	0.89	0.87	0.84
MEAN	0.81	0.81	0.80	0.80	0.82	0.83	0.82

Table C.45: Repeated measures during normal walking without oscillation: mean COP speed.

Subject no	1	2	3	4	5	6	7
1	59.62	61.03	66.30	66.71	67.60	66.13	65.25
2	62.85	61.99	62.67	64.93	60.88	62.98	60.55
3	68.15	66.68	68.66	69.99	66.19	67.28	70.79
4	44.59	45.81	46.57	48.61	46.01	42.36	44.42
5	59.70	61.33	61.12	61.27	59.52	59.00	60.83
6	56.34	59.56	55.27	55.67	53.87	55.40	54.23
7	58.27	56.87	58.13	54.84	52.68	53.62	57.35
8	54.47	56.39	57.89	54.77	57.02	54.52	55.12
9	57.27	59.39	59.56	57.85	57.16	58.06	59.94
10	49.48	48.60	50.00	51.79	51.03	50.18	49.39
11	51.75	53.74	53.63	48.07	51.62	51.01	51.05
12	68.78	67.02	66.53	69.24	71.16	70.19	69.36
13	54.16	56.07	55.29	57.57	57.55	58.18	57.04
14	50.96	53.78	50.69	51.19	51.06	55.15	52.76
15	58.93	57.89	58.89	56.69	57.05	55.99	56.26
16	58.77	62.97	62.78	63.65	62.17	61.56	63.78
17	49.32	47.81	46.63	46.94	44.29	43.30	46.65
18	60.94	61.66	60.01	61.98	61.35	62.76	61.90
19	75.57	71.43	73.95	65.45	68.56	67.67	72.16
20	58.74	56.89	57.89	58.14	60.76	59.84	60.29
21	64.72	59.54	61.09	61.65	61.31	64.03	64.91
22	60.20	61.70	60.67	59.57	57.65	59.28	60.83
23	64.72	63.94	65.61	64.15	65.94	68.91	67.63
24	53.52	55.38	57.60	54.69	55.42	55.42	56.69
25	59.98	60.66	59.18	59.31	64.57	61.56	63.07
26	63.75	63.55	64.92	63.07	61.98	62.84	62.55
27	63.21	59.70	64.62	61.61	65.53	64.33	63.30
28	65.70	64.99	62.99	62.47	60.93	65.48	64.08
29	51.18	49.83	47.86	51.16	49.53	51.15	48.73
30	68.89	66.86	67.96	71.47	72.03	78.59	71.40
31	53.25	53.66	52.85	53.91	52.69	53.51	53.37
32	71.56	68.51	68.60	70.65	69.96	68.87	69.65
33	61.02	60.86	58.94	61.29	60.65	59.44	60.23
34	58.03	60.36	58.55	61.44	61.42	60.85	61.62
35	62.14	58.11	62.42	60.21	59.90	60.12	61.44
36	79.21	82.02	80.98	79.19	80.71	80.86	80.88
37	84.83	81.46	80.55	81.88	82.90	83.09	79.94
38	63.13	64.09	62.16	61.11	63.47	65.51	61.18
39	77.25	79.77	76.24	75.15	76.16	78.19	80.83
40	66.10	66.42	68.45	67.10	67.91	67.11	67.50
41	57.96	58.96	56.72	58.01	58.30	57.49	58.53
42	48.10	50.84	47.64	48.65	48.27	47.03	47.96
43	65.09	62.74	62.84	65.65	63.37	60.03	63.07
44	68.59	67.75	69.56	66.82	64.93	68.14	65.65
45	68.10	65.28	66.68	64.52	66.05	68.94	70.00
46	55.41	56.53	55.35	55.25	55.98	54.50	53.74
47	64.59	64.13	64.34	63.24	63.45	65.66	61.96

48	57.27	59.04	57.42	58.75	55.62	58.16	58.42
49	56.00	58.87	55.95	55.76	54.37	57.47	55.10
50	55.46	54.53	53.98	53.67	51.48	55.46	52.37
51	63.26	62.66	63.04	64.04	63.25	63.42	64.07
52	75.86	65.93	67.56	69.83	72.17	69.61	74.21
53	67.84	67.20	67.90	67.15	67.26	69.78	69.56
54	62.26	61.19	62.57	61.35	60.84	61.09	61.31
55	39.33	41.18	41.90	43.82	42.69	41.97	46.17
56	65.33	64.96	65.38	65.60	64.43	65.14	63.00
57	64.55	66.65	62.70	62.13	60.21	59.89	58.51
58	72.70	71.60	68.66	68.54	70.83	69.80	68.09
59	58.20	55.01	57.19	52.87	57.86	56.02	54.83
60	58.79	58.26	59.05	59.60	62.08	56.81	58.91
61	60.98	61.62	61.87	58.72	60.89	62.41	62.22
62	59.28	56.71	59.15	57.35	57.61	58.10	58.01
63	62.40	59.94	62.94	62.05	62.30	62.32	59.83
64	62.95	61.82	62.99	66.60	65.57	65.58	65.96
65	69.57	70.17	66.29	70.59	67.84	67.57	67.89
66	64.05	61.71	62.29	64.49	64.34	62.64	62.63
67	55.12	55.22	54.61	53.23	55.32	56.07	54.42
68	70.33	64.67	64.50	64.10	63.79	64.32	66.82
69	65.96	55.50	60.10	57.44	64.11	64.92	64.01
70	56.05	57.09	54.82	55.04	55.20	55.75	55.77
71	60.05	60.08	59.68	59.80	59.21	59.28	60.62
72	53.55	52.78	50.58	54.68	54.18	56.46	55.01
73	70.96	69.94	73.04	75.59	67.56	70.71	70.08
74	70.25	73.03	69.48	69.59	70.99	71.90	72.30
75	53.64	53.20	53.08	52.12	52.52	52.92	52.44
76	61.90	62.41	55.82	56.21	58.38	57.96	58.08
77	61.78	62.29	62.98	62.24	60.27	64.59	62.79
78	53.11	55.54	57.73	55.74	56.90	59.43	61.54
79	79.05	75.96	76.78	78.43	78.02	79.01	77.83
80	60.10	61.01	62.03	57.41	58.87	59.33	56.67
81	54.90	52.77	53.15	52.81	52.34	55.02	53.67
82	66.68	64.70	63.69	66.19	66.68	63.40	64.49
83	75.95	74.49	78.14	78.14	80.37	77.03	73.16
84	57.85	56.50	55.99	59.00	61.33	64.94	58.77
85	63.53	65.59	63.07	65.83	66.44	66.26	66.42
86	66.45	65.87	66.75	68.70	68.44	65.74	66.92
87	57.12	56.15	61.40	58.54	59.63	58.90	61.56
88	57.90	57.59	60.03	57.45	57.10	57.37	55.59
89	52.07	51.29	50.32	52.34	52.62	51.04	52.49
90	61.31	62.14	64.42	63.12	60.59	64.84	60.58
91	79.00	76.68	74.89	79.52	76.26	81.45	78.58
92	68.99	70.50	68.50	71.28	73.62	69.65	71.26
93	56.94	55.66	55.41	53.37	55.57	50.98	52.75
94	71.90	71.64	74.22	71.58	66.52	64.99	65.62
95	65.75	65.39	66.21	64.82	65.11	66.08	65.02
96	62.50	62.93	58.25	57.99	56.05	57.14	57.49
97	54.34	53.52	52.85	52.80	56.39	56.03	51.65
98	84.39	82.09	87.58	88.21	90.30	88.63	85.50

99	64.00	64.74	66.48	64.73	68.20	68.99	67.98
100	63.54	67.52	68.06	69.95	66.37	63.70	62.70
MEAN	62.12	61.70	61.81	61.80	61.85	62.14	61.92

Table C.46: COP measures for each age group (G1, G2, G3, G4 and overall) with N number of subjects, as a function of number of repetitions of 1 Hz stimuli. Mean and standard deviations (S.d.) are reported.

Peak-to-peak lateral COP position								
Mean	N	1	2	3	4	5	6	7
G1 (18-24)	25	26.73	22.82	22.94	21.65	22.50	21.88	22.31
G2 (25-45)	25	28.17	25.90	26.41	26.43	25.43	25.94	27.24
G3 (46-59)	26	27.45	27.26	27.41	26.95	25.84	27.52	25.78
G4 (60-70)	24	27.27	29.72	28.70	29.73	27.75	28.75	28.85
Overall	100	27.41	26.40	26.35	26.16	25.36	26.01	26.01
S.d								
G1 (18-24)	25	5.20	4.85	4.68	4.55	5.35	5.36	4.87
G2 (25-45)	25	5.96	5.41	5.15	4.27	3.73	4.83	4.91
G3 (46-59)	26	4.42	5.97	5.00	5.44	5.16	5.60	4.64
G4 (60-70)	24	8.16	6.25	5.71	6.12	5.08	6.26	6.01
Overall	100	5.99	6.09	5.50	5.83	5.15	6.03	5.59
Lateral r.m.s. COP velocity								
Mean	N	1	2	3	4	5	6	7
G1 (18-24)	25	55.80	44.78	43.30	41.36	44.86	40.43	44.79
G2 (25-45)	25	62.69	53.70	58.37	58.68	56.09	54.57	55.07
G3 (46-59)	26	60.12	60.06	58.74	58.67	56.09	59.85	54.06
G4 (60-70)	24	60.75	66.34	64.25	67.91	60.73	62.62	64.47
Overall	100	59.83	56.16	56.11	56.56	54.40	54.34	54.49
S.d								
G1 (18-24)	25	14.84	14.75	11.56	13.37	15.59	12.26	13.73
G2 (25-45)	25	21.70	17.08	19.22	15.45	11.79	14.75	13.86
G3 (46-59)	26	17.91	19.04	18.28	18.21	19.42	19.73	15.13
G4 (60-70)	24	26.83	17.36	23.05	18.25	20.91	15.41	17.69
Overall	100	20.55	18.66	19.76	18.82	17.99	17.78	16.46
Normalized r.ms. Force								
Mean	N	1	2	3	4	5	6	7
G1 (18-24)	25	1.20	1.04	0.97	0.93	0.96	0.92	0.93
G2 (25-45)	25	1.37	1.11	1.05	1.06	1.00	1.02	1.02
G3 (46-59)	26	1.31	1.17	1.11	1.10	1.05	1.09	1.04

G4 (60-70)	24	1.41	1.28	1.22	1.21	1.13	1.15	1.14
Overall	100	1.32	1.15	1.08	1.07	1.04	1.05	1.03
S.d								
G1 (18-24)	25	0.29	0.24	0.18	0.20	0.20	0.19	0.19
G2 (25-45)	25	0.44	0.27	0.26	0.23	0.18	0.23	0.19
G3 (46-59)	26	0.30	0.26	0.25	0.25	0.21	0.24	0.21
G4 (60-70)	24	0.34	0.32	0.27	0.21	0.24	0.29	0.24
Overall	100	0.35	0.28	0.25	0.24	0.21	0.25	0.22
mean COP speed								
Mean	N	1	2	3	4	5	6	7
G1 (18-24)	25	74.52	70.51	69.35	67.89	69.18	68.59	68.42
G2 (25-45)	25	78.47	73.53	74.49	73.29	72.80	72.59	73.10
G3 (46-59)	26	77.79	79.33	76.94	75.07	74.09	76.21	72.98
G4 (60-70)	24	80.92	80.04	79.19	79.88	77.48	79.38	78.81
Overall	100	77.89	75.85	74.97	73.99	73.35	74.16	73.27
S.d								
G1 (18-24)	25	8.11	7.73	9.17	7.96	9.32	8.20	6.66
G2 (25-45)	25	12.95	8.56	10.75	7.96	8.11	8.92	7.43
G3 (46-59)	26	10.12	12.01	13.31	13.11	10.64	12.94	9.86
G4 (60-70)	24	11.76	10.43	9.76	9.86	11.66	10.99	11.17
Overall	100	10.95	10.49	11.34	10.72	10.29	11.05	9.55

Table C.47: COP measures for each age group (G1, G2, G3, G4 and overall) with N number of subjects, as a function of number of repetitions of 0.5 Hz stimuli. Mean and standard deviations (S.d.) are reported.

Peak-to-peak lateral COP position				
Mean	N	1	2	3
G1 (18-24)	25	29.53	29.09	27.73
G2 (25-45)	25	34.30	31.50	30.75
G3 (46-59)	26	36.85	33.20	32.47
G4 (60-70)	24	37.07	35.83	34.36
Overall	100	34.44	32.38	31.31
S.d				
G1 (18-24)	25	4.96	6.45	5.38
G2 (25-45)	25	5.32	4.63	4.16
G3 (46-59)	26	6.39	6.35	5.53
G4 (60-70)	24	5.76	5.79	5.68
Overall	100	6.34	6.27	5.69
Lateral r.m.s. COP velocity				
Mean	N	1	2	3
G1 (18-24)	25	45.12	43.64	42.86

G2 (25-45)	25	55.89	53.99	52.66
G3 (46-59)	26	58.57	55.15	56.15
G4 (60-70)	24	63.40	61.25	59.68
Overall	100	55.70	53.45	52.80
S.d				
G1 (18-24)	25	10.89	11.18	11.72
G2 (25-45)	25	15.16	11.84	13.23
G3 (46-59)	26	14.31	14.69	13.00
G4 (60-70)	24	16.47	16.04	14.51
Overall	100	15.62	14.78	14.38
Normalized r.m.s. Force				
Mean	N	1	2	3
G1 (18-24)	25	1.06	1.00	0.98
G2 (25-45)	25	1.13	1.06	1.05
G3 (46-59)	26	1.24	1.17	1.14
G4 (60-70)	24	1.25	1.21	1.21
Overall	100	1.17	1.11	1.09
S.d				
G1 (18-24)	25	0.26	0.17	0.18
G2 (25-45)	25	0.23	0.21	0.23
G3 (46-59)	26	0.31	0.26	0.27
G4 (60-70)	24	0.25	0.24	0.23
Overall	100	0.27	0.24	0.24
Mean COP speed				
Mean	N	1	2	3
G1 (18-24)	25	71.42	69.93	69.08
G2 (25-45)	25	75.75	74.08	73.67
G3 (46-59)	26	76.55	75.88	76.19
G4 (60-70)	24	78.76	77.78	77.50
Overall	100	75.60	74.40	74.10
S.d				
G1 (18-24)	25	8.34	8.11	7.66
G2 (25-45)	25	7.66	7.74	7.08
G3 (46-59)	26	10.67	10.72	10.11
G4 (60-70)	24	8.69	8.81	8.78
Overall	100	9.19	9.27	8.97

Table C.48: COP measures for each age group (G1, G2, G3, G4 and overall) with N number of subjects, as a function of number of repetitions of 2 Hz stimuli. Mean and standard deviations (S.d.) are reported.

Peak-to-peak lateral COP position				
Mean	N	1	2	3
G1 (18-24)	25	19.87	19.85	19.20
G2 (25-45)	25	24.46	23.09	23.86
G3 (46-59)	26	23.53	24.00	21.74
G4 (60-70)	24	24.86	24.42	25.32
Overall	100	23.16	22.84	22.50
S.d				
G1 (18-24)	25	5.57	4.87	5.74
G2 (25-45)	25	5.30	5.45	5.44
G3 (46-59)	26	5.59	4.80	5.13
G4 (60-70)	24	5.38	6.99	5.51
Overall	100	5.73	5.78	5.84
Lateral r.m.s. COP velocity				
Mean	N	1	2	3
G1 (18-24)	25	46.30	45.10	41.16
G2 (25-45)	25	62.09	58.03	59.32
G3 (46-59)	26	60.86	58.95	53.01
G4 (60-70)	24	63.01	58.01	60.49
Overall	100	58.04	55.03	53.42
S.d				
G1 (18-24)	25	19.27	14.86	13.71
G2 (25-45)	25	23.70	20.04	19.63
G3 (46-59)	26	22.75	15.27	16.43
G4 (60-70)	24	19.97	20.31	22.16
Overall	100	22.28	18.41	19.50
Normalized r.m.s. Force				
Mean	N	1	2	3
G1 (18-24)	25	1.21	1.11	1.02
G2 (25-45)	25	1.29	1.18	1.14
G3 (46-59)	26	1.39	1.29	1.16
G4 (60-70)	24	1.34	1.29	1.17
Overall	100	1.31	1.22	1.12
S.d				
G1 (18-24)	25	0.28	0.24	0.20
G2 (25-45)	25	0.32	0.27	0.23
G3 (46-59)	26	0.39	0.35	0.28
G4 (60-70)	24	0.32	0.24	0.24
Overall	100	0.34	0.29	0.24
Mean COP speed				

Mean	N	1	2	3
G1 (18-24)	25	75.05	73.73	71.57
G2 (25-45)	25	80.69	76.00	75.55
G3 (46-59)	26	79.57	77.61	75.38
G4 (60-70)	24	79.99	78.28	78.21
Overall	100	78.82	76.40	75.15
S.d				
G1 (18-24)	25	8.73	9.77	7.22
G2 (25-45)	25	11.71	6.93	8.48
G3 (46-59)	26	12.54	10.94	11.22
G4 (60-70)	24	10.13	11.08	11.76
Overall	100	10.96	9.83	9.98

Table C.49: Repeated COP measures during normal walking without oscillations, for each age group (G1, G2, G3, G4 and overall) with N number of subjects. Mean and standard deviations (S.d.) are reported.

Peak-to-peak lateral COP position								
Mean	N	1	2	3	4	5	6	7
G1 (18-24)	25	16.51	16.85	16.16	15.87	16.23	16.18	16.46
G2 (25-45)	25	18.98	18.77	19.34	19.16	18.70	18.52	18.38
G3 (46-59)	26	17.36	17.53	17.96	17.63	17.95	17.33	18.25
G4 (60-70)	24	18.87	18.71	18.92	19.12	19.24	19.76	19.38
Overall	100	17.91	17.96	18.09	17.93	18.02	17.92	18.11
S.d								
G1 (18-24)	25	4.12	4.02	3.49	4.19	3.78	3.52	3.91
G2 (25-45)	25	4.83	5.04	4.19	5.11	4.58	4.23	4.42
G3 (46-59)	26	4.46	4.16	4.44	4.84	4.77	4.43	4.42
G4 (60-70)	24	4.05	4.01	4.90	4.89	5.70	4.87	4.78
Overall	100	4.44	4.34	4.39	4.89	4.81	4.42	4.45
Lateral r.m.s. COP velocity								
Mean	N	1	2	3	4	5	6	7
G1 (18-24)	25	29.60	29.51	29.01	29.38	29.04	29.16	29.21
G2 (25-45)	25	36.80	36.18	37.52	35.99	35.51	36.07	35.82
G3 (46-59)	26	33.97	34.18	35.41	34.50	35.66	35.43	35.66
G4 (60-70)	24	37.37	37.12	37.44	38.22	37.77	39.13	38.10
Overall	100	34.40	34.22	34.83	34.48	34.47	34.91	34.68
S.d								
G1 (18-24)	25	8.34	8.15	7.40	8.38	7.84	7.25	7.69
G2 (25-45)	25	11.65	11.86	11.38	11.21	9.61	9.35	9.92
G3 (46-59)	26	11.42	10.86	11.96	12.36	11.63	12.05	11.74
G4 (60-70)	24	10.20	10.31	10.93	11.23	11.99	12.29	10.62
Overall	100	10.78	10.64	10.98	11.22	10.75	10.90	10.50

Normalized r.ms. Force								
Mean	N	1	2	3	4	5	6	7
G1 (18-24)	25	0.76	0.76	0.76	0.77	0.78	0.79	0.78
G2 (25-45)	25	0.78	0.77	0.77	0.76	0.78	0.79	0.80
G3 (46-59)	26	0.82	0.84	0.84	0.83	0.84	0.85	0.84
G4 (60-70)	24	0.88	0.86	0.85	0.86	0.87	0.88	0.87
Overall	100	0.81	0.81	0.80	0.80	0.82	0.83	0.82
S.d								
G1 (18-24)	25	0.14	0.12	0.12	0.14	0.12	0.11	0.10
G2 (25-45)	25	0.14	0.15	0.15	0.13	0.14	0.15	0.15
G3 (46-59)	26	0.19	0.18	0.19	0.20	0.19	0.19	0.18
G4 (60-70)	24	0.20	0.19	0.18	0.16	0.19	0.22	0.20
Overall	100	0.17	0.17	0.17	0.16	0.16	0.18	0.16
Mean COP speed								
Mean	N	1	2	3	4	5	6	7
G1 (18-24)	25	60.93	60.53	60.09	60.46	60.83	61.03	60.76
G2 (25-45)	25	62.08	61.68	61.80	61.38	61.31	61.67	61.60
G3 (46-59)	26	61.72	62.00	62.22	61.93	61.89	62.16	62.36
G4 (60-70)	24	63.83	62.62	63.19	63.46	63.45	63.78	62.97
Overall	100	62.12	61.70	61.81	61.79	61.85	62.14	61.92
S.d								
G1 (18-24)	25	5.64	4.87	5.66	5.49	5.72	5.56	5.63
G2 (25-45)	25	6.03	5.85	5.56	5.45	5.34	5.35	5.88
G3 (46-59)	26	9.98	9.72	10.03	10.39	10.90	11.40	10.56
G4 (60-70)	24	10.17	9.14	9.51	9.93	9.68	9.76	9.10
Overall	100	8.18	7.61	7.93	8.11	8.22	8.39	8.01

Appendix D: Matlab script

D.1. Matlab script for the calculation of objective parameters

The script below shows the Matlab code used to calculate the objective parameters of COP from the raw force data obtained via eight forces sensors embedded inside the treadmill.

```
clc
clear all
close all
tic
loadparameters;
hvlab;

%Define base directory that is used throughout the m-file!!

basedirrawdata='C:\Documents and Settings\Mujde Sari\My
Documents\Office_docs\MYDOCS\PhDDocs_Treadmill PC\';
basedirxls='C:\Documents and Settings\Mujde Sari\My
Documents\Office_docs\MYDOCS\PhD DOCUMENTS\4th experiment\';
%Read parameters from excel file
par_file_name=strcat(basedirxls,'EXP4_OBJECTIVE
PARS_Filt_19April_v1.xlsx');
[par_mat,txtdataparmat,rawdataparmat]=xlsread(par_file_name,'Subj_resp
onses','B2:DE14');
[prob_subj,txtdataprobsubject,rawdatasubject]=xlsread(par_file_name,'S
ubj_responses','J2:DE14');
[direction,txtdatadirection,rawdatadirection]=xlsread(par_file_name,'S
ubj_responses','B20:B119');

freq=par_mat(:,1);
rms_accl=par_mat(:,2);
rms_vel=par_mat(:,3);
rms_disp=par_mat(:,4);
peak_accl=par_mat(:,5);
peak_vel=par_mat(:,6);
peak_disp=par_mat(:,7);

%FOR loop for the subjects
for subject_no=1:100
   %FOR loop for the parameters
 for par_count=1:size(par_name_mat,1)
     close all
        subject_no
        par_count

        f1=freq(par_count);    % 0.5-0.63-0.8-1-1.25-1.6-2 Hz % CHANGE
THE FREQUENCY
        A=rms_accl(par_count)*2;

        %Define the directory paths
```

```
datafilepath_exp=strcat(basedirrawdata,'EXPERIMENT4_walks\SUBJECT',num
2str(subject_no,'%d'),'\',par_name_mat{par_count},'.exp');

datafilepath_gaux=strcat(basedirrawdata,'EXPERIMENT4_walks\SUBJECT',nu
m2str(subject_no,'%d'),'\',par_name_mat{par_count},'.gaux');

        %IF EXP

        if (exist(datafilepath_exp,'file'))

            rawdatadata=importdata(datafilepath_exp,'\t',17);
            raw= rawdatadata.data(:,10:25);
            accelerationdata=raw(:,11);
            is_par_file_exist{par_count,subject_no}=['Subject no '
num2str(subject_no,'%d') '   Par count ' num2str(par_count,'%d') ':
EXP'];

            %IF GAUX
        elseif (exist(datafilepath_gaux,'file'))

            rawdatadata=importdata(datafilepath_gaux,'\t',4);
            raw=rawdatadata.data(2:length(rawdatadata.data),:);
            accelerationdata=raw(:,11);
            is_par_file_exist{par_count,subject_no}=['Subject no '
num2str(subject_no,'%d') '   Par count ' num2str(par_count,'%d') ':
GAUX'];

            %NO FILE OR FILE NAME WRONG!!
        else

        is_par_file_exist{par_count,subject_no}=['Subject no '
num2str(subject_no,'%d') '   Par count ' num2str(par_count,'%d') ': NO
FILE'];

        %no file so skip current FOR loop step!!
        continue;

        end

sign=direction(subject_no);
time=0:0.01:(length(raw)-1)*0.01
%%
%THEORIC ACCELERATION, VELOCITY AND DISPLACEMENT
%INITIAL TIME DURATION FOR ZEROS
if f1==0.5 %equalized for 12 sec
const=1.5;  %(12-(4.5/0.5))/2
final_time=12; %15 seconds collected,
elseif f1==1 %,equalized for 8 sec
const=1.75;
final_time=8; %12 seconds collected
elseif f1==2.0
 const=0.875;%equalized for 4 sec
final_time=4; % 8 seconds collected
end
norm_dur=4.5/f1;
```

```
motion_sim_int=1/256;%time interval for the generated stimuli to be
equalized in 6-axis

%Theoric acceleration, velocity and displacement for any n number of
cycles
t0=0:motion_sim_int:(const-motion_sim_int);
a0=0.*t0;
v0=a0;
d0=a0;

n=4.5; %number of cycles
t1=const:motion_sim_int:((n/f1)+const);%
a1=sign.*(A*sin(2*pi*f1*(t1-const))).*sin(pi*f1*(t1-const)/n);
v1=sign.*((A*(pi*f1/n)/((pi*f1/n)^2-(2*pi*f1)^2))*(-sin(2*pi*f1*(t1-
const)).*cos((pi*f1/n)*(t1-const))+(2*n)*sin((pi*f1/n)*(t1-
const)).*cos(2*pi*f1*(t1-const))));
        d1=sign.*((A/2)*(((cos((2+(1/n))*pi*f1*(t1-const))-
1)/(((2+(1/n))*pi*f1)^2))-((cos((2-(1/n))*pi*f1.*(t1-const))-1)/(((2-
(1/n))*pi*f1)^2))));

t2=(n/f1+const+motion_sim_int):motion_sim_int:final_time;
a2=0.*t2;
v2=a2;
d2=a2;

theoric_acc=[a0 a1 a2];
theoric_velocity=[v0 v1 v2];
theoric_disp=[d0 d1 d2];
theoric_time=[t0 t1 t2];

%RMS and PEAK values of THEORIC ACCELERATION, VELOCITY and
DISPLACEMENT
mean_theoric_acc=mean(a1);
rms_theoric_acceleration=sqrt(sum(a1.*a1)/length(a1))%[m/s2]
peak_theoric_acceleration=max(max(a1),abs(max(-a1)))
CF_theoric=peak_theoric_acceleration/rms_theoric_acceleration

mean_theoric_vel=mean(v1);%[m/s]

rms_theoric_velocity=sqrt(sum(v1.*v1)/length(v1))%[m/s]%DIFFERENTIATED
FROM THEORIC DISPLACEMENT
peak_theoric_velocity=max(max(v1),abs(max(-v1)))
CF_theoric_vel=peak_theoric_velocity/rms_theoric_velocity;

mean_theoric_disp=mean(d1);%[m]
rms_theoric_displacement=sqrt(sum(d1.*d1)/length(d1))%[m]
peak_theoric_displacement=max(max(d1),abs(max(-d1)))

%% MEASURED ACCELERATION, VELOCITY AND DISPLACEMENT
%Auxiliary : Channel 1 connected to acceleration Y, Clippage proglem
occured
        %in some of them
scale_factor=50/10;
acc_measured=accelerationdata;
acc_meas=scale_factor.*acc_measured;%NOT FILTERED
measured_acceleration=acc_meas-mean(acc_meas).*ones(size(acc_meas));%
acc_meas subtracted mean
```

```
time_meas=time;

%FILTERED acceleration
fcut_acc=8;%CUT-off acceleration

measured_acceleration_struc=hvcreate(measured_acceleration, 0.01);
filtered_acceleration=hvfilter(measured_acceleration_struc,
'lobutter',fcut_acc, 2);%filtered acc in structure


measured_velocity_struct=hvintegral(measured_acceleration_struc,1);
measured_displacement_struct=hvintegral(measured_acceleration_struc,2;
measured_velocity=detrend(measured_velocity_struct.y);
measured_displacement=detrend(measured_displacement_struct.y);

figure(1)

plot(time_meas,measured_acceleration,time_meas,filtered_acceleration.y
,'g',theoric_time,theoric_acc,'r')
legend('measured','filtered','desired')
xlabel('Time (s)')
ylabel('Acceleration (ms^-^2)')
title('Measured acceleration')

figure(2)

plot(time_meas,measured_velocity,'g',theoric_time,theoric_velocity,'r'
)
legend('measured','desired')
xlabel('Time (s)')
ylabel('Velocity(ms^-^1)')
title('Measured velocity')

figure(3)

plot(time_meas,measured_displacement,'g',theoric_time,theoric_disp,'r'
)
legend('measured','desired')
xlabel('Time (s)')
ylabel('Displacement(m)')
title('Measured displacement')

%% GATHERED RAW DATA

fs=100; %100 Hz sampling frequency

% FRONT FORCE PLATE IN VOLTS
force1=raw(:,1); %front force plate top-left corner
force2=raw(:,2); %front force plate top-right corner
force3=raw(:,3); %front force plate bottom-right corner
force4=raw(:,4); %front force plate bottom-left corner
% REAR FORCE PLATE IN VOLTS
force5=raw(:,5); %front force plate top-left corner
force6=raw(:,6); %front force plate top-right corner
force7=raw(:,7); %front force plate bottom-right corner
force8=raw(:,8); %front force plate bottom-left corner
```

```
%CONVERT RAW DATA IN VOLTS INTO NEWTON
amp_range=4720;
sens_sensitivity=[4.69;4.66;4.68;4.57;4.35;4.91;4.48;4.27];
F1=force1.*amp_range/(5*sens_sensitivity(1));
F2=force2.*amp_range/(5*sens_sensitivity(2));
F3=force3.*amp_range/(5*sens_sensitivity(3));
F4=force4.*amp_range/(5*sens_sensitivity(4));
F5=force5.*amp_range/(5*sens_sensitivity(5));
F6=force6.*amp_range/(5*sens_sensitivity(6));
F7=force7.*amp_range/(5*sens_sensitivity(7));
F8=force8.*amp_range/(5*sens_sensitivity(8));
total_force=F1+F2+F3+F4+F5+F6+F7+F8;
mean_tot_force=mean(total_force).*(ones(size(total_force)));

%filtered total force
fcut_force=8;
total_force_str=hvcreate(total_force,0.01);
filt_total_force=hvfilter (total_force_str, 'lobutter', fcut_force,
2);

mean_filt_tot_force=mean(filt_total_force.y).*(ones(size(filt_total_fo
rce.y)));

figure(4)
plot(time,F1,'r',time,F2,'g--',time,F3,'g-.',time,F4,'m--',...
time,F5,'b',time,F6,'y--',time, F7,'k-.',time, F8,'c:')
ylim([0 700])
set(gca,'XTick',[0 1 2 3 4 5 6 7 8 9 10 11 12])
legend('Sensor 1','Sensor 2','Sensor 3','Sensor 4','Sensor 5','Sensor
6','Sensor 7','Sensor 8')
xlabel('Time (sec)')
ylabel('Force (N)')
title('Raw force data in Newton')

figure(5)
plot(time,total_force,time,filt_total_force.y,'r')
grid
hold on
plot(time,mean_filt_tot_force,'k','LineWidth',2)
hold off
xlabel('time (sec)')
ylabel('Total Filtered Force (Newton)')
title('Total Filtered force data in Newton')

%CALCULATION OF COP(note: _d means dimensions)
a_d=30.632;%[cm]
c_d=3.015;%[cm]
d_d=65.532;%[cm]

Ax=1.*(F1.*d_d+F2.*d_d+F3.*c_d+F4.*c_d-F5.*c_d-F6.*c_d-F7.*d_d-
F8.*d_d)./total_force;
Az=1.*(F1.*a_d-F2.*a_d-F3.*a_d+F4.*a_d+F5.*a_d-F6.*a_d-
F7.*a_d+F8.*a_d)./total_force;
Midline=mean(Az).*ones(size(Az));%Middle walking line
Vel_Ax=diff(Ax)./(diff(time)');%not filtered
Vel_Az=diff(Az)./(diff(time)');%not filtered
```

```
fcut_COP=8;%cut off for filtering COP before differentiation
[Az_struc] = hvcreate(Az, 0.01);
[Ax_struc] = hvcreate(Ax, 0.01);
Az_filtered=hvlobessel(Az_struc,fcut_COP,2);
Ax_filtered=hvlobessel(Ax_struc,fcut_COP,2);
Vel_Az_filtered= hvdifferential(Az_filtered,1);
Vel_Ax_filtered = hvdifferential(Ax_filtered,1);

%% CALCULATIONS DURING PERTURBATION TIME
start_pert_theoric=t1(1);%start time of theoric perturbation
end_pert_theoric=t1(length(t1));%end time of perturbation
duration_of_perturbation=end_pert_theoric-start_pert_theoric; %value
for the duration of perturbation

%MEASURED ACCELERATION, VELOCITY AND DISPLACEMENT EXTRACTED

measured_acceleration_pert_str=hvextract(measured_acceleration_struc,
duration_of_perturbation, start_pert_theoric);
measured_acceleration_pert=measured_acceleration_pert_str.y;

filtered_acceleration_pert_str=hvextract(filtered_acceleration,
duration_of_perturbation, start_pert_theoric);
filtered_acceleration_pert=filtered_acceleration_pert_str.y;
measured_velocity_structure=hvcreate(measured_velocity,0.01);

measured_velocity_pert_str=hvextract(measured_velocity_structure,
duration_of_perturbation, start_pert_theoric);
measured_velocity_pert=measured_velocity_pert_str.y;
measured_disp_structure=hvcreate(measured_displacement,0.01);
measured_disp_pert_str=hvextract(measured_disp_structure,
duration_of_perturbation, start_pert_theoric);
measured_disp_pert=measured_disp_structure.y;

        %RMS and PEAK values of MEASURED ACCELERATION, VELOCITY and
DISPLACEMENT
        %RMS and PEAK values of THEORIC ACCELERATION, VELOCITY and
DISPLACEMENT
        %theoric acc is a1, theoric vel is v1, theoric disp is d1,
        %mean, rms and peak values were calculated in the first
section

rms_measured_acceleration_pert=sqrt(sum(filtered_acceleration_pert.*fi
ltered_acceleration_pert)/length(filtered_acceleration_pert))%[m/s2]

peak_measured_acceleration_pert=max(max(filtered_acceleration_pert),ab
s(max(-filtered_acceleration_pert)))

CF_measured_pert=peak_measured_acceleration_pert/rms_measured_accelera
tion_pert

rms_measured_velocity_pert=sqrt(sum(measured_velocity_pert.*measured_v
elocity_pert)/length(measured_velocity_pert))%[m/s]

peak_measured_velocity_pert=max(max(measured_velocity_pert),abs(max(-
measured_velocity_pert)))

CF_vel_measured_pert=peak_measured_velocity_pert/rms_measured_velocity
_pert
```

```
rms_measured_disp_pert=sqrt(sum(measured_disp_pert.*measured_disp_pert
)/length(measured_disp_pert))%[m/s]
        peak_measured_disp_pert=max(max(measured_disp_pert),abs(max(-
measured_disp_pert)))

CF_disp_measured_pert=peak_measured_disp_pert/rms_measured_disp_pert

%Az and Ax extracted from filtered versions
Az_pert_str=hvextract(Az_filtered, duration_of_perturbation,
start_pert_theoric);
Ax_pert_str=hvextract(Ax_filtered, duration_of_perturbation,
start_pert_theoric);
Az_pert=Az_pert_str.y;
Ax_pert=Ax_pert_str.y;
time_Az_pert=Az_pert_str.x;
time_Ax_pert=time_Az_pert;

%mean, rms and peak values of lateral COP (Az)
mean_COP_pert=mean(Az_pert);%mean COP position during perturbation
rms_COP_pert=sqrt(sum((Az_pert.*Az_pert)/length(Az_pert)));
COP_meanCOP=Az_pert-mean(Az_pert).*ones(size(Az_pert));

std_COP_pert=sqrt(sum((COP_meanCOP.*COP_meanCOP)/length(COP_meanCOP)))
;%standard deviation of Az-meanAz during perturbation
        max_pos_COP_minus_mean=max(COP_meanCOP);%max positive Az-
meanAz
        max_neg_COP_minus_mean=-max(-COP_meanCOP);%max negative Az-
mean Az

abs_peak_COP_pert_mean=max(abs(max_pos_COP_minus_mean),abs(max_neg_COP
_minus_mean));

peak_to_peak_COP_pert=abs(max_pos_COP_minus_mean)+abs(max_neg_COP_minu
s_mean);

%Mean COP speed and mean lateral speed calculations
for zz=1:(length(Ax_pert)-1)
   Ax_change_pert(zz)=Ax_pert(zz+1)-Ax_pert(zz);
   Az_change_pert(zz)=Az_pert(zz+1)-Az_pert(zz);

mean_COP_speed_each_pert(zz)=sqrt(Ax_change_pert(zz)^2+Az_change_pert(
zz)^2);
end
mean_COP_speed_pert=sum(mean_COP_speed_each_pert);
mean_COP_speed_pert_normalized=mean_COP_speed_pert/norm_dur;


clear mean_COP_speed_each_pert

%Force during perturbation_ total force extracted from total filtered
force
total_force_pert_str=hvextract(filt_total_force,
duration_of_perturbation, start_pert_theoric);
total_force_pert=total_force_pert_str.y;

mean_total_force_pert=mean(total_force_pert);
```

```
force_minus_mean_pert=total_force_pert-
mean_total_force_pert.*ones(size(total_force_pert));

rms_force=sqrt(sum((total_force_pert.*total_force_pert)/length(total_f
orce_pert)));

std_force=sqrt(sum((force_minus_mean_pert.*force_minus_mean_pert)/leng
th(force_minus_mean_pert)));
        max_pos_force_minus_mean=max(force_minus_mean_pert);
        max_neg_force_minus_mean=-max(-force_minus_mean_pert);%minimum
total force during perturbation

abs_peak_force_mean=max(abs(max_pos_force_minus_mean),abs(max_pos_forc
e_minus_mean));


figure(6)
plot(Az_pert,Ax_pert,'.')
hold on

plot(mean_COP_pert.*ones(size(Ax_pert)),Ax_pert,'r','Linewidth',2)
grid
xlabel('Az(cm)')
ylabel('Ax (cm)')
title('COP')


%Lateral COP velocity extracted after filtered
COP_vel_pert_str=hvextract(Vel_Az_filtered, duration_of_perturbation,
start_pert_theoric);
COP_vel_pert=COP_vel_pert_str.y;

mean_COP_vel=mean(COP_vel_pert); %mean Az vel during perturbation
        COP_vel_pert_minus_mean=COP_vel_pert-
mean_COP_vel.*ones(size(COP_vel_pert));

rms_COP_vel=sqrt(sum((COP_vel_pert.*COP_vel_pert)/length(COP_vel_pert)
));

std_COP_vel=sqrt(sum((COP_vel_pert_minus_mean.*COP_vel_pert_minus_mean
)/length(COP_vel_pert_minus_mean)));
max_vel_COP_minus_mean=max(COP_vel_pert_minus_mean);%max positive Az
vel-mean vel during pert
max_neg_vel_COP_minus_mean=-max(-COP_vel_pert_minus_mean);

abs_peak_COP_vel_mean=max(abs(max_vel_COP_minus_mean),abs(max_neg_vel_
COP_minus_mean));

figure(7)
plot(time_Az_pert,Az_pert,'.')
xlabel('time(s)')
ylabel('Lateral COP (cm)')
title('Lateral COP during perturbation')

figure(8)
plot(time_Az_pert,COP_vel_pert_minus_mean,'LineWidth',1.5)
xlabel('Time (sec)')
```

```
ylabel('Lateral COP velocity (cms^-1)')


out_index=par_count+(subject_no-1)*13;

output(out_index,:)=[subject_no par_count
prob_subj(par_count,subject_no)...
 rms_measured_acceleration_pert rms_measured_velocity_pert
rms_measured_disp_pert...
 peak_measured_acceleration_pert peak_measured_velocity_pert
peak_measured_disp_pert...
 mean_COP_pert rms_COP_pert std_COP_pert abs_peak_COP_pert_mean
peak_to_peak_COP_pert...
 mean_COP_vel rms_COP_vel std_COP_vel  abs_peak_COP_vel_mean ...
 mean_total_force_pert  rms_force std_force  abs_peak_force_mean...
 mean_COP_speed_pert_normalized];

phase_string(out_index)={gate_phase};

    end %end of the parameter loop (starts at line 1)

end %end of the subject loop (starts at line 1)


% %definition of startcell, subjects are written one after the other
% %in the SAME SHEET, there is a 2 row offset to account for the
% %header rows!!
%
excel_directory_name=strcat(basedirxls,'EXP4_OBJECTIVE
PARS_Filt_19April_v1.xlsx');
%

[success,message]=xlswrite(excel_directory_name,output2,'Overall','AG)

toc
```

List of references

Alexander, N.B. (1996). Gait disorders in older adults. J. Amer. Geriatr. Soc. 44, 434-451.

Alexander, N.B., Shepard, N., Gu, M.J. and Schultz, A. (1992). Postural control in young and elderly adults when stance is perturbed: kinematics. Journals of Gerontology: Medical Science 47 , M79–M87.

Balogun, J.A., Akindele, K.A., Nihinlola ,J.O. and Marzouk, D.K. (1994). Age related changes in balance performance. Disability and Rehabilitation 16, 58-62.

Baloh, R.W., Jacobson, K.M., Enreitto, J.A., Corona, S. and Honrubia, V. (1998). Balance disorders in older persons: quantification with posturography. Otolaryngol Head and Neck Surg 119, 89-92.

Barak, Y., Wagenaar, R.C., Kenneth, G.H. (2006). Gait Characteristics of Elderly People with a History of Falls: A Dynamic Approach. Physical Therapy, 86, 1501-1510.

Baratto, L., Morasso, P.G.,Re, C., and Spada, G. (2002). A new look at posturographic analysis in the clinical context: sway desnity versus other parameterization techniques. Motor Control 6, 246-270.

Bateni, H. and Maki, B.E. (2005). Review Article: Assistive Devices for Balance and Mobility: Benefits, Demands, and Adverse Consequences. Arch Phys Med Rehabil. 86, 134-145.

Bauby, C.E. and Kuo, A.D. (2000). Active control of lateral balance in human walking. Journal of Biomechanics 33,1433-1440.

Berg, K., Wood-Dauphinee, S., Williams, J.I. and Gayton, D. (1989). Measuring balance in the elderly: Preliminary development of an instrument. Physiotherapy Canada 41, 304–311.

Berger, W., Dietz, V. and Quintern, J. (1984). Corrective reactions to stumbling in man: Neuronal coordination of bilateral leg muscle activity during gait. The Journal of Physiology 357, 109-125.

Berger, W., Trippel, M., Discher, M. and Dietz, V. (1992). Influence of subjects' height on the stabilization of posture. Acta Otolaryngol (Stockh) 112, 22-30.

Berthoz, A. and Pozzo, T. (1988). Intermittent head stabilization during postural and locomotory tasks in humans. In: Amblard, B.,Berthoz, A., Clarac, F. (Eds.), Development, adaptation and modulation of posture and gait. Elsevier, Amsterdam, 189–198.

Bhatt, T., Wening, J.D. and Pai, Y.-C. (2005). Influence of gait speed on stability: recovery from anterior slips and compensatory slipping. Gait and Posture 21, 146-156.

Bhatt, T., Wening, J.D. and Pai, Y.-C. (2006). Adaptive control of gait stability in reducing slip-related backward loss of balance. Exp Brain Res. 170, 61-73.

Black, F.O., Wall, C., Rockette, H.E. and Kitch, R. (1982). Normal subject postural sway during the Romberg test. Amer. J. Otolaryng. 3, 309-318.

Blümle, A., Maurer, C., Schweigart, G. and Mergner, T. (2006). A cognitive intersensory interaction mechanism in human postural control. Exp Brain Res. 173, 357-363.

Bortolami, S.B., DiZio, P., Rabin, E. and Lackner, J.R. (2003). Analysis of human postural responses to recoverable falls. Exp Brain Res. 151, 387–404.

Brady, R.A., Peters, B.T. and Bloomberg, J.J. (2009). Strategies of healthy adults walking on a laterally oscillating treadmill. Gait and Posture 29, 645-649.

Brauer, S., Burns, Y. and Galley, P. (1999). Lateral reach: a clinical measure of medio-lateral postural stability. Physiotherapy Research International 4, 81-88.

British Standards Institution (1987). BS 6841. Measurement and evaluation of human exposure to whole-body mechanical vibration.

Brown, L.A.., Jensen, J.L., Korff, T. and Woollacott, M.H. (2001). The translating platform paradigm: perturbation displacement waveform alters the postural response. Gait and Posture 14, 256-263.

Browning, A.C. (1974). The tolerance of the general public to a speed differential between adjacent floors with special reference to pedestrian conveyors: Exploratory and preliminary experiments, Tech. Report 74076, Farnborough, England, Royal Aircraft Establishment.

Campbell, A.J., Borrie, M.J., Spears, G.F., Jackson, S.L., Brown, J.S. and Fitzgerald, J.L. (1990). Circumstances and consequences of falls experienced by a community population 70 years and over during a prospective study. Age and Ageing 19, 136–141.

Cao, C., Ashton-Miller, J.A., Schultz, A.B., and Alexander, N.B. (1997). Abilities to turn suddenly while walking: Effects of age, gender and available response time. The Journal of Gerontology, Series A, Biological Sciences and Medical Sciences 52, M88-M93.

Cenciarini, M. and Peterka, R.J. (2006). Stimulus-dependent changes in the vestibular contribution to human postural control. J Neurophysiol. 95, 2733–2750.

Chiari, L., Rocchi, L., and Cappello, A. (2002). Stabilometric parameters are affected by anthropometry and foot placement. Clinical Biomechanics 17, 666-677.

Clapp, S. and Wing, A.M. (1999). Light touch contribution to balance in normal bipedal stance. Exp Brain Res. 125m, 521-524.

Courtney, T. K., Sorock, G.S., Manning, D. P., Collons, J.W., and Holbein-Jenny, M.A. (2001). Occupational slip, trip, and fall related injuries- can the contribution of slipperiness be isolated? Ergonomics 44, 1118-1137.

Dean, J.C., Alexander, N.B. and Kuo, A.D. (2007). The effect of lateral stabilization on walking in young and old adults. IEEE Transactions on biomedical engineering 54, 1919-1926.

Dickstein, R. and Laufer, Y. (2004). Light touch and center of mass stability during treadmill locomotion. Gait and Posture 20, 41–47.

Diener, H.C. and Dichgans, J. (1988). On the role of vestibular, visual, and somatosensory information for dynamic postural control in humans. In Pompeiano O., Allum J.H.J. (eds.): Progress in Brain Research 76. New York, Elsevier, 253-262.

Diener, H.C., Dichgans J., Guschlbauer, B., Bacher, M. (1986). Role of visual and static vestibular influences on dynamic posture control. Hum. Neurobiol. 5, 105-113.

Diener, H.C., Dichgans, J., Bruzek, W., and Selinka, H. (1982). Stabilization of human posture during induced oscillations of the body. Exp. Brain Res. 45, 126-132.

Donati, P., Grosjean, A., Mistrot, P. and Roure, L. (1983). The subjective equivalence of sinusoidal and random whole-body vibration in the sitting position (an experimental study using the floating reference vibration method). Ergonomics 26, 251-273.

Duysens, J. and Bloem, B. (2009). Preferring preferred speed, Proceedings of the Nineteenth Conference of the International Society for Posture and Gait Research, Bologna, Italy, June 2009, pp. 40.

Fernie, G.R. and Holliday, P.J. (1978). Postural sway in amputees and normal subjects. J. Bone Jt. Surg. 60-A, 895-898.

Forssberg, H., Grillner, S. and Rossignol, S. (1975). Phase-dependent reflex reversal during walking in chronic spinal cats. Brain Res. 85, 103-107.

Fransson, P.A., Johansson, R., Hafström, A. and Magnusson, M. (2000). Methods for evaluation of postural control adaptation. Gait Posture 12, 14-24.

Geurts, A.C.H., Nienhuis, B. and Mulder, T.W. (1993). Intrasubject variability of selected force-platform parameters in the quantification of postural control. Arch Phys Med Rehabil. 74, 1144–1150.

Goulding, A., Jones, I.E., Taylor, R.W., Piggot, J.M. and Taylor, D. (2003). Dynamic and static tests of balance and postural sway in boys: effects of previous wrist bone fractures and high adiposity. Gait and Posture 17, 136-141.

Graaf, B. De, and Van Weperen, W. (1997). The retention of balance: An explanatory study into the limits of acceleration without losing equilibrium. Human Factors and Ergonomics Society, 39, 111-118.

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

Greenspan, S.L., Myers, E.R., Kiel, D.P., Parker, R.A., Hayes, W.C. and Resnick, N.M. (1998). Fall direction, bone mineral density, and function: risk factors for hip fracture in frail nursing home elderly. Am J Med.104, 539-545.

Griffin, M.J. (1976). Subjective equivalence of sinusoidal and random whole-body vibration. Journal of the Acoustical Society of America 60, 1140-1145.

Griffin M.J. (1990). Handbook of Human Vibration, Academic Press, London. ISBN 0-12-303040-4.

Griffin, M.J. and Whitham, E.M. (1980) Discomfort produced by impulsive whole-body vibration. Journal of the Acoustical Society of America 68, 1277-1284.

Gu, M-J., Schultz, A.B., Shepard, N.T. and Alexander, N.B. (1996). Postural control in young and elderly adults when stance is perturbed: dynamics. Journal of Biomechanics 29, 319-329.

Hahn, M.E. and Chou, L-S. (2003). Can motion of individual body segments identify dynamic instability in the elderly? Clinical Biomechanics 18, 737-744.

Hausdorff, J.M., Rios, D.A. and Edelberg, H.K. (2001). Gait variability and fall risk in community-living older adults: A 1-year prospective study. Arch. Phys. Med. Rehab. 82, 1050-1056.

Hindmarsh, J.J. and Estes, E.H.Jr. (1989). Falls in older persons. Causes and interventions. Arch Intern Med. 149, 2217-2222.

Hof, A. L. (2008). The ‘extrapolated center of mass' concept suggests a simple control of balance in walking. Human Movement Science 27, 112-125.

Hof, A. L., Vermerris, S.M. and Gjaltema, W.A.. (2010). Balance responses to lateral perturbations in human treadmill walking. Journal of Experimental Biology 213, 2655-2664.

Hof, A.L., van Bockel, R.M., Schoppen, T. and Postema, K. (2007). Control of lateral balance in walking: Experimental findings in normal subjects and above-knee amputees. Gait and Posture 25, 250-258.

Holden, M., Ventura, J., and Lackner, J.R. (1994). Stabilization of posture by precision contact of the index finger. Journal of Vestibular Research 4, 285-301.

Horak, F.B. and Nashner, L.M. (1986). Central programming of postural movements: adaptation to altered support-surface configurations. J. Neurophysiology 55, 1369-1381.

Horak, F.B., Shupert, C.L., and Mirka, A. (1999). Components of postural dyscontrol in the elderly: A review. Neurobiol. Aging 10, 727-738.

Howarth, H. V. C. and Griffin, M. J. (1991). Subjective reaction to vertical mechanical shocks of various waveforms. Journal of Sound and Vibration 147, 395-408.

Hue, O., Simoneau, M., Marcotte, J., Berrigan, F., Doré, J., Marceau, P., Marceau, S., Tremblay, A. and Teasdale, N. (2007). Body weight is a strong predictor of postural stability. Gait Posture 26, 32-38.

Imms, H.J. and Edholm, O.G. (1979). The assessment of gait and mobility in the elderly. Age and Ageing 8, 261-267.

International Organization for Standardization (1997). ISO 2631-1. Mechanical vibration and shock—evaluation of human exposure to whole-body vibration—part 1: general requirements.

IPAQ-International Physical Activity Questionnaire, Short last 7 days self-administered format, for use with young and middle-aged adults (15-69 years), (August 2002). https://sites.google.com/site/theipaq/home

ISO 2631-4:2001, Mechanical vibration and shock- Evaluation of Human exposure to whole-body vibration, Part 4: Guidelines for the evaluation of the effects of vibration and rotational motion on passenger and crew comfort in fixed-guideway transport systems.

Jacobson, G.P. (1993). Chapter 12: Practical biomechanics and physiology of balance, by Nashner, L. M. Handbook of Balance Function Testing, 261-279.

Jeka, J.J and Lackner, J.R. (1994). Fingertip contact influences human postural control. Exp Brain Res, 100, 495-502.

Jeka, J.J and Lackner, J.R. (1995). The role of haptic cues from rough and slippery surfaces in human postural control. Exp Brain Res. 103, 267-276.

Jeka, J.J. (1997). Light Touch Contact as a Balance Aid. Physical Therapy, 77, 476-487.

Jongkees, L.B.W., Groen, J.J. (1942). De standvastigheid van de mens [The stability of the human body]. Nederlands Tiidschrift voor Geneeskunde 86, 1401-1407.

Kaufman, K.R., Robert, H.B., Chou, L-S., Rabatin, A., Brown, A.W., Basford, J.R. (2006). Comparison of subjective and objective measurements of balance disorders following traumatic brain injury. Medical Engineering and Physics 28, 234-239.

Kaya, B.K, Krebs, D.E. and Riley, P.O. (1998). Dynamic stability in elders: momentum control in locomotor ADL. J. Gerontol. A. Biol. Sci. Med. Sci. 53, M126-34.

Koozekanani, S. H., Stockwell, C. W., McGhee, R. B. and Firoozmand, F. (1980). On the Role of Dynamic Models in Quantitative Posturography. Biomedical Engineering, IEEE Transactions on BME 27, 605-609.

Lee, H-J. and Chou, L-S. (2006). Detection of gait instability using the centre of mass and centre of pressure inclination angles. Arch .Phys. Med. Rehabil. Brain Res. 87, 569-575.

Lestienne, F., Soechting, J.F. and Berthoz, A. (1977). Postural readjustments induced by linear motion of visual scenes. Experimental Brain Research 28, 363-384.

Lewis, C. H. and Griffin, M. J. (1997). Effect of deck motions on postural stability: the development of models to predict loss of balance during standing and locomotion. ISVR Contract Report No. 97/8, Institute of Sound and Vibration Research, University of Southampton, UK.

Lockhart, T.E., Woldstad, J.C. and Smith, J.L. (2003). Effects of age related gait changes on the biomechanics of slips and falls. Ergonomics 46, 1136-1160.

Luchies, C.W., Schiffman, J., Richards, L.G., Thompson, M.R., Bazuin, D., and DeYoung, A.J. (2002). Effects of age, step direction, and reaction coordination on the ability to step quickly. The Journals of Gerontology Series A, Biological Sciences and Medical Sciences 57, M246-249.

Lundin-Olsson, L., Nyberg, L. and Gustafson, Y. (1997). Stops walking when talking as a predictor of falls in elderly people. Lancet 349, 617.

Lundin-Olsson, L., Nyberg, L. and Gustafson, Y. (1998). Attention, frailty, and falls: the effect of a manual task on basic mobility. J Am Geriatr Soc. 46, 758–61.

MacKinnon, C.D. and Winter, D.A. (1993). Control of whole body balance in the frontal plane during human walking. Journal of Biomechanics 26, 633-644.

Maki, B. E. and Fernie, G.R. (1988). A System Identification Approach to Balance Testing. Chapter 26, 1988 Elsevier Science Publishers B.V. (Biomedical Division), 297-306.

Maki, B.E. (1986). Selection of perturbation parameters for identification of the posture-control system. Medical & Biological Engineering & Computing, 24, 561-568.

Maki, B.E. (1993). Biomechanical approach to quantifying anticipatory postural adjustments in the elderly. Medical & Biological Engineering & Computing 31, 355-362.

Maki, B.E. (1997). Gait changes in older adults: Predictors of falls or indicators of fear? Journal of the American Geriatrics Society, 45, 313-320.

Maki, B.E. and McIlroy, W.E. (1996). Postural control in the older adult. Clinical Geriatric Medicine 12, 635-658.

Maki, B.E. and Ostrovski, G. (1993a). Do postural responses to transient and continuous perturbations show similar vision and amplitude dependence? Journal of Biomechanics 26, 1181-1190.

Maki, B.E. and Ostrovski, G. (1993b). Scaling of postural responses to transient and continuous perturbations. Gait & Posture 1, 93-104.

Maki, B.E., and McIlroy, W.E. (1999). The control of foot placement during compsensatory stepping reactions: does speed of response take precedence over stability? IEEE trans Rehabil Eng. 7, 80-90.

Maki, B.E., Edmondstone, M.A., and McIlroy, W.E. (2000). Age-related differences in laterally directed compensatory stepping behavior. Journal of Gerontology: Medical Sciences 55A, M270-M277.

Maki, B.E., Holliday, P.J., Fernie, G.R. (1987). A posture control model and balance test for the prediction of relative postural stability. IEEE Transactions on Biomedical Engineering 10, 7971-810.

Maki, B.E., McIlroy, W.E., and Perry, S.D. (1996). Influence of lateral destabilization on compensatory stepping responses. J. Biomechanics 29, 343-353.

Manfre, M.J., Yu, G.H., Varma, A.A., Mallis, G.I., Kearney, K., Karageorgis, M.A. (1994). The effect of limited handrail support on total treadmill time and the prediction of VO2 max. Clin. Cardiol.17,445–450.

Mansfield, A., Peters, A.L., Liu, B.A. and Maki, B.E. (2007). A perturbation-based balance training program for older adults: study protocol for a randomised controlled trial. BMC Geriatrics 7, 1-14.

McAndrew, P.M., Dingwell, J.B. and Wilken, J.M. (2010). Walking variability during continuous pseudo-random oscillations of the support surface and visual field. Journal of Biomechanics 43, 1470-1475.

McIlroy, W.E. and Maki, B.E. (1993). Do postural adjustments precede compensatory stepping reactions evoked by perturbation? Neuroscience Letters 164, 199-202.

Melzer, I., Benjuya, N. and Kaplanski, J. (2004). Postural stability in the elderly: a comparison between fallers and non-fallers. Age and Ageing 33, 602–607.

Mergner, T., Schweigart, G., Maurer, C. and Blumle, A. (2006). Human postural responses to motion of real and virtual visual environments under different support base conditions. Experimental Brain Research 167, 535-556.

Miwa, T. (1967). Evaluation methods for vibration effect. Part 1. Measurements of the threshold and equal sensation contours of whole body for vertical and fore and aft vibrations. Industrial Health 5, 183-205.

Moe-Nilssen, R. and Helbostad, J.L. (2005). Interstride trunk acceleration variability but not step-width variability can differentiate between fit and frail older adults. Gait Posture 21, 164-170.

Morioka, M. and Griffin, M. J. (2006). Magnitude-dependence of equivalent comfort contours for fore-and-aft, lateral and vertical whole-body vibration. Journal of Sound and Vibration 298, 755-772.

Murray, M.P., Kory, R.C. and Clarkson, B.H. (1969). Walking patterns in healthy old men. Journal of Gerontology 24, 169-178.

Murray, M.P., Seireg, A. and Scholz, R.C. (1967). Centre of gravity, centre of pressure and supportive forces during human activities. Journal of Applied Physiology 23, 831-838.

Nashner, L.M. (1980). Balance adjustments of humans perturbed while walking. Journal of Neurophysiology 44, 650-664.

Nashner, L.M. (1986). Organization of human postural movements during standing and walking. Neurobiology of Posture and Locomotion, London, MacMillan, 637-648.

Nashner, L.M., Woollacott, M. and Tuma, G. (1979). Organization of Rapid Responses to Postural and Locomotor-Like Perturbations of Standing Man. Exp. Brain Research 36, 463-476.

Nawayseh, N. and Griffin, M.J. (2006). Effect of frequency, magnitude and direction of translational and rotational oscillation on the postural stability of standing people. Journal of Sound and Vibration 298, 725-754.

O'Connor, S. and Kuo, A.D.. (2009). Direction dependent control of balance during walking and standing. Journal of Neurophysiology 102, 1411-1419.

Oborne, D. J. and Boarer, P. A. (1982). Variability in human response to whole-body vibration The effects of instructions. Ergonomics 25, 759-769.

Oborne, D. J. and Clarke, M. J. (1974). The determination of equal comfort zones for whole-body vibration. Ergonomics 17, 769-782.

Oddsson, L.I.E., Wall III, C., McPartland, M.D., Krebs, D.E. and Tucker, C.A. (2004). Recovery from perturbations during paced walking. Gait and Posture 19, 24-34.

Overstall, P.W., Exton-Smith, A.N., Imms, F.J. and Johnson, A.L. (1977). Falls in the elderly related to postural imbalance. Br Med J. 1, 261-264.

Owings, T.M. and Grabiner, M.D. (2004) .Step width variability, but not step length variability or step time variability, discriminates gait of healthy young and older adults during treadmill locomotion. Journal of Biomechanics 37, 935-938.

Pai, Y-C. and Iqbal, K. (1999). Simulated movement termination for balance recovery: can movement strategies be sought to maintain stability in the presence of slipping or forced sliding? J. Biomech. 32, 779-786.

Patla, A.E., Frank, J.S., Winter, D.A., Rietdyk, S., Prentice, S., and Prasad, S. (1993). Age-related changes in balance control system: Initiation of stepping. Clinical Biomechanics 8, 179-184.

Pavol, M.J., Owings,T.M., Foley, K.T. and Grabiner, M.D. (1999). Gait characteristics as risk factors for falling from trips induced in older adults. J Gerontol A Biol Sci Med Sci. 54, M583-M590.

Peterka, R.J. (2002). Sensorimotor integration in human postural control. Journal of Neurophysiol. 88, 63-58.

Peterka, R.J. (2003). Simplifying the complexities of maintaining balance. IEEE Engineering in Medicine and Biology Magazine 22, 63-68.

Pheasant, S. (1988). Bodyspace: anthropometry, ergonomics and design. Taylor & Francis, London.

Prieto, T.E., Myklebust, J.B. and Myklebust, B.M. (1993). Characterization and modeling the postural steadiness in the elderly: a review. IEEE Transaction on Rehabilitation Engineering 1, 26-34.

Prieto, T.E., Myklebust, J.B., Hoffmann, R.G., Lovett, E.G., Myklebst, B.M. (1996). Measures of postural steadiness: differences between healthy young and elderly adults. IEEE Trans Biomed Enfr 43, 956-966.

Prince, F., Winter, D.A., Stergiou, P. and Walt, S.E. (1994). Anticipatory control of upper body balance during human locomotion. Gait and Posture 2, 19-25.

Redfern, M.S. and Schumann, T. (1994). A model of foot placement during gait. Journal of Biomechanics 27, 1339-1346.

Reiher, H. and Meister, F. J. (1931). Die Empfindlichkeit des Menschen gegen Erschtterungen (The sensitiveness of the human body to vibrations, The effect of vibration on people). Forschung 2, 381-386. English Translation: Report No. F-TS-616-RE, Headquarters Air Material Command, Wright Field, Ohio, 1946.

Riley, P.O., Mann, R.W. and Hodge, W.A. (1990). Modeling of the biomechanics of posture and balance. Journal of Biomechanics 21, 591-600.

Rubenstein, L.Z. and Trueblood, P.T. (2004). Gait and Balance Assessment in Older Persons. Annals of Long-Term Care12, 39-45.

Runge, C.F., Shupert, C.L., Horak, F.B. and Zajac, F.E. (1999). Ankle and hip postural strategies defined by joint torques. Gait and Posture 10, 161-170.

Sari, M. and Griffin, M.J. (2009). Subjective assessment of the postural stability of walking subjects exposed to lateral vibration. 44th UK Conference on Human responses to Vibration, 7-9 September 2009, Loughborough University, England.

Shumway-Cook, A. and Woollacott, M.H. (2000). Attentional demands and postural control: the effect of sensory context. J. Gerontol. 55, M10–16.

Shumway-Cook, A., Woollacott, M., Kerns, K.A. and Baldwin, M. (1997). The effects of two types of cognitive tasks on postural stability in older adults with and without a history of falls. J. Gerontol Med Sci. 52A, M232–40.

Stelmach, G.E., Zelaznik, H.N. and Lowe, D. (1990). The influence of aging and attentional demands on recovery from postural instability. Clin Exp Aging, 2, 155–61.

Stevens, S.S. (1975). Psychophysics: introduction to its perceptual, neural, and socialprospects. Oxford, Transaction Publishers.

Suzuki, H., Shiroto, H., Nakagawa, C., Saito, A. and Ohno, H. (2006). Development and utilization of comfort simulator. QR of RTRI 47, 205-210.

Tang, P-F., Woollacott, M.H. and Chong, R.K.Y. (1998). Control of reactive balance adjustments in perturbed human walking: roles of proximal and distal postural muscle activity. Experimental Brain Research 119, 141-152.

Teasdale, N. and Simoneau, M. (2001). Attentional demands for postural control: the effects of aging and sensory reintegration. Gait and Posture 14, 203–210.

Thuong, O and Griffin, M.J. (2010a). The vibration discomfort of standing persons: the effect of body supports. Proc. IMechE, 224, Part F: J. Rail and Rapid Transit, JRRT369.

Thuong, O. and Griffin, M.J. (2010b). Discomfort of standing people exposed to random and transient lateral vibration. 45th United Kingdom Conference on Human Response to Vibration, Institute of Naval Medicine, Alverstoke, Gosport, England, 6-8 September 2010, 181-195.

Thuong, O. and Griffin, M. J. (2011). The vibration discomfort of standing persons: 0.5-16-Hz fore-and-aft, lateral, and vertical vibration. Journal of Sound and Vibration 330, 816-826.

Tinetti, M.E., Speechley, M. and Ginter, S.F. (1988). Risk factors for falls among elderly persons living in community. New England Journal of Medicine 319, 1071–1077.

Tinetti, M.E., Williams, T.F. and Mayewski, R. (1986). Fall risk index for elderly patients based on number of chronic disabilities. American Journal of Medicine 80, 429–434.

Tjernström, F., Fransson, P.A., Hafström, A. and Magnusson, M. (2002). Adaptation of postural control to perturbatuions-a process that initiates long-term motor memory. Gait Posture 15, 75-82.

Townsend, M. A. (1985). Biped gait stabilization via foot placement. Journal of Biomechanics 18, 21-38.

Tremblay, F., Mireault, A-C., Dessureault, L., Manning, H. and Sveistrup, H. (2004). Postural stabilization from fingertip contact: I. Variations in sway attenuation, perceived stability and contact forces with aging. Exp. Brain Research 157, 275-285.

Tucker, C.A., Ramirez, J., Krebs, D.E. and Riley, P.O. (1998). Centre of gravity dynamic stability in normal and vestibulopathic gait. Gait and Posture 8, 117-123.

Tucker, M.G., Kavanagh, J.J., Barrett, R.S. and Morrison, S. (2008). Age-related differences in postural reaction time and coordination during voluntary sway movements. Human Movement Science 27, 728-737.

van der Kooij, H., Jacobs, R. and van der Helm, F. (2001). An adaptive model of sensory integration in a dynamic environment applied to human stance control. Biological Cybernetics 84, 103-115.

Wagenaar, R.C. and Beek, W.J. (2000). Hemiplegic gait: kinematic analysis using walking speed as basis. Journal of Biomechanics 25, 1007-1015.

Westling, G. and Johansson, R.S. (1987). Responses in glabrous skin mechanoreceptors during precision grip in humans. Exp Brain Res 66,128-140.

Winter, D. A. (1995). Human balance and posture control during standing and walking. Gait and Posture 3, 193-214.

Winter, D.A. (1992). Foot trajectory in human gait: a precise and multifactorial motor control task. Physical Therapy 72, 45-56.

Winter, D.A., Patla, A.E., Ishac, M. and Gage, W.H. (2003). Motor mechanisms of balance during quiet standing. Journal of Electromyography and Kinesiology 13, 49–56.

Wolfson, L., Whipple, R., Derby, C.A., Amerman, P., Murphy, T., Tobin, J. N., and Nashner, L. (1992) A dynamic posturography study of balance in healthy elderly. Neurology 42, 2069-75.

Woollacott, M.H. and Tang, P. (1997). Balance control during walking in the older adult: Research and its implications. Physical Therapy 77, 646-660.

Wyllie, I. H. and Griffin, M. J. (2007). Discomfort from sinusoidal oscillation in the roll and lateral axes at frequencies between 0.2 and 1.6 Hz. The Journal of the Acoustical Society of America 121, 2644-2654.

You, J-Y., Chou, Y-L., Lin, C-J. and Su, F-C. (2001). Effect of slip movement of body centre of mass relative to base of support. Clinical Biomechanics16, 167-173.

Yutaka, F., Nagata, T., Ishida, A. and Minamitani, H. (2001). Characteristics of somatosensory feedback in postural control during standing. IEEE Transactions on Neural systems and Rehabilitation Engineering 9, 145-153.