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VISUOMOTOR CONTROL OF STEP DESCENT: THE IMPORTANCE OF VISUAL INFORMATION FROM THE LOWER VISUAL FIELD IN REGULATING LANDING CONTROL

When descending a step from a stationary standing position or during on-going gait, is online visual information from the lower visual field important in regulating prelanding kinematic and landing mechanic variables?

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Abstract

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Visuomotor control of step descent: the importance of visual information from the lower visual field in regulating landing control

Keywords: biomechanics, human movement, locomotion, adaptive gait, step descent, vision, lower visual field, online, feedforward.

The majority of previous research investigating the role of vision in controlling adaptive gait has predominantly focused on over-ground walking or obstacle negotiation. Thus there is a paucity of literature investigating visuomotor control of step descent. This thesis addressed the importance of the lower visual field (lvf) in regulating step descent landing control, and determined when visual feedback is typically used in regulating landing control prior to / during step descent.

When step descents were completed from a stationary starting position, with the lvf occluded or degraded, participants adapted their stepping strategy in a manner consistent with being uncertain regarding the precise location of the foot / lower leg relative to the floor. However, these changes in landing control under conditions of lvf occlusion were made without fundamentally altering stepping strategy. This suggests that participants were able to plan the general stepping strategy when only upper visual field cues were available. When lvf was occluded from either 2 or 1 step(s) prior to descending a step during on-going gait, stepping strategy was only affected when the lvf was occluded in the penultimate step. Findings suggest that lvf cues are acquired in the penultimate step / few seconds prior to descent and provide exproprioceptive information of the foot / lower leg relative to the floor which ensures landing is regulated with increased certainty. Findings also highlight the subtle role of online vision used in the latter portion of step descent to ‘fine tune’ landing control.

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Chapter 1

Introduction

Step descent

1.1 Falls during step / stair negotiation

Normal everyday locomotion encompasses a variety of adaptive gait tasks, which include negotiation of obstacles, irregular terrain, kerbs, steps and / or stairs. With falls on stairs being one of the most frequent causes of falls occurring during locomotion (Startzell *et al.,* 2000), step / stair negotiation is perhaps the most challenging of locomotor tasks encountered in normal everyday life. In the United Kingdom (UK), over 500 deaths and 230,000 accidents result from falls on stairs each year (Consumer Safety Unit DTI, 1997). During 1998, 10,700 people in the United States died as a result from falls on stairs (National Safety Council, 1999). Falls during step / stair negotiation occur in sideways (Greenspan *et al.,* 1998) and / or forward (Roys, 2001) directions. Step descent is inherently more dangerous than step ascent, with accidents reported to occur approximately three times more frequently (in the elderly) and usually resulting with more serious injuries (Tinetti *et al.,* 1988; Templer, 1992; Startzell, 2000; Roys, 2001). Accidents are usually more serious during step descent compared to step ascent because a fall forward during step descent will likely result in falling down the entire flight of stairs. However, a fall forward during step ascent will likely result in falling onto the steps, a fall of approximately 0.5 m, which can be attenuated by the individual putting their hands out in front of them. One potential explanation for the increased risk of falling during step descent compared with step ascent is attributed to reduced stability during stepping down (Zachazewski *et al.,* 1993; Stacoff *et al.,* 1995). Indeed, during step initiation, the centre of pressure (CP) initially moves backward and laterally toward the swing limb. This causes the centre of mass (CM) to move forward and laterally toward the stance limb, subsequently allowing the swing limb to be lifted from the floor. The divergence between CM and CP is larger during step / stair descent compared with step / stair ascent, resulting in increased instability and a higher chance of falling (Zachazewski *et al.,* 1993). In addition, during step / stair descent, the majority of single support time (where only one limb is in contact with the ground) is spent with only the metatarsal heads and toes of the supporting foot in contact with the surface of the step, resulting with a relatively small base of support during the critical period when body weight is supported on just one limb (Roys, 2001). During the period of single support, the trail limb also requires sufficient lower limb strength to control and support the lowering of the whole-body CM while moving forwards and downwards (Lark *et al.,* 2003).

At contact with the lower level, the lead limb has to arrest forward and downward momentum of the body through absorbing the kinetic energy by means of eccentric contractions (Buckley *et al.,* 2005). If the downwards momentum is not attenuated during descent and / or initial landing, the kinetic energy may compromise dynamic stability (Hof *et al*., 2005), resulting in the individual falling forwards down the step / stairs (Roys, 2001). Positioning of the lead foot on the lower level is thus a crucial aspect of step descent, since this determines the quality of the base of support for the ensuing weight-bearing phase (Simoneau, 1991). Predicting when and where contact between the landing limb and the lower level will be made is therefore a critical factor in successful descent (Buckley *et al.*, 2005). The ability to accurately perceive the location of the lower surface relative to the foot is heavily dependent on visual cues obtained from the environment (Buckley *et al.*, 2005, 2008; Cowie *et al.,* 2008). These visual cues assist the nervous system by providing some information about initiating a change in action (Treselian, 1999). For this reason, visual impairment has been strongly associated with an elevated risk of falling (Lord *et al.,* 1991a; Ivers *et al.,* 1998).

1.2 The role of vision in controlling step descent

Worldwide it is estimated that 135 million people are severely visually impaired (Thylefors *et al.,* 1995; Thylefors, 1999), and visual impairment is a well known risk factor for falls in older people (Lord, 2006). In a cross-sectional survey on 3,299 adults (aged 49 years and above), poor visual acuity, reduced visual field, impaired contrast sensitivity and the presence of cataract were strongly associated with falling over a recorded 12-month period (Ivers *et al.,* 1998). In addition, reduced contrast sensitivity was associated with the differences between the number of times older adults (mean age 83 years) fell and did not fall when questioned every month over a one year period (Lord *et al.,* 1991a). Several studies have also concluded that visual impairment is not directly related to falling in the elderly (Campbell *et al.,* 1981; Tinetti *et al.,* 1988).

Campbell *et al.* (1981) highlighted a number of variables in both men and women (aged 65 years and above) associated with an increased risk of falling, including disorders with gait, impaired mental function and use of psychotropic drugs. However, visual impairment was not associated with increased number of falls. Tinetti *et al.* (1988) also highlighted a number of pre-disposing factors for falls in older adults (aged 75 years and above), with the risk of falling not associated with one specific factor such as visual impairment, rather increasing linearly with the number of risk factors identified. However, the aforementioned research (Campbell *et al.,* 1981; Tinetti *et al.,* 1988) only measured one aspect of vision, visual acuity. Measuring visual acuity over-estimates visual function in the 'real-world' because this is not representative of how a person views the world (Helbostad *et al.,* 2009). Indeed, the surrounding environment is not just made up of high contrast, sharp edged objects, rather it is composed of objects with a variety of contrasts, many of which would likely be defined as low contrasts. It is therefore necessary for other clinical visual assessment tests to be completed in addition to visual acuity, when determining the role of visual impairment in falling (Helbostad *et al.,* 2009).

Even when young healthy adults descend steps / stairs, the ability to accurately perceive the precise instant of contact on the lower level is heavily dependent on visual mechanisms (Buckley *et al.,* 2008; Cowie *et al.,* 2008). However, during normal everyday locomotion we often successfully complete step downs in the absence of some / all visual information. For example, when stepping down from a kerb onto the road we typically tend to look if there is any oncoming traffic, subsequently not receiving any visual information about the kerb edge and immediate lower ground area (Geruschat *et al.,* 2003, 2006). In this situation, visual information from the lower visual field (lvf) pertaining to the kerb edge and lower level are likely acquired several steps prior to reaching the kerb edge. This suggests that step descents can be completed successfully without receiving online visual information from the lvf. However, when we do not accurately perceive the height of the lower level we intend to step down onto and landing occurs unexpectedly, we are unable to effectively pre-programme muscle activity and positioning of lower extremity joints (McFadyen and Winter, 1988) which results in a relatively large shock (reaction) force being generated at instant of contact, potentially leading to a fall (Berg *et al.,* 1997) or injury. It is therefore unclear whether visual cues acquired from the lvf immediately prior to / during step descent provide any advantage in terms of regulating landing control. Either visual information from the lvf is gained several seconds prior to descending the step, or visual information from the lvf is required during some / all of the descent to control landing. It is also possible that lvf information is not required during step descent, rather, we utilise visual information from areas of the upper visual field (uvf) to control landing. Presently, it also remains unclear when full field visual information is required prior to / during step descent to ensure landings are safely controlled. Either visual information acquired prior to step descent is used by feedforward visual mechanisms (visual cues acquired in advance of the on-going movement) to control subsequent landing, and / or online visual mechanisms (visual cues acquired concurrently during the on-going movement) are required during some / all of the descent to regulate landing control.

1.3 Purpose of the thesis

Despite the potentially hazardous situations individuals face when descending step / stairs, there is a paucity of literature concerning the role of visuomotor control of step descent. This thesis will therefore determine the importance of the lvf in regulating step descent landing control. In addition, this thesis will determine when prior to / during step descent visual feedback is most important in regulating landing control. To achieve these aims, the specific objectives of the thesis are to:

1) Determine when during step descent visual information is customarily used to control landing.

2) Investigate whether descending a step carrying added mass affects the role of feedforward versus online vision used to control landing.

3) Determine what advantage visual cues acquired from the lower visual field immediately prior to / during step descent provide in terms of regulating landing control.

4) Determine when visual cues are acquired from the lower visual field to regulate step descent landing control.

5) Investigate whether the level of awareness and experience pertaining to a potential visual occlusion during step descent affects landing control.

6) Determine whether the probability of visual occlusion during step descent affects landing control.

1.4 Thesis outline

Prior to investigating the specific objectives of this thesis, a review of the literature is provided (chapter 2). This chapter includes the role of the lvf in the control of locomotion and when visual information is sampled for such control. The methodology chapter (chapter 3) outlines the experimental set up and general techniques used within this thesis. The first experimental chapter (chapter 4: Evidence of a specialised role of the lower visual field in regulating step descent landing control), determines whether lvf information provides any advantage in terms of regulating step descent landing control. In addition, this experimental chapter determines when during step descent visual information is customarily used to regulate landing control. Through occluding visual cues from the lvf, it is possible to determine whether the lvf provides any advantage in regulating landing control. Furthermore, by manipulating the availability of visual information at specific times relevant to step initiation, it is possible to determine when during step descent visual information is customarily used to regulate landing control.

The second experimental chapter (chapter 5: Does the role of feedforward versus online vision used in the control of step descent change when descents are completed carrying added mass?), determined whether the role of feedforward versus online vision used in the control of step descent changes when descents are completed carrying added mass compared to no added mass. This was determined through completing step descents with / without carrying added mass and occluding vision from either, immediately prior to movement initiation or mid-swing onwards during the descent.

Previous research has highlighted the problems multifocal spectacle-wearers face when negotiating steps, stairs and surface height changes (Johnson *et al.,* 2007, 2008; Menant *et al.,* 2009). This is likely because multifocal wearers view a step / stair edge through the lower region of the lens designed for reading, resulting in the lvf including the view of the surface height change and the foot being degraded (i.e. blurred and / or magnified), and thus the exact and relative height of the floor is difficult to judge. Therefore, the third experimental chapter (chapter 6: Use of single-vision distance spectacles improves landing control during step descent in well-adapted multifocal lens-wearers), determined whether step descent landing control in older long-term multifocal wearers is improved when they switch to using single-vision distance lenses, since distance vision in their lvf is improved.

The fourth experimental chapter (chapter 7: Does the probability, awareness and / or experience of visual occlusion during step descent affect the role of online vision in regulating landing control?), determined whether the probability of visual occlusion during step descent affects the role of online vision in regulating step descent landing control. In addition, this experimental chapter examined whether awareness and / or prior experience concerning a potential visual occlusion during step descent affects the utility of online vision in controlling subsequent landing. By conducting two separate experiments, it was possible to determine whether a low (16.7 %, experiment 1) or high (67 %, experiment 2) probability of visual occlusion during step descent affects how participants plan to use online vision to control subsequent landing. Furthermore, by providing participants with increasing levels of awareness and experience concerning a potential visual occlusion (experiment 1), it was also possible to determine their effects on how participants plan to use online vision during step descent to control landing.

The fifth and final experimental chapter (chapter 8: When is lower visual field information acquired to control landing when descending a step during on-going gait?), examined when, prior to step descent during on-going gait, visual cues from the lvf were acquired to regulate landing control. By occluding the lvf from either the penultimate or final step prior to step descent, it was possible to determine when such visual cues were acquired to regulate step descent landing control.

The concluding chapter (chapter 9: Conclusions and future work), summarises the main findings of the series of experiments presented in this thesis. Limitations of the thesis are also presented. Furthermore, recommendations are made for the future in order to gain further insight into visuomotor control of step descent.

Chapter 2

Literature review

Visual control of locomotion

The following chapter will initially present a brief overview of the visual system and progress to provide an extensive review of the literature pertaining to visual control of locomotion. Specifically, this chapter will outline the anatomy and physiology of the visual system and highlight the importance of both eyes simultaneously viewing the visual scene to provide binocular vision. The chapter will also provide a comprehensive overview of 'what' and 'when' visual information is used to control locomotion. Insight will also be provided into the common methods applied in a clinical setting to determine a 'normal' fully functioning eye. This chapter will conclude by providing a rationale for the kinematic modelling techniques employed within the subsequent experimental chapters.

Prior to outlining the anatomy and physiology of the visual system, it is important to acknowledge that movement control is a complex process that involves modification of motor output from multi-sensory information provided by visual, vestibular and somatosensory systems. The importance of receiving multi-sensory information for movement control has been previously highlighted in postural control (Elliott *et al.,* 1995; Anand *et al.,* 2003; Turano *et al.,* 2004). For example, when participants' postural instability significantly increased when vision was degraded, instability further increased with perturbation of the somatosensory and / or vestibular systems (Elliott *et al.,* 1995; Anand *et al.,* 2003; Turano *et al.,* 2004). The somatosensory system includes various receptors that provide information about pressure distribution (cutaneous), muscle tension (Golgi tendon organs), joint angle changes (joint receptors) and muscle length changes (spindles). Cues from the vestibular system provide changes in angular and translational head movements and head position. All of these various cues contribute to the control of locomotion. However, the remainder of this chapter will only focus on the visual contribution to locomotion.

2.1 Anatomy and Physiology of the visual system

Vision results from rays of light which enter the eye being refracted and subsequently converging on the retina (see figure 2.1). The refraction of light as it passes through the eye allows images presented at distance to be precisely focused on the retina. If rays of light converge in front or behind the retina, the image will be blurred and unclear. In a fully functioning eye, rays of light are focused precisely on the retina through cillary muscles located around the margin of the lens contracting and relaxing, thus changing the shape of the lens (Bear *et al.,* 2007).

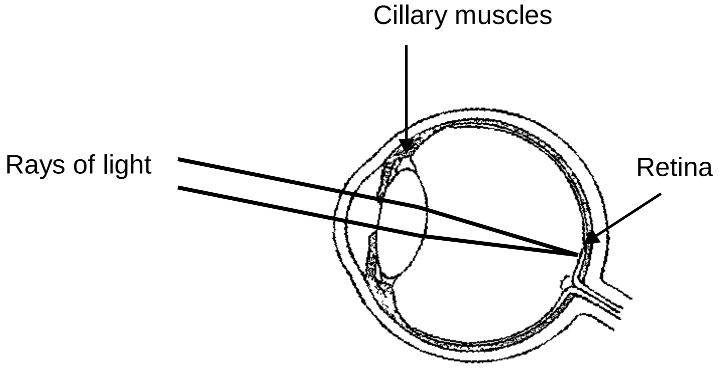

Figure 2.1. The refraction of rays of light passing through the eye ensure the image is focused precisely on the retina (adapted from Color and Light, 2010).

2.1.1 The retina

The role of the retina is to act as a medium for transforming information from light energy into neural energy. As light falls on the retina, it is converted into neural activity by photoreceptor cells.

2.1.2 Photoreceptors

Photoreceptor cells are located throughout the retina with increased density in the fovea (centre of the retina) and lvf (Curcio *et al.,* 1987). With increased density of photoreceptor cells located in the fovea, this subsequently increases the detail of the sampled visual information (Banks *et al.,* 1991). A retina contains over 100 million photoreceptor cells which are responsible for converting light energy into neural activity through producing chemical changes that start a cascade of neural events ending in visual sensation (Bear *et al.,* 2007). Photoreceptor cells are long, thin tubes consisting of an outer segment that contains light sensitive elements, and an inner segment which forms the cell body. The outer segment of the photoreceptor contains ion channels affected by the absorption of light resulting in hyperpolarisation, whereas the inner segment is responsible for sustaining constant electric-current flow around the photoreceptor (Oyster, 1999). Photoreceptors can be categorised into two different classes; rods and cones. Both are densely distributed in the retina, with their long axes parallel to the direction of incident light (Mather, 2006). Rods and cones are differentiated by the shape of their outer segments (see figure 2.2) and their different response properties to visual stimulus. Rod photoreceptor cells perceive images in black and white, respond to movement in the periphery, sample images in low resolution and respond during low levels of light. Cone photoreceptor cells perceive images in colour, in fine detail, sample images that are not moving and are most responsive in high levels of light (Hamel, 2007). Rod photoreceptor cells respond during low levels of light due to their pigment being highly sensitive, whereas cone photoreceptor cells are less sensitive, thus responding during higher levels of light (Wandell, 1995). Due to the poor spatial and temporal characteristics of rod photoreceptors, many aspects of visual performance deteriorate under reduced lighting conditions (Plainis *et al.,* 2005). This has been evidenced through reduced, spatial resolution (Arumi *et al.,* 1997), contrast discrimination (Arend, 1993), accommodation response (Charman, 1996) and increased visual reaction time (Plainis and Murray, 2002). Furthermore, because rod photoreceptor pathways have a slower response time than cone photoreceptor pathways, the ability to respond to rapidly changing viewing conditions at night is significantly impaired (Plainis *et al.,* 2005).

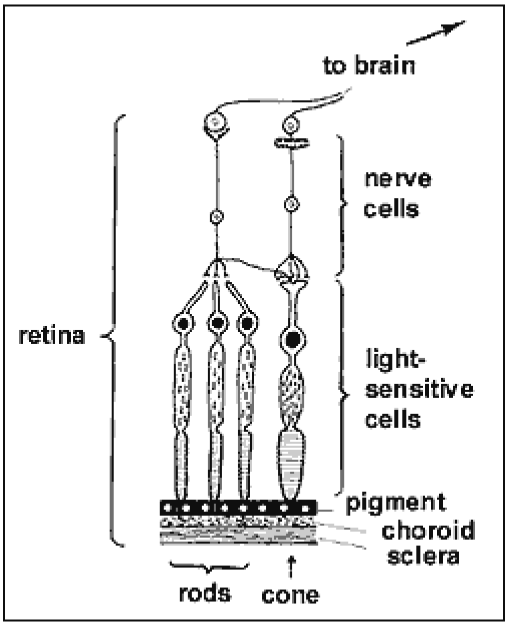

Figure 2.2. Anatomical differences between photoreceptor rods and cones (The human eye, 2010).

2.1.3 Horizontal, bipolar, amacrine and ganglion cells

The connections from the synaptic terminals of the photoreceptors synapse onto the dendritic fields of the bipolar cells and horizontal cells in the outer plexiform layer of the complex retinal circuitry. The main purpose of horizontal cells is to integrate and regulate the input relayed from photoreceptor cells to bipolar cells (Mather, 2006). All visual signals pass through the bipolar cells, as this provides the only link between the outer and inner plexiform layers (Wandell, 1995). There are different types of bipolar cells (midget and diffuse) which determine the information that is relayed from the photoreceptors to the ganglion cells (Bear *et al.,* 2007). Midget bipolar cells are responsible for clarity of the image on the retina, receiving single inputs from cone photoreceptor cells (Wolfe *et al.,* 2006). In contrast, diffuse bipolar cells receive input from a large number of photoreceptor cells where the information is pooled. Pooling of visual information does not occur in the forvea, but in the periphery (Curio and Allen, 1990). With fewer photoreceptor cells located in the periphery, pooling visual information increases visual sensitivity before transmitting a signal to the ganglion cell (Banks *et al.,* 1991).

Bipolar cells are also sub-divided into ‘On bipolars’ and ‘Off bipolars’ of roughly equal portions (Rodieck, 1998). The On and Off bipolars have two distinct pathways which are shared, one pathway for increases in light intensity and the other for decreases in light intensity (Oyster, 1999). The output of both On and Off bipolar cells are received by retinal ganglion cells. Light falling on the receptive field of both On and Off bipolar cells will produce an opposite response in either cell; an On bipolar cell will be inhibited and an Off bipolar cell excited. The opposite is observed under decreased illumination (Hansen *et al.,* 2009). In addition, moving stimuli perceived in the visual field also has an effect on the output of both On and Off bipolar cells. For example, movement perceived in the periphery without stimulation of the receptive field centre has been shown to excite retinal ganglion cells in cats (Noda and Adey, 1974; Fischer *et al.,* 1975; Barlow *et al.,* 1977). However, contrary to the aforementioned studies, recent research has also evidenced situations when movement perceived in the periphery produces inhibition of retinal ganglion cells in cats (Passaglia *et al.,* 2001). When a moving stimuli is perceived by both the periphery and receptive field centre, the cell’s sensitivity in the centre is generally suppressed (Enroth-Cugell and Jakiela, 1980; Cook *et al.,* 1998). This aforementioned research is highlighted to illustrate that depending on what visual stimuli is perceived (i.e. illumination of an object) and where in the visual field this is perceived, this will ultimately have a significant effect on how the visual system processes the stimuli.

Amacrine cells receive inputs from bipolar cells and other amacrine cells, and send signals to bipolar, amacrine and ganglion cells (Wolfe *et al.,* 2006). Amacrine cells serve many functions including modulating the signal carried by bipolar cells and ganglion cells over both short and long periods of time. As the visual signal reaches the retinal ganglion cells, it is converted into neural signals. The density of retinal ganglion cells peaks at about 1 mm from the centre of the fovea, with more peripheral regions showing substantially lower densities (Curcio and Allen, 1990). For example, in a horizontally orientated elliptical ring 0.4 - 2.0 mm from the foveal center, ganglion cell densities reach 32,000 - 38,000 cells / mm^2, which is more than a 300 % increase compared to some regions in the periphery (Curio and Allen, 1990). The reduction in ganglion cell density in the periphery has an impact on colour perception. Indeed colour perception is most accurate in the fovea and declines further towards the periphery (Hansen *et al.,* 2009). Whilst the accuracy of colour perception is diminished in the peripheral visual field, visual perception under reduced levels of luminance is less effected when perceived in the periphery compared to the central field (Mullen, 1991).

Ganglion cells can be classified into four major types based on their anatomical properties: biplexiform, bistratified, midget and parasol ganglion cells. Biplexiform cells are the only type that connects directly to photoreceptors.

Bistratified, midget, and parasol ganglion cells are all influenced directly by the connecting amacrine and bipolar cells. The precise role of these ganglion cells in visual processing remains unclear (Mather, 2006).

2.1.4 Retina to cortex - eye to brain

The previous sub sections in the ‘anatomy and physiology of the visual system’ section, highlighted how light passes through the eye from the retina and is converted into neural signals when reaching the retinal ganglion cells. Upon leaving the retina, the ganglion cells become individual fibres of the optic nerve. When the fibres of the optic nerve reach the optic chiasm, half the fibres from one eye cross over to the other section of the brain and the other half of the fibres remain on the same side of the brain. The split in optic fibres at the optic chiasm occurs according to the visual field in both eyes. This ensures that two neural maps of the same region of visual space project to the same side of the cortex (Bear *et al.,* 2007). From the optic chiasm, the nerve fibres continue through the optic tract to the lateral geniculate nucleus (LGN) where they synapse (Wolfe *et al.,* 2006). The neurons in the LGN then project to the primary visual cortex (located in the occipital lobe of the brain) through optic radiation (Bear *et al.,* 2007). There are four times as many LGN cells per ganglion cell from the fovea, relative to the periphery (Connolly & van Essen, 1984).

2.1.5 Lateral geniculate nuclei (LGN)

The LGN (one for each eye) is divided into six different neuronal layers which receives nerve fibres from both eyes (Palmer, 1999). The nerve fibres from the two eyes are kept separate within the LGN through projections to different surfaces within the LGN (Wolfe *et al.,* 2006). The upper four layers of the LGN have small cell bodies which are termed parvocellular, or P Layers. The two lower layers are larger cell bodies which are termed magnocellular cells, or M layers. Parvocellular cells respond to detail regarding colour and vision, whilst magnocellular cells are more sensitive to movement (Palmer, 1999). The projection from the LGN has a single major synaptic target, the primary visual cortex, however, a precise understanding of the LGN remains largely unknown (Mather, 2006).

2.1.6 The primary visual cortex / striate cortex

The highly folded sheet of nerve cells that forms the outer layer of the cerebral hemisphere is divided into a number of different layers. These layers run parallel to the cortical surface containing many thousands of cortical cells representing each small region of the retina (Mather, 2006). Different fibres from the LGN terminate in different layers of the primary visual cortex (Bear *et al.,* 2007). Other terms used to describe the primary visual cortex include the V1 and striate cortex. As illustrated in figure 2.3, V1 is only the first stage of visual processing. Currently over 30 identified cortical regions receive visual information, which are either connected exclusively to vision or contribute to vision (Felleman and van Essen, 1991). The ratio of V1 cells to incoming LGN projections is ten times larger for foveal than for peripheral vision (Connolly & van Essen, 1984).

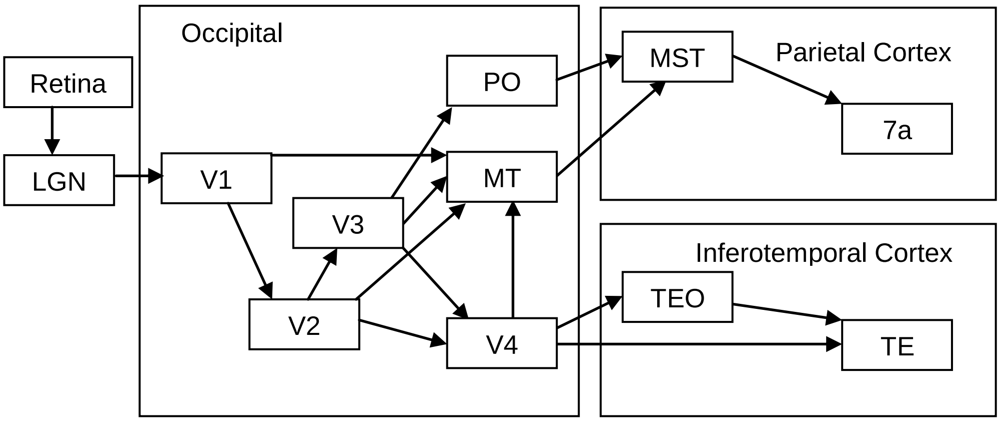

Figure 2.3. Diagrammatical representation of the different cortical areas within the brain, adapted from Vision (2010).

Similar to the LGN, the striate cortex is divided into six layers, which are numbered from 1 to 6. Layer 1 can be found nearest the surface and layer 6 furthest from the surface (Mather, 2006). In addition to the 6 layers, layer 4 is further sub-divided into separate sub-layers called 4A, 4B, 4C∞ and 4Cβ (Mather, 2006). There are a great number of interconnections between the different cortical layers within the striate cortex. A large number of connections can be found between the cells in layer 4 and cells in layers 2 and 3. A large number of connections have also been identified from the superficial layers 2, 3 and 4 down to layers 5 and 6 (Bear *et al.,* 2007). As well as the abundance of interconnections within the striate cortex, fibres also project to several other areas within the brain. Cells in layers 2, 3 and 4B project to the extrastriate cortex, whereas cells in layer 5 project to midbrain structures where the superior colliculus and pons are situated (Mather, 2006).

Within the brain, there appears to be two different levels for the awareness of objects (Goodale *et al.,* 2004). First, there is a low-level mechanism in the colliculus which is responsible for locating visual objects and translating that information into commands that need to be sent to the brainstem to direct eye movements to the objects. Second, there is a higher level mechanism called the cerebral cortex and basal ganglia that 'gives permission' to the colliculus to initiate a saccade only when the target is judged to be worth looking at (Goodale *et al.,* 2004). Part of the basal ganglia cells called the substantia nigra pars reticulata contains neurons which fires tonically most of the time at a high frequency, inhibiting the movement of cells of the colliculus and preventing them from responding to the presence of a visual target. When we are about to make a saccade, these neurons cease firing (temporarily) allowing the saccade to be made (Goodale *et al.,* 2004).

2.1.7 Extrastriate visual cortex

Beyond the striate cortex, a large number of visually responsive cortical areas have been identified due to topological organisation, anatomical connections, and cell response properties. Despite their identification, the functional significance of many cortical areas remains relatively unknown (Mather, 2006). There are two cortical streams of visual processing, one located dorsally from the striate cortex near the parietal lobe, and the other located ventrally near the temporal lobe (Wolfe *et al.,* 2006). Discussion of the role of both dorsal and ventral streams in movement control is provided below (2.1.8 Pathways for perception and action). The area V5 (or MT) located along the dorsal stream has the highest portion of motion and disparity selective cells which are used for visual movement and direction (Mather, 2006). Area V4 is located along the ventral stream and has a higher proportion of colour selective cells than other areas in the brain. Interestingly, the brain has two separate areas for processing colour which has been attributed to the concept colour constancy (Eysenck and Keane, 2005). Colour constancy refers to the manner in which the colour of a surface is perceived when illuminated in different lighting conditions. Even though the physical wavelength of the composition of light reflected from a particular surface can be shown to differ under different conditions, the same object can be recognised across a generic area (Eysenck and Keane, 2005).

A small number of retinal ganglion cells from the optic chiasm project to the superior colliculi (part of the midbrain tectum), which is involved in calculating the direction of visual objects in space and using this information to generate appropriate orienting movements of the eyes and head (Mather, 2006). Neurons in the colliculus have been suggested to fall into two types based on their response properties. These are neurons located near the surface and those located deeper in the colliculus. Neurons near the surface consist of afferent fibres which code the visual location of objects, whereas neurons deeper in the colliculus, projecting down into the brain stem code the movement (Carpenter, 1992). It is also believed that the superior colliculi is involved in integrating visual and auditory signals (Wolfe *et al.,* 2006). Additional projections from the optic chiasm synapse at the hypothalamus which regulates a variety of biological rhythms including the biological clock, and at the Edinger-Westphal nucleus which regulates pupil diameter (Bear *et al.,* 2007).

2.1.8 Pathways for perception and action

Traditionally it was viewed that the computations which use a variety of object parameters to generate the appropriate movements of individual limb(s), when stepping over an obstacle, or when reaching and grasping an object, are processed through a visual stream separate to the stream processing conscious perception of the same object (Goodale and Haffenden, 1998). Ungerleider and Mishkin (1982) first proposed that the dorsal stream processes the location (perception) of an object, while the ventral stream processes other visual attributes (characteristics of the object) including size, shape and colour. These distinct streams were first suggested through Ungerleider and Mishkin's (1982) cerebral cortex work on Macaque monkeys. Lesion studies of the monkey's inferior temporal cortex (damaging the ventral stream) profoundly impaired their ability to discriminate between objects on the basis of visual feature, however they were clearly able to use this information to direct their grasping movements. When testing monkeys with posterior parietal lesions (damaging the dorsal stream), the direct opposite was found; they demonstrated an ability to accurately discriminate between objects on the basis of visual feature, but were unable to use this information to direct their grasping movements (Goodale and Humphrey, 1998). If indeed these results are correct, it appears that the ventral stream plays a specialised role in the identification of objects, whereas the dorsal stream is responsible for locating objects in visual space.

The sub division in parocellular and magnocellular segregation in the primary visual cortex, V2, V4 and middle temporal area is not as clear as previously thought, which has led to the current suggestion that both the dorsal and ventral streams each receives input from mango and parvo pathways (Goodale and Milner, 1992). These findings caused Goodale and Milner (1992) to re-interpret Ungerleider and Mishkin's (1982) proposal by placing less emphasis on the difference in visual information each visual stream receives. Rather, Goodale and Milner (1992) suggested that both visual streams process information about object features and their spatial location but each stream uses the visual information in different ways. They suggested that the ventral stream is responsible for building up a representation of the surrounding world, which includes the identities and properties of objects and other creatures in the environment. In the dorsal stream moment-to-moment information about locations of objects are obtained allowing the control of skilled actions (Milner and Goodale, 1995). Consistent differences in task performance have also been identified in the dorsal and ventral streams governing perception and visually guided motor-based action. For example, when using the Titchener circles illusion (two target circles of equal size are each surrounded by a circular array of either smaller or larger circles), participants usually report that the target circle surrounded by the array of smaller circles appears to be larger than the target surrounded by larger circles. Indeed Aglioti *et al.* (1995) reported that participants' perceptual judgements were strongly affected by the Titchener circles illusion, however, when participants were asked to pick up the target circle, the scaling of their grip aperture was largely determined by the actual size of the target disc and not its perceived illusory size. If indeed, results from this study are correct, conclusions suggest that visual processes mediating perception are very different to those processes controlling action. Additional support for the hypothesised separate visual processes mediating perception and action has been highlighted in studies using participants with neurological disorders. Such research has shown that visually guided motor-based actions are controlled by visual mechanisms located in the cerebral cortex that are quite distinct from those underlying perception of the object. For example, individuals who have suffered damage to the superior portion of the posterior parietal cortex which affects the dorsal stream (i.e. Balint's syndrome), are unable to produce normal scaling of a grasping movement when reaching and grasping objects (Jakobson *et al.,* 1991). This is evidenced through large adjustments made in the aperture of the index finger and thumb the closer the participants hand gets to the object. However, the same individual is able to identify simple line drawings of objects (Jakobson *et al.,* 1991). In contrast, individuals who have suffered cortical damage which affects ventral and lateral areas of the occipital cortex (i.e. visual agnosia), demonstrate the opposite response. These individuals are unable to recognise the size, shape and orientation of visual objects, however demonstrate comparable accuracy to a control group (no cortical damage) in guiding hand and finger movements towards grasping an object (Goodale *et al.,* 1991; Milner *et al.,* 1998). The ability for patients suffering from ventral damage to control multi-limb movements has been demonstrated during adaptive gait. Patla and Goodale (1996) highlighted that individuals with ventral damage were able to successfully negotiate an obstacle placed in their travel path, but were unable to verbally report the height of that same obstacle negotiated.

Some studies report a direct link between perception and action and thus dispute the theory that visual perception and action are separated in the cortical visual system. For example, superimposing a visual illusion pattern on a single step not only induces a perceived increase in step height, but also causes increased maximum lead limb toe clearance (distance between the toe and apex of step) when stepping up onto the step (Elliott *et al.,* 2009). This proposed link between perception and action has also been reported elsewhere (Franz *et al.,* 2000; Dassonville and Bala, 2004; Glover and Dixon, 2004; Li and Matin, 2005). If the research above, which suggests a link between the hypothesised visual processes mediating perception and action are correct, this questions the theory proposed by Goodale and Milner (1992) for two separate and distinct visual streams for visual perception and visuomotor action. Therefore, the proposal by Goodale and Haffended (1998) may be more appropriate; the ventral steam enables the selection of an appropriate movement or action with respect to objects within the world and the dorsal stream is able to regulate online control of such actions. Nevertheless, it is clear that further research is needed to better understand the contribution of the dorsal and ventral streams for visual perception and action.

2.2 What visual information is used for the control of locomotion?

In order to obtain visual cues from within the environment, images must first appear in the field of view. The monocular field of view, or visual field as it is more commonly known, for a normal stationary eye extends to ~60 º superior and ~75 º inferior of the horizontal meridian (figure 2.4a), and ~150 º between nose and temple (figure 2.4b). Each eye consists of a blind spot where no receptors are located. This is the region when the optic nerve leaves the retina (Howard and Rogers, 1995). The projection of the blind spot in the visual field is about 3 º in diameter and appears ~12 - 15 º in the temporal hemifield (Howard and Rogers, 1995). The monocular visual field from either eye overlaps and thus ~114 º of the visual field is visible to both eyes simultaneously (figure 2.4b, Howard and Rogers, 1995).

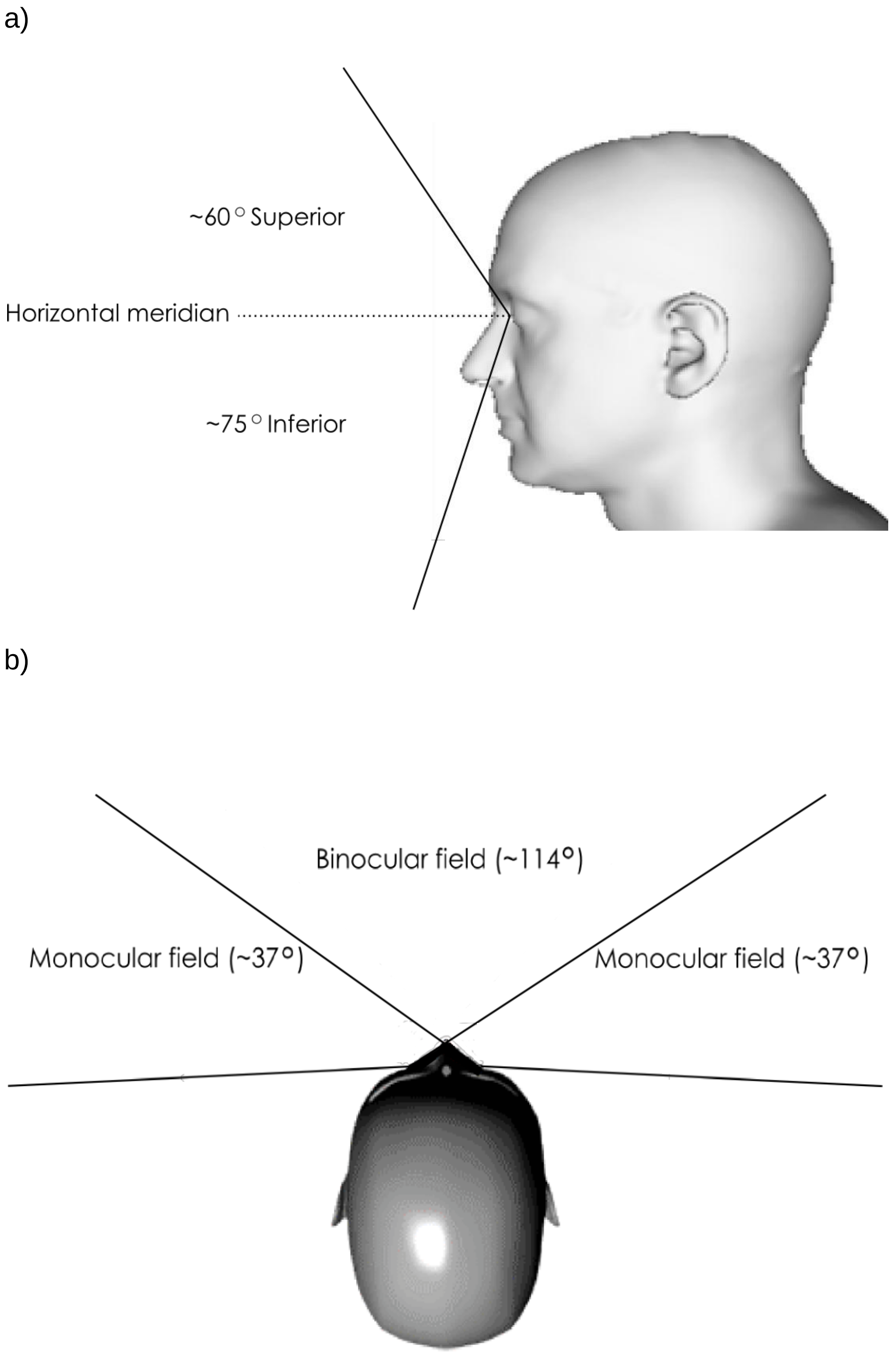

Figure 2.4. Extent of the visual field in healthy stationary eye(s) in a) superior and inferior directions from the horizontal meridian and b) monocularly and binocularly along the transverse plane.

2.2.1 Binocular vision

Vision that incorporates images from the two eyes simultaneously improves the ability to accurately perceive object depth. This is because as each eye forms an image of the object on its retina, they form slightly different or disparate images on each retina, and this slight disparity is the basis for stereopsis or depth. Whilst monocular visual cues (vision from one eye) also contribute to depth perception i.e. looming, motion parallax, occlusion and shadow (Daum and McCormack, 2006), these depth cues are not as advantageous compared to receiving binocular depth cues. For example, compared to binocular vision, the availability of monocular vision (from either left or right eye) several steps before and during obstacle negotiation results with participants exhibiting greater variability in perceptual estimates of obstacle height and increased maximum lead limb toe clearance over the obstacle (Patla*,* 1997; Patla *et al.,* 2002). The increase in toe clearance can be attributed to the degradation of the sensory-to-motor transformation when obtaining visual cues regarding obstacle height through monocular vision, resulting in such a safety response to reduce the risk of obstacle contact (Patla *et al.,* 2002). This safety response to reduce the risk of obstacle contact in monocular vision has been attributed to the inability to accurately determine the height of the obstacle during the approach prior to the obstacle (Patla *et al.,* 2002). Whilst previous research has suggested that the position of the obstacle in space (i.e. its height) will be significantly affected when viewed binocularly, perception of the lateral and fore-aft position of the obstacle will also likely be affected.

Differences in controlling lead limb trajectory have also been found in monocular compared to binocular visual sampling when more complex adaptive locomotion tasks are considered. For example, Hayhoe *et al.* (2009) tasked participants with walking a short path (~7 m) stepping over 2 obstacles of different height (19 cm and 15 cm high respectively, with ~41 cm difference between placement) en route to walking through an aperture 43 cm wide, then walking back along the same path negotiating the same 2 obstacles. Gaze behaviour was also recorded. Compared to the binocular viewing condition, the monocular condition resulted with participants walking slower, subsequently causing travel time to increase ~10 %. In corroboration with previous research (Patla*,* 1997; Patla *et al.,* 2002), the monocular condition resulted in participants increasing lead limb toe clearance during obstacle negotiation. Hayhoe *et al.* (2009) also demonstrated that compared to gaze patterns in the binocular viewing condition, in the monocular condition participants spent longer fixating on key locations during the task (even after accounting for reduced walking speed). This was evidenced by increased duration of fixation on the obstacle furthest from the participant prior to negotiation, which was demonstrated both before and after negotiating the aperture (Hayhoe *et al.,* 2009). This pattern of increased duration of fixation on the obstacle in the monocular viewing condition suggests a strategy to more accurately perceive the position of the obstacle in space (height and lateral and fore-aft position). A similar prolonged fixation pattern has also been reported in reaching movements under monocular conditions (Loftus *et al.,* 2004). When negotiating the aperture, Hayhoe *et al.* (2009) observed similar fixation patterns for both monocular and binocular viewing conditions. Hayhoe *et al.* (2009) suggested that because movement planning to negotiate the aperture would occur after negotiating the second obstacle at ~1 metre away from the aperture, stereo discrimination would be much improved. At this distance residual uncertainty of ± 0.5 cm for a 43 cm gap would be adequate for planning the movement through the aperture under both binocular and monocular viewing conditions (Hayhoe *et al.,* 2009). Adaptations in stepping strategy present during adaptive locomotion when visual information is sampled under monocular vision compared to binocular vision has also been similarly reported amongst those with unequal monocular acuity. For example, compared with age matched normals, older adults with unilateral cataracts increase lead limb toe clearance during obstacle negotiation (Elliott *et al.,* 2000). Also, when stepping up to a new level during on-going gait under conditions of small monocular refractive blur (0.5D and 1.0D), both young and older adults increased lead limb toe clearance and single support time (time when only the trail-limb is in contact with the ground during the step up) when negotiating the raised surface (Vale *et al.,* 2008a, b). However, the latter changes may have been in response to magnification effects of the positive blur lenses, which made the step appear larger (Elliott and Chapman, 2010). Of note, improved depth perception is not the only benefit of binocular vision compared to monocular vision. For example, under binocular vision, visual acuity is 10 - 12 % better and contrast sensitivity ~40 % higher than monocular vision (Campbell and Green, 1965; Home, 1978; Cagenello *et al.,* 1993; Horowitz, 1994).

2.2.2 Optic flow and egocentric-direction strategy

Navigating within an environment requires continuous updated visual information regarding the spatial relationship between oneself and objects within the environment (Turano *et al.,* 2005). This relationship becomes increasingly important when navigating within a potentially hazardous environment (i.e. when faced with multiple objects or surface height change). In the 1950's James Gibson first proposed that information for both perception of self motion and the control of self motion is attained based on visual information available to an eye moving through the environment, which is termed optic flow (Warren, 2009). The ability to control self motion though visually sampling the environment when optic flow is available has been shown to be advantageous compared to when optic flow is not available (e.g. Patla, 1998; Patla and Greig, 2006). For example, compared to when dynamically visually sampling the environment (i.e. using optic flow) prior to negotiating an obstacle with vision occluded several steps before and during obstacle negotiation, in static visual sampling condition (i.e. participants stood still prior to visual occlusion and then walked up to and negotiated the obstacle), foot placement became increasingly variable in the approach phase prior to obstacle negotiation (Patla, 1998; Palta and Greig, 2006). The increasingly variable foot placement caused participants to step over the obstacle from different take off points, which subsequently increased the number of occasions participant's lead foot made contact with the obstacle. Sampling the obstacle dynamically (i.e. moving towards the obstacle) produced a radial pattern of optic flow termed the focus of expansion (FOE). In the dynamic condition, participants were able to use flow information ('re-afference') in the initial steps prior to visual occlusion to update the error in foot placement between the obstacle and self; this process is understood to operate using a perception-action loop (Warren *et al.,* 1998). However, in the static condition participants did not receive initial FOE information and were subsequently unable to update any error between self and obstacle prior to visual occlusion. A similar ability to eliminate foot placement error using 're-afference' has been previously reported in skilled long jumpers as they approach the takeoff board (Lee *et al.,* 1982).

During locomotion individuals do not fixate the same object within the environment i.e. individuals scan the environment for potential hazards. If optic flow is dependent on the FOE for adjusting movements to reach a desired goal as Warren *et al.* (1998) hypothesised, then fixating away from the target will change the optic flow field and should affect the ability to maintain heading direction. Schubert *et al.* (2003) hypothesised that in such situations individuals would increase lateral sway due to impaired postural stability and heading direction. Participants tracked a visual target which moved in the horizontal plane when walking on a treadmill at a constant speed. Compared to when no optic flow pattern was available, the availability of minimal optic flow velocity (which was less than treadmill speed) dramatically reduced participant's lateral sway. However, because increasing optic flow velocity to match treadmill speed had no further improvement on reducing participant's lateral sway, Schubert *et al.* (2003) suggested that in this particular condition, this indicated that the FOE within the flow field cannot be a dominant cue to control heading direction. Indeed the conclusion from Schubert *et al.* (2003) that the FOE obtained through optic flow may play little / no role in controlling heading direction in certain situations has received additional support (Rushton *et al.,* 1998; Warren *et al.,* 2001; Harris and Bonas, 2002; Macuga *et al.,* 2006). Warren *et al.* (2001) demonstrated by using a 3-D immersive environment, that when the availability of optic flow information was diminished, heading direction can still be obtained through the observer perceiving the location of the goal with respect to the body, termed egocentric-direction strategy. However, with each 3-D immersive environment that included more optic flow cues, participants exhibited responses that indicated a greater reliance on using optic flow strategy. Since similar heading performance in a 3-D immersive environment can be achieved through using either an optic flow or egocentric-direction strategy (Warren *et al.,* 2001; Macuga *et al.,* 2006), this further suggests that heading perception is not fully dependant on optic flow and such models that exclusively depend upon optic flow strategy may be somewhat limited (Macuga *et al.,* 2006).

The dissociation between optic flow and egocentric-direction strategy has also been demonstrated through using displacing prisms (Rushton *et al.,* 1998; Warren *et al.,* 2001; Harris and Bonas, 2002). Such prisms shift the image of the world on the retina by an amount corresponding to the power of the prism. Thus viewing through the prisms subsequently causes objects that are actually positioned to one side of the participant's body midline to appear straight ahead. If an egocentric-direction strategy is used to reach a target when viewing through the prisms, this will cause individuals to produce a constant heading error between the actual target and perceived target, resulting in the individual walking a curved path. However, if an optic flow strategy is used, viewing through the prisms should not affect heading error (apart from on the first step, when flow is not available), because displacing prisms do not change differential properties of the FOE (Harris and Bonas, 2002). In conditions when participants walked towards a luminous target and were unable to use optic flow information (lights turned off, thus egocentric-direction strategy required) or when exposed to a high number of optic flow cues (full lighting), Harris and Bonas (2002) demonstrated that participants walked a similar curved path in both lighting conditions, suggesting a reliance on the egocentric-direction not optic flow strategy. This conclusion supports the earlier work of Rushton *et al*. (1998) who also demonstrated that when participants viewed through displacing prisms and optic flow information was available, they relied on egocentric-direction strategy, as evidenced by walking a curved path to reach the target location. Whereas results which used a 3-D immersive environment to disassociate between optic flow and egocentric-direction strategy demonstrated that an environment which included more optic flow cues caused participants to exhibit responses that indicated a greater reliance on using optic flow strategy (Warren *et al.,* 2001), this was not observed when using the displacing prisms (Harris and Bonas, 2002). Rather, when viewing through the displacing prisms, participants relied on egocentric-direction strategy to regulate heading, irrespective of the amount of optic flow information available. Collectively, these results could be interpreted to further suggest how heading perception is not always fully dependant on optic flow. However, Warren *et al.* (2001) suggested that using displacing prisms introduce optical distortion that warps the optic flow pattern, which likely causes individuals to depend more on an egocentric-direction strategy to regulate heading perception. In a second experiment, Warren *et al.* (2001) measured heading direction whilst participants wore displacing prisms and viewed the same 3-D immersive environment as in his first experiment. Results of this second experiment demonstrated that the effects of heading error with each environment that included more optic flow cues were similar to results in the first experiment but were attenuated when wearing the prisms (Warren *et al.,* 2001). Since the effects of heading error in the 3-D immersive environment were significantly smaller when participants wore the prisms, this suggests that the prismatic distortion caused participants to rely more on egocentric-direction strategy (Warren *et al.,* 2001). However, because the increased amount of optic flow available moderately influenced walking trajectory, this also suggests that optic flow strategy had some influence on controlling heading direction (Warren *et al.,* 2001).

The aforementioned research demonstrates the on-going debate whether optic flow or egocentric-direction strategy is the dominant visual cue used to control heading. Conclusions likely suggest that individuals rely on the most salient strategy available to control heading i.e. when optic flow is reduced or distorted, behaviour appears to be governed by an egocentric-direction strategy, whereas when optic flow information is available, this strategy is favoured to guide heading. This debate highlights the flexibility and robust nature of the visual system for the control of heading.

2.2.3 The visual field

Whilst the visual field is often divided into two sections, the central visual field and the peripheral visual field (Fortenbaugh *et al*., 2007), there are a variety of definitions in the literature concerning where this division occurs. The central visual field can extend from 5 °, encompassing the macular field (the point of fixation with maximum visual acuity, McClure, 1988) to a maximum of 30 ° (the extent of the central visual field often measured clinically using automated permeters i.e. frequency doubling perimetry or Humphrey Field Analyser). The visual field can also be divided along the transverse plane into the upper visual field (uvf) and the lower visual field (lvf) (Darker and Jordan, 2004). To investigate what visual information within the visual field directly contributes to the control of locomotion, the effects of visual field restriction from a fundamental and clinical perspective have been explored.

The superiority of the lvf compared to the uvf has been highlighted in discriminating visual stimuli based on visual acuity, contrast, and motion (Darker and Jordan, 2004; Levine and McAnany, 2005), texture segregation (Graham *et al.,* 1999) and attentional resolution (He *et al.,* 1997). In addition, compared to the uvf, the latency of voluntary reaction times to visual stimuli is shorter (Payne, 1967; Maehara *et al.,* 2004), and visually guided pointing movements are faster and more accurate when performed in the lvf (Danckert and Goodale, 2001; Khan and Lawrence, 2005; Krigolson and Heath, 2006). However, this bias towards processing visual information within the lvf has not been observed in all tasks. When participants were required to determine the orientation of a single grating patch flanked by four patches (two on either side) at a different orientation (see figure 2.5), there was no difference between task success when presented in the uvf or lvf (He *et al.,* 1997, experiment 1).

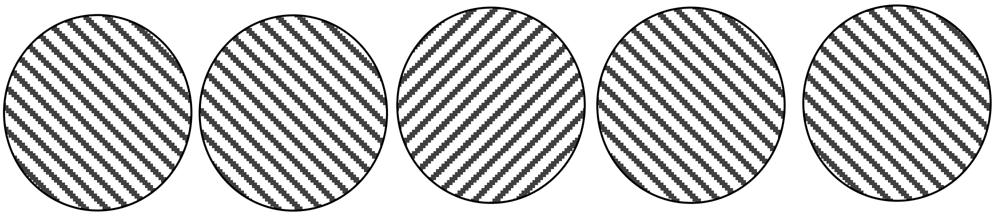

Figure 2.5. Middle grating patch flanked by four grating patches (two on either side) at a different orientation, adapted from He *et al.* (1997).

In addition, similarities in participant's response accuracy for identifying words and non-words in the uvf and lvf have also been reported (Darker and Jordan, 2004). In some instances the uvf is even superior to the lvf; when discriminating visual stimuli based on apparent distance from the observer (Levine and McAnany, 2005) and the speed at which visually guided saccades are triggered (Heywood and Churcher, 1980; Dafoe *et al.,* 2007). It therefore appears that the superiority of the lvf when compared to the uvf is only apparent during visual processing of tasks that require a high level of precision during movement and / or increased attention. Indeed, there are a greater number of cone photoreceptor cells and LGN cells in the lvf compared with the uvf (Connolly and van Essen, 1984; Curcio and Allen, 1990) which will facilitate faster processing of increased spatial resolution and colour vision (Darker and Jordan, 2004). For example, He *et al.* (1997, experiment 2) tasked participants with either maintaining fixation 10 º above or below the centre of a rectangular area whilst nine green moving balls were presented inside the rectangle. At the beginning of each trial, two of the balls turned red for one second and then turned back to green. Participants were required to track the two previously highlighted red balls whilst maintaining fixation on the pre specified location.

After five seconds the balls stopped moving and the participant was required to indicate which two balls were initially highlighted. He *et al.* (1997, experiment 2) demonstrated that compared to tracking the moving balls in the uvf (gaze fixation below the rectangle), when participants tracked the balls in the lvf (gaze fixation above the rectangle), they were ~30 % more accurate in identifying the pre specified red balls.

The advantage of the lvf compared to the uvf in tasks that require a high level of precision, as highlighted above (He *et al.,* 1997, experiment 2; Danckert and Goodale, 2001; Khan and Lawrence, 2005; Krigolson and Heath, 2006) were limited to either single upper-limb movements, or required no movement from the individual. It is therefore likely that during adaptive locomotion, which requires control of multiple limb movements, the advantages in regulating movement control when visual information is acquired from the lvf compared to uvf would be further evidenced. However, since movement control is often completed in the absence of visual feedback of the lower-limbs, and / or immediate floor area i.e. when descending a step whilst carrying a laundry basket (visual feedback of the lower-limbs and / or of the area on the ground where we intend to land is occluded), this lvf advantage may not be as apparent in adaptive locomotion as first suggested. Occluding online visual information from the lower limbs and ground ~1.2 m immediately in front of the participant (i.e. lvf) had little effect on task success if exproprioceptive visual information (relation of body segments to the environment) was available through head position cues (Rietdyk and Rhea, 2006; Graci *et al.,* 2010). Since participants were able to successfully negotiate the obstacle under condition of lvf occlusion, this suggests that expproprioceptive information regarding the orientation of the head in space (uvf cues) can be used to compensate for lack of information regarding the lvf. However, under such lvf occlusion conditions, participants were unable to ‘fine tune’ lower-limb trajectory as evidenced by increased lead limb stride length and lead limb toe clearance during obstacle crossing (Rietdyk and Rhea, 2006; Graci *et al.,* 2010). Increasing stride length values and lead limb toe clearance during obstacle crossing under conditions of visual occlusion have been previously suggested as safety strategies to reduce the risk of tripping (Patla and Rietdyk, 1993). When the lvf was occluded and no head position cues were available, trail limb toe clearance increased. The increase in trail limb toe clearance value when the lvf was occluded is surprising because during obstacle negotiation, as the trail limb crosses the obstacle the limb is behind the individual and subsequently out of sight. Therefore lvf occlusion should have no effect on trail limb toe clearance. However, a significant correlation was found between trail foot horizontal distance (horizontal distance of the toe from the obstacle) and trail limb toe clearance (Rietdyk and Rhea, 2006). This suggests that trail limb toe clearance is dependent upon horizontal placement prior to the obstacle (Rietdyk and Rhea, 2006). This correlation was confirmed by similar trail foot horizontal distance placement values in full vision and lvf occluded conditions when position cues were available (Rietdyk and Rhea, 2006). Whilst Graci *et al.* (2010) observed decreases in trail foot placement when the position cue was available, values did not return to full vision values. The most plausible explanation for the differences in trail foot placement between studies is likely attributed to the different methodologies used. For example, Graci *et al.* (2010) used smaller obstacle heights compared to Rietdyk and Rhea (2006) (obstacles 4 and 8 cm high compared to 10, 20 and 30 cm high respectively). In addition, participants sampled the environment using monocular vision (Graci *et al.,* 2010) compared to binocular vision; indeed the advantage of sampling the environment through binocular vision compared to monocular vision has been previously highlighted in this chapter (sub section 2.2 Binocular vision).

The importance of lvf information for regulating movements which require a high level of precision has also been reported in negotiating multi-surface terrain (negotiating solid, compliant, rocky, irregular, tilt and slippery surfaces; Marigold and Patla, 2008). Compared with full field vision available throughout the trial, occluding the lvf (resulting in the lower limb and ~2 steps in front of the participant being occluded) caused participants to flex their head significantly more and alter gait patterns, reducing speed and step length (Marigold and Patla, 2008). Increasing head flexion allowed more immediate locations in the travel path to be viewed, which was likely a strategy to compensate for occluded lvf information. Furthermore, since the immediate ground area and lower limb was occluded, exproprioceptive information regarding the foot relative to the floor was unavailable. This resulted in participants adapting their gait by reducing speed and step length and ensuring the CM remained closer to the stance limb. These adaptations in stepping strategy subsequently reduced the chance of falling forward if a ‘miss step’ occurred over the uneven terrain. A similar cautious stepping strategy has been reported when participants were tasked with stepping over an obstacle which varied in height and / or width (from trial-to-trial, Patla and Rietdyk, 1993), and during over ground walking when the entire peripheral visual field was occluded (Graci *et al.,* 2009). Upon first inspection, in may appear surprising that Graci *et al.* (2009) only observed adaptations in participants gait when the entire peripheral visual field was occluded compared to full field vision and there was no difference when the lvf or uvf was occluded. However, since participants were only required to walk along a 7 m even terrain, the lack of differences in lvf or uvf conditions is likely attributed to the task demand. The importance of online lvf information in controlling adaptive gait has also been reported by Marigold *et al.* (2007). Participants were required to walk upon a moving treadmill and step over an obstacle that was randomly released onto the treadmill they were walking. The time available to step over the obstacle varied from 200 - 450 ms. Participants were required to either fixate the area immediately in front of them whilst walking, or fixate approximately two steps ahead. In the latter condition, participants were told to maintain fixation two steps ahead until they detected the release of the obstacle. After the obstacle was released, they were free to fixate where they liked. Results highlighted that participants rarely (only 18 % of trials) redirected their gaze when the obstacle was released and participants were fixating 2 steps ahead, despite having sufficient time before obstacle contact. This suggests that during locomotion, lvf cues are used to detect the presence of a floor based obstacle and initiate suitable avoidance strategy (Marigold *et al.,* 2007). The ability to monitor lower limb trajectory through visual cues obtained in the lvf has been further evidenced by Patla and Vickers (1997). Indeed when participant’s gaze was monitored as they walked up to and negotiated an obstacle, there were no fixations on the lower limb.

Clinical research on the effects of visual field occlusion on adaptive gait compares stepping responses from individuals with permanent visual field loss to age matched individuals who have no visual field loss. ‘Functional’ visual field loss can be caused through ocular diseases such as glaucoma and / or retinal disease in both eyes, or a tumour or vascular occlusion along the cerebral visual pathway (Coleman *et al.,* 2007). Monocular visual field loss can be compensated by a normal visual field in the other eye (Wood and Troutbeck, 1995). However, binocular visual field loss has been associated with frequent falls and decreased quality of life (Ivers *et al.,* 1998; Sherwood *et al.,* 1998; Klein *et al.,* 2003) even in the presence of good visual acuity (Rubin *et al.,* 2001). For example, visual field loss is associated with both multiple falls and fractures (Ivers *et al.,* 1998; Klein *et al.,* 2003). Visual field loss is also a significant predictor of experiencing an automobile crash (Rubin *et al.,* 2007). For example, drivers who missed more that 20 points in the visual field test (indicating a poor visual field) were associated with an increased crash risk (Rubin *et al.,* 2007). A degraded residual visual field is also a significant predictor of mobility performance (Marron and Bailey, 1982; Brown *et al.,* 1986; Lovie–Kitchin *et al.,* 1990; Haymes *et al.,* 1996; Geruschat *et al.,* 1998; Kuyk *et al.,* 1998; Turano *et al.,* 1999). For example, Turano *et al.* (1999) previously highlighted that individuals with glaucoma reduce mean walking speed by ~10 % and are twice as likely to stumble or walk into an approaching pedestrian, compared to normal vision subjects. Since individuals with advanced glaucoma have significantly reduced peripheral vision, the reduction in walking speed highlighted above, was likely a strategy to maximise the amount of visual information that could be sampled during each step. This would subsequently increase the effectiveness of route planning i.e. the ability to successfully avoid an oncoming pedestrian.

2.3 When is visual information used for the control of locomotion?

The research outlined in the aforementioned sub section (2.2.3) highlighted that visual cues obtained from the central and uvf are predominantly involved in planning the general stepping strategy during adaptive locomotion, whereas visual cues obtained from the lvf enable increased level of accuracy during the movement. So that movement planning can occur, we need to view task-relevant areas from within the environment. This may occur through directly fixating on a particular object / area, or acquiring this visual information from the periphery (i.e. lvf). The majority of visual cues viewed from within the environment are predictable i.e. when required to step over an obstacle, we view the obstacle (Patla and Vickers, 1997). However, what is not so obvious is when these visual cues are viewed during locomotion and / or are of increased importance for movement control; the following sub section will provide an overview of this.

2.3.1 Over-ground walking

The ability to use visual information to plan future stepping actions (termed feedforward visual control) has been demonstrated during over-ground walking. For example, when participants were tasked with walking without vision and stopping at pre viewed locations, they demonstrated high levels of precision with stopping up to a distance of around 12 m (Thomson, 1983, experiment 1), or until approximately 8 secs had passed (Thomson, 1983, experiment 2). Similarly, during over-ground walking where participants had no prior experience of the terrain, when participants self selected when they received visual feedback, they sampled the environment for ~10 % of the travel time (Patla *et al.,* 1996). Collectively, this aforementioned research suggests that intermittent visual sampling allows future stepping actions to be accurately internalised, however there is a limit before visual update is required. This type of intermittent visual sampling occurs in everyday tasks e.g. as we look around the environment to admire the scenery whilst walking. When task demand is increased such that precise foot placement is required during each step through relatively large lateral movements of the body, foot placement on the ‘stepping stones’ has been shown to be under feedforward visual control (Hollands *et al.,* 1995; Hollands and Marple-Horvat, 1996, 2001). For example, when analysing gaze fixations when negotiating the stepping stones in the dark with LEDs (light emitting diodes) positioned at the centre of each stepping stone, participants fixated to the next target at the end of the current stance of that stepping limb (Hollands and Marple-Horvat, 1996). These results are consistent with previous work highlighting that participants performing the same stepping task under conditions of full lighting demonstrate a feedforward fixation strategy; participants consistently fixate the next target just before they lifted their foot to be repositioned, that is towards the end of that limb’s stance phase (Hollands *et al*., 1995). Since movement control of the lead limb onto the stepping stone was controlled through feedforward visual sampling, it is therefore not surprising that occluding the LEDs in the latter part of participants stance phase resulted with adaptations in stepping strategy, evidenced through participants increasing stance duration (Hollands and Marple-Horvat, 1996, 2001). Since gaze fixations were the same when the stepping stones were negotiated both with and without full lighting, this indicates that only providing target location is sufficient to ensure precise foot placement (Hollands and Marple-Horvat, 1996).

The utility of feedforward visual sampling to ensure appropriate movement planning may be linked to the physical limitations of the human body. For example, participants are unable to initiate a change in direction during the on-going step when cued to respond to 30 ° or 60 ° directional changes (Patla *et al.,* 1991). This inability to respond to the required change in direction did not mean participants attempted to alter direction and did not achieve the required amount of direction change, rather, participants simply continued along their same travel path unable to initiate a response. However, the required change in direction is achieved if cued from 2 steps in advance (Patla *et al.,* 1991). Altering direction of locomotion requires either crossing the lead limb over the trail limb or stepping in a direction which moves the lead limb away from the trail limb. In either situation, to increase stability and avoid the risk of falling, prior to changing direction the CM must be positioned within the base of support of the intended stance limb. If participants attempted to change direction when cued during the on-going step prior to addressing the constraints of the CM, the instability created would likely result in a fall (Lyon and Day, 1997). This provides a likely explanation why Patla *et al.* (1991) observed that participants were unable to initiate a response when cued with direction change during the on-going step. However, when cued from 2 steps in advance, this allowed sufficient time to displace the CM prior to initiating a direction change. Since the human body requires advanced planning for subsequent movement control, visually sampling the environment in a feedforward manner allows upcoming potential hazards to be identified with sufficient time to implement appropriate avoidance strategies. Indeed, the importance of feedforward visual sampling has been highlighted in planning subsequent steps in the gait cycle when running over uneven terrain (Warren *et al.,* 1986). This was evidenced by altering the vertical impulse applied to the ground during each stance phase to allow adjustments in step length based on upcoming changes in the uneven terrain. Indeed, Marigold and Patla (2007) also demonstrated that gaze fixations when negotiating uneven terrain were directed to areas eventually stepped on. Furthermore, Marigold and Patla (2007) also demonstrated that participants repeatedly fixated the upcoming ground terrain, which is indicative of online visual control of locomotion (visual information sampled concurrently during the on-going movement) which is beneficial for ensuring precise control of foot placement. A similar repeated pattern of fixation to control foot placement has also been observed when stepping on raised blocks (Patla and Vickers, 2003). Ensuring precise control of foot placement subsequently reduces the risk of falling, through placing the foot in an inappropriate or undesirable area. The ability to precisely control foot placement using online vision has been further evidenced by increases in foot placement error when stepping to a stationary target and vision was occluded at foot-off (Reynolds and Day, 2005). This suggests that visual information can be used during the swing phase of the limb to update movement control (Reynolds and Day, 2005).

2.3.2 Obstacle negotiation

The role of feedforward visual information regulating movement control has also been highlighted during obstacle negotiation. When approaching an obstacle, participants fixate on the obstacle on average 2 steps before negotiating the obstacle (Patla and Vickers, 1997). However, during obstacle negotiation participants fixate on the ground immediately ahead of the obstacle, where the lead limb will subsequently contact the floor (Patla and Vickers, 1997). When required to step over obstacles of increased height (obstacles 1 cm, 15 cm, or 30 cm high negotiated), the increased obstacle height only resulted in increased frequency of obstacle fixation. Whilst the increased frequency of obstacle fixation allowed participants to accurately determine the height of the obstacle, ultimately the same feedforward pattern of visual sampling was used irrespective of obstacle height. The negligible effect of increasing obstacle height and by implication task demand on the pattern of visual sampling is somewhat surprising considering that the aforementioned section (2.2.1 over-ground terrain) highlighted consistent differences in visual sampling (feedforward compared to online) dependent upon task demand (Hollands *et al*., 1995; Marigold and Patla, 2007). These results could be interpreted to suggest that irrespective of obstacle height, visual information is sampled in a feedforward manner. However, because Patla and Vickers (1997) only varied the height of the obstacle and not location, there was no increased demand on visual processing and subsequent need for online visual sampling.

Other studies undertaken to determine how vision is used for successful obstacle crossing indicate that despite vision being occluded from up to 4 steps before the obstacle and remaining occluded until both lead and trail limbs have crossed the obstacle, successful obstacle negotiation is still possible (Patla *et al.,* 1998; Mohagheghi *et al.,* 2004). In contrast when visual information is occluded from five or more steps before the obstacle, crossing success is significantly reduced (Patla and Greig, 2006). This aforementioned research (Patla *et al.,* 1998; Mohagheghi *et al.,* 2004; Patla and Greig, 2006) suggests that occluding vision in advance of two strides (4 steps) before the obstacle significantly impacts the nervous system's ability to utilise feedforward visual cues to integrate an efference copy of the motor command to plan the general stepping pattern during obstacle crossing (Marigold, 2008). Despite being able to successfully negotiate obstacles and step / stairs through receiving feedforward visual information, occluding vision in the last 2 steps before the obstacle significantly affects the ability to 'fine tune' lower-limb trajectory and as a result lead limb toe clearance is increased compared to when vision is available throughout (Patla *et al.,* 1998; Mohagheghi *et al.,* 2004). However, lead limb toe clearance is unaffected when vision is occluded during the last step before the obstacle compared to when vision is available throughout (Patla *et al.,* 1998; Mohagheghi *et al.,* 2004). This suggests that 'fine tuning' of lower-limb trajectory occurs through online visual information acquired in the penultimate step prior to the point of crossing.

When vision is returned to the participant as they step over the obstacle with their lead limb, such that online vision is available to 'fine tune' lead limb trajectory, lead limb maximum toe elevation decreases, however, trail limb trajectory is unaffected. Conclusions from these aforementioned studies (Patla *et al.,* 1998; Mohagheghi *et al.,* 2004; Patla and Greig, 2006) indicate that whilst the lead and trail limbs benefit from feedforward visual sampling to plan the general stepping pattern required to successfully negotiate an obstacle, 'fine tuning' lower-limb trajectory only occurs in the lead limb when online vision is available in the penultimate step prior to crossing and / or during the latter period of obstacle crossing. Indeed, the ability for participants to alter their lead limb trajectory online in response to sudden increases in obstacle height have been reported when the ipsilateral or contralateral limb is placed before the obstacle (Patla, 1997), or during the initial part of the swing phase (Quevedo *et al.,* 1997).

The majority of the literature highlighted in this chapter has focused on over-ground walking and the relative importance of sensory integration in controlling such movements. Obstacle crossing and multi-surface terrain walking have also been covered in this literature review, which are understandably more challenging than over-ground walking, but are nevertheless simple tasks. Perhaps the most challenging of locomotor tasks encountered in daily everyday life is step / stair descent i.e. falls occur most frequently on stairs (Startzell *et al.,* 2000) and higher impact forces on landing are observed compared to level walking. However, there is a paucity of literature pertaining to the visual control of drop landing and step descent. Nevertheless, this limited research will be discussed in turn.

2.3.3 Drop landing

A drop landing refers to a descent which involves a period of 'free fall', whereby the trail limb is unable to exert control on the body during descent. Receiving feedforward visual information pertaining to the drop height prior to initiating the movement compensates effectively for occluding vision throughout the descent (Libermann and Goodman, 2006), and even with no prior visual knowledge of drop height and with vision occluded during the descent, participants are able to adopt a suitable preparatory strategy through proprioceptive and vestibular sensory mechanisms (Santello *et al.,* 2001). Despite participants being able to regulate landing under no vision condition from heights ranging from 15 cm (Liebermann and Goodman, 2006) to 80 cm (Santello *et al.,* 2001), these tasks are quite different from a controlled step descent where the trail limb is able to exert control on the body during the descent.

2.3.4 Step descent

When feedforward visual information of the step height prior to descent is not available, participants are still able to scale movements of the lead leg to ensure safe and controlled landing when full binocular vision is subsequently made available at movement initiation (Cowie *et al.,* 2008). However, when vision was unavailable prior to and during step descent participants were unsure of the exact height of the lower level, so were unable to effectively scale movements of the lead leg to the step height in preparation for landing (Cowie *et al.,* 2008). Buckley *et al.* (2008) also demonstrated that blurring (achieved using light scattering lenses which cause diffuse blur) or occluding vision prior to movement initiation caused participants to adopt a cautious strategy of ‘sitting back’ on the trail limb so they could use the lead limb to ‘probe’ for the ground and gain somatosensory feedback regarding floor height prior to committing to the step down. Such adaptations in stepping strategy when visual information is degraded or occluded during step descent indicates that during step descent, online vision is used to precisely control landing. However, since vision was occluded / degraded several seconds prior to movement initiation, it is also possible that the adaptations in stepping strategy could be attributed to the occlusion of feedforward vision in the period immediately prior to step descent. These findings also support work highlighting that if step descents are performed with vision occluded and there is no prior knowledge of step height, anticipatory lower limb muscle activity preceding ground contact is reduced or even absent, which results in ‘softer’ landings and an increased reliance on the stance limb to control descent (Craik *et al.,* 1982). Whilst the aforementioned research highlights the importance of visual information immediately prior to / during step descent to ensure step descent landings are appropriately controlled, the point in time during step descent when visual cues are used by the visuomotor system to regulate landing control remains unclear.

2.4 Clinical visual assessments

To ensure that any conclusions drawn in subsequent experimental chapters can be attributed to experimental manipulations rather than deficiencies in the visual system, the following clinical visual assessment tests were conducted and used as a basis for participant inclusion / exclusion: visual acuity, stereopsis and contrast sensitivity. The rationales for their inclusion are provided below.

Visual acuity is a measurement of the resolving power of the eye and its ability to discriminate detail. This measurement is the most commonly measured aspect of visual function and is used as an essential reference scale to define the suitability of a person to drive, whether someone is visually impaired, or legally blind (Elliott and Flanagan, 2007). Traditionally visual acuity has been measured using a Snellen notation where the numerator denotes the distance of the subject from the chart and the denominator represents the letter size that can just be perceived at that distance (Bailey, 2006). However, more recently, logarithm of the minimum angle of resolution (logMAR) charts have been considered the standard measure employed in clinical research (Ferris and Bailey, 1996) since these tests provide a more reliable and discriminative measure of visual acuity (Lovie-Kitchin, 1988). Whilst a logMAR score of 0.00 (Snellen equivalent 6 / 6) is said to equate to ‘normal’ vision, it has been shown that many young adults have better acuity than this (Elliott *et al.,* 1995). Since the surrounding environment is not just made up of high contrast, sharp edged objects, rather, composed of objects with a variety of contrasts, many of which would likely be defined as low contrasts, it has been suggested that measuring visual acuity over-estimates visual function in the ‘real-world’ because this is not representative of how a person views the world (Helbostad *et al.,* 2009). It is therefore necessary for other clinical visual assessment tests to be completed in addition to visual acuity (Helbostad *et al.,* 2009).

The disadvantage of visually sampling the environment through monocular or unequal binocular vision compared to equal binocular vision has been previously highlighted in this chapter. It is therefore important to measure participant's ability to perceive depth. Stereopsis (the process in visual perception leading to sensation of depth from binocular vision) is the most reliable source of depth and is measured by the threshold value to detect position differences know as stereoacuity. Stereoacuity is described as the minimum angle of separation that two objects can be brought together, whilst maintaining a perception of depth. The most frequently used clinical tests include the random-dot tests of the TNO test, the Lang I and II tests, and the Random-dot E test (Von Noorden and Campos, 2002). Normal stereoacuity values range from 5 - 60 seconds of arc. Depth perception tests often use the random-dot stereogram because they convey no visual information other than random noise, if seen monocularly. However, if binocularly fused, vivid depth perceptions occur (Breyer *et al.,* 2006).

Contrast sensitivity is important for detecting edges such as when crossing door sills and negotiating kerbs (Helbostad *et al.,* 2009). Contrast sensitivity is defined as the reciprocal value of a contrast threshold; contrast threshold is the smallest amount of contrast (difference in luminance) required to see a target or an object. There is now considerable evidence to support the importance of contrast sensitivity assessment on functional vision, with studies showing a significant link to the probability of falling, activities of daily living, control of balance, reading and driving (Leat and Woodhouse, 1993; Turano *et al.,*1994; Wood and Troutbeck, 1994; Lord and Dayhew, 2001; Haymes *et al.,* 2002;

Anand *et al.,* 2003). Despite the numerous tests available to measure contrast sensitivity, the Pelli-Robson letter contrast sensitivity chart (Pelli *et al.,* 1998) is often preferred because measurements are quick and simple and easy for patients to understand, with results being repeatable (Elliott and Bullimore, 1993; Reeves *et al.,* 1993).

2.5 Biomechanical modelling

Biomechanical modelling facilitates increased understanding of human movement (Robertson *et al.,* 2004) through providing insight into the relationships among variables and often indicating how these relationships are governed (Nigg, 2007). Biomechanical modelling occurs through two different categories; rigid body models or mass-spring models (Robertson *et al.,* 2004). Rigid body models consider part or all of the body as a set of rigid segments controlled by joint movements (Robertson *et al.,* 2004). Mass-spring models comprise one or more masses linked to one or more springs (Robertson *et al.,* 2004). Within this thesis both a rigid body and mass-spring model are used for calculating CM (2.5.1) and vertical stiffness (2.5.2) (of the lower extremities) respectively.

2.5.1 Centre of mass (CM)

The point about which the body's mass is evenly distributed is referred to as the CM. The CM is a theoretical position which changes from instant to instant during movement (Hamill and Knutzen, 1995). The most widely used approach to determining the trajectory of the CM in 3-D space is from full body kinematics. Since full body kinematics is calculated based on the weighted sum of the CM of every segment of the body, prior to critiquing the different methods of determining the trajectory of the CM, a brief review of how body segment parameters are estimated will be provided. Some body parts consist of several segments i.e. the foot can be divided into two segments due to the movement available at the metatarsal-phalangeal joint. However, it is common to model most parts of the body as single rigid segments (Robertson *et al.,* 2004). Through assuming single rigid segments, this allows body segment parameters to be quantified easier. Quantifying body segment parameters has been described in cadaver studies (Dempster, 1955), scanning and imaging techniques (Zatsiorsky and Seluyanov, 1983), kinematic measurements (Hatze, 1975) and mathematical modelling (Hanavan, 1964). Dempster's work on cadavers in 1955 produced equations for proportionally determining the body segment parameters needed to biomechanically analyse human movement. This was achieved through cadavers being segmented and their length, masses and volumes being recorded. This allowed Dempster to calculate the location of the centre of gravity (using a balancing technique) and the moment of inertia (using a pendulum technique) for each segment (Robertson *et al.,* 2004). An alternative method of estimating body segment parameters involves scanning living individuals using a radiation technique. Zatsiorsky and Seluyanov (1983) used gamma mass scanning to estimate the mass, CM and principal moments of inertia in 3-D for a 15 segment model (Robertson *et al.,* 2004). The development of kinematic techniques provides indirect measurements of each segment. For example, Hatze (1975) developed an oscillation technique that defines the mass, CM and moment of inertia of segments of the extremities.

However, since this technique is unable to measure an individual's trunk and results are greatly affected by different levels of muscle activation, this approach is somewhat limited. Full body kinematic modelling for 3-D analysis was pioneered by Hanavan in 1964. Hanavan made the assumption that mass was uniformly distributed within each segment and that segments were rigid bodies that could be represented by geometric shapes. By taking additional anthropometric measures of the participant being modelled, such as mid-thigh circumference, malleous height, knee diameter and bi-acromial breadth, equations were developed to compute the three primary moments of inertia.

A number of authors (Shimba, 1984; Crowe *et al.,* 1993; Whittle, 1997) have suggested that the "gold standard" in determining the trajectory of the CM in 3-D space is to doubly integrate the three components of the ground reaction force with respect to time:

$$\Delta CM_i = \int_i \int \frac{F_i}{m} \cdot dt,$$

Where F_i (i = x, y, z) is the ith component of the ground reaction force, m is the total body mass and t is time. Not only is this process time consuming, but this method requires multiple force plates and the known initial conditions (Eames *et al.,* 1999). In a laboratory with only 2 force plates (which is commonly the case), it is not possible to get data for an entire gait cycle as the force under a limb during two of the three periods of double support is not known. For these reasons, this particular method has been used mainly for validation of other CM modelling techniques. An alternative method which avoids the use of multiple force plates for modelling CM is to assume that the CM is a fixed point on the body. Static studies using adult participants have reported that the CM is approximately 60 % of total body height (Hensinger, 1986). It is therefore possible to estimate the trajectory of the CM from either placing a single marker on the sternum, or by using three markers to define the pelvis segment and calculating the geometric centre from this (Eames *et al.,* 1999). However, the criticism of using either the sternum or pelvis to estimate CM location is that both these approaches do not consider how CM location is influenced by relative movement of other body segments i.e. raising both arms in the air will raise the CM, however no difference in CM height would be observed using these methods. A widely used approach which accounts for individual limb movements when calculating the position of the CM is from full body kinematics. The marker set most frequently used to calculate full body kinematics is the Helen Hayes marker set (Fukuchi *et al.,* 2010), originally developed at the Helen Hayes Hospital by Kadaba *et al.* (1990). This marker set consists of a 13 body segment full body set of 28 retro-reflective markers (figure 2.6). Validation of the Helen Hayes marker set for calculating the CM from full body kinematics compared to calculating the CM position from force plates (as well as a fixed point on the pelvis, geometric centre) has been previously highlighted (Eames *et al.,* 1999). Eames *et al.* (1999) calculated the total body CM as the weighted sum of the CM of every segment of the body:

$$CM_i = \frac{\sum_j m_j \cdot P_{ij}}{\sum_j m_j},$$

Where m_j is the mass of segment j, and P_{ij} is the ith component (i = x, y, z) of the position vector of its CM. Values for segment mass and position of CM were obtained using the anthropometric regression equations highlighted by Dempster (1955).

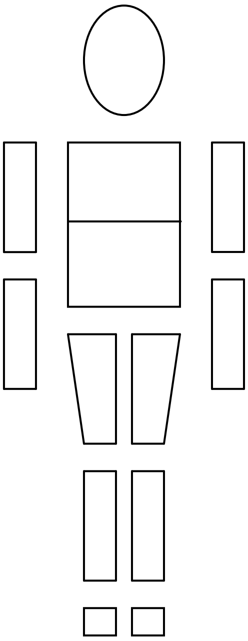

Figure 2.6. The 13 segment model used by Eames *et al.* (1999).

Assessment of the total excursion of the CM produced by each technique (force plates, fixed point and the body segment model) in all three orthogonal components was compared. Results highlighted that the body segment model and force plates methods showed very similar results, whereas the accuracy of calculating CM trajectory using the fixed point was significantly reduced compared to the force plates method. Previous comparisons between force plates calculations and a segmental model have found significant differences between the two (Saini *et al.,* 1998). However, the segmental model used by Saini *et al.* (1998) consisted of 7 segments, 6 segments used for the lower body kinematics and 1 for the upper body (which was assumed to be 1 rigid segment). In contrast, Eames *et al.* (1999) used a 13 segment model (7 upper and 6 lower body segments). The difference between models in both studies likely suggests that if using a segmental model to analyse CM trajectory, the importance of arm swing, trunk and head movements must be separately considered (Eames *et al.,* 1999). It is relevant to note that the limitation of attaching markers to define a body segment model is that anthropometric landmarks lie on the surface of the body and are often removed from the actual joint centre of rotation by various layers of tissue. Thus the position of each segment can only be approximated due to the difficulty locating key anthropometric landmarks (Robertson *et al.,* 2004). In order to reduce placement error, added care must be taken during marker placement and when recording anthropometric measures. This thesis will use the segmental model developed by Helen Hayes. In all but the last study, participants were required to perform single step descents from a stationary standing position (with arms by their side). In the last study, participants performed a step down during on-going gait again with arms down by their side. In all studies it was assumed that the arms would have a minimal influence on the CM and the arms were therefore considered to be part of the trunk.

2.5.2 Vertical stiffness

The concept of stiffness is based on Hooke's law (Butler *et al.,* 2003). Hooke's law explains that the force (*F*) required to deform a material is related to the force / spring constant (*k*) and the distance (*x*) the material is deformed (Butler *et al.,* 2003):

$$F = kx$$

In terms of the human body, stiffness can be described from the level of a single muscle fibre, to modelling the entire body as a mass and spring (Butler *et al.,* 2003). Whilst Latash and Zatsiorsky (1993) suggest that an accurate model must account for all of the components that contribute to stiffness (for a review see Latash and Zatsiorsky, 1993), a model that accounts for all the components that influence motion is very complicated and becomes impractical (Butler *et al.,* 2003). Thus a simple approach is often used whereby the leg is modelled as a spring supporting the mass of the body. This model is termed a 'spring mass model' (figure 2.7). Use of the spring mass model is not without controversy (Hortobágyi and DeVita, 1999, 2000) and limitations (Arampatzis *et al.,* 1999; Latash and Zatsiorsky, 1993) i.e. modelling lower extremity stiffness as a spring-mass system assumes that the stiffness generated by the tissues surrounding the ankle, knee and hip joint occurs in a linear fashion, however this is not the case (Latash and Zatsiorsky, 1993). Nevertheless, despite the limitations of modelling the entire leg as a single spring mass model, such an approach has been used within the literature to understand human movement control (Arampatzis *et al.,* 1999).

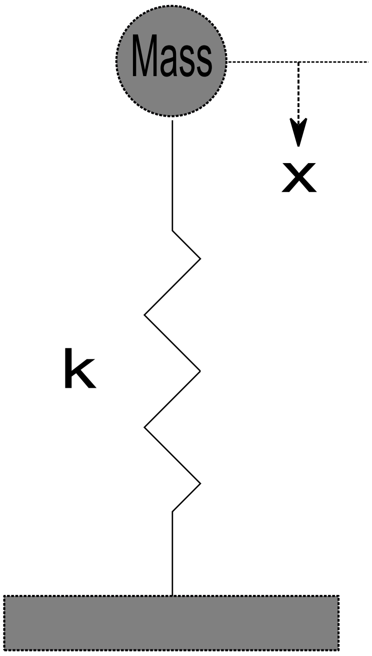

Figure 2.7. Spring and mass model used to calculate vertical stiffness. The vertical impact of the mass deforms the spring (k) allowing the calculation of stiffness based on the distance the spring (x) is deformed. Adapted from Butler *et al.* (2003).

There are several different calculations of lower body extremity stiffness including vertical, leg and joint stiffness. Vertical stiffness is often used to describe linear movements that occur in the vertical direction such as running (McMahon *et al.,* 1987; Nigg and Liu, 1999), hopping (Farley *et al.,* 1991) and step descent (Hortobágyi and DeVita, 1999). During running or hopping, results consistently highlight that lower extremity stiffness increases as velocity of the activity increases, which may be necessary to resist collapse of the limb during the early phase of landing and allow for maximum energy return during the propulsive phase (Arampatzis *et al.,* 1999, 2001). During step descent, increases in lower body extremity stiffness have been observed in elderly compared to young women. This increase subsequently resulted in elderly women reducing joint excursions as well as muscle loading, which was attributed to reduced muscle strength (Hortobágyi and DeVita, 1999). While a certain level of stiffness may be necessary for performance, too much or too little stiffness may lead to injury (Butler *et al.,* 2003). Since increasing stiffness typically reduces lower extremity excursions and increases peak force, this typically increases loading rates which has been associated with increased shock to the lower extremity (Henning and Lufatune, 1991). However, too little stiffness may allow for excessive joint motion, resulting in soft tissue injury (Butler *et al.,* 2003). This suggests that there may be an ideal range of stiffness which is most effective for performance whilst minimising the risk of injury.

Since this thesis tasks individuals with step descent under various experimental conditions, only vertical stiffness models relevant to step descent will be discussed from here on in. Hortobágyi and DeVita (1999) calculated lower body extremity stiffness for the initial landing period as participants completed step descents. Lower extremity stiffness (*K*) was defined as:

$$K = \frac{F_{\max}}{X_{\max}}\ (\mathrm{kN/m})$$

Where Force ($F_{\max}$) was calculated as the maximal value of the resultant ground reaction force vector applied under the foot and directed from the 5th metatarsal head to the hip joint during initial landing (time interval between ground contact and the instant of maximum knee flexion). Displacement (X_{max}) was determined as the maximum shortening of the distance between the 5th metatarsal head and hip joint during initial landing. A criticism of the lower body extremity stiffness model used by Hortobágyi and DeVita (1999) is that it does not take into account the role of the trail leg in regulating landing control. During step descent, both lead and trail limbs are in contact with the floor when the lead limb contacts the lower level (period defined as double support). Therefore the trail limb must exert some influence on landing stiffness. However, Hortobágyi and DeVita (1999) suggested that this influence from the trail limb was minimal. In an attempt to also determine the contribution of the trailing limb to lower body extremity stiffness in step descent, Buckley *et al.* (2005) adapted a model developed by McMahon and Cheng (1990). The amount of bodyweight supported on the trail leg at the end of the initial landing period (initial landing period defined from contact with the lower level to lead limb peak knee flexion) was also evaluated (Buckley *et al.,* 2005). This model is described by McMahon and Cheng (1990):

$$K_{vert} = \frac{Fz_{\text{max}}}{\Delta l} \quad \text{(kN/m)}$$

The lower extremities were modelled as a linear spring, with the length of the spring determined as the height of the CM above the ground and spring loading determined from the vertical ground reaction force (Buckley *et al.,* 2005).

Vertical stiffness (K_{vert}) was computed for the initial landing phase, as the ratio of maximum vertical ground reaction force (Fz_{max}) to the change in length of the spring (Δl) i.e. the vertical displacement of the CM. Through determining the change of the spring length through the CM, this considers the role of the trail leg since increasing weight on the trail leg during initial landing would reduce vertical displacement of the CM thus affecting vertical stiffness. Indeed Buckley *et al.* (2005) reported that under conditions when vertical stiffness was reduced, the amount of bodyweight supported on the trail leg increased. Differences between how Buckley *et al.* (2005) and Hortobágyi and DeVita (1999) interpret the contribution of the trail limb to landing stiffness may be attributed to their methodologies. Buckley *et al.* (2005) used smaller step heights (7.5 cm, 14.6 cm and 21.8 cm) compared to a much higher step height (~33 cm) by Hortobágyi and DeVita (1999). When descending smaller step heights, the trail limb would be able to exert more control on the body during step descent. Since step heights used in the subsequent experimental chapters of this thesis closely reflect those used by Buckley *et al.* (2005), the same vertical stiffness model will be used to calculate lower limbs stiffness.

Chapter 3

General methods

All the experiments included in this thesis were collected in the Vision and Mobility / Biomechanics Research Laboratory at the University of Bradford. The following chapter provides a detailed description of this laboratory, including the set up and equipment used for the experimental work undertaken. In addition, this chapter details the criteria for participant's inclusion, the various modelling techniques employed during data collection, the analysis of data and treatment of extreme data scores identified in the subsequent analysis of the data. Details of specific methodologies, including data and statistical analyses and procedures for each experiment can be found within the appropriate chapter (chapters 4 - 8).

3.1 Participants

The majority of participants used in the subsequent experimental chapters (4, 5, 7 and 8) were recruited from the student population at the University of Bradford. The remaining participants recruited were either family or friends. Each participant was recruited through opportune sampling and was a young and healthy adult. In chapter 6, participants were healthy older adults recruited by a previous Ph.D student. For further information pertaining to the participants used, see chapter 6 or Johnson (2008). The remainder of this sub-section will only discuss the recruitment process for participants used in chapters 4, 5, 7 and 8. Prior to taking part in the study, participants received an information sheet outlining the aim of the particular study they were being asked to take part in and what would be required from them (for an example see appendix 1). Upon agreeing to take part in the study, their health was assessed via a self-report questionnaire (see appendix 2). Participants were excluded from the study if they reported any history of neurological, musculoskeletal or cardiovascular disorders which could affect their balance or gait, or history of eye disorders including amblyopia or strabismus. Participants who did not engage in moderate or high physical activity (Allied Dunbar Fitness Survey, 1992) for at least three times per week for 30 + minutes were also excluded. This ensured all participants were able to complete the experimental protocol in full, without being overly fatigued. Prior to the day of data collection, participants were informed that data collection would occur during a single session and they were required to wear shorts, t-shirt and appropriate footwear (flat soled shoes used for everyday wear). Participants agreed to refrain from alcohol intake during the 24 hours prior to testing. Studies were conducted in accordance with the declaration of Helsinki and ethical approval was obtained from the University of Bradford research ethics committee. Written consent was obtained from each participant prior to testing (see appendix 3).

3.2 Visual screening

To assess each participant's suitability for the study, three separate vision tests were conducted. Vision tests comprised of visual acuity, contrast sensitivity and stereopsis. The rationale for their inclusion has been previously discussed in chapter 2.3. The total number of letters correctly read (visual acuity and contrast sensitivity), or the total number of screening plates correctly identified (stereopsis) were added together and a final score was given for each individual vision test, which were based on the pre-determined scoring systems (see below for specific details) for each individual test. Participants with a score above 0.0 logMAR (visual acuity) or 120 secs of arc (stereoacuity) or below 1.65 log units (contrast sensitivity) were excluded. These visual assessments ensured that participants scored within the limits of healthy eyes (Vale *et al.,* 2008a) so subsequent findings could be attributed to experimental manipulations not inadequate vision.

3.2.1 Visual Acuity

Binocular visual acuity was measured using the Early Treatment Diabetic Retinopathy Study (ETDRS) logMAR (log of the Minimum Angle of Resolution) chart, at a 4.0 m working distance with luminance of 130 cd/m^2 (Ferris and Bailey, 1996). The chart consisted of five Sloan letters per line and a logarithmic progression of letter size (see figure 3.1). Each completed line of the ETDRS chart corresponded to a pre assigned logMAR score. The interline size progression was 0.1 log units and if only part of the line was completed, a score of 0.02 log units was assigned to each successful letter read. Visual acuity threshold was determined when four out of five letters were read incorrectly on a line (Carkeet, 2001). Once participants reached what they thought were the smallest letters they could read, they were encouraged to attempt to determine additional letters. Participants were encouraged to attempt to identify further letters as some individuals are more cautious than others, only identifying letters that are clear and easy to read (Elliott and Flanagan, 2007).

Figure 3.1. ETDRS chart used to assess visual acuity.

3.2.2 Contrast sensitivity

Binocular contrast sensitivity was measured using the Pelli-Robson contrast sensitivity test chart at 1.0 m working distance and chart luminance of 200 cd/m^2. The wall chart measured 90 x 60 cm and comprised 8 lines of letters of size 4.9 cm x 4.9 cm. Each line consists of 6 letters with the first 3 letters (a triplet) on the left having greater contrast than the 3 letters (a triplet) on the right (see figure 3.2). The first triplet had a log sensitivity value of 0.00. Each subsequent triplet of letters corresponds to a 0.15 log unit decrease in contrast. To improve reliability and sensitivity of the measurement (Elliott, 2006) a scoring rule of 0.05 log unit was assigned per letter correctly read (including the letter ‘C’ read as an ‘O’). Contrast sensitivity threshold was determined when two out of three letters were read incorrectly. When participants were unable to see any further letters on the chart, the next lower contrast triplet was pointed to and participants were asked to keep looking at this area for ~20 seconds to see if the letter(s) became visible. Participants were encouraged to guess if they were uncertain of the letter.

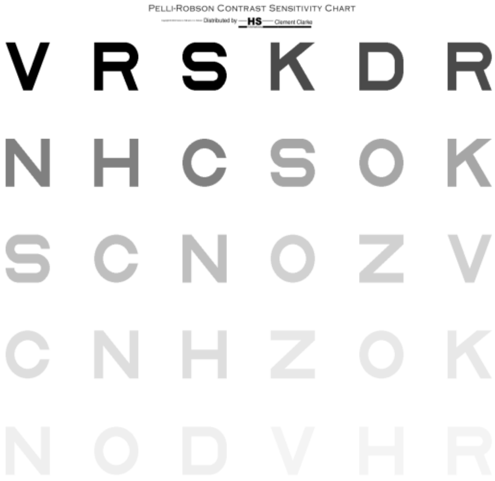

Figure 3.2. Pelli-Robson contrast sensitivity chart (Pelli *et al.,* 1988).

3.2.3 Stereopsis

Stereopsis was measured using the TNO Random-dot stereogram test at a testing distance of 40 cm and positioned at an angle so that the book was parallel to the plane of the participant’s face. The TNO test consists of red and green anaglyphs and requires the use of red and green filter goggles to create a stereoscopic image due to the disparity caused by the red and green dots (see figure 3.3). The TNO test measures 1,980 seconds of arc in the screening plates and from 480 to 15 seconds of arc on the quantitative section. During the quantitative section participants were required to accurately identify where wedges had been removed from either, the top, bottom, left or right of circles. Participants were given as much time as they required to view each particular plate. If only one of the two plates for a particular stereo level was identified, participants were given a second attempt. However, if the same plate was wrongly identified on the second attempt, the previous correctly identified stereo level was recorded.

Figure 3.3. The random-dot stereogram test used to assess stereopsis.

3.3 Movement laboratory set up and equipment

The five experimental chapters presented in this thesis were all undertaken in the Vision and Mobility / Biomechanics Research Laboratory at the University of

Bradford. The equipment used to collect data comprised of an 8[1] camera motion analysis system (Vicon MX3, Oxford Metrics Ltd), a computer Data station with software suitable for data processing, calibration equipment and reflective markers. The motion analysis system recorded participants 3-D segmental kinematics and force data (collecting at 100 Hz) when stepping from one level down to another following the protocols of each particular study.

The cameras were either wall or ceiling mounted at approximately 2.3 m above the floor and were positioned around the laboratory to view the participant from all angles (see figure 3.4). The laboratory which measured 5.77 m x 7 m x 2.77 m was illuminated via 6 fluorescent lighting tubes which were mounted within the ceiling (~2.8 m above the floor). The fluorescent lighting was the only form of lighting within the laboratory as black-out roller blinds were used to occlude external lighting. The luminance over the step area, measured using a photo-meter (CS-100 Minolta Co Ltd.), was approximately 400 lux (taken at participant’s chest height), and the luminance of the floor and top surface of the step was 15 and 30 cd/m² respectively.

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[1] Data collected in chapter 6 used a 5 camera motion analysis system and were collected by a previous Ph.D student. However, the previously collected data were re-analysed and interpreted independently using a different approach. Further information detailing the laboratory set up can be found in Johnson (2008).

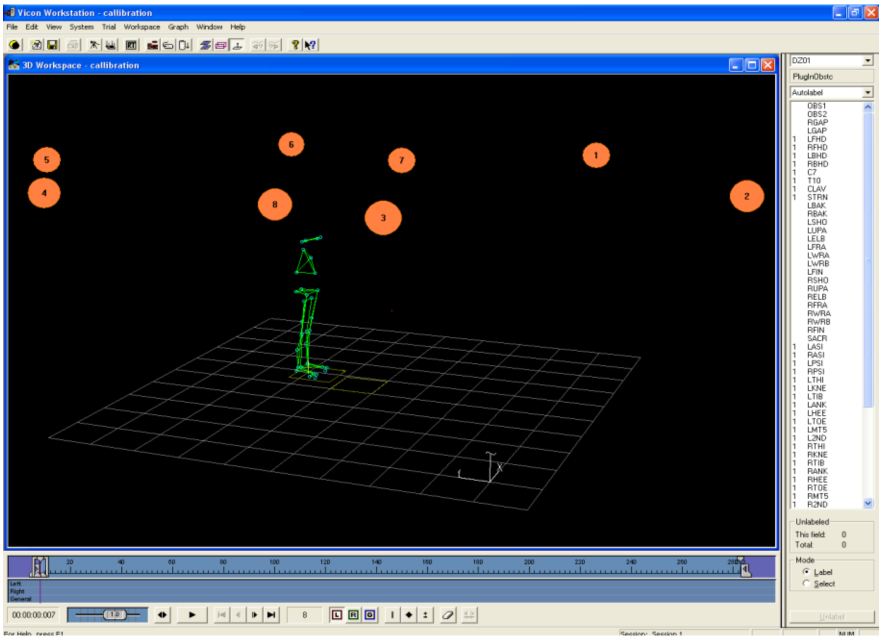


Figure 3.4. Position of the 8 Vicon cameras (orange circles) around the Vision and Mobility / Biomechanics Research Laboratory.

3.3.1 Step heights used

Government standards and current literature on step heights commonly faced within the environment were reviewed to determine the step heights for the experimental chapters. The following heights were selected as they reflect the environmental demands participants face in everyday life. Particular focus identified kerb heights, individual step heights on stairs and stepping on / off busses. There are no government guidelines concerning kerb heights within the UK (search conducted at department for transport, 2010). However, the requirements for kerb heights used to raise the footpath from the road around a bus stop specify a height between 140 - 160 mm (The Public Service Vehicles Regulations, 2002). Due to the lack of appropriate guidelines regarding kerb height, a range of 40 kerbs from around the local area were also measured (table 3.1).

Table 3.1. The mean, median and range of forty kerb heights from around the city of Bradford.

Mean	Median	Range
79 mm	85 mm	20 -150 mm

Powell-Smith and Billington (1986) highlighted that UK building regulations require individual step heights for indoor stairs to range from a minimum of 150 mm to a maximum height of either 220 mm (private stairs), 190 mm (common stairs) or 180 mm (institutional stairs). Step height regulations for escalators used in a public place specify that each individual step height must range from 210 - 240 mm to allow the escalator to be used as an emergency exit when stationary (inclusive mobility, 2010a). When alighting from a bus, a standard bus with no ‘kneeling’ mechanism leaves a step down height of 250 mm from the roadway (inclusive mobility, 2010b). Since the average kerb height around a bus stop ranges from 140 - 160 mm, step downs from a bus range between 90 - 110 mm.

Step descents completed from an increased height present a greater threat to stability. Therefore, to increase the likelihood of observing any effects of experimental conditions (i.e. visual field, timing of visual occlusion and / or added weight conditions) within the following experimental chapters, pilot work was conducted to determine a step height which reached the upper limit whereby participants could control step descent. 15 % (± 1 cm) participant’s total body height (15%bodyHt) was chosen, as heights above this potentially involve a period of ‘free fall’, whereby the trail limb is unable to exert control on the body during descent. Previous research has calculated step height based on a percentage of participant’s leg length (calculated from the anterior superior iliac spines to the medial malleoli, Cowie *et al.,* 2008) or total body height (Hortobágyi and DeVita, 1999, 2001). The accuracy of precisely determining leg length using the approach described by Cowie *et al.* (2008) will be significantly affected by participant’s body type (i.e. ectomorph, mesomorph, endomorph or a combination). For example, an endomorph body type will have increased amounts of adipose tissue located around their abdominal region, thus increasing the difficulty of accurately locating their anterior superior iliac spines. Therefore, to ensure that the highest step height was accurately calculated for each participant irrespective of body type, a percentage of participants total body height was chosen in favour of percentage leg length. Based on the above literature and also added constraints of using 18 mm thickness medium density fibreboard (MDF) to construct the steps, the following step heights were used:

- Low step height - 146 mm

  Kerb heights around a bus stop (range from 140 - 160 mm).
- High step height - 218 mm

  Escalators in a public place range from 210 - 240 mm and the maximum height of indoor private stairs is 220 mm.
- 15%bodyHt - 15 % (± 1 cm) of participant total body height

Steps were constructed from sheets of MDF, which were bonded together to create a solid block with standing area 464 mm x 508 mm. Steps were covered with a 2 mm thick vinyl material, to match the surrounding floor.

3.3.2 PLATO liquid crystal goggles

Data collected in chapters 4, 5 and 7 required participants to wear PLATO (Portable Liquid Crystal Apparatus for Tachistoscopic Occlusion) goggles (Translucent Technologies, Canada, figure 3.5) throughout the experiment. The PLATO lenses are constructed with specially designed liquid crystal cells, powered by an electrical field applied across the two glass plates. Under command of a switching or electronic timing circuit, the experimenter is able to change the state of the cells from transparent to opaque. In the open state, looking through the lenses is like looking through clear glass. In the closed state, the lenses scatter light and thus take on a translucent milky texture, which prevents the participant from receiving visual information. The participant's eye nevertheless remains illuminated and thus does not have to re-adapt to light when the lenses 're-open'.

Figure 3.5. View of the PLATO goggles in both translucent and opaque conditions.

The response time of the lenses is approximately 4 ms to reach the open state and 3 ms to reach the closed state (PLATO Visual Occlusion Spectacles, 2010). Power is supplied to the spectacles by a specially designed compact battery operated power supply, which fits into the participant’s shorts pocket or can be clipped onto their belt. Black cloth material was attached to the frames to eliminate visual feedback from around the edges of the goggles. Participants familiarised themselves with the goggles in the transparent and opaque conditions prior to the experiment (except in chapter 7). The size of the frames surrounding the lenses of the PLATO goggles (see figure 3.5) would have occluded the outer peripheral regions of the participant’s visual field. To determine the extent of visual field loss when wearing the goggles, a sample of participants (n = 5) were tested both with and without wearing the goggles. Binocular Estermann visual fields, the standard used by the UK Driving and Vehicle Licensing Agency (DVLA) to determine safety to drive were assessed using the Humphrey Field Analyzer (Carl Zeiss Meditec, Inc., Dublin, CA). The output from the visual field test of one participant both with and without wearing the goggles is illustrated below in figure 3.6. Results from the visual field test highlighted that some or all of the outer most peripheral regions of the visual field were occluded when wearing the goggles. Across the participants tested when wearing the goggles, visual field was reduced to no less than 70 ° in the horizontal field in both eyes and 30 ° superior and 50 ° inferior of the horizontal meridian. Whilst these results clearly demonstrate that wearing the goggles reduced participant’s field of view, a 120 ° or more of continuous horizontal visual field with no significant field defect 20 ° above or below the horizontal meridian is the required standard to have sufficient field of vision to be safe to drive (UK DVLA visual standards, 2010) (further details of a ‘normal' full visual field can be found in chapter 2.2). Since the horizontal visual field of the participants tested remained ≥ 140 ° and no defects were within 30 ° of the horizontal meridian, wearing the goggles was not thought to have a significant effect on the ability to sample visual cues from within the environment. Of note, the back vertex distance (distance between the lenses of the goggles and the participant’s eye) will vary between participants, thus the amount of peripheral visual field occlusion will vary slightly. However, it is highly unlikely that the back vertex distance will change to such an extent that continuous horizontal field of view will reduce to less than 120 ° or encroach within 20 ° of the horizontal meridian.

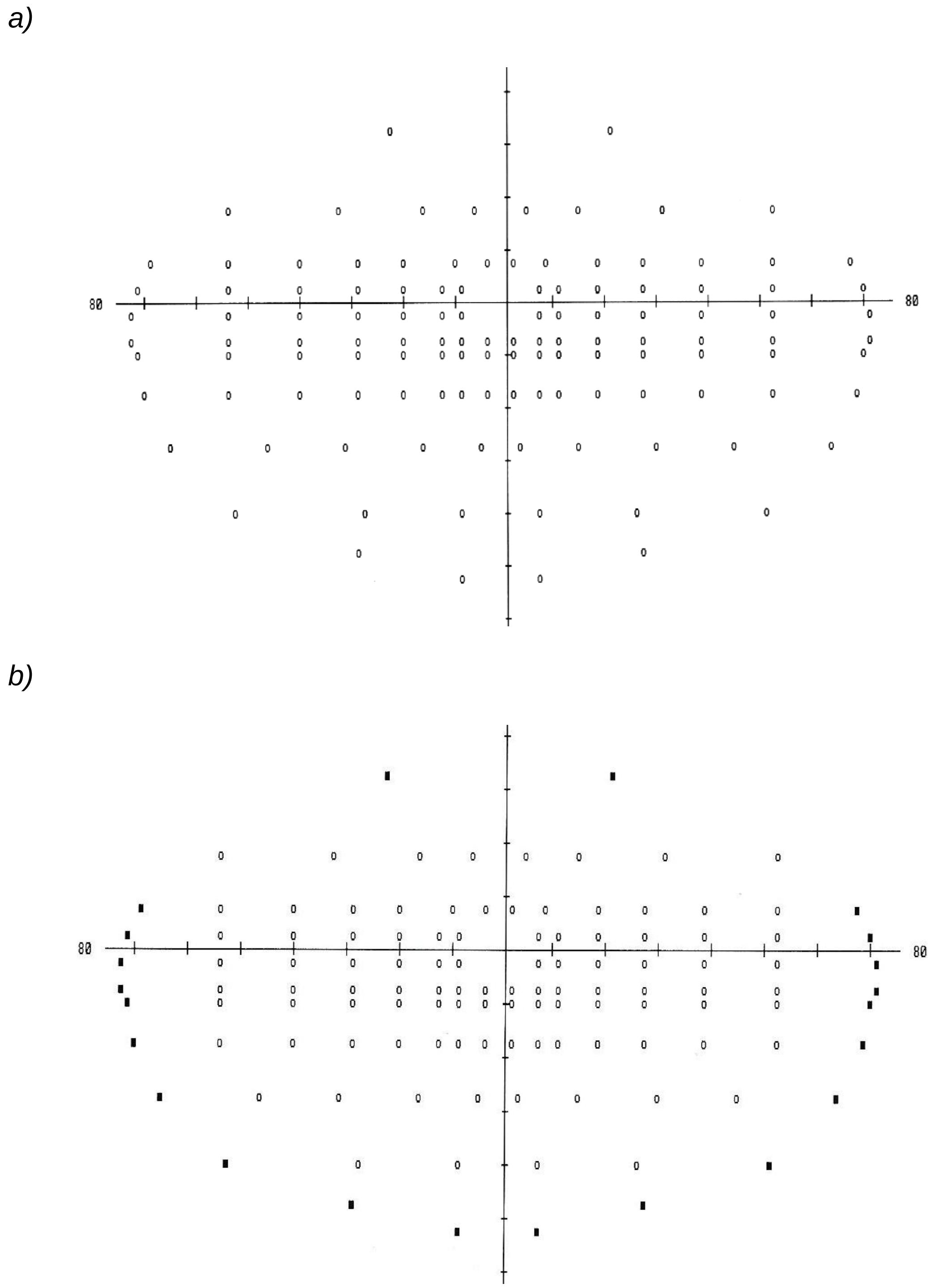


Figure 3.6. Exemplar data of the extent of a participant’s visual field tested a) without wearing the PLATO goggles and b) wearing the PLATO goggles. The hollow black circles highlighted in the figure above represent areas seen by the participant. The black rectangles located towards the edges of the peripheral visual field (b) represent areas not seen by the participant when wearing the PLATO goggles.

3.3.3 Pressure mat and variable delay switch

To control when the lenses of the PLATO goggles were triggered to change from transparent to opaque (at toe-off or mid-swing) a pressure mat (Lafayette Instrument, USA, figure 3.7a) placed directly under the ball / toes of the lead limb foot was connected ‘in series’ to a variable delay switch that was set to an appropriate delay for each trial (figure 3.7b). Participants were instructed to place their lead foot on the pressure mat, between two black stripes of tape placed on the mat (figure 3.7a) which was found during pilot work, to be the most sensitive part of the mat. Timings of delays (table 3.2) for both low and high step heights were calculated based on average single support times obtained from the same step heights descended with a similar participant population (Jones *et al.,* 2005; Lythgo *et al.,* 2007; Buckley *et al.,* 2008). Pilot work was undertaken to determine the time delay for the highest step height (15%bodyHt). This also provided an opportunity to further confirm the accuracy of single support timings calculated for both low and high step heights.

a)

b)

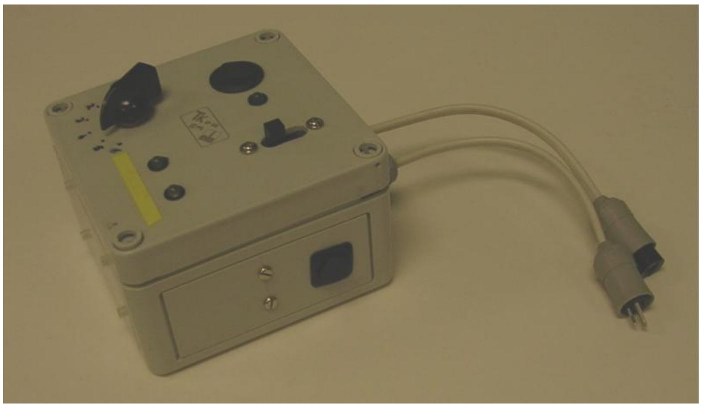


Figure 3.7. a) Pressure mat which was connected to a b) variable delay switch to occlude vision (online) from set points onwards during step descent.

Table 3.2. Timing delays used at each step height to occlude vision from mid-swing onwards during step descent. These timings were used in chapters 4, 5 and 7.

Low step	High step	15%bodyHt
300 ms	300 ms	350 ms

3.3.4 Liquid crystal sheet

The liquid crystal (LCD) sheet (manufacturer unknown, figure 3.8) used in chapter 8 operates largely in the same manner as the PLATO goggles. The sheet consisting of liquid crystal cells, was powered by an electrical field applied across each end. The response time of the LCD sheet was ~5 ms to reach both opaque and translucent states. The sheet was powered by a 9 v rechargeable battery. When opaque, the LCD sheet eliminated all visual information whilst maintaining ambient light levels so that no dark adaptation occurred (figure 3.8).

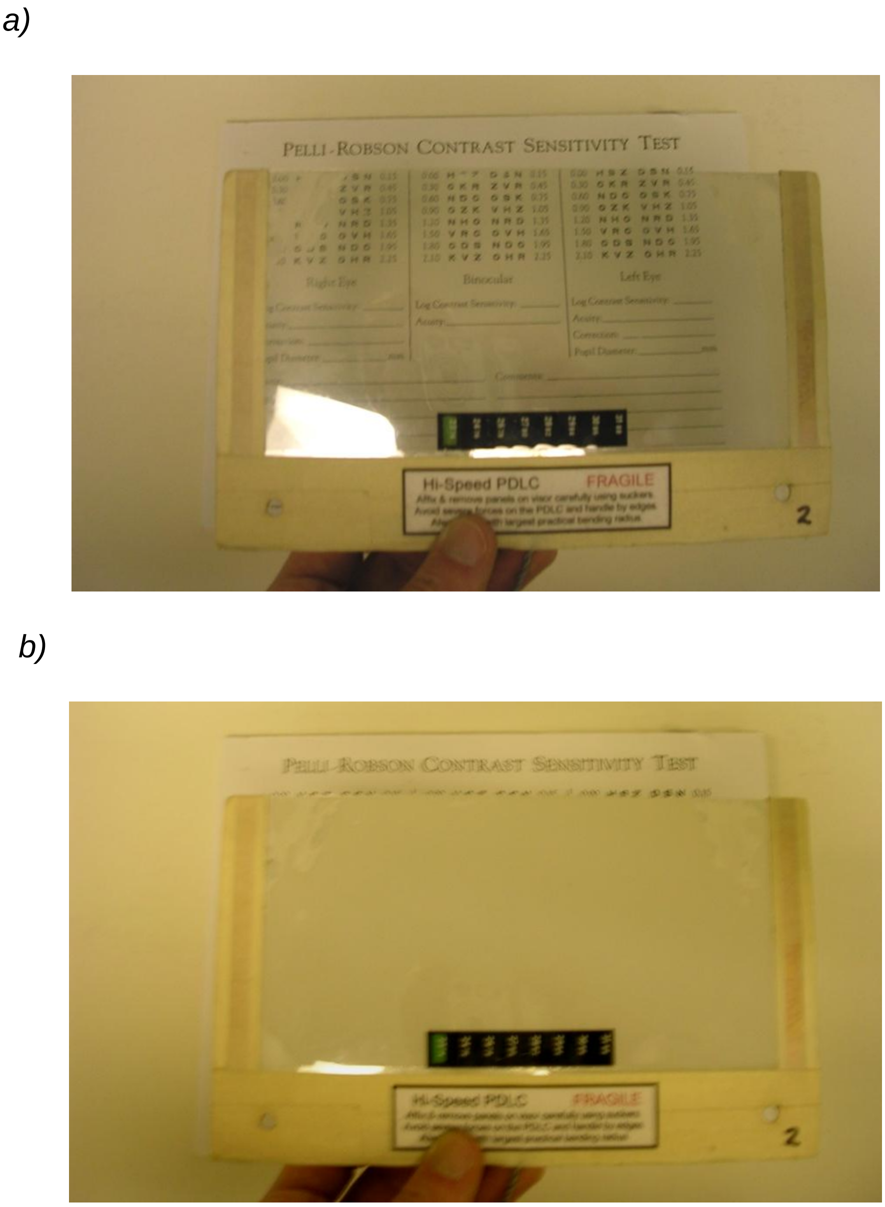


Figure 3.8. LCD sheet in the a) transparent condition and b) opaque condition which eliminates all visual information.

3.3.5 Motion capture / analysis system

Kinematic data were collected (100 Hz) using 8 MX3 Vicon cameras (Oxford Metrics, Ltd) which used digitally controlled strobes that emit infrared light. The infrared light from the cameras illuminated the retro-reflective markers attached to the participant. The markers reflected the light back into the camera which was processed through a filter which only allowed light of one particular wavelength to be focused on the camera's sensitive plate. Information which was reflected back from the markers to the camera converted the pattern of light into an electronic video signal, which was transmitted to the data station where it was processed into a 2-D record of marker location at any given time. From the 2-D images collected, and using the camera parameters obtained through the DynaCal calibration procedure, Workstation was able to reconstruct a 3-D location of each marker for each movement frame. The Vicon system then links the locations of each marker together to form continuous trajectories. This process describes the path each individual captured marker trajectory has taken during the capture period, representing how the markers have moved over time.

Prior to data collection, calibration of a pre defined capture volume took place which defined the area where motion data were to be collected (figure 3.9). This calibration process determined the positions of the cameras relative to each other and the laboratory origin, which enabled the orthogonal axes within the capture volume to be defined. Calibrating the system defined the capture volume and the relative position and orientation of the cameras. Calibration was undertaken using the DynaCal calibration procedure (Vicon: Oxford Metrics LTD, Oxford, UK) which consisted of both a dynamic and static process. The static calibration uses an L-Frame (46 x 55 cm) to define the origin (i.e. 0,0,0 co-ordinates) of the capture volume. The origin for the co-ordinates was placed at the corner of force platform 1 (figure 3.10). This calculated the origin of the capture volume and determined the orientation of the 3-D Workspace. Dynamic calibration involves moving a ‘T’ shaped calibration wand (88 cm length x 41 cm) continuously throughout the whole volume allowing each camera to calculate the position and orientation relative to the origin (figure 3.10). Following calibration, a .CP file containing the parameters of all the cameras was automatically processed in Workstation which was subsequently used in the reconstruction of the 3-D marker trajectories in the dynamic trials. The reconstruction errors of the 3-D marker locations averaged 0.47 mm (range 0.19 - 1.17 mm).

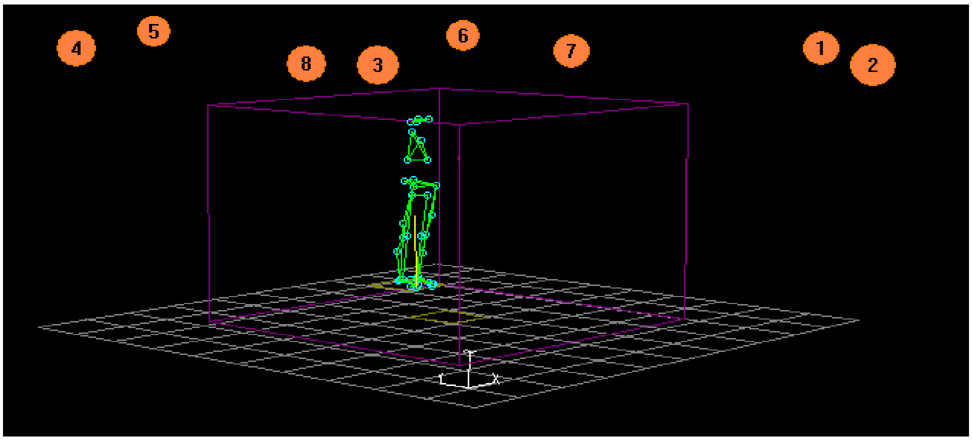


Figure 3.9. Pre defined capture volume (area inside the purple box) where cameras (orange circles) recorded data from the retro-reflective markers attached to the participant.

Once the capture volume was defined and the calibration was complete, participants had 27 passive reflective spherical markers located at the following anatomical landmarks (placed either directly on to the skin, or onto clothing): anterio- and posterio-lateral aspects of the head, vertebrae C7 and T10, jugular notch, xiphoid process, anterior superior iliac spines, sacrum (placed on a 5 cm long wand), and bi-laterally on the lateral aspects of the thigh and shank, lateral femoral condyles, lateral malleoli, calcanei, superior aspects of end of hallux and second and fifth metatarsal heads (see figure 3.11). The sacral marker was attached via a plastic wand to minimise the obstruction of the marker from clothing during data collection. 6 mm diameter markers were attached to the feet and 14 mm markers were attached elsewhere on the participant's body. Smaller markers (6 mm) were chosen for the feet since this allowed 3 markers (superior aspects of end of hallux and second and fifth metatarsal heads) to be placed on the foot in close proximity without markers 'merging' during data capture. Placing these markers on the foot subsequently allowed the foot to be modelled as two segments (hind / mid foot and forefoot), rather than as one entire 'foot' segment.

Figure 3.10. The L-Frame placed at the origin of the capture volume and the 'T' shaped wand used for static and dynamic calibration respectively.

Once all markers had been located on the participant, anthropometric measurements including, participant knee and ankle width (measured in the frontal place), height, mass, inter-ASIS distance and leg length were recorded (see appendix 4). A subject calibration file was then created in the Workstation operating software, by recording a static trial for each participant, of them standing in a stationary position with arms slightly out from their side (this ensured the arms did not cover the markers placed on the anterior superior iliac spine). Once the static trial data were collected, the individual marker points were assigned their respective anatomical locations within the software. Manually labelling the markers in this way during the static calibration provided the Workstation software with the ability to auto-label the captured marker trajectories in subsequent trials. Any captured trajectories which were incorrectly labelled in the auto-label function were manually corrected during data processing. A cubic spline interpolation was used to ‘fill’ gaps in trajectories of less than six time intervals. Gaps of more than five time frames were interpolated by copying the trajectory of a marker that was attached to the same limb segment; this method produced the most realistic trajectory. However, if a realistic trajectory was not produced when copying the trajectory from the same limb segment, the gap was not filled.

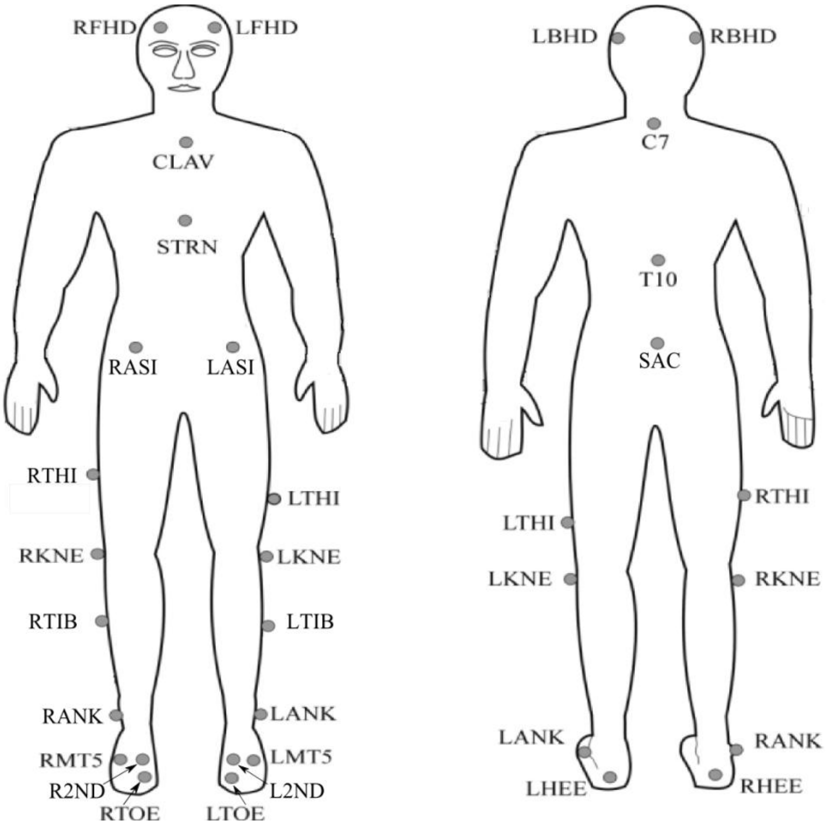


Figure 3.11. Anatomical locations of the placement of the retro-reflective markers. To ensure that marker placement did not interfere with participant's gait, markers were placed on lateral aspects of segments (adapted from the Vicon Motion Systems Preparation Manual, 2010).

Based on pilot work (see sub section 3.3.6), the 3-D markers were processed using a smooth filtering routine with a predicted mean squared error (MSE) value of 10 and processed through the Vicon animation pipeline. These data, along with the previously collected anthropometric data, were then used to define a 3-D linked segment model of the participant (figure 3.12).

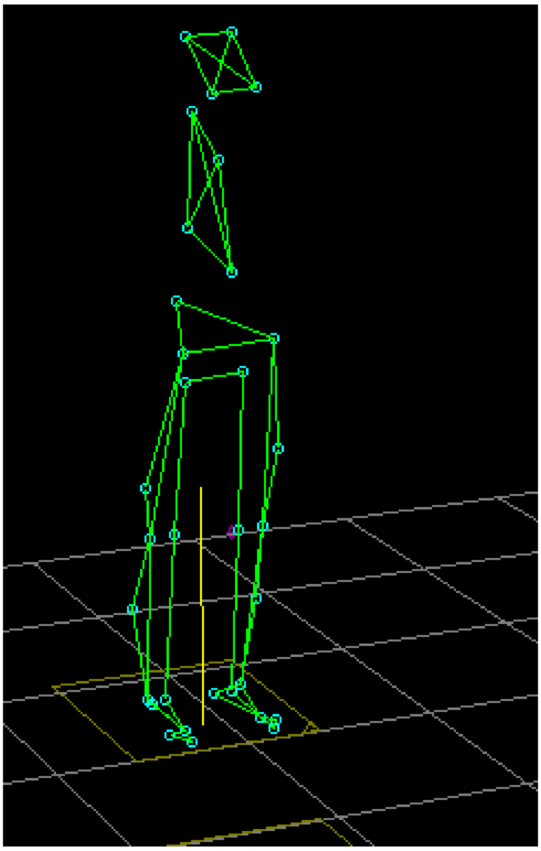


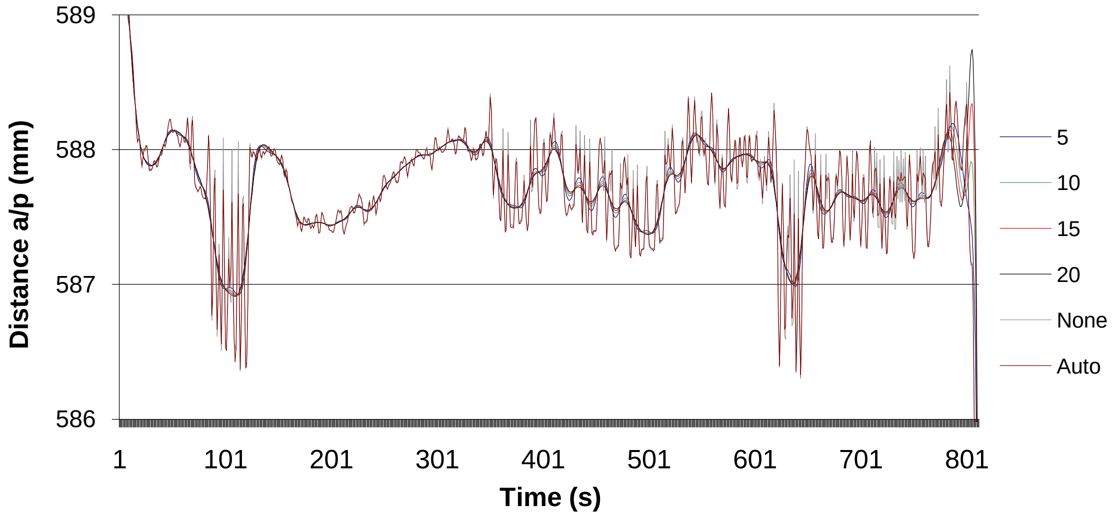
Figure 3.12. Three dimensional segment model of the human body.

3.3.6 Smooth filtering routine

Pilot work was undertaken to determine the most suitable MSE smooth filtering value to apply to the processed data. This consisted of analysing different marker trajectories across a range of markers, from a number of different trials, whereby a participant completed a single step descent which was processed using a range of smoothing options (no smoothing, auto function and MSE value ranging from 5 - 20). When the automatic smoothing option was applied to the data, the trajectory was similar to the no smoothing option and failed to filter out the noise inherent within the data. Using an MSE smooth filtering value ranging from 5 - 20 filtered out this noise. Removing the noise inherent within the data allowed stricter criteria to be used to determine key points within the movement. For example, the exemplar data presented below in figure 3.13a were used to calculate lead limb toe-off and subsequent single support time. Through filtering out the noise in the data (in this case removing ~2 mm a/p displacement), the criteria for defining lead limb toe-off could be set to when the a/p displacement of the second metatarsal head marker was greater than 3 mm for 5 consecutive frames.

Using an MSE smooth filtering value of 5 over filtered the data, subsequently removing key parts of the trajectory. The exemplar data presented in figure 3.13b below illustrates that filtering the data using an MSE value of 5 (blue line) removed the peak immediately prior to toe-off. In contrast, using a MSE value of 20 resulted in over compensating key parts of the trajectory. For example, in figure 3.13b an MSE value of 20 (black line) produced a peak at toe-off higher than the ‘none’ filtered data (grey line). Whilst an MSE value of 10 (green line) and 15 (red line) presented in the figure below appear very similar, in a number of trials analysed, an MSE value of 15 filtered the data in a similar manner to an MSE value of 20. Therefore an MSE value of 10 was applied to the experimental data as this option filtered the data in a manner best representative of expected movement trajectory.

a)



b)

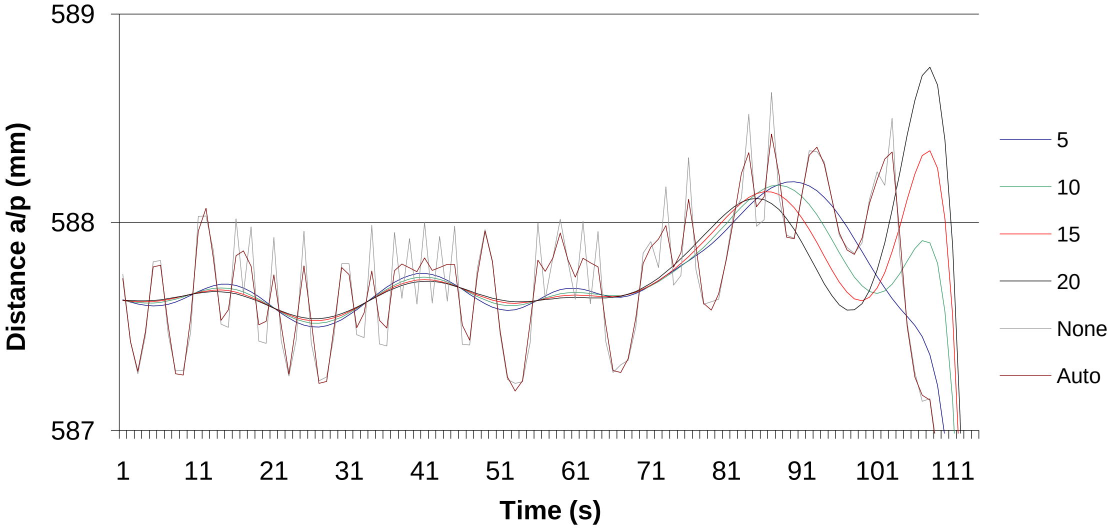


Figure 3.13. Exemplar data recorded from the a/p trajectory of the right foot second metatarsal head representing how movement trajectory is affected dependent upon the filtering option used. Filtering options included no smoothing (grey line, none), auto smoothing (maroon line), MSE smoothing option of ‘5’ (blue line), ‘10’ (green line), ‘15’ (red line) and ‘20’ (black line). Figure a) plots data from standing stationary on the upper step to the period immediately after right foot toe-off b) magnified view of toe-off.

3.3.7 Force platforms

Two AMTI OR6-7 (Advanced Mechanical Technologies Inc., Boston, USA) force platforms (figure 3.14) were mounted flush with the surface of the laboratory floor. The two force platforms each had a top surface area measuring 464 x 508 mm located in the middle of the laboratory adjacently positioned with a 3.2 mm gap between them and a 10 - 20 mm gap between the edge of each force platform and the surrounding laboratory floor, as per the manufacturer's instructions. The top surface of the force platform along with the surrounding floor was covered in a 2 mm thick foam backed vinyl floor covering to prevent slipping. Due to interference from the output connectors on the force platform, one force platform was rotated 180 º and positioned oppositely to the other so that both platforms could be positioned as close as possible but leaving the required 3.2 mm gap. To ensure that both force platforms had the same co-ordinate connection, the polarity of one platform (platform 2) was reversed (process occurred within the operating software).

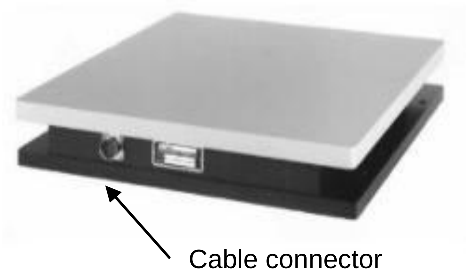


Figure 3.14. AMTI model OR6-7 force platform.

Any forces and moments acting on the force platforms' rigid upper plate were measured using electrical resistance strain gauges which were attached to a load cell located at each corner of the force platform. The gauges form six Wheatstone bridges, three of the output signals are proportional to the forces parallel to the three axis and the other three outputs are proportional to moments about the three axis. The force platform measured the orthogonal ground reaction force components along the X, Y and Z axis and the moments about the three axis producing a total of six outputs, allowing a full understanding of the forces (both magnitude (size) and direction) being applied to the object. Force is a vector quantity having both magnitude and direction; the magnitude measured in Newtons (N). The force generated to rotate the body about some point, termed the moment, was calculated as the product of the force and the distance from the point to the direction of the force.

The output of the force plates were sent via a six-channel strain gauge amplifier (AMTI MSA-6) which provided amplification for each channel of the force plate to the Data station. The X, Y and Z coordinates were positioned such that origin of the global reference system was located at the bottom left hand corner of force platform 1. The positive Y coordinate pointed towards the second force platform and the positive Z axis pointed up from the floor. The positive X axis was positioned to run parallel along the force plate (figure 3.15). The force and moment output were amplified through a six channel amplifier (one for each platform) that incorporates a low-pass (1000 Hz) filter for each channel. The outputs from the amplifiers were passed through an analogue-to-digital converter (16 bit). The Vicon Workstation software was then used to process the digital output (sampled at 100 Hz) to determine centre of pressure (CP) data, within the laboratory coordinate system. The CP is defined as the point of location of the GRF vector, representing the weighted average of the total pressures in contact with the ground. During normal bipedal standing, the net CP lies between the two feet. The Vicon Workstation software also used a zero sample range, which incorporated an analogue zero base-level calibration procedure for each force platform that was performed before each data collection session. This was obtained by entering a range of frames during which the analogue input was zero for all force plate channels. During data collection, a push-button system located on each amplifier was used to auto-zero the platform (which used data from the zero base-level calibration) every time the step height was changed. This ensured the zero-voltage reference included the weight of the step.

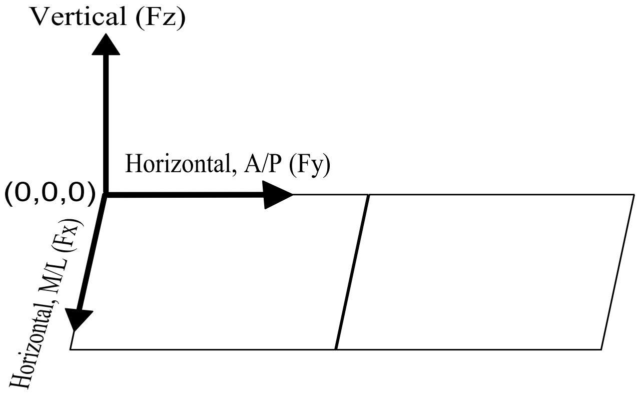


Figure 3.15. Origin of the laboratory coordinate system (0,0,0) located at the front left of force platform 1.

3.4 Centre of pressure (CP) calculations

Output signals of force and moment profiles were used to calculate the location of the centre of pressure (CP) in the x and y directions relative to the origin of the force platform at each time point. For example, the CP coordinate (x) was calculated as follows:

$$CP(x) = \frac{(-My - (z_{off} .Fx))}{Fz}$$

Where My = moment about the y axis, z_{off} = vertical offset from the top plate to the origin of the force platform, $F\text{x}$ and $F\text{z}$ = forces along the x and z axes. The y coordinate of the CP was calculated based upon the moment about the x axis and the forces acting in the y and z direction.

3.5 Modelling

A detailed critique of the various modelling techniques available and the rationale for using the following models are detailed in the chapter 2.

3.5.1 Modelling of centre of mass (CM)

Three dimensional total body CM was calculated using the anthropometric regression equation highlighted by Dempster (1955). This allowed whole body CM to be calculated as the weighed sum of all segment CM locations.

$$CM_j = \frac{\Sigma_j m_j \cdot p_{ij}}{\Sigma_j m_j}$$

Where m_j is the mass of the segment *j*, and P_{ij} is the *i*th component (*i* = x, y, z) of the position vector of its CM (Eames *et al.,* 1999). Monitoring whole-body CM displacement provided a global picture of body segment movement synergy. Whole-body CM trajectory data, along with trajectory data of certain other body markers were exported in ASCII format for further data analysis.

3.5.2 Landing limb effective vertical stiffness

Landing limb was modelled as a linear spring with its initial length defined as the height of the CM at the instant of ground contact, and with the maximum vertical ground reaction force of landing during the initial ground contact phase equating to the applied spring load (figure 3.16). Vertical stiffness (k_{vert}) was then calculated as the ratio of the maximum vertical ground reaction force (Fz_{max}) to the spring's displacement, i.e. the vertical displacement of the CM during the initial contact period (Buckley *et al.,* 2005).

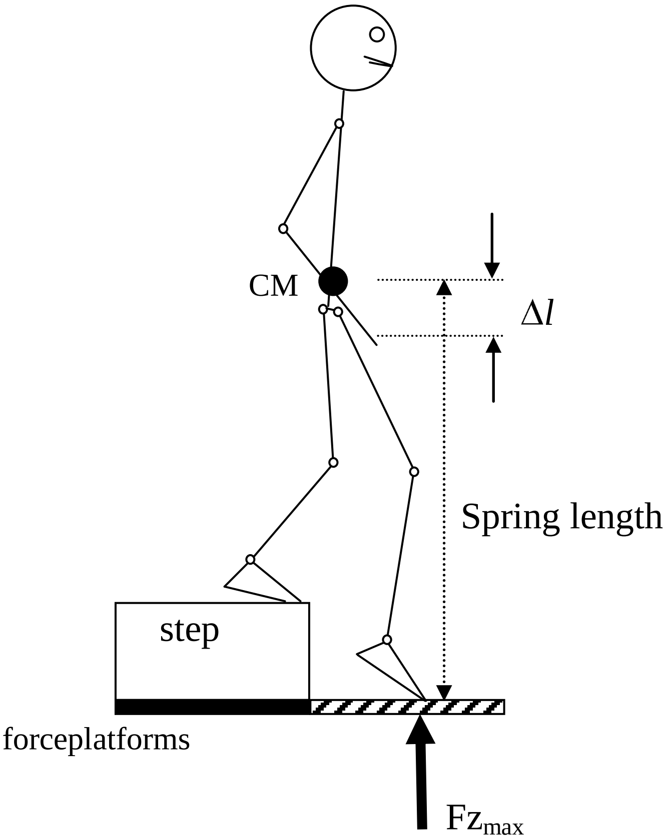


Figure 3.16. As illustrated in Buckley *et al.* (2005), vertical stiffness was computed as ratio of maximum vertical GRF to change in height of CM within initial contact period.

3.5.3 Kneedrop

Prelanding kinematic measures analysed included “kneedrop”, which is the distance dropped by the knee from its peak height to the point where the lower limb (calf) segment reaches its maximum forwards swing before its polarity is reversed (Cowie *et al.,* 2008). To measure kneedrop, the saggital plane angular displacement (‘swing’) relative to the lead-limb lower leg segment and the vertical was measured during the step descent. In all step descents, the same characteristic pattern was observed in which the leg swung outward reaching its peak and then swung back towards the body in preparation for landing. Swing peak was defined as the point where the lower limb (calf) segment reaches its maximum forward swing before its polarity is reversed. The distance that the knee had descended vertically from its maximum height to the instant of swing peak was defined as kneedrop (figure 3.17).

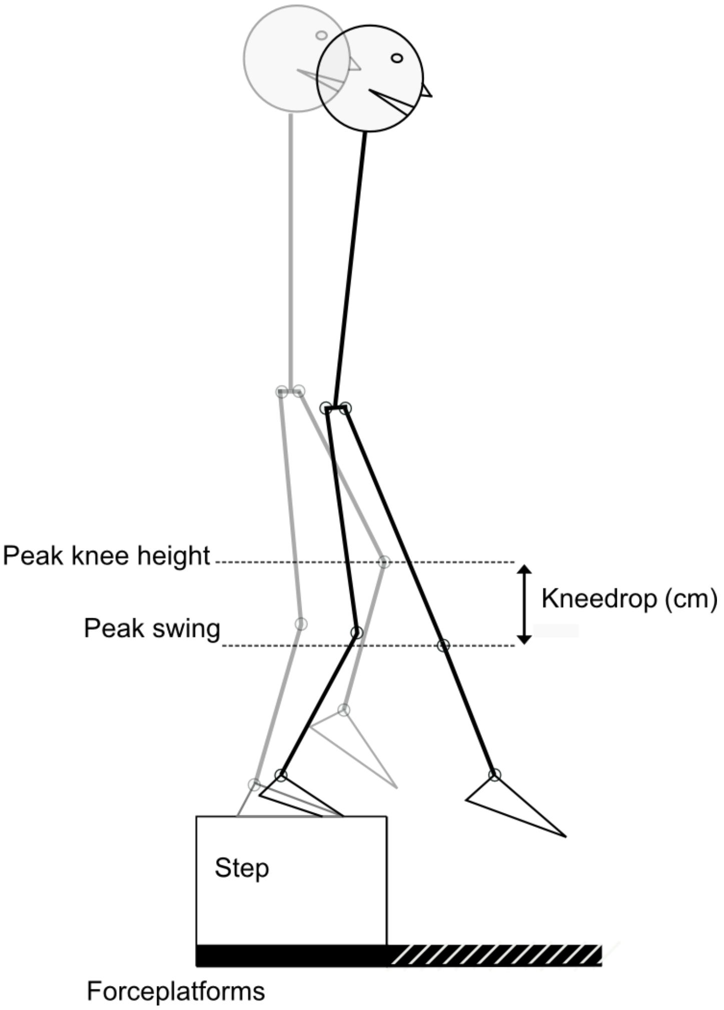


Figure 3.17. Calculation of kneedrop, determined by the vertical displacement of the knee from its peak height to the moment of peak swing. Adapted from Cowie *et al.* (2008).

3.6 General protocol

In all but the last experimental chapter (details of the protocol used in the last experimental chapter can be found in chapter 8), step downs were performed from a stationary standing position on top of a block that was placed over a force platform (AMTI OR6-7, Advanced Mechanical Technologies Inc., Boston, USA). Participants stepped down onto an adjacent force platform coming to an immediate rest on the lower level with feet side-by-side. Participants remained stationary for approximately 5 sec prior to performing the stepping movement. This allowed onset of movement to be clearly identified within the kinematic / kinetic data analysed, e.g. to distinguish CP movement involved in a step initiation from that which occurs as a consequence of normal body sway. Starting position on top of the block was feet comfortably apart, and toes as close to the front edge as possible (figure 3.18).

Figure 3.18. Starting position attained by each participant in chapters 4, 5 and 7 prior to stepping down onto the lower level.

To reduce learning from somatosensory and / or proprioceptive feedback received by stepping directly up on to a block, starting positions in chapters 4 and 5 were instead attained by participants being asked to walk up to the block, from approximately 3 m away, using a number of ‘stepping stones’ which were randomly varied in height from trial to trial. The stepping stones were covered with the same coloured vinyl material and had the same surface area as the blocks, figure 3.19a. Due to the practicalities associated with changing the stepping stones after every trial, participants in chapter 7 instead negotiated 4 blocks of different height (in a random order) that were placed adjacent to each other (figure 3.19b). The random order was achieved by first numbering each block from 1 to 4. Numbers 1 - 4 (representing each block height) were then put into separate rows within the same column in Microsoft excel and the '=Rand()' function assigned to each number in the adjacent column. The '=Rand()' function randomly produces a number between 0 and 1 (using 9 decimal places). Both columns were highlighted and assorted in ascending order based on the random values. The order of the numbers 1 - 4 was then recorded, which denoted the order to negotiate the blocks. This process was repeated for each trial and for each participant.

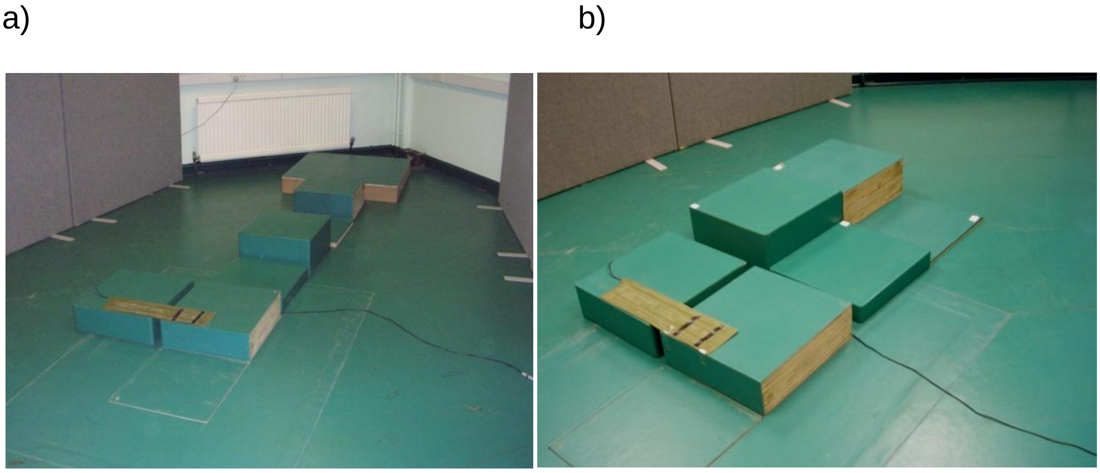


Figure 3.19. The 'stepping stones' participants negotiated a) in chapters 4 and 5, and b) in chapter 7.

In chapters 4, 5 and 7 participants were instructed, on attaining their starting position, to look at a visual stimulus consisting of a straight white line (1 cm thickness, 50 cm in length) placed on the floor 1.5 m directly in front and perpendicular to the participant. Participants stepped down leading with their preferred limb (determined during practice trials). In all but the first experiment (chapter 4), participants were free to choose where they looked whilst completing each step descent. In chapter 4 participants were asked to maintain fixation on the white line throughout the step descent as this ensured that the lvf remained occluded i.e. ensured participants did not flex their head to gain visual information from the foot or immediate lower floor area (specific detail can be found in chapter 4). Adequate rest periods were provided in all experiments to ensure the participant did not become fatigued.

It is recognised that step descents during normal everyday life can be completed using a heel or toe landing technique. For example, step descents of 10 cm or less are usually completed using a heel landing (van Dieën *et al.,* 2008). However, when step height increases, participants switch to a toe landing (Freedman and Kent, 1987). When participants are unable to see the height of the step descent, they adopt a toe landing, even when the descent is only 5 cm (van der Linden *et al.,* 2007). A toe landing is preferred compared to a heel landing when stepping from an increased or unknown height because a toe landing increases dynamic stability (Freedman and Kent, 1987). For example, less kinetic energy is gained during the descent when landing occurs on the toes because ground contact occurs earlier compared to landing on the heel (van Dieën *et al.,* 2008). This subsequently reduces the vertical impact forces generated at landing (van Dieën *et al.,* 2008). In the following experimental chapters, participants will complete step descents in conditions of reduced dynamic stability (i.e. visual occlusion). Furthermore, the minimum step height used is 14.6 cm (see section 3.5.1 for further details on step heights used). Therefore it was expected that step descents would be completed using a toe landing technique, and this was indeed found to be the case for the majority of trials. If participants used a heel landing technique in more than 10 % of the trials collected, this suggested that they adopted a different stepping strategy to others within the study and thus their entire data set was excluded from the subsequent data analysis. Of the 65 participants used in this thesis, 3 participants were excluded. None of the participants included in the data analysis in this thesis adopted a heel landing technique in any of the trials completed.

3.7 Data analysis

In all of the following experimental chapters (4 – 8), after processing the data via Pipeline (in Workstation), co-ordinate marker, ground reaction force, CM and CP data were exported in ASCII format and were analysed by visual basic macros. These macros allowed a number of key points from within the data, across hundreds of trials, to be identified and recorded relatively quickly. A further macro was used to collate the data into a format ready for subsequent statistical analysis. In all experimental chapters (4 – 8) the testing protocol was designed to ensure that a macro could easily be written to analyse the data. For example, in chapters 4 – 7, once participants had adopted their start position (further details see 3.6. General protocol), they were required to stand stationary on the upper level for ~5 seconds prior to descending the step. This ensured a base-line was recorded so that the macro could precisely determine movement initiation and lead limb toe-off, used to calculate double support and single support time respectively. An example of a macro used to analyse data collected in chapter 4 (experimental chapter 1) can be found in appendix 5. A section of this macro used to calculate peak vertical contact force (pvcf) is highlighted in table 3.3a. Table 3.3b illustrates the output from the macro produced in Microsoft Excel. An explanation of how the macro calculates pvcf follows. In the column adjacent to vertical force plate 2 data (F_ForcePlate2:Z, in table 3.3b), the macro finds the row where instant of landing occurs on the lower level. In table 3.3b the macro would find column C, row 3. The macro then finds the maximum value in F_ForcePlate2:Z data between start and end of the landing period (column B, between rows 3 and 9, table 3.3b). The maximum value is filled in the column adjacent to F_ForcePlate2:Z data between the start and end of the landing period. The macro then selects the row where instant of landing occurs on the lower level, 2 columns from F_ForcePlate2:Z data (column D, row 3, table 3.3b) and subtracts the value in the same row of the previous column from the value in the same row of F_ForcePlate2:Z data (i.e. column C, row 3 subtracted from column B, row 3 in table 3.3b). This equation is filled between start and end of the landing period 2 columns from F_ForcePlate2:Z data (column D, table 3.3b). The macro then finds the first ‘0’ within this column and moves along the same row, 2 columns to the left. The macro then records the value and location in this column, which is subsequently called ‘pvcf’. Of note, peak vertical contact force is calculated in this way to allow ‘time to pvcf’ to be easily calculated. The macro also calculates other key points including movement initiation and lead limb toe-off (see figure 3.20).

Table 3.3. a) sample of a visual basic macro written to determine peak vertical contact force with b) part of the output produced in Microsoft Excel.

a)

peak vertical contact force	=SELECT("R"&start_FP2&"c8")
	=FORMULA("=max(r"&start_FP2&"C7:r"&End_FP1&"c7)")
	=FILL.AUTO("RC:R"&End_FP1&"C")
	=SELECT("R"&start_FP2&"c9")
	=FORMULA("=(rc[-2]-rc[-1])")
	=FILL.AUTO("RC:R"&End_FP1&"C")
	=FORMULA.FIND("0",2,1,1,1,FALSE)
	=SELECT("rc[-2]")
pvcf	**=ROW(ACTIVE.CELL())**
	=SELECT("Rc[-1]")
	=FORMULA("pvcf")
	=FONT.PROPERTIES(,,,,,,,,,3)

b)

	A	B	C	D
1		F_ForcePlate2:Z		
2		0		
3		157.22	1349.4	-1192.18
4		389.52	1349.4	-959.88
5		624.6	1349.4	-724.8
6		768.96	1349.4	-580.44
7		1124.2	1349.4	-225.2
8	pvcf	1349.4	1349.4	0
9		1106.4	1349.4	-243

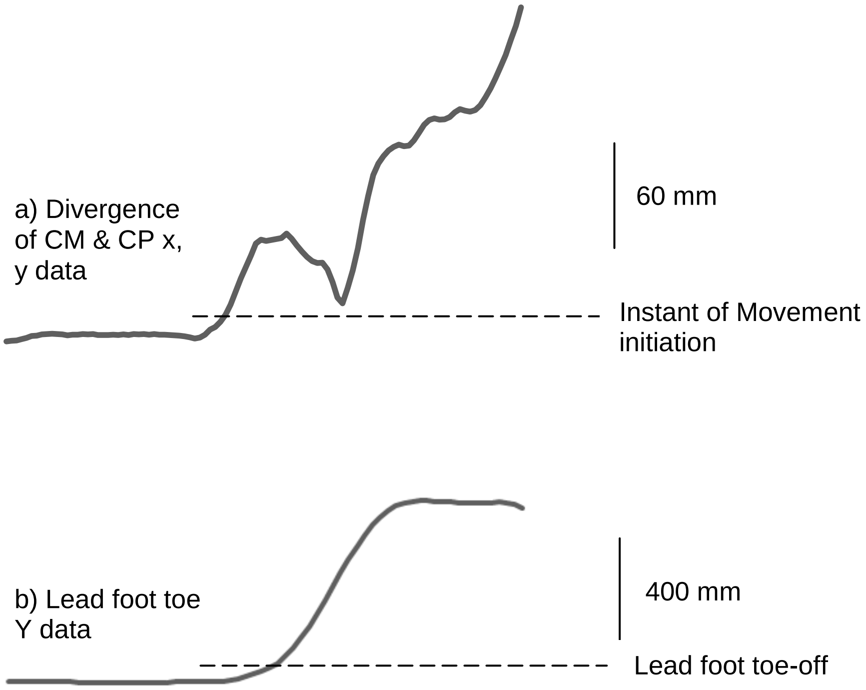


Figure 3.20. Data plotted to illustrate a) divergence of CM and CP data used to determine instant of movement initiation and b) lead foot toe-off, used to determine start of single limb support time.

3.8 Data inspection - dealing with extreme data points

The most common measure used to describe the typical score within any data set is the mean (Dancey and Reidy, 2004). The standard deviation (SD) is also often provided along with the mean as this provides an indication of the range of values within the data set, highlighting how much the scores vary around the mean (Clarke and Cooke, 1998). It is important to determine if the maximum and / or minimum values within the data set are considered anomalies or outliers, as including these values in the statistical analysis may affect the output (i.e. cause false conclusions to be drawn from the data). For example, a single outlier is capable of considerably changing how the data are interpreted. Consider figure 3.21 below. The inclusion of the suspected outlier (value 8,1) in figure 3.21a has a significant effect upon the best fit line. This is highlighted by the alteration of the best fit line when the outlier is removed (figure 3.21b).

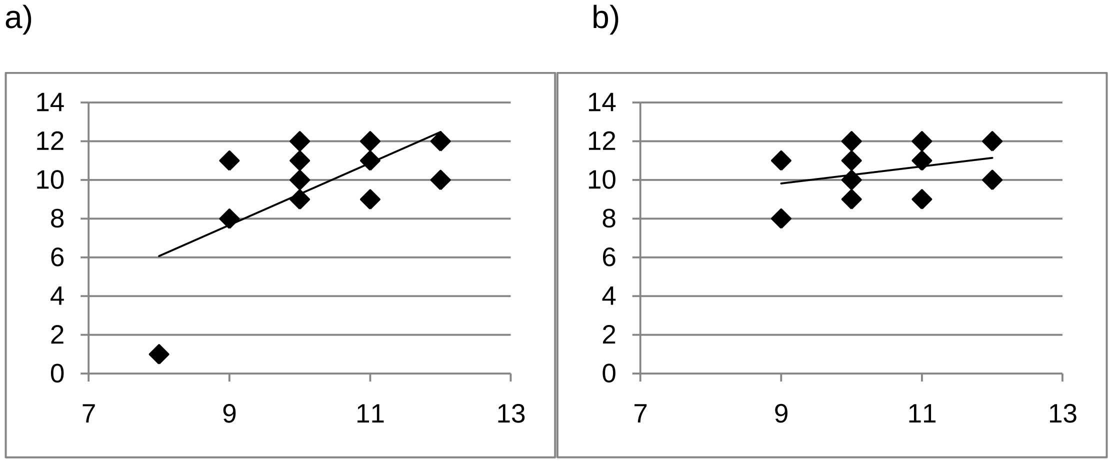


Figure 3.21. The interpretation of data can be significantly affected by a) the inclusion or b) removal of an outlying data point.

Due to the difficulties defining an outlier in multidimensional data, there is no general rule to follow (Ryan, 2007). However, there are a number of processes that can be undertaken to help determine the likelihood of correctly identifying a true outlier within the data set. A relatively simple approach of identifying a potential outlier is to graph the entire data set. Through graphing the data, this will highlight individual values that do not appear to fit into the trends of the data set (Clarke and Cooke, 1998). An alternative approach is to calculate how many standard deviations a particular value lies above or below the mean. This is called calculating the standard normal distribution (SND). To calculate the SND, scores in the sample need to be transformed to standard normal scores. This is achieved by subtracting the mean from the relevant data point and dividing by the SD. The result is termed the *Z* score and is expressed in SD units:

$$Z = \frac{value - mean}{SD}$$

Plotting the *Z* scores of an entire set of normally distributed data should profile a Gaussian distribution or, as it is more commonly known, a ‘bell curve’. The standard normal distribution is one with a mean of 0 and a standard deviation of 1 (figure 3.22). The area under the curve represents probability: 68.26 % of data will lie within 1 standard deviation of the mean, 95.44 % within 2 standard deviations, and 99.14 % within 3 standard deviations. Alternatively, there is less than a 5 % chance that a chosen data point will lie outside 2 standard deviations of the mean, and less than 1 % chance that it will lie outside 3 standard deviations.

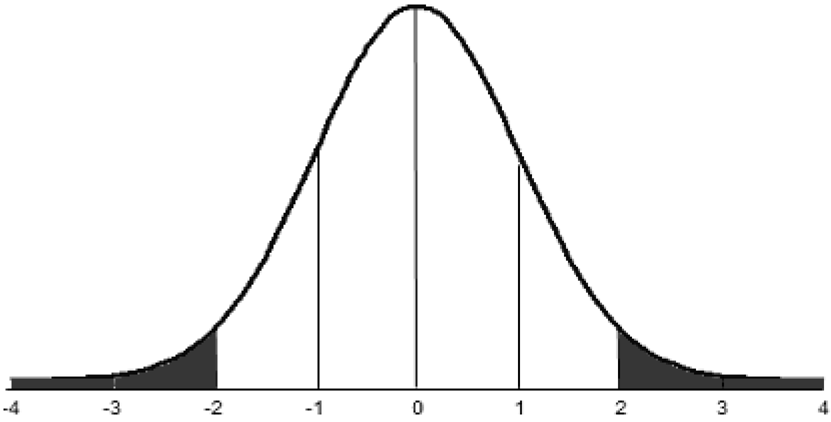


Figure 3.22. The standard normal distribution curve.

Once the *Z* score of each data point is calculated, any value that lies more than two and a half standard deviations (in either positive or negative direction) from the mean can be considered an outlier and removed from the data set (Frank and Althoen, 1994; Clarke and Cooke, 1998; Griffiths *et al.,* 1998). This approach is most appropriate when the distribution of the data is unimodal and symmetrical (Frank and Althoen, 1994). If the data are skewed, identifying outliers that are 1.5 times greater than the mid-spread may be more appropriate (Frank and Althoen, 1994; Clarke and Cooke, 1998).

If the data set does not produce a bell shape curve, this suggests that the data are skewed and are not drawn from a normally distributed population (Hinton *et al.,* 2004). If the majority of scores fall to the right of the mean, data are considered to be negatively skewed (figure 3.23a). However, if the data falls to the left of the mean, the data are considered to be positively skewed (figure 3.23b). Using statistical analysis software it is possible to obtain a measure of skewness. A value of skewness around + 1 or - 1 are extreme deviations from normality and usually indicate that the data are not from a normally distributed population (Dancey and Reidy, 2004).

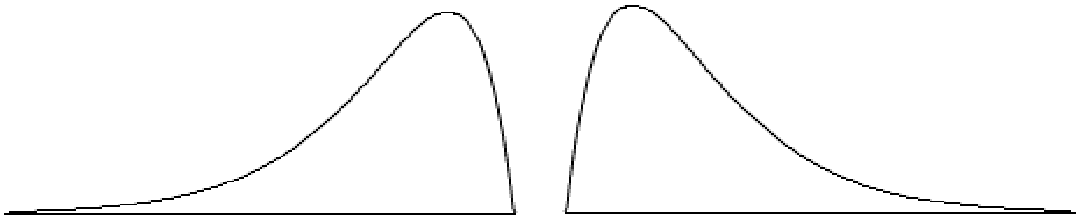


Figure 3.23. Examples of (a) negatively skewed and (b) positively skewed data.

An alternative approach to calculating *Z* scores is to produce a box-plot of the data (Kinnear and Gray, 2000). A box-plot graphically displays five important pieces of information about the data (figure 3.24). The horizontal black line running through the box itself represents the median or middle value of the data when arranged in order of magnitude. The advantage of using the median value compared to the mean is that the median is less affected by extreme data points. The median divides the data into two halves, so half the data lie above the median and the other half below. The box itself represents the spread of the middle half of the data, containing a quarter of the values in the upper quartile and a quarter of the values in the lower quartile; the portion of the distribution falling between the 25th and 75th percentiles. The lines extending from the shaded box are commonly referred to as 'whiskers' and connect the highest and lowest scores that are not considered to be outliers (Hinton *et al.,* 2004). Outlying data points are considered in excess of 1.5 box lengths from the 25th or 75th percentiles (Griffiths *et al.,* 1998; Kinnear and Gray, 2000; Dancey and Reidy, 2004; Hinton *et al.,* 2004). These are highlighted by individual data points in figure 3.24.

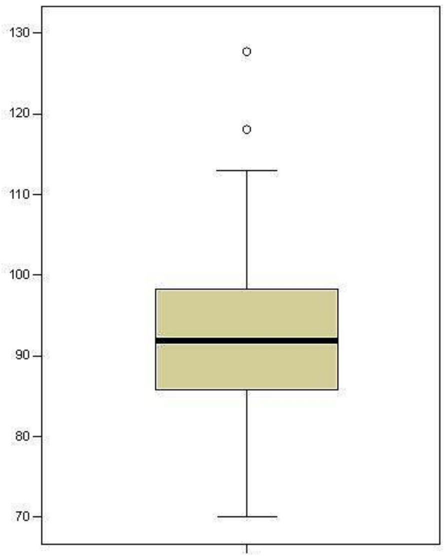


Figure 3.24. Box-plot highlighting the median (horizontal black line), upper and lower quartile (brown shaded area), the whiskers and two outlying data points.

One limitation of using the box-plot is that it can be difficult to determine when data are not from a normally distributed population. For example, both box plots below (figure 3.25a) appear to be from normally distributed data, however the histogram output for these data sets are very different (figure 3.25b).

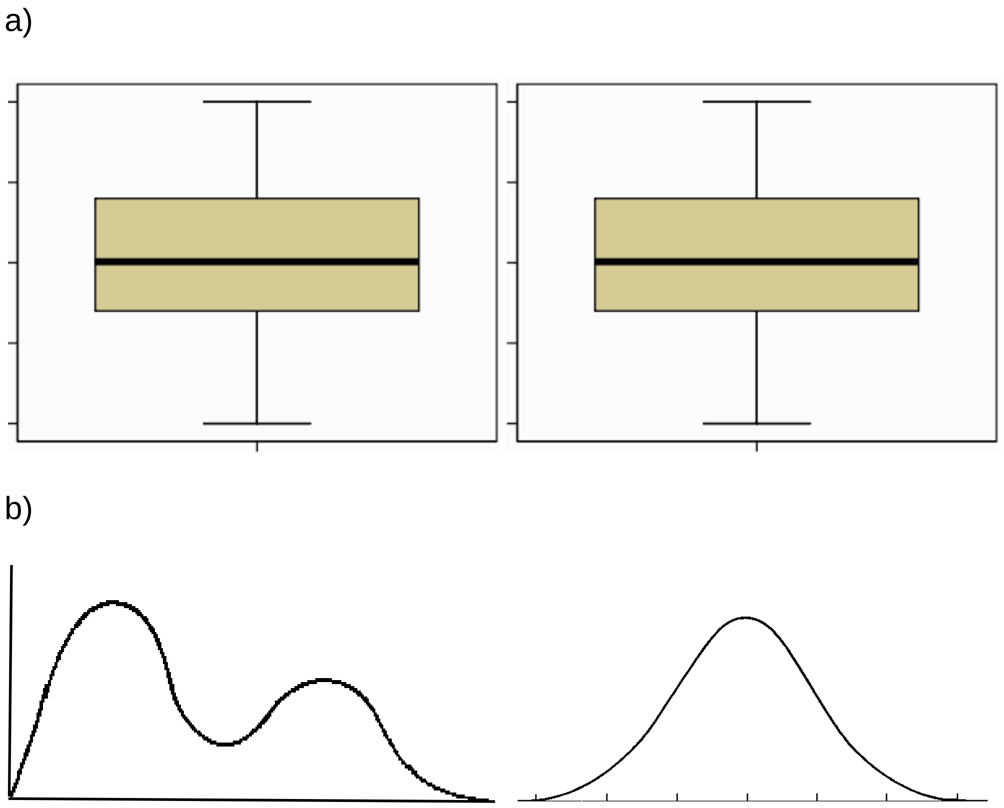


Figure 3.25. a) Producing a box-plot of the data may lead to false conclusions being drawn from the data set, as evidenced by b) plotting the data using a histogram.

Despite bimodal distribution (figure 3.25b) being rare (Hinton *et al.,* 2004), it is clear that each data set needs to be analysed using a variety of approaches to ensure any outlying data points are not included in the subsequent statistical analysis. In this thesis, graphically interpreting the data set and *Z* scores were used when considering potential outlying data points. Indeed, any data point(s) that did not initially appear to fit into the trends of the data set when graphed, or *Z* scores higher than 3 standard deviations were retained for further analysis. Once a potential outlying data point had been identified the following aspects were considered:

- The raw data of that particular trial was checked and the variable calculated by hand (if needed) to determine if this was the correct value.
- Did something different happen during data collection in that particular trial? i.e. if the experimenter failed to zero the force plate after placing the step on it, the value obtained for body weight supported on the trail leg would be much larger than normal.
- Is this value supported by other variables? i.e. if peak vertical ground reaction force generated during initial landing on the lower level was increased, a reduction in body weight supported on the trail leg during initial landing would support the reason for this value.
- Is there a logical / theoretical reason for this value?

If the potential outlying data point could not be explained after considering the above points, the particular value was excluded from the analysis. From the 3,121 trials collected in this thesis, only 5 trials (< 1 %) were considered outliers and were subsequently excluded. The trials excluded did not differ between experimental conditions (i.e. visual field, timing of visual occlusion and added mass conditions). Of note, ~30 additional trials in the thesis were excluded from analysis due to key markers falling off during the trial.

Chapter 4

Experiment 1

Evidence of a specialised role of the lower visual field in regulating step descent landing control

The work presented in this chapter has been published as:

Timmis MA, Bennett SJ, and Buckley JG, (2009). Visuomotor control of step descent: evidence of specialised role of the lower visual field, *Experimental Brain Research,* **195(2),** 219-227.

This work was also disseminated as a poster presentation at:

- The International Society for Posture and Gait Research, Bologna, Italy (2009).
- The Symposium on Gait, Posture and Balance: Function, Dysfunction and Rehabilitation, University of Birmingham, UK (2008).

4.1 Introduction

Research investigating how vision is used to control step descent has shown that if vision is blurred or occluded prior to movement initiation, participants land with reduced force and lower extremity stiffness (Buckley *et al.,* 2005, 2008). This is indicative of a 'softer' landing, which is a consequence of participants adopting a cautious stepping strategy of ‘sitting back’ on the trailing limb, using the lead limb to ‘probe’ for the ground and not fully committing to transferring weight to the lead limb until somatosensory feedback confirms they have made contact with the lower level (Buckley *et al.,* 2008). Movement control of this type is suggested to use continuous comparison of expected feedback and real-time sensory feedback in order to obtain an error signal to adjust the motor output (Miall and Wolpert, 1996). ‘Probing’ for the ground with the lead-limb when stepping down under diminished vision conditions, may therefore be a strategy to acquire load feedback earlier than was initially estimated using the impoverished visual feedback. These findings support early work showing that if step descents are performed with vision occluded and there is no prior knowledge of step height, anticipatory lower limb muscle activity preceding ground contact is reduced or even absent, which results in ‘softer’ landings and an increased reliance on the stance limb to control descent (Craik *et al.,* 1982). A similar use of vision has been reported in tasks that involve negotiation of floor-based obstacles. For example, when online vision is not available, participants’ lead and trail foot placement prior to and during obstacle crossing are farther from the obstacle, demonstrating a lack of ‘fine tuning’ of limb trajectory during the movement (Patla, 1997, 1998; Patla *et al.,* 2002; Mohagheghi *et al.,* 2004).

When moving around within the environment, step downs are often performed successfully and safely in the absence of visual feedback of the feet or lower-limbs, and / or of the area on the ground where we intend to land. For example, when we step down from a kerb onto the road, we typically tend to look sideways to see if there is any oncoming traffic (Geruschat *et al.,* 2003, 2006). In other situations, we are able to successfully descend a step whilst carrying relatively large objects such as a laundry basket, which occlude vision of the lower limbs and feet. This implies that in such situations visual information regarding the surface height change that is gained during the approach to the step / kerb edge is used by feedforward processes to regulate landing control, and that any contribution from online visual information comes from areas of the uvf, such as those relating head position to the environment. The question remains, however, when information regarding the surface height change is available in the lvf during the step down, which is normally the case because gaze tends to be directed one or two walking steps ahead during adaptive locomotion (Patla and Vickers, 1997), does it provide any advantage in terms of regulating landing control? In other words, does information available in the lvf contribute to control of daily locomotor tasks such as stepping down from a kerb to cross a road?

To date, the only study to manipulate the availability of lvf information during step descent showed that when it was occluded there was significantly less pre-contact muscle activity present (Craik *et al.,* 1982). However, as there was no concurrent change in landing control (i.e. there was no significant change in rate of ankle motion or rate of force application following landing) it is not clear why this change in muscle activation occurred and what it achieved. Therefore, the present study was designed to further examine the visuomotor control of step descent following manipulation of online visual information from lvf. Furthermore, by manipulating the availability of visual information from either full or upper visual field (lvf already occluded) for specific periods relative to step initiation, the aim of the present study was to determine when during step descent visual information from lvf is customarily used to ‘fine tune’ landing.

4.2 Method

4.2.1 Participants

10 healthy adults (4 male, 6 female), age 24.4 ± 9.4[2] years (mean ± SD), height 175 ± 11.0 cm and mass 68.86 ± 15.3 kg, were recruited using the same inclusion / exclusion criteria described in the general methods (see sub section 3.1). The tenets of the Declaration of Helsinki were observed and the experiment gained approval from the local Research Bioethics Committee. Written informed consent was obtained from each participant prior to undertaking the study.

4.2.2 Visual assessment

Binocular visual acuity (VA), binocular contrast sensitivity (CS) and stereoacuity were each assessed using the methods described previously (see sub section 3.2). All participants recorded values within the limits of healthy eyes (Vale *et al.,* 2008a), with measures of VA, CS, and stereoacuity of -0.1 ± 0.1 logMAR, 1.9 ± 0.1 log units, and 46.5 ± 30.4 secs of arc respectively. Although we didn’t

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[2] The high standard deviation in participant age was due to one participant being considerably older than the others within the group. To see if this particular individual behaved in the same manner as the others, statistical analyses (see below) were run with and without inclusion of the data for this individual. There was no difference in the significant effects when this individual’s data was removed from the analysis. Thus the data presented are for all 10 participants.

objectively assess visual field, all participants self-reported they had no visual field restrictions.

4.2.3 Protocol

Participants undertook repeated step downs from three block heights: 14.6 cm, 21.8 cm and 15 % (± 1 cm) of participant's height (15%bodyHt). Further detail pertaining to the blocks can be found in the methods section (sub section 3.3.1). Prior to attaining the start position on the final block edge, participants were required to negotiate a series of 'stepping stones'. Further detail regarding the 'stepping stones' is described in the general methods (see sub section 3.6). Step downs were performed adhering to the protocol highlighted in sub section 3.6. Illumination over the stepping area was 400 lux (taken at participant's chest height) and the luminance of the floor and top surface of the block was 15 and 30 cd/m^2 respectively.

Throughout the experiment, participants wore PLATO goggles (Translucent Technologies, Canada). Details of the goggles can be found in the general methods (sub section 3.3.2). Participants completed the step downs with and without visual feedback available online, and with and without visual field being restricted. Visual field restriction involved obstructing lvf by placing black card across the lower half of the spectacles with its upper edge in line with the middle of the pupil. When the goggles remained translucent during the trial, visual feedback was available throughout. When the goggles switched to opaque, from lead limb toe-off or from approximately 50 % of lead-limb swing time onwards (mid-swing) using the method previously described in sub section 3.3.3, visual feedback was unavailable. Step height (14.6 cm, 21.8 cm and 15%bodyHt) changed randomly every 9 trials. Within the 9 trials, each vision condition (available throughout, available up to toe-off, available up to mid-swing) was completed three times in random order. This procedure was completed in both visual field (full, restricted) conditions (random order), for a total of 54 trials.

To determine if participants altered their stepping strategy based upon the probability of visual occlusion, step descents were also completed with vision unperturbed and the prior instruction that vision would not be occluded during the descent (zero probability block). Step height changed randomly every three trials, with three trials collected at each height, for a total of 9 trials. The zero probability block was only completed in full field vision condition as this reflects participant's habitual stepping response. The zero probability block was randomly inserted within the study. Participants were informed once trials in the zero probability block had been collected and they were re-entering the main part of the study i.e. when there was a possibility (67 % chance) of vision being occluded during the descent. NB. Participants were not informed of the percentage of visual occlusion during the main part of the study. Comparison between full vision trials, completed when there was a high probability (67 % chance) of visual occlusion and zero probability of visual occlusion during step descent, are reported in chapter 7 (Does the probability, awareness or experience of visual occlusion during step descent affect the role of online vision in regulating landing control?). A total of 63 trials were collected in chapter 4.

An 8 camera 3-D motion analysis system (Vicon MX3, Oxford Metrics Ltd) was used to record (at 100 Hz) segmental kinematics as participants stepped down from one level to another. Data were collected during a single testing session for each participant, with adequate rest periods provided to prevent fatigue. Participants wore shorts, t-shirt and flat soled shoes. Reflective spherical markers where attached and anthropometric measurements taken are described in the methods section (3.3.5). Joint angles were defined as the resultant angle between two adjacent segments. Knee, ankle and head angular displacement data, ground reaction force data (including CP co-ordinates) from each force platform, and the co-ordinate data for whole-body CM, knee, ankle and all foot markers were exported (at 100 Hz) for further analysis.

4.2.4 Data analysis

Visuomotor control of step descent was evaluated by determining prelanding kinematic measures and landing mechanic variables for the initial contact period. Prelanding kinematic measures included kneedrop and time of kneedrop as a percentage swing time (see methods section for further details of kneedrop parameter, 3.5.3). The following measures were also determined for the instants of landing: lead limb knee (θ_{knee}) and ankle (θ_{ank}) joint angular displacement, anterior-posterior (a/p) stepping distance, a/p position of CM within base of support (CM-positioning), and a/p and (downward) vertical CM velocity. θ_{knee} and θ_{ank} joint angular displacement were determined as the change in joint angle at the instant of landing relative to their average position when standing stationary on the upper step. Stepping distance was measured as the a/p distance between the marker of the second metatarsal head on the trail and lead limb at the instance of lead limb contact on the lower level. CM-positioning was measured as a percentage of a/p stepping distance from the trail limb. Instant of landing was defined as the instant when the vertical ground reaction force (GRF) on the lead limb first became greater than 20 N.

To determine if participants flexed their head during the descent in an attempt to receive visual information from the foot and / or lower surrounding floor area when the lvf was occluded, a retrospective analysis of head pitch angles was conducted. If head flexion was found to exceed 10 degrees (relative to head angle determined for quiet standing), the trial was excluded from the analysis (this only occurred for 2 trials across the 630 trials completed by all participants).

The mechanics of initial landing, defined as the period up to lead-limb maximum knee flexion, were evaluated by determining the peak vertical ground reaction force (Fz_{peak}), peak angular velocity at the knee (ω_{knee}) and ankle ($_{ank}\omega$) joint, lower extremity stiffness (l.x.s), and amount of bodyweight still being supported by the trailing limb (bodyWt sup landing, Buckley *et al.,* 2008). Time from movement initiation (MI) to lead limb toe-off (double support, DS), lead limb toe-off to foot contact (single support, SS) and foot contact to contra-lateral limb toe-off (weight transfer, WT) were also evaluated. Peak ω_{knee} and $_{ank}\omega$ determined how the lead limb was loaded during the landing phase. MI was defined from when the resultant x, y coordinates of the CM and CP first diverged by greater than 10 mm for 5 consecutive frames. Lead limb toe-off was defined from when the a/p displacement of the second metatarsal head marker was greater than 3 mm for 5 consecutive frames. Trail limb toe-off was defined from when the vertical GRF on the upper block first dropped below 20 N.

4.2.5 Statistical analysis

Each calculated dependant variable was analysed using separate x 2 Visual Field (full, restricted) x 3 Step Height (low, high, 15%bodyHt) x 3 Vision Condition (no visual occlusion, vision occluded from toe-off, vision occluded from mid-swing) ANOVA, with repeated measures on each factor. Level of significance was accepted at $p < 0.05$, and post-hoc analyses were performed using Tukey's honest significant difference (HSD) test.

4.3 Results

4.3.1 Prelanding kinematics

At the instant of landing, θ_{knee} and θ_{ank}, a/p and vertical CM velocity, a/p stepping distance and CM-positioning, were unaffected by visual field or vision condition. Except for vertical CM velocity and a/p stepping distance, all of these dependent variables were significantly affected by step height ($p < 0.04$); there were no significant interactions between factors (table 4.1). Each variable increased with increasing step height.

Table 4.1. Prelanding kinematics: Group mean (± 1 SD) ankle and knee angle (θ), a/p and vertical CM velocity, a/p stepping distance and CM-positioning, at instant of landing across visual field and vision condition.

	full visual field			**restricted visual field**			**sig fact**
	no-disr	**toe-off**	**mid-swing**	**no-disr**	**toe-off**	**mid-swing**	
θ_{ank} (deg)	-29.4 (6.5)	-28.6 (5.9)	-29.0 (5.8)	-28.5 (6.9)	-28.7 (6.7)	-29.0 (6.4)	H
θ_{knee} (deg)	8.9 (5.5)	8.9 (5.7)	8.7 (5.3)	9.3 (5.9)	8.9 (4.8)	9.0 (5.1)	H
a/p CM vel (cm/s)	-45.2 (5.5)	-46.2 (5.2)	-44.7 (5.5)	-42.7 (6.9)	-43.6 (6.1)	-42.8 (5.3)	H*
Vert CM vel (cm/s)	-49.3 (10.1)	-49.7 (9.8)	-48.9 (8.9)	-46.1 (8.7)	-46.2 (7.7)	-45.6 (8.4)	n/a
a/p stepping dist (cm)	40.2 (3.6)	40.7 (4.0)	40.0 (3.8)	39.8 (3.5)	40.1 (3.2)	40.1 (3.4)	n/a
CM-position (% step dist)	47.2 (5.3)	49.0 (5.8)	47.3 (5.3)	47.5 (7.2)	47.9 (6.2)	48.0 (6.8)	H*

Data are averaged across step heights. There were no significant differences between visual field or vision condition. There were no interactions between factors. Factors found to be significantly affected by step height (H) are shown by capital letter (*p* < 0.05) and asterisks (*p* < 0.001). Ankle angle (-ve) indicates the amount of plantar-flexion and knee angle indicates amount of knee flexion relative to standing.

Time of kneedrop was significantly affected by visual field (*p* < 0.001), vision condition (*p* < 0.003), and step height (*p* < 0.001), but there were no significant interactions between factors. Kneedrop occurred earlier when visual field was restricted, when vision was removed at mid-swing onwards compared to when it was available throughout (*p* < 0.04) or removed at toe-off (*p* < 0.002, table 4.1), and occurred earlier with increasing step height (figure 4.1).

Kneedrop distance was significantly affected by visual field (*p* < 0.007), vision condition (*p* < 0.04), and step height (*p* < 0.001), but there were also significant visual field-by-step height (*p* < 0.003) and vision condition-by-step height (*p* < 0.004) interactions. Post-hoc testing showed that the increase in kneedrop distance with increasing step height was less evident when visual field was restricted, and that the reduction in kneedrop distance when vision was removed at mid-swing onwards was greater when stepping from 15%bodyHt compared to when stepping from low and high step heights (figure 4.1).

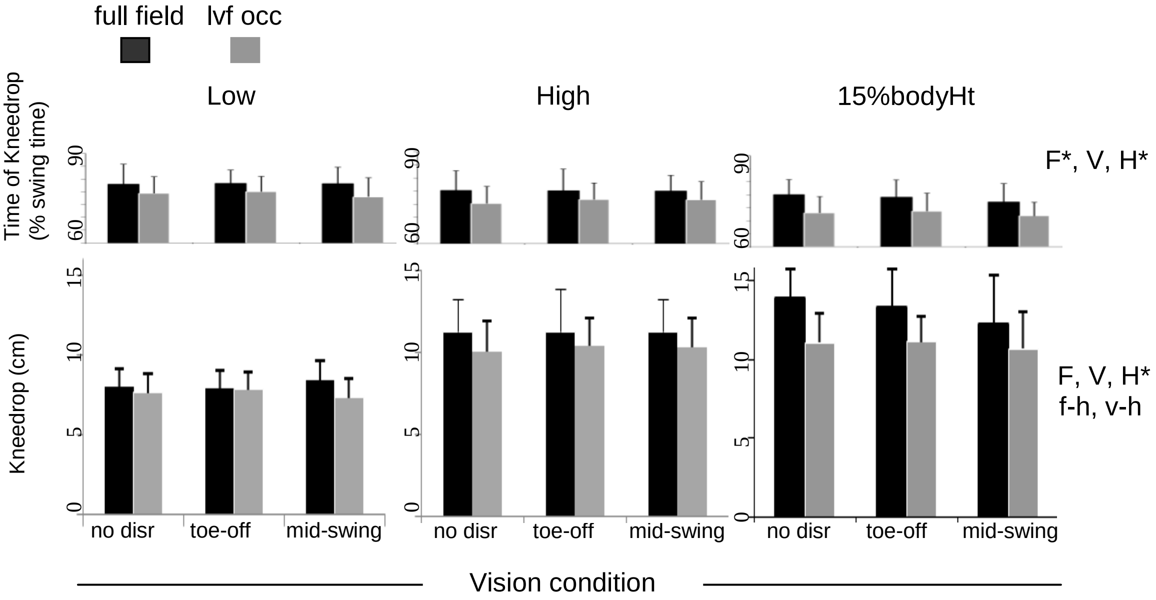

Figure 4.1. Group mean kneedrop time and distance for the 2 field, 3 vision and 3 step height conditions. Factors found to be significant are shown by letter (p < 0.05) and asterisk (p <0.001), for step height (H), field (F) and vision condition (V). Interactions between factors are shown by lower case letters.

4.3.2 Landing mechanics

ω_{knee} and $_{ank}\omega$, and Fz_{peak} during initial landing were unaffected by vision condition, but were significantly affected by visual field ($p < 0.03$) and step height ($p < 0.001$). The magnitude of each variable decreased when lvf was occluded and increased with increasing step height. l.x.s was unaffected by visual field or vision condition, but was significantly reduced with increasing step height ($p < 0.001$).

BodyWt sup landing was unaffected by visual field or vision condition ($p > 0.05$), was affected by step height ($p < 0.006$), and there was also a significant visual field-by-step height interaction ($p < 0.02$). The reduction in body weight supported on the trail leg with increasing step height was more pronounced when visual field was restricted. Consequently, there was significantly more bodyweight on the trail leg at the end of the initial landing period when stepping from the low step height under restricted field conditions (table 4.2).

Table 4.2. Landing mechanics: Group mean (± 1 SD) lower extremity stiffness (l.x.s), ankle and knee angular velocity (ω), peak vertical force (Fz_{peak}) during initial landing and body weight supported on trail leg (bodyWt sup) at end of initial landing, across visual field and vision condition.

	full visual field			**restricted visual field**			**sig fact**
	no-disr	**toe-off**	**mid-swing**	**no-disr**	**toe-off**	**mid-swing**	
l.x.s	22.9	23.7	23.0	24.3	24.7	24.3	H*
($kN.m^2$)	(13.2)	(14.7)	(13.1)	(12.5)	(12.1)	(13.3)	
$_{ank}\omega$	224	219	217	212	208	212	F, H*
($deg. s^{-1}$)	(52)	(54)	(48)	(62)	(50)	(60)	
ω_{knee}	103	107	105	90	93	91	F, H*
($deg. s^{-1}$)	(34)	(37)	(30)	(31)	(27)	(30)	
Fz_{peak}	916	930	918	899	908	884	F, H*
(N)	(195)	(186)	(188)	(210)	(208)	(211)	
bodyWt	43	46	44	43	53	42	H, f/h
sup (N)	(44)	(51)	(54)	(47)	(66)	(53)	

Data are averaged across step heights to illustrate the effects of visual field. There were no effects of vision condition. Factors found to be significant are shown by capital letter ($p < 0.05$) and asterisks ($p < 0.001$), for visual field (F) and step height (H). Interactions between factors are denoted by lower case letter.

4.3.3 Temporal parameters

DS time was unaffected by vision condition, step height or visual field. SS time was significantly affected by step height ($p < 0.001$) and visual field ($p < 0.05$), but was unaffected by vision condition; there were no significant interactions between factors. SS time increased when visual field was restricted and increased with increasing step height. WT time was unaffected by vision condition, but was decreased with increasing step height (*p* < 0.001), and there was also a significant visual field-by-step height interaction (*p* <0.05), which indicated the reduction in WT time with increasing step height was more pronounced when visual field was restricted (table 4.3).

Table 4.3. Temporal parameters: Group mean (± 1 SD) double support (DS), single support (SS), and weight transfer (WT) times across visual field and vision condition.

	full visual field			**restricted visual field**			**sig fact**
	no-disr	**toe-off**	**mid-swing**	**no-disr**	**toe-off**	**mid-swing**	
DS (s)	0.45 (0.09)	0.45 (0.08)	0.45 (0.08)	0.47 (0.06)	0.46 (0.06)	0.46 (0.06)	n/a
SS (s)	0.60 (0.09)	0.61 (0.11)	0.61 (0.10)	0.64 (0.08)	0.64 (0.08)	0.64 (0.09)	F, H*
WT (s)	0.20 (0.05)	0.20 (0.06)	0.20 (0.05)	0.20 (0.05)	0.19 (0.05)	0.20 (0.06)	H*, f/h

Data are averaged across step heights to illustrate the effects of visual field. There were no effects of vision condition. Factors found to be significant are shown by capital letter (*p* < 0.05) and asterisks (*p* < 0.001), for visual field (F), and step height (H). Interactions between factors are denoted by lower case letter.

4.4 Discussion

Step downs can occur during on-going gait or from a stationary position. For example, prior to stepping down from a kerb to cross the road, individuals may have to pause momentarily whilst waiting for an opportunity to cross safely (for a discussion of the differences between these visual sampling situations [static and dynamic] see sub section 2.2.2 Optic flow and egocentric-direction strategy). In both the static and dynamic situations, step downs are performed successfully and safely, often in the absence of visual feedback of the feet or lower-limbs, and / or of the area on the ground where we intend to land. Therefore, the present study investigated the contribution of information from lvf to the planning / control of step descent from a stationary standing position, and when during step descent these visual cues are typically used. Findings indicate significant differences in landing control (mainly limited to the mechanics of the initial landing phase, with only subtle changes in prelanding kinematics) for step descents performed in conditions permitting access to full visual field compared to uvf alone, which highlights the importance of lvf information to the control of step descent. The removal of full or upper visual field (lvf already occluded) from beginning of swing or mid-swing onwards caused limited effects that were restricted to measures relating to kneedrop (reflecting subtle changes in lead-limb prelanding kinematics), which indicates that visual information is mainly used for movement planning. There were also expected effects of step height.

4.4.1 Importance of lvf information

When lvf was occluded, SS time increased, kneedrop decreased and occurred earlier in the descent, and Fz_{peak} and ω_{knee} and ω_{ank} during initial landing were reduced. These differences suggest that participants were unable to effectively use visual cues from areas of the uvf, such as those relating head position to the environment, to plan / control landing in the same manner as occurred under full field vision. However, these changes in landing control were made without fundamentally altering stepping strategy. For instance, there was no change in l.x.s, body weight supported on the trail leg, or CM-positioning when lvf was restricted. The implication, therefore, is that participants were able to use visual cues from the uvf to effectively plan the general stepping pattern, but that in the absence of visual cues from lvf, stepping strategy was modulated in a manner that is consistent with participants being uncertain regarding precise location of the foot / lower leg relative to the lower floor level.

It is relevant to note, however, that alterations to stepping strategy when vision from the lvf is occluded has been observed in other forms of locomotion. For example, negotiating irregular terrains when lvf information was unavailable caused participants to reduce gait speed and step length; allowing increased control of foot placement (Marigold and Patla, 2007). In addition, stepping over an obstacle when lvf information was unavailable resulted in participants increasing the horizontal distance between the foot and the obstacle at foot placement (presumably by shortening step length) in both the lead and trail limbs (Rietdyk and Rhea, 2006). In the present study, the adaptations in landing control found without any obvious alterations in stepping strategy when lvf was restricted, suggest that information from lvf is particularly important in determining the precise instant of landing when stepping down. The changes found in certain landing mechanic variables highlight that without information from lvf, landing control was altered to bring about a ‘softer’ and more ‘cautious’ landing. It is likely that such alterations reflect a drive towards increased safety, as landing unexpectedly on an incorrectly prepared limb might otherwise result in a relatively large shock (reaction) force being generated at instant of contact. Most of us, at one time or other, will have experienced such a landing when stepping down from a kerb that we had not anticipated. The shock force generated travels up the leg to the base of the spine and is experienced as an uncomfortable ‘jolt’ to the lower back.

Given that participants were able to effectively plan stepping strategy when only uvf cues were available (lvf occluded), it follows that this was based on visual exproprioceptive information regarding head position relative to the environment. The utility of head-obstacle exproprioceptive information in guiding foot placement during obstacle negotiation under lvf occluded conditions has recently been demonstrated (Rietdyk and Rhea, 2006). These authors showed that the presence of vertical poles (2 m high), to indicate obstacle position, enabled placement of the lead and trail foot to return (decrease) to values recorded under full vision conditions. Of further interest to the present study is that Rietdyk and Rhea (2006) also found that lead-limb toe clearance was increased under lvf occluded conditions, and was unaffected by the presence of the positional cues on obstacle position. This indicates that participants were unable to ‘update’ lower-limb trajectory when exproprioceptive information regarding foot position relative to the obstacle was not available from lvf. Findings that in the present study, under lvf occluded conditions participants were unable to 'update' exproprioceptive information of the foot / lower leg relative to the floor in the same manner as occurred under full field vision, further strengthens the suggestion that visual exproprioceptive information regarding foot position relative to the environment is required to 'update' lower-limb trajectory (Rhea and Rietdyk, 2007; Graci *et al.,* 2010).

4.4.2 When is vision customarily used during step descent to plan / regulate landing control?

Compared to when vision was available throughout, the occlusion of visual information available from either full or uvf (i.e. lvf occluded) from toe-off or mid-swing onwards caused very few differences in landing control. The only exceptions were that kneedrop occurred earlier in the descent, which resulted in kneedrop distance also being reduced (table 4.1, figure 4.1). Previous research has shown that when vision is occluded prior to MI, during step descent participants are unable to scale kneedrop parameters to stepping height (Cowie *et al.,* 2008). They also adopt a strategy of sitting back on the trailing limb and use their lead-limb to probe for the ground (indicated by increased prelanding ankle plantar-flexion), while not committing to weight transfer until somatosensory feedback from the lead-limb confirms landing has occurred (Buckley *et al.,* 2008). In the present study, adaptations to stepping strategy were not observed when information available from either full or uvf (lvf already occluded) was restricted after MI (i.e. beginning of swing or mid-swing onwards). In combination with previous work (Buckley *et al.,* 2008; Cowie *et al*., 2008), the present findings suggest that visuomotor control of step descent that relates to stepping strategy predominantly occurs prior to or during MI (i.e. feedforward control, with stepping dynamics determined in a ballistic manner). This is consistent with work on target-directed locomotion, which has shown that the occlusion of vision during the swing phase of the locomotor cycle has no effect (Hollands and Marple-Horvat, 1996; Patla *et al.,* 1996). Furthermore, the finding that only the magnitude and timing of kneedrop was altered when vision was unexpectedly occluded from mid-swing onwards, suggests that the precise control of landing only requires subtle 'fine tuning' using online vision in the latter portion of the descent phase. Such late online 'fine tuning' has been shown to occur when stepping onto floor-mounted targets (Reynolds and Day, 2005).

Occluding vision several seconds prior to MI has been shown to result in participants pre-planning not to receive visual information during step descent, and instead relying more on feedforward mechanisms or other sensory modalities such as proprioception, while at the same time altering stepping strategy (Buckley *et al.,* 2008). In the present study, because vision was occluded unexpectedly on a trial-by-trial basis after MI, it is possible that participants planned for the worse-case scenario (i.e. vision unavailable) and used only feedforward mechanisms. However, the finding that participants exhibited different measures related to kneedrop depending on the vision condition (see figure 4.1), is indicative of subtle online corrections. Therefore, it would seem that while participants relied heavily on feedforward mechanisms, they also pre-planned to use visual feedback during step descent to ‘fine tune’ landing when it was available.

Although it was found that kneedrop parameters were affected by occlusion of vision after MI, it is unclear how such subtle ‘fine tuning’ affected landing. Kneedrop represents the instance when the lead-leg shank segment reaches its maximum forwards swing before its polarity is reversed. Thus one might expect a change in kneedrop would coincide with a corresponding change in ankle angle. As highlighted above, there was no significant change in ankle angle when vision was occluded following MI. However, a subsidiary analysis showed that there was a trend of increasing foot angle relative to the floor across vision conditions (p = 0.07). This trend suggests that when vision was occluded from mid-swing onwards, kneedrop was altered in order to ensure landing occurred more ‘on the toes’. In this respect, increasing foot angle would mean the heel would need to travel further in the vertical direction to obtain a foot-flat position following foot contact with the ground. Controlled lowering of the foot (and by implication the CM) is a way of attenuating the force of landing (Buckley *et al.,* 2005; van Dieën *et al.,* 2008). Thus the trend of increasing foot angle across vision conditions tends to suggest that the changes in kneedrop parameters when vision was occluded from mid-swing onwards, occurred to ensure that landing unexpectedly was not associated with an increased peak landing force (landing force was constant across vision conditions). Additional work is required to confirm this hypothesis.

During gait, an unexpected step down, or step onto level ground, which leads to a difference between expected and actual loading at foot contact, can trigger fast corrective muscle synergies in ipsilateral and contralateral leg muscles (van der Linden *et al.,* 2007). This response ensures stability is maintained as forward momentum of the body is being controlled. It has been suggested that an efference copy of the motor command is used to predict sensory consequences of the ongoing action, which is then compared to the actual sensory feedback, enabling any error signal (possibly in the cerebellum; Miall and Wolpert, 1996) to trigger fast responses (van der Linden *et al.,* 2007). In contrast with this previous work, the perturbation experienced in the present study (i.e. occluding vision unexpectedly during step descent), had little effect on landing control; although it is possible that there were subtle changes in muscle activation (which were not assessed), even despite landing stiffness being unaffected (table 4.2). The lack of affect on landing control can be explained by the fact that the perturbation in the present study was limited to the visual modality and did not involve any unexpected change in the height of step descent. Visual information during step descent was only required for subtle ‘fine tuning’, and there should have been little or no discrepancy between the expected proprioceptive feedback and the proprioceptive feedback that was received during descent.

4.5 Conclusion

To conclude, findings indicate that the visuomotor control of step descent takes advantage of information from the lvf when available, and that such information is predominantly used during movement planning. Indeed, when information from either full or uvf was available after toe-off, there was only subtle ‘fine tuning’ of landing control in the latter portion of the descent phase. These findings suggest that in the customary situation where we do not directly look at our feet when descending steps and stairs, information from the lvf is acquired to ‘update’ expproprioceptive information of the foot / lower leg relative to the floor, which subsequently allows landing to be controlled with increased level of certainty.

Chapter 5

Experiment 2

Does the role of feedforward versus online vision used in the control of step descent change when descents are completed carrying added mass?

5.1 Introduction

Whilst walking with heavy shopping bags, for example from the supermarket to either the car or home, individuals will likely negotiate a variety of environmental demands including steps, stairs and kerb edges. Step downs in such situations can occur during on-going gait, or can occur from a stationary position. For example, prior to stepping down from a kerb to cross the road, individuals typically pause momentarily whilst waiting for an opportunity to cross safely. When descending a step carrying added mass, there will be increased downward momentum. If the increased momentum is not controlled or attenuated at landing, this may compromise stability (Hof *et al.,* 2005), resulting in the individual falling forwards (Roys, 2001) or generating a relatively large shock (reaction) force at instant of contact causing discomfort, and increasing the risk of injury. To attenuate the increased momentum generated during descent, landing control is likely adapted. For example, when descending a step wearing a weighted jacket (which increased body mass by 20 %), peak impact forces during landing have been shown to be invariant with those when stepping without wearing the jacket (Spanjaard *et al.,* 2008). Such adaptations in landing control require that the instant of contact on the lower level is accurately judged.

Previous research has shown when descending a step without added mass, that the instant of ground contact is determined using both feedforward and online visual mechanisms (see chapter 4; Buckley *et al.,* 2008; Cowie *et al.,* 2008). For example, when visual information is occluded in the few seconds immediately prior to step descent, participants adapt their stepping strategy by preparing for landing earlier during the descent (Cowie *et al.,* 2008) and adopt a ‘softer’ landing (Buckley *et al.,* 2008). In contrast, when visual information is occluded from mid-swing onwards during step descent, only slight adaptations in landing control are evident which suggest that visual cues acquired during the latter portion of the descent are used to subtly ‘fine tune’ landing (chapter 4). Collectively, these results suggest that visuomotor control of step descent appears predominantly biased towards feedforward visual mechanisms. However, in situations when task demand is increased, online visual mechanisms become increasingly important (Reynolds and Day, 2005; Marigold and Patla, 2008). For example, during negotiation of multi-surface terrain with the lvf occluded, online visual cues pertaining to the lower limb and immediate floor area are unavailable, while feedforward visual cues from the uvf ~2 steps in advance are. In such situations, participants adopt a more cautious gait strategy, evidenced by reduced gait speed and step length, a strategy to allow for increased control of foot placement on the uneven terrain (Marigold and Patla, 2008). Furthermore, when participants are required to precisely step onto a floor mounted target, the occlusion of online vision from lead limb toe-off results in a significant decrease in the accuracy of foot placement (Reynolds and Day, 2005). When descending a step carrying added mass, the requirement to attenuate the increased momentum generated during the descent increases the demands of the task. In the present study, the effect of added mass was hypothesised to change the role of feedforward versus online vision used in the control of step descent.

5.2 Method

5.2.1 Participants

10 healthy adult participants (5 male, 5 female), age 22.3 ± 4.2 years (mean ± SD), height 167.9 ± 8.2 cm and mass 71.7 ± 16.2 kg, were recruited using the inclusion / exclusion criteria described in the general methods (see sub section 3.1). The tenets of the Declaration of Helsinki were observed and the experiment gained approval from the local Research Bioethics Committee. Written informed consent was obtained from each participant prior to undertaking the study.

5.2.2 Visual assessment

Binocular visual acuity (VA), binocular contrast sensitivity (CS) and stereoacuity were each assessed using the methods described in the general methods (see sub section 3.2). All participants recorded values within the limits of healthy eyes (Vale *et al.,* 2008a), with measures of VA, CS, and stereoacuity of -0.1 ± 0.1 logMAR, 1.9 ± 0.1 log units, and 48.5 ± 13.3 secs of arc respectively.

5.2.3 Protocol

Participants undertook repeated step downs from three block heights (sub section 3.3.1): 14.6 cm, 21.8 cm and 15 % (± 1 cm) of participant's height (15%bodyHt). Prior to attaining the start position on the final block, participants were required to negotiate a series of 'stepping stones' (see sub section 3.6). On attaining the start position, participants were instructed, using the verbal command 'step' to initiate a step down on to the lower level. Step downs were performed adhering to the protocol highlighted in sub section 3.6. Illumination over the stepping area was 400 lux (taken at participant's chest height) and the luminance of the floor and top surface of the block was 15 and 30 cd/m^2 respectively. Prior to negotiating the 'stepping stones', participants were given a canvas bag (manufactured for carrying two bowls balls) to hold in each hand containing 0, 7.5, or 15 % of participant's total body mass.

Participants were instructed that throughout the trial they were required to hold the bags with arms down by their side, assuming a position similar to carrying shopping bags. To minimise fatigue, upon completing each trial the experimenter took the bags from the participant while they returned to the start of the stepping stones. Throughout the experiment, participants wore PLATO goggles (Translucent Technologies, Canada, sub section 3.3.2) which were used to manipulate when visual feedback was available. The goggles switched to opaque from the period immediately prior to movement initiation (MI, manually switched to coincide with verbal command) or from approximately 50 % of lead-limb swing time onwards (mid-swing, see sub section 3.3.3 for description of switching method). In occluding vision simultaneously with the command for participants to step, visual feedback was unavailable from the period immediately prior to MI throughout step descent. Step height (14.6 cm, 21.8 cm and 15%bodyHt) changed randomly every 9 trials. Each vision condition (no occlusion (i.e. full vision), occlusion from the period immediately prior to MI, occlusion from mid-swing) and mass condition (no added mass, 7.5 % added mass, 15 % added mass) were completed three times in random order, for a total of 81 trials.

An 8 camera 3-D motion analysis system (Vicon MX3, Oxford Metrics Ltd) was used to record (at 100 Hz) segmental kinematics as participants stepped down from one level to another. Participants wore shorts, t-shirt and flat soled shoes. Reflective spherical markers were attached and anthropometric measurements taken as described in the methods section (3.3.5). Joint angles were defined as the resultant angle between two adjacent segments. Knee, ankle and head angular displacement data, ground reaction force data (including CP co-ordinates) from each force platform, and the co-ordinate data for whole-body CM, knee, ankle and all foot markers were exported (at 100 Hz) for further analysis.

5.2.4 Data analysis

Visuomotor control of step descent was evaluated by determining prelanding kinematic measures and landing mechanic variables for the initial contact period. Prelanding kinematic measures included anterior-posterior (a/p) and medio-lateral (m/l) CM velocity at the instant of lead limb toe-off, kneedrop and time of kneedrop as a percentage swing time (see methods section for further details of kneedrop parameter, 3.5.3). The following measures were also determined for the instants of landing: lead limb knee (θ_{knee}) and ankle (θ_{ank}) joint angular displacement, a/p stepping distance, a/p and (downward) vertical CM velocity and the amount of bodyweight supported by the trailing limb (bodyWt sup fz cont) normalised to individual body weight. θ_{knee} and θ_{ank} joint angular displacement were determined as the change in joint angle at the instant of landing relative to their average position when standing stationary on the upper step. Stepping distance was measured as the a/p distance between the marker of the second metatarsal heads on the trail and lead limbs at the instance of lead limb contact on the lower level. Instant of landing was defined as the instant when the vertical ground reaction force (GRF) on the lead limb first became greater than 20 N.

The mechanics of initial landing, defined as the period up to trail limb toe-off, were evaluated by determining the peak vertical GRF normalised to individual body weight ($Fz_{peak\ norm}$), peak knee (θ_{knee}) and ankle (θ_{ank}) angular joint displacement and peak angular velocity of the knee (ω_{knee}) and ankle (ω_{ank}) joint. Time from MI to lead limb toe-off, lead limb toe-off to foot contact (single support, SS), foot contact to contra-lateral limb toe-off (weight transfer, WT), step time and time from contact with the lower level to Fz_{peak} (time to Fz_{peak}) were also evaluated. Step time was calculated from MI to trail limb toe-off. Peak θ_{knee} and θ_{ank} and ω_{knee} and ω_{ank} determined how the lead limb was loaded during the landing phase. MI onset was defined from when the resultant x, y coordinates of the CM and CP first diverged by greater than 10 mm for 5 consecutive frames. Lead limb toe-off was defined as the instant when the a/p displacement of the second metatarsal head marker was greater than 3 mm for 5 consecutive frames from its average location when standing stationary. Trail limb toe-off was defined from when the vertical GRF on the upper block first dropped below 20 N.

5.2.5 Statistical analysis

Each calculated dependant variable was analysed using separate x 3 Mass (no added mass, 7.5 % added mass, 15 % added mass) x 3 Vision Condition (available throughout, vision occluded from the period immediately prior to MI, vision occluded from mid-swing) x 3 Step Height (low, high, 15%bodyHt) ANOVA, with repeated measures on each factor. Level of significance was accepted at $p < 0.05$, and post-hoc analyses were performed using Tukey's HSD.

5.3 Results

The effects of mass, vision and step height upon the prelanding kinematics, the mechanics of landing and temporal parameters are summarised below in tables 5.1, 5.2 and 5.3 respectively.

5.3.1 Prelanding kinematics

m/l and a/p CM velocity at lead limb toe-off were unaffected by vision, but m/l CM velocity was affected by step height ($p < 0.04$) and a/p CM velocity was affected by mass ($p < 0.04$, table 5.1). a/p CM velocity was unaffected by height and m/l CM velocity was unaffected by mass; there were no significant interactions between factors. Lateral CM velocity was higher when participants stepped from high compared to low step height, and anterior CM velocity increased in the 15 % added mass compared to no added mass condition (table 5.1).

Table 5.1. Prelanding kinematics: Group mean (± 1 SD) a/p and m/l CM velocity at lead limb toe off (a/p, m/l CM vel toe-off), kneedrop, time of kneedrop, step distance, and ankle and knee angle (θ), a/p, m/l and vertical CM velocity and body weight supported on the trail leg (bodyWt sup) at instant of landing across mass and vision condition.

	0 added mass			**7.5 % added mass**			**15 % added mass**			**sig fact**
	No disr	**MI**	**mid-swing**	**No disr**	**MI**	**mid-swing**	**No disr**	**MI**	**mid-swing**	
a/p CM vel toe-off (cm/s)	-115.5	-115.5	-115.9	-123.4	-121.4	-121.4	-123.2	-124.0	-119.6	M
	(23.7)	(27.7)	(21.9)	(26.1)	(31.3)	(26.9)	(30.6)	(34.9)	(28.4)	
m/l CM vel toe-off (cm/s)	123.1	120.1	123.6	126.8	126.7	125.3	126.3	125.7	126.8	H
	(17.9)	(17.5)	(18.4)	(18.1)	(16.8)	(18.6)	(24.5)	(19.4)	(19.4)	
Kneedrop (cm)	12.5	12.5	12.2	13.1	13.0	12.6	13.5	12.8	12.8	M, V
	(2.5)	(2.5)	(2.5)	(2.3)	(2.7)	(2.3)	(2.5)	(2.5)	(2.5)	
Time of kneedrop (% ss)	62	62	63	64	63	62	64	64	64	n/a
	(6)	(5)	(6)	(5)	(6)	(6)	(5)	(5)	(6)	
a/p step distance (cm)	398.2	402.5	401.1	403.2	406.3	400.2	406.6	401.6	400.9	m/v
	(35.6)	(30.1)	(28.5)	(27.9)	(28.4)	(28.5)	(32.9)	(27.2)	(31.3)	
θ_{ank} (deg)	-31.5	-30.6	-30.9	-30.4	-30.6	-31.0	-31.6	-30.9	-31.5	n/a

	(4.22)	(3.55)	(3.36)	(3.59)	(3.76)	(4.01)	(3.74)	(3.44)	(3.88)	
θ_{knee} (deg)	7.8	7.9	7.5	8.4	7.7	7.9	7.8	7.9	8.1	n/a
	(3.29)	(3.54)	(3.37)	(3.75)	(3.83)	(3.80)	(3.84)	(3.83)	(3.77)	
Vert CM vel (cm/s)	-481.0	-490.4	-469.3	-477.8	-464.5	-470.9	-470.3	-450.1	-450.0	n/a
	(116.2)	(116.5)	(108.3)	(113.2)	(134.3)	(126.5)	(113.5)	(107.1)	(111.6)	
a/p CM vel (cm/s)	-439.8	-447.4	-437.7	-448.1	-447.9	-435.4	-448.6	-441.9	-435.2	V
	(69.1)	(70.4)	(66.0)	(65.6)	(71.8)	(69.1)	(73.0)	(66.3)	(74.9)	
m/l CM vel (cm/s)	-147.4	-144.53	-142.88	-151.44	-146.50	-150.22	-148.07	-149.40	-143.67	n/a
	(23.4)	(27.6)	(28.1)	(26.5)	(25.0)	(23.0)	(24.8)	(28.5)	(25.2)	
bodyWt sup (%)	87	87	85	89	90	89	92	92	93	M
	(9)	(8)	(9)	(9)	(10)	(10)	(11)	(11)	(10)	

Data are averaged across step heights to illustrate the limited effects of mass and vision condition. Factors found to be significant are shown by capital letter ($p < 0.05$), for mass (M), vision condition (V), and step height (H). There were no interactions between factors. Ankle angle (-ve) indicates the amount of plantar-flexion and knee angle indicates amount of knee flexion relative to standing.

Kneedrop was unaffected by step height, but was significantly affected by mass ($p < 0.05$) and vision condition ($p < 0.05$); there was no significant interaction between factors. Post hoc analysis revealed that kneedrop reduced when vision was occluded from mid-swing onwards compared to when it was available throughout and increased with 15 % added mass and 7.5 % added mass compared to no added mass condition (table 5.1). Time of kneedrop was unaffected by vision or step height, but there was a trend of being affected by mass ($p < 0.07$). This trend highlighted that kneedrop occurred later in the step descent when stepping with 15 % added mass compared to 7.5 % and no added mass conditions (table 5.1).

At the instant of landing, bodyWt sup fz cont was significantly affected by mass ($p < 0.05$), whereas a/p CM velocity was unaffected (table 5.1). None of these dependant variables were significantly affected by step height, and only a/p CM velocity was significantly affected by vision ($p < 0.05$); there were no significant interactions between factors. bodyWt sup fz cont significantly increased in 15 % added mass compared with no added mass condition, which ensured that $Fz_{peak\ norm}$ remained invariant. Anterior CM velocity was significantly reduced when vision was occluded from mid-swing compared to full vision or vision occluded from the period immediately prior to MI (table 5.1).

m/l and vertical CM velocity, θ_{knee} and θ_{ank} at the instant of landing were unaffected by mass, vision or step height conditions ($p > 0.05$). Step distance was significantly affected by a mass-by-vision interaction ($p < 0.05$). Post hoc analysis revealed that participants stepped further in the full vision condition carrying added mass compared to full vision condition carrying no added mass (table 5.1).

5.3.2 Landing mechanics

During initial landing θ_{knee} and θ_{ank}, $Fz_{peak\ norm}$ and ω_{ank} and ω_{knee} were unaffected by mass, vision or step height condition ($p > 0.05$); there were no significant interactions in any of these variables (table 5.2).

5.3.3 Temporal parameters

Time from MI to lead toe-off, SS time, time to Fz_{peak}, WT time and step time were unaffected by vision or step height conditions. Of these variables only step time was significantly affected by mass ($p < 0.05$), and there were no significant interactions between factors (table 5.3). Post hoc analysis revealed that step time increased in the 15 % added mass compared to 7.5 % and no added mass conditions.

Table 5.2. Landing mechanics: Group mean (± 1 SD) ankle and knee angle (θ), ankle and knee angular velocity (ω), peak vertical force (normalised to individual body weight, $Fz_{peak\ norm}$) during initial landing across mass and vision condition.

	0 added mass			**7.5 % added mass**			**15 % added mass**			**sig fact**
	No disr	**MI**	**mid-swing**	**No disr**	**MI**	**mid-swing**	**No disr**	**MI**	**mid-swing**	
θ_{knee} (deg)	23.8 (9.2)	23.5 (8.9)	23.6 (8.8)	25.1 (9.3)	24.5 (10.6)	22.1 (7.7)	23.3 (10.4)	24.4 (10.0)	23.4 (10.3)	n/a
θ_{ank} (deg)	3.1 (4.8)	3.3 (4.7)	2.5 (4.5)	4.0 (5.5)	3.4 (6.8)	3.2 (4.8)	2.7 (6.3)	3.1 (5.7)	3.5 (5.4)	n/a
ω_{knee} (deg. s^{-1})	108.8 (49.4)	112.8 (48.0)	110.8 (46.1)	113.0 (48.3)	110.0 (52.0)	105.7 (49.3)	104.6 (52.7)	104.9 (50.3)	100.1 (48.9)	n/a
ω_{ank} (deg. s^{-1})	257.7 (51.8)	253.9 (58.0)	253.4 (50.1)	255.0 (52.1)	255.9 (60.0)	243.5 (49.6)	256.0 (56.5)	252.4 (52.5)	248.5 (51.1)	n/a
$Fz_{peak\ norm}$ (%)	135 (23)	133 (21)	133 (25)	135 (24)	136 (26)	133 (25)	139 (22)	136 (23)	135 (21)	n/a

There were no significant effects on any of the landing mechanic variables analysed. Ankle angle (+ve) indicates the amount of dorsi-flexion and knee angle indicates amount of knee flexion relative to standing.

Table 5.3. Temporal parameters: Group mean (± 1 SD) time from movement initiation to lead limb toe-off (MI to lead toe-off), single support (SS), weight transfer (WT), step time and time to peak vertical force (Fz_{peak}) times across mass and vision condition.

	0 added mass			**7.5 % added mass**			**15 % added mass**			**sig fact**
	No disr	**MI**	**mid-swing**	**No disr**	**MI**	**mid-swing**	**No disr**	**MI**	**mid-swing**	
MI to lead toe-off (s)	0.52 (0.04)	0.53 (0.03)	0.52 (0.04)	0.53 (0.04)	0.53 (0.05)	0.52 (0.04)	0.53 (0.05)	0.53 (0.04)	0.54 (0.05)	n/a
SS (s)	0.58 (0.08)	0.57 (0.08)	0.57 (0.08)	0.58 (0.07)	0.57 (0.08)	0.57 (0.08)	0.57 (0.08)	0.57 (0.09)	0.57 (0.08)	n/a
WT (s)	0.23 (0.03)	0.23 (0.05)	0.24 (0.06)	0.24 (0.05)	0.25 (0.08)	0.23 (0.04)	0.24 (0.05)	0.24 (0.05)	0.24 (0.05)	n/a
Step time (s)	1.33 (0.14)	1.23 (0.15)	1.23 (0.14)	1.25 (0.12)	1.35 (0.14)	1.32 (0.14)	1.34 (0.13)	1.34 (0.14)	1.25 (0.13)	M
time to Fz_{peak} (s)	0.22 (0.03)	0.23 (0.04)	0.22 (0.04)	0.22 (0.04)	0.24 (0.07)	0.23 (0.04)	0.23 (0.05)	0.23 (0.05)	0.23 (0.05)	n/a

Data are averaged across step height to illustrate the limited effect of mass and is denoted by capital letter (M) ($p < 0.05$). There were no effects of step height or vision condition. There were no interactions between factors.

5.4 Discussion

The present study determined whether the role of feedforward versus online vision used in regulating step descent landing control changes when descents are completed carrying added mass. Compared to when vision was available throughout step descent, occluding vision from immediately prior to MI had no effect on landing control irrespective of added / no added mass condition. This finding likely suggests that feedforward visual cues are acquired in advance of MI. Compared to when vision was available throughout step descent and vision was occluded from mid-swing onwards during the descent, the same subtle adaptations in landing control were evident across added / no added mass conditions. These subtle adaptations in landing control were evidenced through reduced kneedrop and anterior CM velocity at the instant of landing when vision was occluded from mid-swing onwards during the descent compared to vision available throughout (see table 5.1). These findings provide additional support for the subtle role of online vision 'fine tuning' step descent landing control (as per chapter 4). There were no significant vision-by-mass interactions indicating any differences in step descent landing control when vision was occluded from mid-swing onwards during the descent and descents were completed in added / no added mass conditions. This suggests that the role of online vision used in regulating step descent landing control remains invariant irrespective of added / no added mass condition. Findings suggest that the role of feedforward versus online vision used in regulating step descent landing control does not change when descents are completed carrying added mass. Further interpretations of these findings are provided below.

Previous research investigating visuomotor control of step descent has demonstrated that step descent landing control appears predominantly biased towards feedforward visual mechanisms (see chapter 4; Buckley *et al.,* 2008; Cowie *et al.,* 2008). However, in situations when task demand is increased, such as negotiating uneven terrain or during a precision stepping task, online visual mechanisms become increasingly important (Reynolds and Day, 2005; Marigold and Patla, 2008). Descending a step carrying added mass increases the demands of the task, which in the present study, was hypothesised to change the role of feedforward versus online vision used during step descent to regulate landing control. If the role of online vision increased when step descents were completed carrying added mass, the adaptations in landing control between vision available throughout and occlusion from mid-swing onwards would be greater when descents were completed carrying added mass compared to carrying no added mass. When descents were completed carrying no added mass and vision was occluded from mid-swing onwards during the descent, compared to descents completed carrying no added mass and vision available throughout, kneedrop and anterior CM velocity at the instant of landing decreased (see table 5.1). These findings provide further evidence that step descent landing control only requires online vision during the latter portion of step descent to subtly ‘fine tune’ landing (as per chapter 4). The same subtle adaptations in step descent landing control were observed when descents were completed carrying added mass and vision was occluded from mid-swing onwards during the descent, compared to descents completed carrying added mass and vision available throughout (see table 5.1). In addition, there were no significant vision-by-mass interactions indicating any differences in step descent landing control when vision was occluded from mid-swing onwards during the descent and descents were completed in added / no added mass conditions. The only significant vision-by-mass interaction was found for step distance; participants stepped further in the full vision condition carrying added mass compared to full vision condition carrying no added mass (table 5.1). These findings suggest that the role of online vision in visuomotor control of step descent remains invariant irrespective of added / no added mass condition. Such findings are likely attributed to the adaptations that occurred when descents were completed carrying added mass irrespective of vision condition. Compared to when step descents were completed with no added mass, descents completed carrying added mass resulted in no subsequent increase in landing force on the lower level (table 5.2). Therefore landings in the added mass conditions must have been controlled to ensure that $Fz_{peak\ norm}$, along with ω_{ank} and ω_{knee} during initial landing and θ_{knee} and θ_{ank} at the instant of and during initial landing remained invariant compared to no added mass condition (table 5.1 and 5.2). Such similarities between landings in the added / no added mass conditions clearly suggests that when descents were completed carrying added mass, participants adapted their landing control to attenuate the increased momentum generated during the descent. Attenuating the increased momentum ensured no increase in risk of injury when landing on the lower level carrying added mass, which is evidence of participants adopting a safety stepping strategy. The ability to maintain such a safety stepping strategy when descending the step carrying added mass was achieved through increasing the body weight supported on the trail leg during step descent (table 5.1) and increasing step time (table 5.3). These adaptations ensured that participants increased the amount of control available during the descent, which subsequently ensured landing force was similar in added / no added mass conditions. Since the increased demands of the task when descending the step carrying added mass was attenuated, this meant that there was no increased demand placed on the visual system to regulate step descent landing control. Subsequently, this meant that online visual mechanisms did not become increasingly important when step descent were completed carrying added mass.

It is interesting to note, that the safety stepping strategy observed in the present study when descents were completed carrying added mass is similar to the adaptations in step descent landing control previously reported when vision was occluded or blurred prior to step descent (Buckley *et al.,* 2008). Compared to when vision was available throughout, Buckley *et al.* (2008) highlighted that the occlusion or blurring of vision in the few seconds immediately prior to step descent resulted in (amongst other changes) participants increasing the amount of body weight supported on the trail leg at the end of the initial landing period and increasing step time. This safety strategy demonstrated under conditions of visual occlusion subsequently ensured there was no increase in landing force under conditions of uncertainty regarding the precise location of the lower floor level (Buckley *et al.,* 2008). It is possible that in the present study, since participants had already adopted a safety stepping strategy when descending the step carrying added mass, there was no scope to further adapt the stepping strategy when vision was occluded from mid-swing onwards during the descent. Therefore, when step descents were completed carrying added mass, the adaptations in stepping strategy under such conditions possibly 'masked' any effects of occluding online vision during the descent. However, the effect of added mass failed to 'mask' the subtle effect of occluding vision from mid-swing onwards during step descent. When vision was occluded from mid-swing onwards, kneedrop and anterior CM velocity at the instant of contact was reduced. However, when step descents were completed carrying added mass, there was no effect on a/p CM velocity and an opposite effect on kneedrop (kneedrop increased, see table 5.2). It therefore appears unlikely that in the present study the effect of added mass 'masked' the effects of occluding online vision during the descent.

The previous experimental chapter (chapter 4) concluded that the few differences observed in landing control when vision was occluded from toe-off or mid-swing onwards during step descent suggested that step descent landing control was regulated through visual cues obtained predominantly prior to or during MI. In the present study, compared to when step descents were completed and vision was available throughout, occluding vision from immediately prior to MI had no effect on landing control, irrespective of mass condition. Since landings remained invariant when vision was occluded from the period immediately prior to MI compared to vision available throughout, results suggest that feedforward visual cues contributing to visuomotor control of step descent are acquired predominantly prior to MI and not during MI. These findings are consistent with previous research highlighting the adaptations in step descent landing control when vision is occluded or degraded in the few seconds prior to MI during step descent (Buckley *et al.,* 2008; Cowie *et al.,* 2008).

In the present study, it was possible for the role of feedforward versus online vision to remain invariant and instead the importance of vision per se to increase when descents were completed carrying added mass. Such findings would have been evidenced through significant differences in step descent landing control when descents were completed under conditions of added mass compared to no added mass when vision was occluded from the period immediately prior to MI and from mid-swing onwards during the descent. However, with no significant differences in landing control when descents were completed with / without added mass and vision was occluded from the period immediately prior to MI or from mid-swing onwards, the role of vision per se was not increased.

5.5 Conclusion

The present study determined whether the role of feedforward versus online vision used to regulate step descent landing control changes when descents are completed carrying added mass. Compared to when vision was available throughout the descent, the occlusion of vision from mid-swing onwards resulted in the same subtle adaptations in landing control across added / no added mass conditions. There were no significant vision-by-mass interactions indicating any differences in step descent landing control when vision was occluded from mid-swing onwards during step descent and descents were completed in added / no added mass conditions. This suggests that the role of online vision in regulating step descent landing control remains invariant irrespective of added / no added mass condition. Such findings are likely attributed to the adaptations that occurred when descents were completed carrying added mass irrespective of vision condition. When step descents were completed carrying added mass, participants attenuated the increased momentum generated during the descent to ensure no increase in landing force on the lower level compared with no added mass condition. Since the increased demands of the task when descending the step carrying added mass was attenuated, this meant that there was no increased demand placed on the visual system to regulate step descent landing control. Subsequently, this meant that online visual mechanisms did not become increasingly important when step descents were completed carrying added mass.

These findings suggest that the role of feedforward versus online vision used in regulating step descent landing control remains invariant when descents are completed carrying added mass.

Chapter 6

Experiment 3

Use of single-vision distance spectacles improves landing control during step descent in well-adapted multifocal lens-wearers

The work presented in this chapter has been published as:

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6.1 Introduction

Epidemiological research has shown that multifocal spectacle-wearers (bifocal and progressive addition lenses; PALs) are more than twice as likely to fall compared with non-multifocal spectacle wearers (Lord *et al.,* 2002) with this risk further increasing when negotiating stairs (Davies *et al.,* 2001; Lord *et al.,* 2002). Negotiating steps, stairs and surface height changes may be particularly problematic for multifocal spectacle-wearers because they are likely to view the step / stair edge through the lower region of the lens designed for reading, which is typically focussed at about 40 cm (16”) (figure 6.1). The lvf including the view of any surface height change and the foot are therefore blurred and thus the exact and relative height of the floor is difficult to judge (Lord *et al.,* 2002). The additional dioptric power in the reading section of the lenses will also magnify objects such that step / stair edges will appear higher and closer than they actually are. Such effects, when presented acutely, have been shown to significantly affect an individual’s gait when walking onto a raised surface (Elliott and Chapman, 2010). Multifocal wearers likely adapt to the blur / magnification effects with time. Even so, long term multifocal wearers still display increased variability in foot positioning when walking up to (Johnson *et al.,* 2007) and increased toe clearance variability when stepping onto (Johnson *et al.,* 2008) a raised surface, and make more accidental contacts with it (Johnson *et al.,* 2007, 2008; Menant *et al.,* 2009) compared to when wearing single-vision distance spectacles. Multifocals have been shown to have no effect on standing postural stability (Johnson *et al.,* 2009).

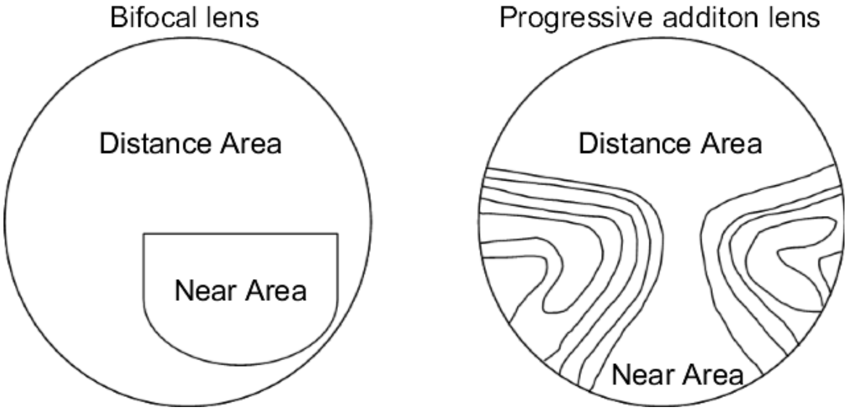

Figure 6.1. Areas of bifocal and PALs that provide clear vision for distant and near objects. The PAL also includes a corridor of clear vision for objects at intermediate distances and the peripheral areas of distortion.

In older adults, falls occur about three times as often during stair descent compared to stair ascent (Tinetti *et al.,* 1988; Startzel *et al.,* 2000) and falls on stairs are a leading cause of accidental death, multiple injuries and hospitalisation in older people (Tinetti *et al.,* 1988; Startzel *et al.,* 2000). In addition, vision is believed to play a major role in the successful negotiation of stairs (Startzel *et al.,* 2000), yet surprisingly, given the high percentage of elderly individuals who wear multifocal spectacles (Lord *et al.,* 2002), no previous studies have reported whether their use causes difficulties when descending steps or stairs. Previous work has highlighted that estimating the precise height of the lower surface and / or the foot's position relative to it, is dependent on visual information gained prior to MI (see chapter 5, Buckley *et al.,* 2008; Cowie *et al.,* 2008) likely acquired from the lvf (chapter 4). Indeed if the lvf is occluded prior to step descent, participants adapt their stepping strategy by moving their landing limb into place earlier during the descent and reducing vertical impact forces during the initial contact period; but make no alteration in stepping strategy (chapter 4). These changes are likely due to a lack of exproprioceptive visual information (foot position relative to the environment) so that participants were unable to modulate landing in the same way as occurred when they had access to full field vision (chapter 4). When descending step / stairs wearing multifocal lenses, the lower floor area and foot would become blurred and magnified when viewed through the lower portion of the lens. This would result in uncertainty in determining the precise instant of foot contact, which was hypothesised to lead to reduced landing control and / or increased landing control variability, either of which could affect safety.

The focus of the present study was to determine whether step / stair descent control in older long-term multifocal wearers is improved when they wear single-vision distance lenses. Specifically, the aim was to determine in habitual multifocal lens wearers, when stepping down from various heights, if landing became less variable and / or more controlled when using single-vision distance spectacles compared to multifocals.

6.2 Methods

The following data were collected by a previous Ph.D student (for further details see Johnson, 2008). However, the raw data from Vicon was re-analysed using a more detailed approach and in some instances, trials had to be re-processed.

6.2.1 Participants

Twenty community-dwelling participants (12 female and 8 male, mean age 71.9 ± 4.2 years, range 62 - 80 years; height: 1.65 ± 0.08 m; BMI: 26.2 ± 3.5 kg/m^2) were recruited as per the inclusion / exclusion criteria reported by Johnson (2008). All participants were independently mobile, able to follow simple instructions, and according to self-report, suffered no neurological, musculoskeletal or cardiovascular disorders that could interfere with balance control or stepping. Those with vestibular disturbances, diabetes, a history of falling in the previous year, or taking medications that could affect balance or vision were excluded. Physical activity levels were determined by self-report using the activity scale of the Allied Dunbar National Fitness Survey (Allied Dunbar Fitness Survey, 1992). All participants engaged in light to moderate physical activities including for example, gardening, light house work and dancing for at least 30 minutes, five days a week. Participants had normal healthy eyes, determined by a full eye examination including ocular screening using slit lamp biomicroscopy, tonometry, indirect ophthalmoscopy, central visual field screening, and binocular vision assessment. Participants had habitually worn multifocal spectacles for at least three years (median 13 years, range 3 – 30 years). Nine wore PAL and eleven wore bifocal spectacles and this included a variety of different types of bifocal and PALs. Seven participants were myopes and thirteen were hyperopes. Median distance spectacle spherical equivalent power was + 2.00 DS, range - 4.75 DS to + 5.75 DS and the median reading addition required was + 2.25 DS, range +1.75 DS to + 2.75 DS.

The tenets of the Declaration of Helsinki were observed and the experiment gained approval from the local Research Bioethics Committee. All participants gave written informed consent and were asked to refrain from alcohol intake during the evening before testing.

Each participant had three pairs of spectacles made for them: bifocals, PALs and single-vision distance, using the refractive error determined from their own spectacles using focimetry. Each participant was provided with slightly different frames and sizes to ensure optimal fit, but the three pairs of spectacles used by each participant were identical in frame style and size and were fitted to ensure the same back vertex distance and pantoscopic angle. The bifocal type was a 28 mm diameter D-segment and the PALs were Norville NCF5 (The Norville Group Ltd., Gloucester, UK), a commonly used PAL in the UK that uses a compromise hard-soft design. All PALs were positioned with the fitting cross-alignment at the centre of the pupil in primary gaze and the top of the bifocal segment aligned with the participant's lower lid.

6.2.2 Clinical evaluation

To assess how vision was affected by the different portions of the multifocal lenses, binocular visual function was measured with (1) near, (2) intermediate (calculated at 50 % of the reading addition power), and (3) distance refractive corrections using full aperture trial frames. Contrast sensitivity (CS) was measured using the Pelli-Robson chart (Pelli *et al.,* 1988) using a by-letter scoring system and a chart luminance of 200 cd/m² (Elliott *et al.,* 1991); Visual acuity (VA) was measured using high (90 %) and low (25 %) contrast Regan logMAR charts (Hazel and Elliott, 2002) with a letter-by-letter scoring system and chart luminance of 160 cd/m² (Ferris and Bailey, 1996); and depth perception with the Howard-Dohlman apparatus (mean of three trials). To determine visual function at a distance that would be encountered when negotiating steps and kerbs in the "real world" (Patla and Vickers, 2003), visual assessments were undertaken at a distance that was equivalent to the distance (average, 1.4 m) between each participant's eye and the floor level when standing on a 15 cm high block. LogMAR and depth perception (stereoacuity) scores were then derived by incorporating a correction factor for each participant's working distance. CS, VA and depth perception scores for the three refractive prescriptions are presented in table 6.1. These scores indicate that vision was significantly worse when viewing through the near compared with both the distance and intermediate prescriptions.

As plantar cutaneous sensation plays an important role in postural control (Lord *et al.,* 1991b; Melzer *et al.,* 2004), sensitivity of the soles of participant's feet was assessed by determining the ability to detect a 10 g force applied to five key sites (hallux, 1st, 3rd, and 5th metatarsal heads and heel) using a monofilament (Bailey instruments Ltd., Manchester, UK, Simoneau *et al.,* 1991). Sixteen participants had normal sensation, and four had reduced sensation at one or two sites tested on the forefoot. The inability to detect monofilament appeared to be due to callus formation. In all cases when the skin was tested immediately adjacent to the callused area sensation was present. Functional mobility was assessed using the timed up-and-go test (TUG, Podsiadlo and Richardson, 1991). Participants took 8.2 ± 1.2 seconds to complete this test classifying them as functionally independent and non-fallers (Shumway-Cook *et al.,* 2000).

Table 6.1. Visual function (at 1.4 m) test results for distance, intermediate and near refractive prescriptions.

Test	**Distance**	**Intermediate**	**Near**
High-contrast visual acuity (logMAR)	- 0.08 (0.05)	-0.02 (0.13)	0.34 (0.19)^{D,I}
Low-contrast visual acuity (logMAR)	0.01 (0.07)	0.10 (0.16)	0.54 (0.15)^{D,I}
Contrast sensitivity (log)	1.90 (0.07)	1.87 (0.10)	1.73 (0.17)^{D,I}
Depth perception (min arc)	11.8 (7.8)	20.0 (14.2)	42.2 (25.9)^{D,I}

Data are expressed as the mean (± 1 SD).

Significant difference between distance and near (^{D}) and intermediate and near (^{I}) ($p < 0.001$).

6.2.3 Step descent protocol

From a stationary standing position on top of a block that was placed over a force-platform, participants stepped down onto an adjacent force platform. The force-platforms (AMTI OR6-7, Advanced Mechanical Technologies Inc., Boston, USA) measured (at 100Hz) the contact forces between the foot and the ground. A five-camera, 3-D motion analysis system (Vicon 250; Oxford Metric Ltd., Oxford, UK) was used to simultaneously record (at 50 Hz) body segment kinematics as participants completed each step down. Three block heights were used, equating to those of a kerb (7.5 cm), a stair riser (15.0 cm), and stepping from a bus (22.0

cm); obstacles frequently encountered in daily life (Powell-Smith and Billington, 1986). Blocks were constructed from medium density fibreboard of 1.8 cm thickness, which were bonded together to create a solid block with standing area 46.4 cm x 50.8 cm. Each block was covered with coloured vinyl material that matched the surrounding floor. Room illuminance, measured at head height, was approximately 300 lux, and the luminance of the floor and top surface of the step was 30 cd/m^2 measured using a photometer (CS-100; Minolta Co. Ltd., Osaka, Japan).

Starting position on top of the block was feet positioned a comfortable width apart and the tips of their shoes aligned directly behind the leading edge of the block. After approximately five seconds in this position (looking straight ahead), participants were instructed to ‘step down’ in a single step at their own comfortable speed coming to a stationary standing position on the lower level with their feet side by side. Participants were free to choose where they looked when stepping. Participants undertook a familiarisation trial at each block height wearing their own spectacles. For each block height (low, medium, high), trials were repeated whilst wearing single-vision distance, PAL or bifocal spectacles. Participants were not informed which pair of spectacles they had been given. All trials were repeated three times with the order of spectacle condition and block height randomised (height was ‘blocked’ in three’s due to practicalities associated with changing the step), totalling 27 trials. Participants led with the same self-selected limb in all trials. Any trial that was not completed according to these instructions was discarded and repeated. An assistant stood close-by to ensure that participants did not fall if they should stumble. Participants had a seated rest each time block height was changed to minimise the onset of fatigue.

For each participant, data were collected during a single two-hour testing session. Participants wore their own shorts, t-shirt and low-healed comfortable shoes. The five cameras, which were either wall or ceiling mounted, where positioned with approximately 70 ° separation and encircled the stepping area. Reflective spherical markers where attached and anthropometric measurements taken are described in the methods section (3.3.5).

The 3-D marker trajectory data were filtered and processed as described in the methods section (3.3.5) to define a 3-D linked-segment model of the participant incorporating whole-body CM location. Joint angles were defined as the resultant angle between two adjacent segments. Knee, ankle and head flexion-extension angular displacement data, and the 3-D co-ordinate data for whole-body CM, and knee, ankle and all foot markers were exported (at 50 Hz) and the 3-D ground contact force data from each force platform (including magnitude and the co-ordinates of its instantaneous location were exported (at 100 Hz) for further analysis.

6.2.4 Data analysis

The analysis predominantly concentrated on prelanding kinematics and the mechanics of landing. Prelanding kinematic measures included, head pitch angle, lead limb knee (θ_{knee}) and ankle (θ_{ank}) joint angular displacement, a/p position of CM within base of support (CM-positioning), and a/p, m/l and vertical (downward) CM velocity for the instant of landing. Head flexion angle at lead limb heel-off and at instant of landing was calculated to check participants did not flex their head differently across spectacle conditions at any point prior to or during step descent. Instant of landing was defined as the instant when the vertical ground reaction force (GRF) on the lead limb first increased beyond 20 N.

The mechanics of landing were assessed from the instant of landing up to the instant of trail limb toe-off. Trail limb toe-off was defined as the frame the vertical contact force on the force platform participants stepped from first dropped below 20 N. The mechanics of landing were evaluated by determining the peak vertical ground reaction force (Fz_{peak}), peak angular velocity at the knee (ω_{knee}) and ankle (ω_{ank}) joint and peak vertical CM velocity. Peak ω_{knee} and ω_{ank} joint determined how the lead limb was loaded during the landing phase.

Time from MI to lead limb toe-off (double-limb support, DS), lead limb toe-off to foot contact (single-limb support, SS), foot contact to trail limb toe-off (weight transfer, WT) and time to peak ground reaction force (time to Fz_{peak}) were also evaluated. MI was defined from when the resultant x, y coordinates of the CM and CP first diverged by greater than 20 mm for 5 consecutive frames. Lead limb toe-off was defined from when the a/p displacement of the second metatarsal head marker was greater than 3 mm for 5 consecutive frames.

6.2.5 Statistical analysis

For each outcome variable data were averaged across repetition, and analysed using the following separate (for each outcome variable analysed) two factors:

1. Spectacle: Three levels, bifocal, PAL and single-vision distance
2. Block height: Three levels, low, medium and high

For each variable, a repeated measures ANOVA was used to assess statistical significance for each factor. Level of significance was accepted at $p < 0.05$, and post-hoc analyses were performed using Tukey’s HSD test.

6.3 Results

Variability was assessed by determining the standard deviation in all variables across each spectacle condition. No significant differences in variability were found. Therefore the remainder of the results section will only consider differences in each variable’s mean across conditions.

Head angle at lead limb heel-off was significantly affected by block height ($p < 0.001$) but there was no effect of spectacle condition and no significant interactions. Individuals increased head flexion at high compared to low and medium block heights.

6.3.1 Prelanding kinematics

θ_{knee} and θ_{ank}, and m/l and vertical CM velocity at the instant of landing were significantly affected by spectacle condition ($p < 0.03$, table 6.2). All these dependent variables (except θ_{knee}) increased significantly with increasing block height ($p < 0.04$). There were no significant interactions across conditions. θ_{ank} and vertical CM velocity decreased and lateral CM velocity increased when wearing single-vision distance spectacles compared with bifocals and PALs. θ_{knee} decreased when wearing single-vision distance spectacles and PALs compared with bifocals.

Table 6.2. Prelanding kinematics: Group mean (± 1 SD) head angle, ankle and knee angle (θ), CM-positioning, and vertical, a/p and m/l CM velocity at instant of landing across spectacle condition.

	Single vision	**PAL**	**Bifocal**	**Significant**
Head angle (deg)	-29.7 (13.9)	-30.3 (13.4)	-30.5 (13.0)	n/a
θ_{ank} (deg)	-31.4 (7.5)	-32.0 (7.5)^{S}	-31.9 (7.2)^{S}	H, V
θ_{knee} (deg)	6.7 (4.1)	6.6 (4.4)^{B}	7.1 (4.5)^{S,P}	V
CM-positioning (% step dist)	33.6 (6.4)	34.0 (6.1)	33.7 (6.7)	H*
Vert CM vel (mm/s)	327 (109)	339 (111)^{S}	342 (107)^{S}	H*, V*
m/l CM vel (mm/s)	135 (36)	123 (39)^{S}	125 (39)^{S}	H*, V*
a/p CM vel (mm/s)	377 (65)	382 (64)	382 (67)	H*

Data are averaged across block height to illustrate the effects of spectacle condition. Factors found to be significant are shown by capital letter ($p < 0.05$) and asterisks ($p < 0.001$) for spectacle type (V) and block height (H). There were no significant interactions between factors. Significant difference between conditions are illustrated by upper case letters: single (^{S}) and PAL (^{P}) and Bifocal (^{B}).

Head angle, a/p CM velocity and CM-positioning at the instant of landing were unaffected by spectacle condition. All these variables except for head angle were significantly affected by block height ($p < 0.001$). There were no significant interactions across conditions. a/p CM velocity increased (in forwards direction) and the CM was positioned further forward within base of support with increasing block height.

6.3.2 Landing mechanics

ω_{knee} and ω_{ank}, vertical CM velocity and Fz_{peak} during landing were significantly affected by block height ($p < 0.001$). Only ω_{ank} and vertical CM velocity were significantly affected by spectacle condition ($p < 0.03$), both decreasing when wearing single-vision distance spectacles compared with bifocals (table 6.3). Each variable increased with increasing block height. There was a significant spectacle-by-block height interaction ($p < 0.04$) for Fz_{peak} with an increase for single-vision distance spectacles compared with bifocal but only for the medium block.

Table 6.3. Landing mechanics: Group mean (± 1 SD) peak vertical force (Fz_{peak}), vertical CM velocity (Vert CM vel) and peak ankle and knee angular velocity (ω) during landing across spectacle condition.

	Single	**PAL**	**Bifocal**	**Significant**
Fz_{peak} (N)	861 (242)	857 (247)	854 (242)	H*, h-v
Vert CM vel (cm/s)	-351 (128)	-359 (128)	-362 (127)^{S}	H*, V
ω_{ank} (deg. s^{-1})	252 (85)	256 (87)	258 (83)^{S}	H*, V
ω_{knee} (deg. s^{-1})	82 (39)	86 (42)	90 (40)	H*

Data are averaged across block height to illustrate the effects of spectacle condition. Factors found to be significant are shown by capital letter ($p < 0.05$) and asterisks ($p < 0.001$) for spectacle type (V) and block height (H). Interactions between factors are denoted by lower case letter. Significant difference between conditions are illustrated by upper case letters: single (^{S}) and PAL (^{P}) and Bifocal (^{B}).

6.3.3 Temporal parameters

DS, SS and WT times and time to Fz_{peak} were significantly affected by block height ($p < 0.001$). Only SS time and time to Fz_{peak} were significantly affected by spectacle condition ($p < 0.03$, table 6.4). There was also a significant spectacle-by-block height interaction ($p < 0.05$) for time to Fz_{peak}. SS time was increased when wearing single-vision distance spectacles compared to bifocals and PALs, whereas time to Fz_{peak} was reduced when wearing single-vision distance spectacles compared with bifocals but only when stepping from the low block height. SS time increased with each step height, and DS time increased when descending high compared with medium and low block heights. WT and time to Fz_{peak} reduced with increasing block height.

Table 6.4. Temporal parameters: Group mean (± 1 SD) for double support (DS), single support (SS), time to peak vertical force (time to Fz_{peak}) and weight transfer (WT) time across spectacle condition.

	Single	**PAL**	**Bifocal**	**Sig fact**
DS (s)	0.389 (0.074)	0.395 (0.077)	0.393 (0.079)	H
SS (s)	0.659 (0.098)	0.646 (0.096)^{S}	0.646 (0.101)^{S}	H*, V
time to Fz_{peak} (s)	0.191 (0.080)	0.195 (0.079)	0.204 (0.090)^{S}	H*, V, h-v
WT (s)	0.250 (0.067)	0.247 (0.061)	0.260 (0.077)	H*

Data are averaged across block height to illustrate the effects of spectacle condition. Factors found to be significant are shown by capital letter ($p < 0.05$) and asterisks ($p < 0.001$) for spectacle type (V) and block height (H). Interactions between factors are denoted by lower case letter. Significant difference between single conditions are illustrated by upper case letters: (^{S}) and PAL (^{P}) and Bifocal (^{B}).

6.4 Discussion

Head flexion magnitudes and lack of any differences in head flexion prior to and during step descent across spectacle conditions suggests that participants viewed the immediate lower floor area through the bottom portion of each prescribed lens. Thus, when wearing single-vision distance spectacles, participants would likely have been more certain about the precise height of the lower floor owing to having a non-blurred and / or non-magnified view of the foot, step edge and immediate floor area. In contrast, when wearing multifocals and particularly bifocals, the near portion of the spectacles blurred and magnified their vision in the lvf (confirmed by the significant reductions in CS, VA and depth perception when participant's vision was assessed at a distance of ~1.4 m wearing the near prescription lens compared to intermediate or distance lens; table 6.1). Unlike single vision-distance lenses, multifocal lenses create prismatic diplopia / jump (bifocals) and peripheral distortions (PALs). There were expected effects of block height (see chapter 4; Buckley *et al.,* 2008) but as these effects were generally consistent across spectacle condition they are not discussed.

Findings suggest that compared to using multifocal spectacles (bifocals or PALs), use of single-vision distance spectacles increased the certainty regarding the precise height of the lower floor. Findings for prelanding kinematics and the mechanics of landing indicate that when wearing single-vision distance spectacles participants had an increased SS time, a reduced vertical CM velocity (at instant of contact and during landing), and a reduced ω_{ank} during landing (table 6.2 and 6.3).

Despite significant reductions in vertical CM velocity when wearing single-vision distance spectacles, there was no change in Fz_{peak} during landing across spectacle conditions. At first these two findings seem inconsistent. However, the reduced ω_{ank} and reduced time to Fz_{peak} when wearing single-vision distance spectacles compared to multifocals indicates that the reduced landing momentum was attenuated over a shorter period than that observed wearing multifocals, which is why Fz_{peak} values were similar to those observed when wearing multifocals (table 6.3). The reduced vertical CM velocity and increased SS time when wearing single-vision distance spectacles suggests landing occurred in a more controlled manner, and as a result ω_{ank} during landing was reduced. In contrast, wearing multifocals participants tended to 'drop' onto the lower level, which caused a significant increase in all the above variables (except SS time which was reduced). The present study's finding of adapted landing control when wearing multifocals are consistent with those from chapter 4, indicating how step descent is affected by occlusion of the lvf. The present and earlier (chapter 4) study suggest that uvf information (e.g. visual exproprioceptive information regarding head position relative to the environment) can be used to effectively plan stepping strategy, but that exproprioceptive information of the foot relative to the floor (i.e. lvf information) is required for the precise control of landing.

A sideways fall during step / stair descent has previously been highlighted as one of the highest risk factors for hip fractures (Greenspan *et al.,* 2003), and it is known the elderly have reduced m/l balance control (Mille *et al.,* 1993) and experience more sideways falls during step / stair descent compared to the young (Lord *et al.,* 1993). As an increase in lateral CM velocity at the instant of landing would increase the chance of the CM moving outside the base of support at landing (Hof *et al.,* 2005), it is likely that being uncertain regarding the precise location of the lower floor height would result in individuals attempting to reduce their lateral CM velocity. This may explain why in the present study lateral CM velocity was higher when participants wore single-vision distance lenses, where an ability to precisely control landing meant there was little need to reduce lateral CM velocity, as was evident in the multifocal condition.

In the present study the hypothesised reduction in variability when wearing single-vision distance spectacles compared to multifocals was not observed. This could be attributed to the instructions given to each participant. Participants were instructed to attain a start position with toes in line with the block's edge; thus they would have been aware of the precise location of the block's edge and could therefore plan their stepping pattern accordingly. Future research investigating the effects of multifocal use on step descent should consider tasking participants with descending steps during walking.

Improvements in landing control were more pronounced when switching from bifocal to single-vision distance lenses in comparison to switching from PALs. Bifocal lenses provide a blurred and magnified image beyond about 40 cm when looking through the lvf, diplopia when viewing at the bifocal edge, and image jump when the eyes move across the bifocal edge (Walsh, 2009). PALs do not present diplopia or image jump. However, they do provide a blurred and magnified image beyond about 40 cm when looking through the lowest part of the visual field and the peripheral parts of PAL lenses are subject to distortions (fig 6.1). Nonetheless the upper section of the narrow corridor of the lvf (i.e. mid height of the lens) is focussed at intermediate distances between 50 cm and 2 m, where the lower floor level (forward of the immediate floor area) may have been viewed. The more pronounced improvements in landing control when switching from bifocal to single-vision distance lenses in comparison to switching from PALs suggests that prismatic diplopia / jump caused greater uncertainty than peripheral distortions did, or that the intermediate distance portion of PALs provided more visual information regarding floor height (average 1.4 m) than that obtained when wearing bifocals. However, the strength of any conclusions regarding the differences between bifocals and PALs is limited by the small number of participants included and requires further investigation.

6.5. Conclusion

In summary, when older adult long-term multifocal wearers used single-vision distance spectacles, control of step descent was improved. This was attributed to participants being more certain about the precise height of the lower floor level owing to a view of the lvf that was not blurred or magnified, with no image diplopia or jump and no peripheral distortions. In contrast, when wearing multifocals, participants tended to ‘drop’ onto the lower level rather than having a controlled landing. The study suggests that step descent is more controlled when wearing single-vision distance spectacles compared to either bifocals or PALs. This highlights the need for randomised controlled trials to determine whether falls rates can be reduced when older frail multifocal wearers use single-vision distance spectacles during everyday locomotion (Haran *et al.,* 2009).

Chapter 7

Experiment 4

Does the probability, awareness and / or experience of visual occlusion during step descent affect the role of online vision in regulating landing control?

7.1 Introduction

Findings from chapters 4 and 5 suggest that when step descents are completed from a stationary standing position, visual cues acquired predominantly prior to MI are used to regulate landing control. Indeed, when vision is occluded from mid-swing onwards during step descent, the subtle adaptations in landing control suggest that online vision is only required in the latter portion of the descent to ‘fine tune’ landing (see chapters 4 and 5). Importantly, in chapters 4 and 5 there was a high probability (67 %) of visual occlusion during step descent. It is possible, therefore, that the high probability of visual occlusion during the descent may reflect a drive towards increasing reliance on feedforward mechanisms, and hence an intentional reduction on the reliance for online vision during the descent to regulate landing control. An implication of such a finding is that when there is a high probability of visual occlusion during step descent and descents are completed under conditions of full vision, participant’s stepping strategy will be significantly different compared to their habitual stepping response (i.e. where there is zero probability of visual occlusion during the descent). This strategy would be consistent with previous research investigating control of single upper-limb movements, which has demonstrated that when participants are uncertain whether online vision will be available during the subsequent trial, they plan for the 'worst-case scenario' (Zelaznik *et al.,* 1983) and adopt a more feedforward mode of control (Elliott *et al.,* 1999). If the reliance on online vision in regulating step descent landings is reduced when there is a high probability of visual occlusion during the descent, reducing the probability of visual occlusion during step descent could minimise the likelihood of participants adopting such a strategy. Currently, however, there is no research investigating whether the probability of visual occlusion during step descent affects the role of online vision in regulating landing control.

Whilst the present experimental chapter investigates the influence of probability on the motor control processes regulating step descent landing control in stepping, this chapter also investigates whether participant awareness and experience pertaining to a potential visual occlusion during step descent affects the role of online vision in regulating landing control. In chapters 4 and 5, participants were made aware of the objectives of the study and were provided with a detailed explanation of the protocol prior to testing. Whilst ethical rules dictate that such information is provided to the participant, research has highlighted that the level of participant awareness pertaining to the experiment has an effect on their response during testing (Morin *et al.,* 2009). Morin *et al.* (2009) demonstrated that compared to when participants were unaware that their gait was being recorded whilst running on a treadmill, when they knew a measurement was being (or about to be) measured they modified their running pattern (evidenced through higher leg stiffness and stride frequency). When participants were informed of the parameter (leg stiffness) being measured, these same parameters were further modified. In other situations, when participants were aware of a potentially low friction surface they were required to walk across, they adopted a more cautious walking pattern to reduce the likelihood of slipping (Cham and Redfern, 2002). This cautious walking pattern was highlighted through increasing the friction between the shoe / floor (by adopting a flatter foot angle) and reducing joint moments when walking over the potentially slippery surface. Furthermore, it has been shown that prior awareness of a potentially low friction surface can also improve participant’s ability to successfully recover from a slip (Marigold and Patla, 2002).

Whilst prior awareness of a perturbation influences the mechanisms of motor control, there is an even greater effect on the measured response when participants have prior experience of a perturbation (Heiden *et al.,* 2006; Siegmund *et al.,* 2006). For example, participants who were aware of a potentially low friction surface (but had no experience of walking across the surface) adopted a flatter foot angle with the floor and reduced knee angle at heel strike when walking, but did not otherwise alter their foot / floor interaction. Only after slipping on the low friction surface did further adaptations occur, which were evidenced through increased lower limb muscle activity (Heiden *et al.,* 2006).

The combined effects of prior awareness and experience of being tripped on subsequent no trip trials has also been reported. Compared to pre-test values (i.e. trials performed prior to experiencing the first trip) in the subsequent trial performed after experiencing a trip (no trip trial) there was a significant increase in minimum toe-height (foot clearance at mid-swing) and muscle activation (co-contraction) in hamstrings, quadriceps and tibialis anterior muscles (Pijnappels *et al.,* 2001, 2006). Then, after several unperturbed trials, spatial parameters and muscle activity decreased although not to the level of pre-test values (Pijnappels *et al.,* 2001, 2006). Perhaps surprisingly, adaptations in spatial parameters and / or muscle activity after several unperturbed trials have not been reported in slip research. Indeed, the analysis of previously published slip data did not focus on trends after several ‘no slip’ trials (Heiden *et al.* 2006), and instead only analysed individual trials immediately before or after participants experienced a low friction surface. Nevertheless, it is highly unlikely that the trends observed by Pijnappels *et al.* (2001, 2006) after several ‘no trip’ trials would be observed after several ‘no slip’ trials due to the increased risk of potential injury when experiencing a slip compared to a trip. The implication is that task demand likely influences the effect of prior awareness and / or experience, such that locomotor tasks with a potentially lower risk of injury will be less influenced by awareness and / or experience.

The aims of this experimental chapter was to determine: 1) if awareness and / or prior experience concerning a potential visual occlusion during step descent affects the role of online vision in regulating landing control; and 2) whether the probability of visual occlusion during step descent affects the role of online vision in regulating landing control. The first of two experiments was designed to satisfy the first aim and in addition determine whether a low probability of visual occlusion during step descent causes participants to reduce the role of online vision in regulating landing control, compared to the situation in which there is zero probability of visual occlusion. The second experiment was designed to determine whether a high probability of visual occlusion during step descent affects the role of online vision in regulating landing control, compared to the situation in which there is zero probability of visual occlusion. The role of awareness and / or prior experience concerning a potential visual occlusion during step descent is not explored in the second experiment. In the present experimental chapter, the probability of visual occlusion was hypothesised to affect the role of online vision in regulating step descent landing control. It was also hypothesised that awareness and / or prior experience concerning a potential visual occlusion during step descent would affect the role of online vision in regulating landing control.

7.2 Experiment 1

Participants initially completed repeated step downs without any awareness or experience pertaining to visual occlusion during step descent. Participants were then given increasing levels of knowledge (awareness followed by experience) of a potential visual occlusion. Visual occlusion occurred from mid-swing onwards during step descent. Each visual occlusion occurred in a 1:5 ratio (perturbation:no perturbation), which resulted in a 16.7 % probability of visual occlusion during each trial (after the first visual occlusion). The following hypotheses were formulated in relation to the first experiment: 1) if awareness of a potential visual occlusion during step descent reduces the role of online vision in regulating landing control, step descents completed under full vision condition when participants are aware of a potential visual occlusion (full vision aware) will be significantly different compared to step descents completed when unaware of a potential visual occlusion (unaware). However, the finding of no significant difference between step descents completed in full vision aware and unaware conditions will highlight little / no effect of prior awareness of a potential visual occlusion on the role of online visual processing: 2) if experience of visual occlusion during step descent affects the role of online vision in regulating landing control, step descents completed under full vision condition after experiencing the first visual occlusion will be significantly different compared to step descents completed without experience of a visual occlusion. No significant differences in step descent landing control between prior experience and no prior experience conditions will highlight little / no effect of prior experience of visual occlusion on the role of online visual processing: 3) finally, if a low probability of visual occlusion during step descent affects the role of online vision in regulating landing control, after the first visual occlusion, when descents are completed with full vision available throughout, there will be significant differences in step descent landings compared to the situation in which there is zero probability of visual occlusion. However, if a low probability of visual occlusion during step descent has no effect on the role of online vision, landing control will remain invariant throughout the experiment.

7.3 Methods

7.3.1 Participants

10 healthy adults (5 male and 5 female), age 22.3 ± 2.2 years (mean ± SD), height 175 ± 7.5 cm and mass 68.72 ± 7.7 kg, were recruited using the same inclusion / exclusion criteria as previously reported in the general methods (see sub section 3.1). The tenets of the Declaration of Helsinki were observed and the experiment gained approval from the local Research Bioethics Committee. Written informed consent was obtained from each participant prior to undertaking the study.

7.3.2 Visual assessment

Binocular visual acuity (VA), binocular contrast sensitivity (CS) and stereoacuity were each assessed using the approach described in the general methods (see sub section 3.2). All participants recorded values within the limits of healthy eyes

(Vale *et al.,* 2008a), with measures of VA, CS, and stereoacuity of - 0.26 ± 0.14 logMAR, 1.95 ± 0.05 log units, and 45.0 ± 16.04 secs of arc respectively.

7.3.3 Protocol

Careful consideration was given to standardise the technical and environmental parts of the protocol. Participants reported individually to the laboratory and were in contact with a single experimenter. They were given exactly the same instructions by the experimenter. Participants undertook repeated step downs from three block heights (see section 3.3.1): 14.6 cm, 21.8 cm and 15 % (± 1 cm) of participant’s height (15%bodyHt). Prior to attaining the start position on the block, participants were required to negotiate a series of ‘stepping stones’ (see section 3.6). On attaining the start position, participants were instructed to initiate step descent under the verbal command of ‘step’. Step downs were performed adhering to the protocol highlighted in section 3.6. Illumination over the stepping area was 400 lux (taken at participant’s chest height) and the luminance of the floor and top surface of the block was 15 and 30 cd/m^2 respectively.

Throughout the experiment, participants wore PLATO goggles (Translucent Technologies, Canada). Details of the goggles can be found in the general methods (sub section 3.3.2). Participants completed step downs with and without visual feedback available online. When the goggles switched to opaque, from approximately 50 % of lead-limb swing time onwards (mid-swing) using the method previously described in sub section 3.3.3, visual feedback was unavailable. Prior to data collection, participants were informed that the reason for wearing the goggles was that the study was ‘interested in the effect of step descent landing control when descending steps of different heights with part of the peripheral visual field occluded’. Information pertaining to the amount of peripheral visual field occluded by the goggles can be found in the general methods (sub section 3.3.2). Participants were unaware that the goggles could be switched from transparent to opaque until informed by the experimenter. Participants undertook 12 pre-test trials. During the initial 9 pre-test trials (pre-test 1), block height changed randomly every 3 trials. After pre-test 1 was completed, participants were informed that ‘the study is now interested in the effects of step descent landing control when vision may or may not be occluded during the descent’. Participants removed the goggles and handed them to the experimenter to be shown the opaque condition, thus at this stage participants only had prior awareness of visual occlusion. Vision was not occluded during the remaining 3 pre-test trials (pre-test 2). Pre-test 2 trials were collected at the high block height (21.8 cm).

During the main test (NB. participants were unaware that there was a pre-test and main test phase) at each block height (14.6 cm, 21.8 cm and 15%bodyHt) vision was randomly occluded (from mid-swing onwards) 3 times (random order) using a 1:5 ratio (perturbation:no perturbation). In an attempt to minimise learning effects from repeated step downs from the same block height whilst being constrained with the practicalities associated with changing the block, block height changed randomly every 6 trials. Visual occlusion was limited to a maximum of two perturbations per 6 trials at one block height and could not be consecutive. In addition, if the perturbation trial occurred in the last trial prior to changing block height, an additional trial (full vision) at the same height was inserted immediately after the perturbation trial. This allowed comparisons between the perturbation and subsequent full vision trial at the same block height. The first visual occlusion always occurred in the first 6 trials (random order), which were collected at the highest block height (15%bodyHt). This was based on the hypothesis that any differences in stepping strategy would be most prominent at the highest block height. To ensure that the first visual occlusion trial (no prior experience) could be compared to equal occlusion trials with prior experience at each block height, 10 perturbation trials were collected in total. The first visual perturbation in the study was labelled “no prior experience occlusion trial” and the subsequent 9 perturbation trials (3 at each block height) were labelled “prior experience occlusion trials”. This resulted in a minimum of 67 (maximum of 76) trials being collected (table 7.1).

Table 7.1. Total number of trials collected for each participant at each step height.

Pre-test 1	
No awareness or experience	
Low step	3
High step	3
15%bodyHt	3
Pre-test 2	3

Awareness, no experience	
High step	
Main test	
Full vision	
Low step	15
High step	15
15%bodyHt	15
No prior experience occlusion trial	
15%bodyHt	1
Prior experience occlusion trial	
Low step	3
High step	3
15%bodyHt	3
Possible inserted trials (full vision)	9
Total number of trials	67 - 76

An 8 camera 3-D motion analysis system (Vicon MX3, Oxford Metrics Ltd) was used to record (at 100 Hz) segmental kinematics as participants stepped down from one level to another. Participants wore shorts, t-shirt and flat soled shoes. Reflective spherical markers were attached and anthropometric measurements taken as described in the methods section (3.3.5). Joint angles were defined as the resultant angle between two adjacent segments. Knee, ankle and head angular displacement data, ground reaction force data (including CP co-ordinates) from each force platform, and the co-ordinate data for whole-body CM, knee, ankle and all foot markers were exported (at 100 Hz) for further analysis.

7.3.4 Data analysis

Visuomotor control of step descent was evaluated by determining prelanding kinematic measures and landing mechanic variables for the initial contact period. Prelanding kinematic measures included kneedrop and time of kneedrop as a percentage swing time (see methods section for further details of kneedrop parameter, 3.5.3). The following measures were also determined for the instants of landing: lead limb knee (θ_{knee}) and ankle (θ_{ank}) joint angular displacement, a/p stepping distance, a/p position of CM within base of support (CM-positioning), a/p, m/l and (downward) vertical CM velocity and the amount of bodyweight supported by the trailing limb (bodyWt sup fz cont). θ_{knee} and θ_{ank} were determined as the change in joint angle at the instant of landing relative to their average position when standing stationary on the upper step. Stepping distance was measured as the a/p distance between the marker of the second metatarsal head on the trail and lead limb at the instance of lead limb contact on the lower level. CM-positioning was measured as a percentage of a/p stepping distance from the trail limb. Instant of landing was defined as the instant when the vertical ground reaction force (GRF) on the lead limb first became greater than 20 N.

The mechanics of initial landing, defined as the period up to lead-limb maximum knee flexion, were evaluated by determining the peak vertical GRF (Fz_{peak}), peak knee (θ_{knee}) and ankle (θ_{ank}) angular displacement and peak angular velocity of the knee (ω_{knee}) and ankle (ω_{ank}) joint. Peak a/p, m/l and (downward) vertical CM velocity and amount of bodyweight still being supported by the trailing limb at the end of initial landing (bodyWt sup landing) were also recorded. Time from MI to lead toe-off (double support, DS), lead limb toe-off to foot contact (single support, SS), foot contact to contra-lateral limb toe-off (weight transfer, WT) and time from contact with the lower level to Fz_{peak} (time to Fz_{peak}) were also evaluated. ω_{knee} and ω_{ank} determined how the lead limb was loaded during the landing phase. MI onset was defined from when the resultant x, y coordinates of the CM and CP first moved greater than 10 mm for 5 consecutive frames from its average location when standing stationary. Lead limb toe-off was defined from when the a/p displacement of the second metatarsal head marker was greater than 3 mm for 5 consecutive frames from its average location when standing stationary. Trail limb toe-off was defined from when the vertical GRF on the upper block first dropped below 20 N.

7.3.5 Statistical analysis

Occluding vision unexpectedly during step descent does not present the same threat to stability as being unexpectedly slipped or tripped. Therefore, in order to ensure that there could be some benefit of being made aware of and / or having experienced the visual perturbation, it was deemed suitable in the current study to only analyse step descent data completed at the highest step height (15%bodyHt).

Processed data were analysed using the most appropriate statistical approach for each comparison. Level of significance was accepted at $p < 0.05$, and post-hoc analyses (where necessary) were performed using Tukey's HSD.

1. A *t*-test for dependent samples compared pre-test 1 trials when descents were completed from 15%bodyHt (no awareness or experience) to main test full vision trials when descents were completed from 15%bodyHt prior to the first visual occlusion (awareness + no experience). NB. each participant experienced a minimum of one full vision awareness + no experience trial at 15%bodyHt prior to the first visual occlusion.

2. A *t*-test for dependent samples compared pre-test 1 trials when descents were completed from 15%bodyHt (no awareness or experience) to main test full vision trials, when descents were completed from 15%bodyHt after experiencing the first visual occlusion (awareness + experience).

3. A x 3 vision condition (full vision (FV) trial immediately before visual occlusion, visual occlusion trial, FV trial immediately after visual occlusion) x 2 test (early, late) ANOVA was used to determine if there were any adaptations in step descent landing control when descents were completed under conditions of visual occlusion compared to vision available throughout, and whether this was modified as a function of the number of trials completed NB. the third level of analysis only included data from the first visual occlusion trial, the FV trial immediately before and after the first visual occlusion (early test condition) and the last visual occlusion trial and

the FV trial immediately before and after the last visual occlusion (late test condition).

7.4 Results

1) Does prior awareness of visual occlusion during step descent affect the role of online vision in regulating landing control?

A *t*-test for dependent samples revealed no significant difference between no awareness or experience trials (pre-test 1) and awareness + no experience trials (main test) for any of the pre or initial landing or temporal parameters measured ($p > 0.05$). The output can be found in appendix 6.

2) Does prior awareness and experience of visual occlusion during step descent affect the role of online vision in regulating landing control?

A *t*-test for dependent samples revealed no significant differences between no awareness or experience trials (pre-test 1) and awareness + experience trials (main test full vision trials) for any of the pre or initial landing parameters measured ($p > 0.05$). The only significant difference was found for SS time ($p < 0.01$), which was reduced when participants had prior awareness + experience of vision being occluded during step descent compared to no awareness or experience. The output can be found in appendix 7.

3) When there is a low probability (16.7 %) of visual occlusion during step descent, does landing control alter when comparing the full vision trial immediately before or after visual occlusion to the occlusion trial, and is this modified as a function of the number of trials completed?

Prelanding kinematics, the mechanics of landing and temporal parameters are summarised below in tables 7.2a, b and c respectively. The output can be found in appendix 8.

7.4.1 Prelanding kinematics

Kneedrop and time of kneedrop were unaffected by vision condition ($p > 0.05$). At the instant of landing, θ_{ank} and θ_{knee}, vertical CM velocity, a/p stepping distance and bodyWt sup fz cont, were also unaffected by vision condition ($p > 0.05$). a/p and m/l CM velocity and CM-positioning were significantly affected by vision condition ($p < 0.04$, table 7.2a). Post hoc analysis revealed that anterior and lateral CM velocity reduced when vision was occluded during step descent compared to both the FV trial immediately before and after occlusion. In addition, the CM was positioned closer to the trail leg when vision was occluded during step descent compared to both the FV trial immediately before and after occlusion. None of the prelanding variables were significantly affected by test condition; there were no significant interactions between conditions.

7.4.2 Landing mechanics

ω_{ank}, vertical CM velocity and Fz_{peak} during initial landing were unaffected by vision or test condition ($p > 0.05$). ω_{knee} and bodyWt sup landing were, however, significantly affected by test condition ($p < 0.05$, table 7.2b). m/l CM velocity and peak θ_{knee} and θ_{ank} were significantly affected by vision condition ($p < 0.01$), whereas a/p CM velocity was significantly affected by vision and test condition ($p < 0.007$). Post hoc analysis revealed that anterior and lateral CM velocity and θ_{knee} and θ_{ank} all reduced when vision was occluded during step descent compared to both the FV trial immediately before and after occlusion. Furthermore, post hoc analysis revealed that anterior CM velocity and ω_{knee} significantly increased in the late compared to the early test condition. bodyWt sup landing was reduced in the late compared to the early test condition. There were no significant interactions between conditions.

7.4.3 Temporal parameters

DS, SS and WT time and time to Fz_{peak} were unaffected by vision condition ($p > 0.05$). Only SS and WT time were significantly affected by test condition ($p < 0.05$, table 7.2c). Post hoc analysis revealed that both SS and WT time were reduced in the late compared to early test condition; there were no significant interactions between conditions.

Table 7.2. a) Prelanding kinematics: Group mean (± 1 SD) kneedrop, time of kneedrop, ankle and knee angle (θ), body weight supported on the trail leg (bodyWt sup), a/p step distance, CM-positioning, a/p, m/l and vertical (downward) CM velocity. b) Landing mechanics: Group mean (± 1 SD), peak knee and ankle angle (θ), peak vertical force (Fz_{peak}), ankle and knee angular velocity (ω), a/p and m/l CM velocity during initial landing and body weight supported on the trail leg (bodyWt sup) at end of initial landing. c) Temporal parameters: Group mean (± 1 SD) double support (DS), single support (SS), weight transfer (WT), time to peak vertical force (time to Fz_{peak}).

		Early Vision			**Late Vision**		**Sig**
	occ -1	**occ**	**occ +1**	**occ -1**	**occ**	**occ +1**	
a) Prelanding kinematics							
Kneedrop (cm)	13.0	13.6	12.7	13.7	14.2	13.9	n/a
	(3.7)	(3.9)	(4.1)	(5.1)	(4.5)	(3.7)	
Time of Kneedrop (% ss)	75	75	75	75	78	77	n/a
	(7)	(8)	(6)	(7)	(8)	(6)	
θ_{ank} (deg)	-31.7	-32.2	-31.8	-33.9	-32.2	-32.3	n/a
	(6.4)	(6.8)	(8.3)	(6.5)	(8.0)	(6.8)	
θ_{knee} (deg)	11.6	9.8	12.1	8.1	8.9	10.4	n/a
	(5.4)	(5.6)	(5.4)	(4.7)	(5.8)	(8.3)	
bodyWt sup (N)	512	521	542	512	535	528	n/a
	(122)	(144)	(108)	(116)	(122)	(122)	
a/p step distance (cm)	389.9	391.5	408.2	410.0	395.8	397.6	n/a
	(33.9)	(39.5)	(39.4)	(55.0)	(38.3)	(47.6)	
CM-	36.6	34.6	35.2	39.1	36.1	36.5	V

positioning (%)	(4.8)	(4.3)	(5.0)	(2.6)	(3.7)	(3.9)	
a/p velocity CM (cm/s)	-454.8	-425.7	-456.2	-491.1	-446.0	-477.6	V
	(54.6)	(67.9)	(63.4)	(61.5)	(74.1)	(71.6)	
m/l velocity CM (cm/s)	-162.9	-138.4	-152.0	-159.8	-141.4	-178.6	V
	(42.0)	(27.9)	(42.6)	(39.7)	(25.9)	(38.1)	
vertical velocity CM (cm/s)	-543.8	-538.6	-559.9	-565.6	-576.5	-558.3	n/a
	(115.2)	(146.7)	(106.9)	(105.1)	(117.9)	(109.5)	
b) Landing mechanics							
Peak θ_{ank} (deg)	-35.8	-33.8	-35.5	-38.5	-35.5	-37.0	V
	(6.0)	(5.2)	(6.6)	(6.8)	(6.1)	(5.5)	
Peak θ_{knee} (deg)	-23.9	-21.7	-23.4	-25.0	-23.6	-24.7	V
	(12.2)	(13.1)	(11.3)	(13.0)	(13.6)	(13.4)	
Fz_{peak} (N)	968	956	918	982	985	1008	n/a
	(189)	(139)	(168)	(115)	(134)	(148)	
ω_{ank} (deg. s^{-1})	264.2	257.6	265.6	287.2	270.5	285.3	n/a
	(44.5)	(52.3)	(53.9)	(57.2)	(54.6)	(57.8)	
ω_{knee} (deg. s^{-1})	109.5	94.1	114.3	130.1	117.9	124.0	T
	(43.0)	(51.7)	(38.9)	(36.1)	(49.0)	(48.7)	
a/p velocity CM (cm/s)	-178.2	-157.8	-179.9	-174.0	-153.8	-190.7	V, T
	(42.2)	(27.9)	(48.1)	(37.5)	(33.3)	(41.3)	
m/l velocity CM (cm/s)	-178.2	-157.8	-179.9	-174.0	-153.8	-190.7	V
	(42.2)	(27.9)	(48.1)	(37.5)	(33.3)	(41.3)	

bodyWt sup (N)	41 (67)	83 (98)	73 (94)	31 (62)	39 (52)	23 (45)	T
c) Temporal							
DS (s)	0.41 (0.15)	0.45 (0.16)	0.43 (0.15)	0.46 (0.05)	0.39 (0.17)	0.42 (0.15)	n/a
SS (s)	0.71 (0.15)	0.67 (0.12)	0.70 (0.14)	0.62 (0.08)	0.66 (0.17)	0.63 (0.14)	T
WT (s)	0.20 (0.07)	0.23 (0.07)	0.22 (0.06)	0.19 (0.05)	0.18 (0.04)	0.19 (0.04)	T
time to Fz_{peak} (s)	0.17 (0.03)	0.16 (0.04)	0.16 (0.03)	0.17 (0.04)	0.15 (0.03)	0.17 (0.03)	n/a

Factors found to be significant are shown by capital letter ($p < 0.05$) for vision (V) and test (T) condition. There were no interactions between factors. Ankle angle (negative) indicates the amount of plantar-flexion and knee angle indicates amount of knee flexion relative to standing. NB. Occ-1 refers to the FV trial immediately prior to visual occlusion and Occ+1 the trial immediately after visual occlusion.

7.5 Discussion

Findings highlight that awareness in the absence of experience pertaining to a potential visual occlusion during step descent had no effect on the role of online vision in regulating landing control. This was evidenced by the finding of no significant difference in landing control when participants were only made aware (no experience) of a potential visual occlusion during the descent compared to no awareness or experience of a potential visual occlusion. Furthermore, step descents completed under conditions of full vision with prior awareness and experience pertaining to a potential visual occlusion during step descent were comparable to those completed with no awareness or experience. The implication is awareness and experience of a potential visual occlusion had minimal affect on the role of online vision in regulating landing control. The only analysed variable that was significantly affected by prior awareness and experience of visual occlusion was SS time ($p < 0.01$), which was reduced compared to no awareness or experience condition. The previous experimental chapter (chapter 6) demonstrated that decreased SS time was indicative of participants exhibiting reduced control when stepping down onto the lower level. However, in the previous chapter, decreased SS time was accompanied with increased vertical CM velocity (at instant of contact and during landing) and increased ω_{ank} during landing. In the present study, since SS time was the only variable significantly affected when step descents were completed under conditions of full vision with prior awareness and experience pertaining to a potential visual occlusion during step descent, it is unlikely that these factors caused participants to reduce control when stepping down onto the lower level. Rather, it is more probable that this result is a chance finding reflecting a type II error (error rate value equals 1.1) due to the high number of variables analysed (Fowler *et al.,* 2002).

The comparison between step descents completed under conditions of awareness and experience and no awareness or experience was also made to determine whether a low probability (16.7 %) of visual occlusion during step descent (full vision awareness and experience trials) compared to zero probability of visual occlusion (no awareness or experience trials) affects the role of online vision in regulating landing control. As discussed above, landing control remained invariant (only SS time was affected) across these conditions. It is relevant to note that occluding online vision during step descent compared to full field vision available throughout resulted in subtle adaptations in landing control. This highlights that participants planned to use online vision (when available) during step descent to regulate landing control. Indeed, the third level of analysis revealed that when vision was occluded during step descent, compared to the FV trial immediately before and after visual occlusion, participants reduced lateral CM velocity at the instant of and during initial landing, reduced anterior CM velocity and ensured the CM was positioned closer to the support limb at the instant of landing and reduced peak θ_{knee} and θ_{ank} during initial landing. These adaptations in landing control under conditions of visual occlusion provide further support for the subtle role of online vision 'fine tuning' step descent landing control (see chapters 4 and 5). Findings from experiment 1 therefore suggest that when there is a low probability of visual occlusion during step descent and descents are completed under conditions of full vision, participants plan to use online vision in the same manner compared to their habitual stepping response (i.e. where there is zero probability of visual occlusion during the descent) to subtly 'fine tune' landing control. It is important to emphasise that the awareness and experience condition comprised of an average of all full vision awareness and experience trials from the highest step height (15%bodyHt). Since the present experiment used a 1:5 ratio (perturbation:no perturbation), it is possible that averaging all full vision awareness and experience trials might have masked any subtle changes in step descent landing control in the trial immediately after visual occlusion. This effect would be consistent with previous research which has shown that while spatial parameters are significantly affected immediately after participants experience a trip, there is a return towards pre-test conditions after several unperturbed trials (Pijnappels *et al.,* 2006). The third level of analysis was designed to address this issue and highlighted that landing control remained invariant between the FV trial immediately before and after visual occlusion. The implication of this finding is that when there is a low probability of visual occlusion during the descent and participants complete step descents under conditions of full vision, participants plan to use online vision during the descent to regulate landing control in the same manner as their habitual stepping response. Of note, output from the third level of analysis concerning the effects of test condition is discussed in the general discussion.

7.6 Experiment 2

Findings from the previous experiment suggest that a low probability (16.7 %) of visual occlusion during step descent has little / no effect on the role of online vision in regulating step descent landings. However, the previous experimental chapters in this thesis, and indeed previously published research, (Buckley *et al.,* 2008; Cowie *et al.,* 2008), used a higher probability (67 %) of occluding vision during step descent. This second experiment therefore determines whether a high probability (67 %) of visual occlusion during step descent affects the role of online vision in regulating landing control, compared to the situation in which there is zero probability of visual occlusion. The question that this experiment was designed to address: 1) when there is a high probability (67 %) of visual occlusion during step descent, does the role of online vision in regulating landing control change when descents are completed with full vision available throughout compared to zero probability of visual occlusion during the descent? If a high probability of visual occlusion during step descent causes participants to reduce the role of online vision in regulating landing control when descents are completed with full vision available throughout, there will be significant differences in landing control compared to the situation in which there is zero probability of visual occlusion. However, if a high probability of visual occlusion during step descent has no effect on the role of online vision, landing control will remain invariant throughout the study. Findings reported here are part of data collected in chapter 4 (sub section 4.2.3).

7.8 Method

Information pertaining to the participants, visual assessments, protocol and dependent variables can be found in chapter 4. However, to explain briefly, in chapter 4, participants undertook repeated step downs from three block heights (14.6 cm, 21.8 cm and 15%bodyHt). Step downs were completed with and without visual feedback available online, and with and without visual field being restricted. Step height (14.6 cm, 21.8 cm and 15%bodyHt) changed randomly every 9 trials. Within the 9 trials, each vision condition (available throughout, available up to toe-off, available up to mid-swing) was completed three times in random order. This procedure was completed in both visual field (full, restricted) conditions (random order). Only step descents completed from the highest block height (15%bodyHt) under conditions of full visual field and vision available throughout were used in the present experiment to comprise the high probability (67 % visual occlusion) condition. Only step descents completed from the highest block height (15%bodyHt) were analysed to increase the likelihood of observing any adaptations in stepping strategy. This resulted in a total of 3 trials being analysed per participant (n = 10) in the high probability condition.

Step descents were also completed with vision unperturbed and with the prior instruction that vision would not be occluded during the descent (zero probability condition). Step height changed randomly every three trials, with three trials collected at each height (14.6 cm, 21.8 cm and 15%bodyHt), for a total of 9 trials. The zero probability condition was randomly inserted within the study. Only step descents completed from the highest block height (15%bodyHt) were analysed to ensure a fair statistical comparison between the high probability condition and zero probability condition. This resulted in a total of 3 trials being analysed per participant (n = 10) in the zero probability condition.

7.8.1 Statistical analysis

t-test for dependant samples compared full vision trials in which there was a high probability (67 %) of visual occlusion during the descent to trials in which there was zero probability of visual occlusion. Level of significance was accepted at $p < 0.05$.

7.9 Results

Of the 20 *t*-tests conducted, there was a significant difference in time of kneedrop and SS time ($p < 0.03$). Compared to descents completed when there was zero probability of visual occlusion during the descent, a high probability resulted in reduced SS time and increased time of kneedrop. The output can be found in appendix 9. The prelanding kinematics, mechanics of landing and temporal parameters are summarised below in table 7.3a, b and c respectively.

Table 7.3. a) Prelanding kinematics: Group mean (± 1 SD) kneedrop and time of kneedrop, ankle and knee angle (θ), body weight supported on the trail leg (bodyWt sup), a/p step distance, CM-positioning, a/p and vertical (downward) CM velocity, b) Landing mechanics: Group mean (± 1 SD) lower extremity stiffness (l.x.s), peak knee and ankle angle (θ), peak vertical force (Fz_{peak}) and ankle and knee angular velocity (ω) during initial landing and body weight supported on the trail leg (bodyWt sup) at the end of initial landing, c) Temporal parameters: Group mean (± 1 SD) double support (DS), single support (SS), weight transfer (WT), time to peak vertical force (time to Fz_{peak}).

	Test		
	Zero probability	**High probability**	**Sig**
a) Prelanding kinematics			
Kneedrop (cm)	14.6 (1.7)	15.2 (1.9)	n/a
% kneedrop (% ss)	85 (3)	87 (2)	T
θ_{ank} (deg)	-29.7 (5.7)	-30.0 (6.0)	n/a
θ_{knee} (deg)	11.0 (5.7)	9.2 (5.8)	n/a
BodyWt sup (N)	511 (171)	515 (191)	n/a
a/p step distance (cm)	400.7 (35.5)	409.3 (35.0)	n/a
CM-positioning (%)	46.1	47.0	n/a

	(3.8)	(4.2)	
	-472.1	-491.1	
a/p velocity CM (cm/s)	(57.5)	(58.2)	n/a
	-589.8	-601.2	
vertical CM velocity (cm/s)	(109.3)	(94.7)	n/a
b) Landing mechanics			
	36	29	
BodyWt sup (N)	(45)	(33)	n/a
	8.4	8.1	
l.x.s (Kn m^2)	(4.6)	(4.5)	n/a
	7.0	6.9	
Peak θ_{ank} (deg)	(4.8)	(4.3)	n/a
	29.4	27.9	
Peak θ_{knee} (deg)	(7.7)	(7.0)	n/a
	1048	1030	
Fz_{peak} (N)	(252)	(199)	n/a
	274.9	278.9	
ω_{ank} (deg. s^{-1})	(49.1)	(49.2)	n/a
	134.2	136.2	
ω_{knee} (deg. s^{-1})	(48.3)	(39.7)	n/a
c) Temporal Parameters			
	0.41	0.41	
DS (s)	(0.08)	(0.08)	n/a
	0.68	0.63	
SS (s)	(0.11)	(0.08)	T

	0.21	0.20	
WT (s)	(0.06)	(0.05)	n/a
	0.13	0.14	
time to Fz_{peak} (s)	(0.04)	(0.05)	n/a

Factors found to be significant are shown by capital letter ($p < 0.05$). There were no interactions between factors. Ankle angle (negative) indicates the amount of plantar-flexion and knee angle indicates amount of knee flexion relative to standing.

7.10 Discussion

Findings highlight that step descents completed under conditions of full vision but with a high probability (67 %) of visual occlusion during the descent, were controlled in a manner similar to step descents completed with zero probability of visual occlusion. SS time and time of kneedrop were the only variables to be significantly affected by a high probability of visual occlusion during step descent (table 7.3). Compared to zero probability of visual occlusion, when there was a high probability of visual occlusion SS time reduced and time of kneedrop increased. Since time of kneedrop variable is calculated as a percentage SS (swing) time, any significant changes in SS time would likely be reflected in changes in time of kneedrop (as reported in chapter 4). Thus, with SS time being significantly reduced when there was a high probability of visual occlusion during the descent, this caused time of kneedrop to occur later in the descent. However, as discussed in experiment 1, it remains unclear why changes in SS time were not reflected in subsequent changes in landing kinematics. Indeed, even subtle increases in ankle plantar-flexion or decreases in knee flexion at the instant of landing would account for significant differences in SS time, although this was not observed (*p* > 0.1, see appendix 9). Overall, then, the findings suggest that when there is a high probability of visual occlusion during step descent and descents are completed under conditions of full vision, participants plan to use online vision during the descent to control landing in the same manner as their habitual stepping response. The implication of this finding is that step descent paradigms which use a high probability of occluding online vision during the descent has little / no effect on how online vision is used to control landing in subsequent full vision trials.

7.11 General Discussion

The present experimental chapter sort to determine how awareness and experience concerning a potential visual occlusion during step descent affected the role of online vision in regulating landing control. In addition, this experimental chapter also determined whether the probability of visual occlusion during step descent affected the role of online vision in regulating landing control. Findings highlight that prior awareness only, or prior awareness and experience of a potential visual occlusion during step descent has little / no effect on the role of online vision in regulating landing control (see experiment 1). Furthermore, findings highlight that irrespective of whether there is a high or low probability of visual occlusion during descent, when descents are completed under conditions of full vision, compared to zero probability of visual occlusion, the role of online vision in controlling step descent landing remains invariant (see experiment 1 and 2). This suggests that participants plan to use online vision during step descent to control landing irrespective of the probability of visual occlusion. Findings also provide additional support that online vision is only used in the latter portion of step descent to subtly ‘fine tune’ landings. Interpretations of these results are discussed below.

Providing participants with prior awareness and experience pertaining to a potential visual occlusion during step descent has little / no effect on the role of online vision in regulating landing control (experiment 1). These results are contrary to previous research investigating the effects of increasing levels of awareness and / or experience of a potential slip / trip on participant’s spatial parameters and muscle activity (Pijnappels *et al.,* 2001, 2006; Heiden *et al.,* 2006; Siegmund *et al.,* 2006). For example, prior awareness and / or experience of a potential trip caused participants to alter their gait to reduce the likelihood of falling through adapting several spatial and muscle parameters. These adaptations included increasing step width, ankle dorsi-flexion throughout the entire swing phase, which subsequently increased minimum toe height from the floor and increased lower limb muscle activation (Pijnappels *et al.,* 2001, 2006). Furthermore, prior awareness and / or experience of a potentially low friction surface caused participants to reduce the chance of slipping though adopting a flatter foot angle with the floor and reduced knee angle at heel strike and increased lower limb muscle activity (Heiden *et al.,* 2006). One possible explanation for the differences in findings between the present study and previous slip / trip research may be attributed to the demands of the task. For example, Pijnappels *et al.* (2001, 2006) demonstrated that after participants experience a trip, in the subsequent trial (no trip trial) kinematic parameters and muscle activity significantly increase compared to pre-test values (i.e. trials performed prior to experiencing the first trip). Then, after several unperturbed trials, spatial parameters and muscle activity decrease although not to the level of pre-test values (Pijnappels *et al.,* 2001, 2006). In the present study, despite participants descending a step height which reached the upper limit whereby the descent could be controlled (15%bodyHt - further details see methods section 3.3.1), the trends observed by Pijnappels *et al.* (2001, 2006) after several 'no trip' trials were not observed after several 'no visual occlusion trials' (i.e. FV trials). This was evidenced by no significant differences in stepping strategy in the FV trial immediately before and after the first or last visual occlusion (table 7.2). The implication of these findings is that a potential visual occlusion during step descent does not present such a threat to stability compared to experiencing a potential slip / trip. It is therefore possible that task demand influences the effect of prior awareness and / or experience, and locomotor tasks with a reduced threat to stability will be less influenced by awareness and / or experience.

The present experimental chapter also investigated whether a high (67 %) or low (16.7 %) probability of visual occlusion during step descent affected the role of online vision in regulating landing control. Irrespective of whether there was a high (67 %, experiment 2) or low (16.7 %, experiment 1) probability of visual occlusion, step descents completed under conditions of full vision did not differ compared to when there was zero probability of visual occlusion. These findings are inconsistent with previous research investigating control of single upper-limb movements, which have demonstrated that when participants are uncertain whether online vision will be available during the subsequent trial, they plan for the ‘worst-case scenario’ (Zelaznik *et al.,* 1983) and adopt a more feedforward mode of control (Elliott *et al.,* 1999). The inconsistency between studies may be attributed to the role of online vision in controlling the movement in each task. For example, the aforementioned upper limb studies focused on rapid manual aiming movements, which appear to be predominantly biased towards online visual processing. This has been evidenced through reduced accuracy and end-point constancy when online vision is unavailable during the aiming movement (for a review, see Elliott *et al.,* 1999). A similar role of online vision has also been evidenced during a precision stepping task; the accuracy of foot placement is reduced when online vision is occluded at the point of foot-off (Reynolds and Day, 2005). In contrast, visuomotor control of step descent appears predominantly biased towards feedforward visual processing (chapters 4 and 5). Indeed online vision is only used in the latter portion of the descent to subtly ‘fine tune’ landing (chapters 4 and 5). In the present experimental chapter, it is possible that participants planned to use online vision during the descent irrespective of the probability of visual occlusion, as the consequence of visual occlusion during the descent presented little threat to stability and thus the ability to precisely regulate the movement. However, during precision movement tasks, as the role of online vision is increased, the consequence of planning to use online vision during the movement and it being unavailable will have significant implications on the ability to precisely regulate the movement. The implication of this finding is that step descent paradigms which use a high probability of occluding online vision during the descent has little / no effect on how online vision is used to control landing in subsequent full vision trials. Future work is required to investigate whether the probability of visual occlusion affects how participants plan to use online vision to control the movement when the role of visuomotor control is predominantly biased towards online visual processing i.e. during a precision stepping task. In a precision stepping task one would expect a high probability of visual occlusion during the movement to have a significant effect on movement control compared to low or zero probability of visual occlusion.

Compared to when vision was available throughout, in experiment 1, occluding vision from mid-swing onwards during step descent caused very few differences in landing control. The only significant differences in landing control were evidenced through reduced anterior and lateral CM velocity at the instant of and during initial landing, the CM being positioned closer to the support leg (CM-positioning) and reduced peak θ_{knee} and θ_{ank} angle during initial landing (table 7.2). Such adaptations in landing control are reflective of a more cautious stepping strategy. Indeed ensuring the CM remains closer to the support limb and reducing anterior and lateral CM velocity at the instant of and during initial landing is likely a strategy to ensure that dynamic stability is maintained well within the margins of safety (Hof *et al.,* 2005). However, the adaptations in stepping strategy when vision was occluded from mid-swing onwards did not result in participants fundamentally altering their stepping strategy i.e. there was no increase in body weight supported on the trail leg or reduced vertical CM velocity at the instant of or during initial landing or reduced Fz_{peak}. Such fundamental adaptations in stepping strategy have been reported when vision is occluded immediately prior to step descent (Buckley *et al.,* 2008). The present study's finding of subtle adaptations in landing control when vision was occluded from mid-swing onwards during the descent are consistent with those from chapters 4 and 5, indicating how online vision is used in the latter portion of step descent to 'fine tune' landing control.

In experiment 1, there were subtle adaptations in step descent landing control when comparing early and late test conditions. Compared to the early test condition, in the late condition ω_{knee} and anterior CM velocity during initial landing increased and SS time, WT time and bodyWt sup landing reduced (table 7.2). Such adaptations in landing control may suggest that participants were becoming familiar to the task and were thus exhibiting learning effects. If indeed participants were becoming familiar with the task, the role of online vision in regulating landing control would likely be reduced as participants would be able to effectively rely on somatosensory and / or proprioceptive feedback received from previous step downs. Reduced reliance on visual feedback to regulate stepping strategy as a function of repetition has been similarly evidenced when participants negotiated a raised surface under conditions of monocular and binocular refractive blur (Heasley *et al.,* 2004; Vale *et al.,* 2008b). In experiment 1, any change in the role of online vision in regulating step descent landing control across the study would be evidenced through significant vision-by-test interactions. Indeed the adaptations in stepping strategy in the first visual occlusion trial when compared to the FV trial immediately before and after the first visual occlusion, would be significantly different to the last visual occlusion trial compared to the FV trial immediately before and after the last visual occlusion. There was no significant vision-by-test interactions in any of the dependent variables analysed. This demonstrates that the subtle adaptations in step descent landing control across early and late test conditions were not attributed to participants reducing reliance on online vision as they became familiar to the task. It is possible that the adaptations in step descent landing control across early and late test conditions are attributed to fatigue, which subsequently resulted in participants exhibiting reduced control during step descent. This was evidenced through reduced SS time, which resulted with participants landing with increased ω_{knee} and anterior CM velocity during initial landing and reduced bodyWt sup landing. However, with no such increases in vertical CM velocity and Fz_{peak}, this suggests that the effects of fatigue were very subtle as participants were still able to maintain a high level of control during the descent.

7.12 Conclusion

The present experimental chapter sought to determine how awareness and experience regarding a potential visual occlusion during step descent affected the role of online vision in regulating landing control. In addition, the chapter also determined whether a high (67 %) or low (16.7 %) probability of visual occlusion during step descent affected the role of online vision in regulating landing control. Findings demonstrate that providing participants with prior awareness only, or prior awareness and experience pertaining to a potential visual occlusion during step descent has little / no effect on the role of online vision in regulating landing control (experiment 1). Whilst these findings are inconsistent with previous slip / trip research investigating the effects of awareness and / or experience, such differences are possibly attributed to the demands of the task. This likely suggests that locomotor tasks with a reduced threat to stability are less influenced by awareness and / or experience. Findings also highlight that participants plan to use online vision during step descent to control landing irrespective of the probability of visual occlusion. Since visuomotor control of step descent appears predominantly regulated through feedforward visual processing, it is possible that participants planned to use online vision during step descent irrespective of the probability of vision being available, as the consequence of visual occlusion during the descent presented little threat to stability and thus the ability to precisely regulate the movement. The implication of this finding is that step descent paradigms which use a high probability of occluding online vision during the descent has little / no effect on how online vision is used to control landing in subsequent full vision trials. The present study's findings are also consistent with those from chapters 4 and 5, indicating how online vision is used in the latter portion of step descent to 'fine tune' landing control.

Chapter 8

Experiment 5

When is lower visual field information acquired to control landing when descending a step during on-going gait?

8.1 Introduction

During normal everyday locomotion we frequently have to negotiate obstacles, steps / stairs and / or multi-surface terrain. Locomotion during such adaptive gait tasks is predominantly regulated through visual information obtained through feedforward processes (i.e. visual information acquired in advance of the on-going movement). Studies undertaken to determine when vision is used for successful obstacle crossing indicate that, despite vision being occluded from up to 4 steps before the obstacle and remaining occluded until both lead and trail limbs have crossed the obstacle, successful obstacle negotiation is still possible (Patla, 1998, experiment 1; Mohagheghi *et al.,* 2004). In contrast when visual information is occluded from five or more steps before the obstacle, crossing success is significantly reduced (Patla and Greig, 2006). This aforementioned research (Patla, 1998, experiment 1; Mohagheghi *et al.,* 2004; Patla and Greig, 2006) suggests that occluding vision in advance of two strides (4 steps) before an obstacle significantly impacts the nervous system's ability to utilise feedforward visual cues to integrate an efference copy of the motor command to plan the general stepping pattern during obstacle crossing (Marigold, 2008). A similar use of feedforward visual processing has been reported in tasks that involve step descent from a stationary standing starting position. For example, when visual information is occluded in the few seconds prior to step descent, participants are unable to scale kneedrop parameters to step height (Cowie *et al.,* 2008), land with reduced anticipatory lower limb muscle activity (Craik *et al.,* 1983), and reduced force and lower extremity stiffness (Buckley *et al.,* 2008). Such landings are observed as a consequence of participants increasing reliance on the stance limb and using the lead limb to 'probe' for the ground and not fully committing to transferring weight to the lead limb until somatosensory feedback confirms that contact has been made with the lower level (Buckley *et al.,* 2008; Craik *et al.,* 1983).

Despite being able to successfully complete a number of adaptive gait tasks when only feedforward visual information is available, under such conditions the ability to 'fine tune' lower-limb movements is significantly reduced. For example, under conditions of lvf occlusion during obstacle negotiation, when the lower limb(s) and obstacle are occluded online from 2 or 1 step(s) prior to crossing (whilst uvf cues for feedforward control are undisrupted), the ability to 'fine tune' lower-limb trajectory is significantly affected (Patla*,* 1998, experiment 2; Rietdyk and Rhea, 2006; Rhea and Rietdyk, 2007; Graci *et al.,* 2010). This is evidenced through significant increases in lead foot horizontal distance before the obstacle and toe clearance over it (Patla*,* 1998, experiment 2; Rietdyk and Rhea, 2006; Rhea and Rietdyk, 2007; Graci *et al.,* 2010). The inability to 'fine tune' stepping strategy under conditions of lvf occlusion has also been reported in tasks that involve negotiation of multi-surface terrain (Marigold and Patla, 2008) and step descent from a stationary standing starting position (see chapter 4). Under conditions of lvf occlusion, such that the lower limb and the immediate ground area ~2 steps ahead are unavailable during negotiation of multi-surface terrain, participants alter their gait pattern by reducing gait speed and step length, to allow for more precise control over foot placement on such terrain (Marigold and Patla, 2008). When the lvf is occluded prior to step descent from a stationary standing starting position, such that the view of the lower leg and immediate lower floor area is occluded prior to and during the descent, participants adopt a stepping strategy consistent with being uncertain regarding precise floor height (chapter 4). Only subtle adaptations in stepping strategy are evident if full or uvf (lvf already obstructed) is occluded from mid-swing onwards during the descent (chapter 4). This suggests that the contribution of lvf information to landing control occurs predominantly prior to step descent movement initiation (see chapter 4 and 5).

Presently, it remains unclear whether lvf cues sampled 2 or 1 step(s) prior to descending a step during on-going gait contribute to landing control. To this end, the present study occluded the lvf from the instant of heel strike 2 steps (penultimate) or 1 step (final) prior to descending a step during on-going gait. Such lvf occlusions were compared to descents completed with full field vision available throughout (see figure 8.1). If lvf cues contribute to landing control in the penultimate step prior to step descent, occluding lvf from 2 steps prior to descent will have a significant effect on landing control. However, if lvf cues contribute to landing control in the final step prior to step descent, occluding lvf 1 step prior to descent will have a significant effect on landing control. Finally, if occluding lvf from either 2 or 1 step(s) prior to step descent has no effect on landing control, then this will indicate that lvf cues acquired in advance of the penultimate step prior to step descent are used to control landing.

8.2 Methods

8.2.1 Participants

12 healthy adults (6 male and 6 female), age 22 ± 2.5 years (mean ± SD), height 175.7 ± 8.5 cm and mass 68.2 ± 8.1 kg, were recruited using the same inclusion / exclusion criteria as described in the general methods (see sub section 3.1).The tenets of the Declaration of Helsinki were observed and the experiment gained approval from the local Research Bioethics Committee. Written informed consent was obtained from each participant prior to undertaking the study.

8.2.2 Visual assessment

Binocular visual acuity (VA), binocular contrast sensitivity (CS) and stereoacuity were each assessed using the approach described in the general methods (see sub section 3.2). All participants recorded values within the limits of healthy eyes

(Vale *et al.,* 2008a), with measures of VA, CS, and stereoacuity of - 0.23 ± 0.06 logMAR, 1.95 ± 0.02 log units, and 39 ± 14.5 secs of arc respectively.

8.2.3 Protocol

Participants walked along a 146 mm raised surface before stepping down onto the floor level and continued to walk for approximately 4 to 5 steps along the laboratory floor (see figure 8.1). Start position, from the upper level, was randomly varied by between 4 or 5 steps to ensure participants did not adopt a repeated motor strategy to negotiate the step. A force-platform mounted in the floor collected ground reaction force (GRF) data (at 100 Hz) for the step down onto the floor. The raised surface was constructed from plywood and covered in the same green vinyl as the surrounding floor. The laboratory was well lit with ambient illuminance of 400 lux measured at eye level.

A force sensitive resistor (FSR, Delsys, Boston, USA) was attached on the sole of each participant's footwear, 1 cm anterior and 1 cm lateral of the midpoint of the shoe's posterior border. An additional FSR was attached to the sole of the right foot 1 cm distal of the metatarsophalangeal joint line at the midline of the foot. Participants wore safety goggles (Protector Safety, England) with a translucent (LCD) sheet (manufacturer unknown) attached across the lower half of the goggles (further description of the LCD sheet in provided in the methods section 3.3.3). Signals from the FSRs were fed to a control box which was used to switch the LCD sheet from transparent to opaque (instantaneously occluding the lvf) from either heel strike 2 or 1 step(s) from the step descent and then from opaque to transparent at lead (right) foot (toe) contact with the lower level (see figure 8.1). Participants were required to descend the step leading with their right leg. Any trial that was not completed according to these instructions was discarded and repeated.

Visual perturbations (lvf occluded from either 2 or 1 step(s) prior to step descent) were randomly presented with a 1:5 ratio (perturbation:no perturbation). In addition, to ensure participants used visual information to determine the height of the lower level rather than using somatosensory feedback from previous trials, a number of ‘dummy trials’ were also completed. These involved increasing the height of the raised surface by + 15 mm (to give a height of 161 mm) every third trial (the first dummy trial was completed after the first (real) trial to ensure that the last (real) trial was not a dummy trial). No data were collected during dummy trials and participants were advised that the height of the raised surface would be varied throughout the study. lvf occlusion trials (occlusion from either 2 or 1 step(s) prior to step descent) were repeated 3 times, and thus, with the inclusion of 6 ‘dummy trials’, participants completed a total of 36 trials.

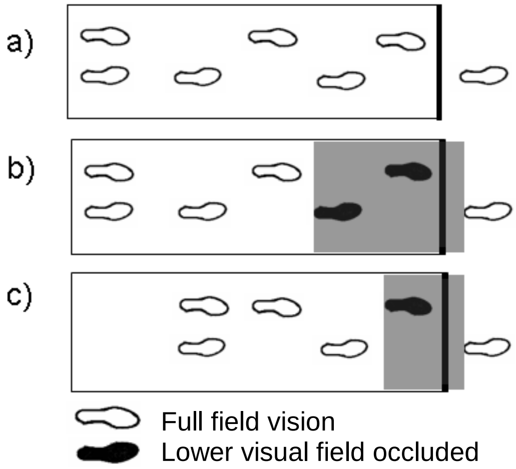

Figure 8.1. Illustration of all testing conditions a) start position 5 steps away from step descent with no visual occlusion, b) start position 5 steps away from step descent with lower visual field occlusion 2 steps prior to descent c) start position 4 steps away from step descent with lower visual field occlusion 1 step prior to descent.

Kinematic data were collected (at 100 Hz) using an 8 camera 3-D motion analysis system (Vicon MX3, Oxford Metrics Ltd). Further details of the laboratory set up, including the camera system can be found in the general methods (3.3.5). Data were collected during a single testing session for each participant, with adequate rest periods provided to prevent fatigue. Participants wore shorts, t-shirt and flat soled shoes. Reflective spherical markers were attached and anthropometric measurements taken as described in the general methods (3.3.5). Knee, ankle and head angular displacement data, ground reaction force data (including CP co-ordinates) from the force platform, and the co-ordinate data for whole-body CM, knee, ankle and all foot markers and the markers placed on the front edge of the raised surface were exported (at 100 Hz) for further analysis.

8.2.4 Data analysis

The role of lvf cues during adaptive gait was evaluated by determining changes in kinematic / gait measures, and landing mechanic variables for the initial contact period following lvf occlusion. Gait measures for either the final step before the step edge (final step) or for the step onto the lower level (step down) included, peak CM (a/p) velocity and minimum head pitch angle, trail-foot placement distance (a/p direction) from the step edge, lead-foot minimum horizontal and vertical heel clearance over the step edge and kneedrop, and time of kneedrop as a percentage swing time (see methods section for further details of kneedrop parameter, 3.5.3). Minimum head pitch angle was measured to determine if participants flexed their head prior to or during descent, in an attempt to receive visual information from the lower limb and / or lower surrounding floor area when the lvf was occluded. Minimum horizontal and vertical heel clearance over the step edge was defined as the minimum distance between the heel marker and the apex of the step as it crossed the vertical and horizontal position of the apex of the step respectively; measured to understand how lvf occlusion affects the ability to 'fine tune' lower limb trajectory. The following gait / kinematic measures were also determined for the instant of landing: lead limb knee (θ_{knee}) and ankle (θ_{ank}) joint angular displacement, a/p, m/l and (downward) vertical CM velocity and lead-foot placement distance (a/p direction) from edge of step at contact with the lower level. θ_{knee} and θ_{ank} joint angular displacements were determined as the change in joint angle at the instant of landing relative to their average angle when standing stationary on the upper level. The distance that the foot landed beyond the step edge was measured as the a/p distance between the marker on the lead limb heel and the leading edge of the upper level. Instant of landing was defined as the instant when the vertical GRF on the lead limb first became greater than 20 N.

The mechanics of initial landing, defined as the period up to maximum knee flexion, were evaluated by determining the peak vertical ground reaction force (Fz_{peak}), peak knee (θ_{knee}) and ankle (θ_{ank}) joint angular displacements, peak angular velocity at the knee (ω_{knee}) and ankle (ω_{ank}) joint and peak a/p, m/l and (downward) vertical CM velocity. ω_{knee} and ω_{ank} were calculated to determine how the lead limb was loaded during the initial landing phase. Lead limb toe-off to ipsilateral foot contact (single support, SS), foot contact to contra-lateral limb toe-off (weight transfer, WT) and time to peak ground reaction force (time to Fz_{peak}) were also evaluated. Lead limb toe-off was defined as the instant when the a/p velocity of the second metatarsal head marker first increased above 150 mm/s and trail limb toe-off was calculated using the same criteria.

8.2.5 Statistical analysis

To determine whether occluding lvf cues from either 2 or 1 step(s) prior to step descent had a significant affect on landing control, the following comparisons were undertaken:

1. lvf occluded from 2 steps prior to step descent, compared to full field vision available throughout.

2. lvf occluded from 1 step prior to step descent, compared to full field vision available throughout.

Data were analysed with a generalized estimating equation, random effects, population-averaged model (Stata ver. 8.0; Stat Corp., College Station, TX). This multivariate statistical model was obtained using the ‘xtreg’ command that uses the generalized least squares (GLS) random-effects estimator, to produce a matrix-weighted average of the between-subjects and within-subject output. Given the experimental design an exchangeable correlation structure was judged to be appropriate, and due to the exploratory nature of the study no type I error adjustment of the alpha level was deemed necessary. Thus, level of significance was set at $p < 0.05$.

8.3 Results

Compared to full field vision available throughout, there was no effect of occluding lvf from 1 step prior to descent in any dependant variable ($p > 0.09$).

8.3.1 Gait / kinematic measures

Compared to full field vision available throughout, occluding lvf from 2 steps prior to descent led to the following changes; mean lead-foot vertical heel clearance significantly increased $z = 4.11$, $p < 0.001$, as did horizontal heel clearance $z = 2.70$, $p < 0.007$. There was approximately 14 % (~7 mm) and 11 % (~17 mm) increase in vertical and horizontal heel clearance respectively when lvf was occluded from 2 steps prior to descent (table 8.1). Kneedrop significantly decreased $z = -2.16$, $p < 0.03$ and timing of kneedrop occurred significantly earlier $z = -2.82$, $p < 0.005$ (figure 8.2) when lvf was occluded from 2 steps prior to descent. Trail-foot placement prior to step descent, minimum head flexion and walking velocity at the final step or step down were unaffected when lvf was occluded from 2 steps prior to descent ($p > 0.62$).

When the lvf was occluded from 2 steps prior to descent, compared to full field vision available throughout, at the instant of landing θ_{knee} significantly increased $z = 2.17$, $p < 0.03$ and lateral CM velocity significantly decreased $z = -3.24$, $p < 0.001$ (table 8.1). θ_{ank}, lead-foot placement, vertical and a/p CM velocity were unaffected when lvf was occluded from 2 steps prior to descent ($p > 0.17$).

8.3.2 Landing mechanics

When the lvf was occluded from 2 steps prior to descent, compared to full field vision available throughout, during initial landing peak θ_{knee} and θ_{ank} significantly increased; $z = 2.40$, $p > 0.02$ and $z = 2.17$, $p < 0.03$ respectively (figure 8.3). Lateral CM velocity significantly decreased when lvf was occluded from 2 steps prior to descent $z = -2.18$, $p < 0.03$ (table 8.2). ω_{ank}, ω_{knee}, $Fz_{peak,}$ vertical and a/p CM velocity were unaffected when lvf was occluded from 2 steps prior to descent ($p > 0.13$).

8.3.3 Temporal parameters

When the lvf was occluded from 2 steps prior to descent, compared to full field vision available throughout, WT time significantly increased z = 2.50, $p < 0.01$ (table 8.3). SS time and time to Fz_{peak} were unaffected when lvf was occluded from 2 steps prior to descent ($p > 0.70$).

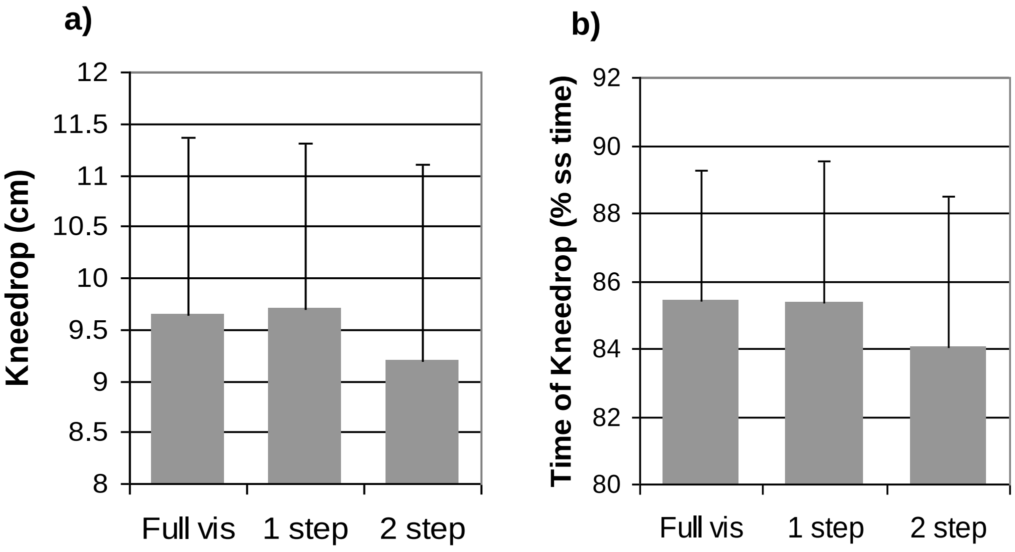

Figure 8.2. Group mean (SD) a) kneedrop distance and b) time kneedrop in full vision (Full vis), lvf occluded from 1 step prior to stepping down (1 step) and lvf occluded from 2 step prior to stepping down (2 step). See text for statistical comparisons.

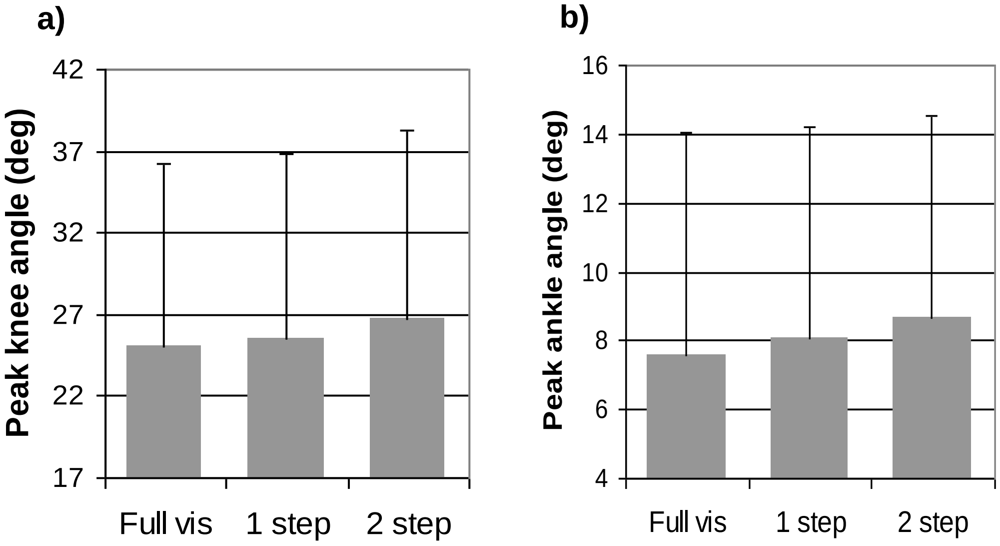

Figure 8.3. Group mean (SD) a) peak knee and b) ankle angle during initial landing in full vision (Full vis), lvf occluded from 1 step prior to stepping down (1 step) and lvf occluded from 2 step prior to stepping down (2 step). See text for statistical comparisons. NB, the zero reference ankle and knee angles were those determined for stationary standing.

Table 8.1. Prelanding kinematics: Group mean (± 1 SD) horizontal heel clearance, vertical heel clearance, final foot placement from step edge on upper level (Trail-foot placement), lead foot placement from the step edge on the lower level (Lead-foot placement), head flexion prior to step descent (head flex prior to step), head flexion during step descent (head flex descent) walking velocity (Walking velocity prior to step), walking velocity during step descent (Walking velocity descent), and ankle and knee angle (θ), vertical and a/p and m/l CM velocity at instant of landing in full vision, 1 step and 2 step lvf occlusion conditions.

	Vision Condition		
	Full vision	**1 step**	**2 step**
Horizontal heel clearance (mm)	152.5 (59.8)	154.6 (55.2)	169.3 (71.2)*
Vertical heel clearance (mm)	53.4 (17.9)	54.2 (17.4)	60.7 (21.1)*
Trail-foot placement (mm)	-139.7 (52.2)	-138.7 (42.7)	-137.3 (36.3)
Lead-foot placement (mm)	431.3 (71.3)	430.7 (63.9)	438.8 (70.9)
Head flex prior to step (deg)	2.0 (17.2)	2.9 (18.8)	2.0 (16.0)
Head flex final step (deg)	-0.4 (18.3)	-0.2 (19.7)	-1.3 (18.3)
Walking velocity prior to step (cm/s)	956.7 (136.7)	955.0 (132.8)	962.7 (131.8)
Walking velocity descent (cm/s)	876.9 (152.7)	865.8 (154.0)	868.6 (160.9)
θ_{ank} (deg)	-23.6 (4.6)	-24.4 (5.0)	-23.8 (4.6)
θ_{knee} (deg)	9.3 (5.9)	9.0 (5.9)	10.2 (6.4)*
Vertical CM velocity (cm/s)	-443.8 (129.1)	-442.9 (134.1)	-461.8 (136.6)
a/p CM velocity (cm/s)	860.8 (182.7)	845.7 (186.0)	857.1 (190.1)
m/l CM velocity (cm/s)	103.3 (36.9)	94.7 (34.8)	83.6 (39.9)*

Significant differences to full vision condition are shown by an asterisk * ($p < 0.05$). Ankle angle (negative) indicates the amount of plantar-flexion and knee angle indicates amount of knee flexion relative to standing.

Table 8.2. Landing mechanics: Group mean (± 1 SD) ankle and knee angular velocity (ω), peak vertical contact force (Fz_{peak}), vertical and a/p and m/l CM velocity during initial landing in full vision, 1 step and 2 step lvf occlusion conditions.

	Vision Condition		
	Full vision	**1 step**	**2 step**
ω_{ank} (deg s^{-1})	272 (74)	276 (77)	281(87)
ω_{knee} (deg s^{-1})	139 (85)	138 (80)	145 (85)
Fz_{peak} (N)	935 (171)	949 (183)	922 (163)
Vertical CM velocity (cm/s)	-445.2 (131.9)	-443.9 (131.5)	-463.7 (138.0)
a/p CM velocity (cm/s)	999.5 (182.9)	975.5 (195.8)	997.3(197.8)
m/l CM velocity (cm/s)	115.7 (41.5)	111.3 (38.9)	102.1(42.2)*

Significant differences to full vision condition are shown by an asterisk * ($p < 0.05$). Ankle angle (negative) indicates the amount of plantar-flexion and knee angle indicates amount of knee flexion relative to standing.

Table 8.3. Temporal parameters: Group mean (± 1 SD) single support (SS), weight transfer (WT) times and time to peak vertical contact force (time to Fz_{peak}) in full vision, 1 step and 2 step lvf occlusion conditions.

	Vision Condition		
	Full vision	**1 step**	**2 step**
WT time	0.09 (0.03)	0.09 (0.03)	0.09 (0.03)*
SS time	0.65 (0.08)	0.65 (0.09)	0.66 (0.10)
time to Fz_{peak}	0.15 (0.04)	0.15 (0.04)	0.15 (0.04)

Significant differences to full vision condition are shown by an asterisk * ($p < 0.05$).

8.4 Discussion

The present study investigated whether lvf cues sampled 2 or 1 step(s) prior to descending a step during on-going gait contribute to landing control. Findings indicate significant differences in landing control (evident in prelanding kinematic measures and landing mechanics) when step descents were completed under conditions of lvf occlusion from 2 steps prior to negotiating the descent compared to full field vision available throughout. There were no significant differences in any of the dependent variables analysed when descents were completed under conditions of lvf occlusion from 1 step prior to the descent compared to full field vision available throughout. Significant differences in landing control when lvf was occluded from 2 steps prior to the descent during on-going gait, highlights that lvf cues obtained in the penultimate step prior to step descent are used in a feedforward manner to 'update' lower-limb trajectory to ensure landings are controlled with increased certainty. With no significant differences in landing control when lvf was occluded from 1 step prior to the descent, this suggests that the nervous system was able to effectively utilise feedforward visual cues in advance of 1 step prior to the descent, to ensure that landing was precisely controlled.

Compared to full field vision available throughout, when lvf was occluded from 2 steps prior to step descent, kneedrop decreased and occurred earlier in the descent, lateral CM velocity at the instant of and during initial landing decreased, θ_{knee} at the instant of contact increased and peak θ_{knee} and θ_{ank} during initial landing and WT time increased. These differences in stepping strategy when lvf was occluded from 2 steps prior to descent suggest that participants were unable to effectively use visual cues from areas of uvf to plan / control landing in the same manner as occurred under full field vision. Indeed the adaptations in stepping strategy when lvf was occluded from 2 steps prior to step descent are indicative of participants being uncertain regarding precise floor height, as evidenced by preparing for landing earlier (figure 8.2) and being more cautious upon landing on the lower level (figure 8.3, tables 8.1, 8.2 and 8.3). However, the adaptations in landing control when lvf was occluded from 2 steps prior to step descent were made without fundamentally altering stepping strategy. For example, there was no change in trail or lead-foot placement at the instant of landing, Fz_{peak}, ω_{ank}, ω_{knee}, vertical or a/p CM velocity during initial landing. These results demonstrate that lvf cues available in the penultimate step prior to step descent are typically used in a feedforward manner to ‘update’ expproprioceptive information of the foot / lower leg relative to the floor, and as a result allow individuals to plan to control landing with increased level of certainty. Indeed, such findings are consistent with previous research which demonstrated that, when lvf is occluded prior to step descent from a stationary standing starting position, visual expproprioceptive information from the head in space was sufficient to plan the general stepping pattern, however, was unable to fully compensate for loss of visual information pertaining to the lower limbs and floor (chapter 4).

When the lvf was occluded from 2 steps prior to step descent, compared to full field vision available throughout, vertical and horizontal heel clearance significantly increased. Such increases have been previously reported as a safety strategy reflective of an inability to ‘fine tune’ lower limb trajectory (Patla, 1998). Indeed the inability to ‘fine tune’ lower limb trajectory under conditions of lvf occlusion has been similarly reported during obstacle negotiation (Rietdyk and Rhea, 2006; Rhea and Rietdyk, 2007; Graci *et al.,* 2010). It is possible that when the lvf was occluded from 2 steps prior to step descent, the inability to ‘fine tune’ lower limb trajectory may have attributed to the subsequent changes in landing control on the lower level. For example, the 14 % increase in vertical heel clearance (table 8.1) would have resulted in the lead limb descending onto the lower level from a greater height. Increasing the height that the lead limb descended may have contributed to the subsequent increase in θ_{knee} and θ_{ank} during initial landing (figure 8.2). However, since there was no subsequent increase in Fz_{peak}, ω_{ank}, or ω_{knee} during initial landing (table 8.2), this confirms that the adaptations in step descent landing control were a direct result of participants altering the mechanisms controlling landing.

When lvf was occluded from 1 step prior to descent, compared to full field vision available throughout, stepping strategy remained invariant. These results (or lack of) are inconsistent with previous findings (Patla, 1998, experiment 2). Patla (1998, experiment 2) tasked participants with negotiating an obstacle under conditions of lvf occlusion, such that the view of the obstacle was occluded in the final step prior to negotiation. Under such conditions participants were unable to ‘fine tune’ lower limb trajectory (Patla, 1998, experiment 2). One possible explanation for the inconsistencies in results is likely attributed to the amount of lvf that was occluded. In the present study, when lvf was occluded from 1 step prior to step descent, this occurred at the precise instant of trail limb heel contact with the ground. In contrast, Patla (1998, experiment 2) occluded the lvf 1 step ahead using basketball goggles which were worn prior to the start of the trial (thus lvf was occluded 1 step ahead throughout the entire trial). However, since slight variations in the position of lvf occlusion from the basketball goggles on the pupil will alter the amount of lvf occluded (i.e. placing the upper edge of the lvf occlusion higher on the pupil will increase the amount of lvf occluded), it is possible that Patla (1998, experiment 2) occluded the lvf in advance of one step. It is interesting to note that the adaptations in lower limb trajectory reported by Patla (1998, experiment 2) when lvf was occluded 1 step in advance were similar to findings reported by Rietdyk and Rhea (2006) and Rhea and Rietdyk (2007) when lvf was occluded ~2 steps ahead (lvf also occluded using basketball goggles). Nevertheless, results from the present study highlight the nervous system’s ability to utilise feedforward visual cues in advance of the final step, to integrate an efference copy of the motor command to precisely control future stepping actions (Marigold, 2008).

It is relevant to note that whilst step descents completed from a stationary standing starting position utilise online vision during the latter portion of the descent to subtly ‘fine tune’ landings (see chapter 4), such results were not observed in the present study when descents were completed during on-going gait. Indeed any indication in the present study that visual cues were required during step descent to ‘fine tune' landing would have been reflected by significant differences in stepping strategy when lvf was occluded from 2 and 1 step(s) prior to descent, compared to full field vision available throughout. The inconsistent findings between chapter 4 and the present study are likely attributed to the benefits dynamic visual sampling of the environment affords compared to static visual sampling (Patla and Greig, 2006).

8.5 Conclusion

In summary, when lvf was occluded from 2 steps prior descending a step during on-going gait, participants adapted their stepping strategy in a manner consistent with being uncertain regarding precise floor height. This was evidenced by preparing for landing earlier during the descent and being more cautious upon landing on the lower level. With such adaptations in landing control made without fundamentally altering stepping strategy, this demonstrates that lvf cues available in the penultimate step prior to step descent, are typically used in a feedforward manner to 'update' exproprioceptive information of the foot / lower leg relative to the floor, and as a result allow individuals to plan to control landing with increased level of certainty. When the lvf was occluded from 1 step prior to descending the step, there were no differences in landing control compared to when vision was available throughout. These findings suggest that, during normal everyday adaptive locomotion where we look approximately 2 steps in advance (Patla and Vickers, 1997), visual information from the lvf is used in the penultimate step prior to step descent during on-going gait to precisely control subsequent landing control on the lower ground level. To increase our understanding of when lvf cues are acquired in the penultimate step prior to step descent, future research should investigate the effects of occluding the lvf from different time points during the penultimate step prior to step descent (i.e. from mid-swing).

Chapter 9

Conclusions and future work

9.1 General Discussion

The literature reviewed in chapter 2 highlighted that the majority of previous research investigating the role of vision in controlling adaptive gait has predominantly focused on over-ground walking or obstacle negotiation. Thus, there is limited research investigating the role of vision in controlling gait during descent of steps and / or stairs. This is somewhat surprising considering that step / stair descent is perhaps the most challenging of locomotor tasks encountered in normal everyday life. The challenge individuals face during step / stair descent has been previously evidenced with accidents reported to occur approximately three times more frequently (in the elderly) and usually resulting with more serious injuries compared to step / stair ascent (Tinetti *et al.,* 1988; Templer, 1992; Startzell, 2000; Roys, 2001). As previous research has shown that the lower visual field (lvf) is important to the control of adaptive gait (Rietdyk and Rhea, 2006; Marigold *et al.,* 2007; Graci *et al.,* 2010), the purpose of the experiments presented in this thesis was to determine the importance of visual information from the lvf in regulating step descent landing control. In addition, the thesis also determined when prior to / during step descent visual feedback from the lvf is typically used in regulating landing control.

The protocol used in this thesis tasked participants with single step descent rather than descending multiple steps. The rationale behind such an approach is that it was possible to frequently change step height and thus avoid participants adopting a repeated motor strategy by descending the same step height throughout each study. There was also a safety aspect in that asking older participants to descend several steps would be potentially dangerous and would necessitate the use of a safety harness for ethical reasons. The use of a safety harness may affect how older adults descend stairs due to the psychological impression of safety provided by the device (Freitas *et al.,* 2005). Furthermore, descending a single step minimised the risk of participants becoming fatigued during each study, despite the high number of trials collected in some studies (range of 36 - 81 trials collected across studies). The main findings from each experiment are summarised below with a discussion of a synthesis of findings across the various studies.

Chapter 4 determined that when step descents were completed from a stationary standing position under conditions of lvf occlusion, compared to full field vision available throughout, participants adopted a cautious stepping strategy which was attributed to uncertainty regarding precise location of the foot / lower leg relative to the lower floor level. This cautious stepping strategy was evidenced through participants preparing for landing earlier during the descent and ensuring that a ‘softer’ landing occurred. However, these changes in landing control under conditions of lvf occlusion were made without fundamentally altering stepping strategy. The implication is that participants were able to plan the general stepping strategy when only upper visual field (uvf) cues were available, presumably using exproprioceptive information regarding head position relative to the environment. However, when lvf information was available prior to / during step descent, participants were able to perceive the height of the lower floor level with greater accuracy and also receive exproprioceptive information of the foot / lower leg relative to the floor, and as a result were able to regulate landing with increased levels of certainty.

Compared to step descents completed with vision available throughout, full field visual occlusion from beginning of swing (toe-off) or mid-swing onwards resulted in minimal differences in stepping strategy. The only adaptations in stepping strategy were evidenced in reductions in the magnitude and timing of kneedrop (which highlighted that participants prepared for landing earlier during the descent, Cowie *et al.,* 2008) when vision was occluded from mid-swing onwards. These findings (chapter 4) suggest that visual cues acquired in the latter part of step descent were used to subtly 'fine tune' landing.

Collectively, findings from chapter 4 indicate that visuomotor control of step descent utilises visual cues from the lvf to determine the position of the foot / lower leg in relation to the height of the lower floor level and this information allows landing to be regulated with increased level of certainty. Since the occlusion of vision (full or uvf) from either immediately prior to or during step descent caused very few differences in landing control, this suggests that visual cues acquired from lvf to regulate step descent landing control occurs predominantly prior to or during movement initiation, and that online vision is used only in the latter portion of the descent phase to subtly ‘fine tune’ landing.

Chapter 6 extended knowledge concerning the role of lvf in regulating step descent landing by determining the effects on landing control when descents were completed with lvf information degraded by wearing multifocal spectacles. These spectacles include a lower section of the lenses focussed for near work at about 40 cm and provide a blurred and / or magnified image of objects beyond that distance, particularly in older adults. Findings highlighted that wearing multifocal spectacles resulted in adaptations in step descent landing control in a manner consistent with participants being uncertain regarding precise floor height. Such adaptations in stepping strategy were evidenced through participants ‘dropping’ onto the lower level when wearing bifocal spectacles, rather than landing in a controlled manner as occurred when wearing single-vision distance spectacles. However, the adaptations in landing control when lvf was degraded were made without fundamentally altering stepping strategy. Overall the findings of chapter 6 corroborate those of chapter 4; that uvf information can be used to effectively plan the general stepping strategy, but visual information from the lvf which likely pertains to expproprioceptive information of the foot / lower leg relative to the floor is required to precisely determine lower floor height, which allows landing to be regulated with increased levels of certainty.

The adaptations that occurred in step descent landing control when lvf was occluded (chapter 4) or degraded (chapter 6) serendipitously provided different levels of adaptation in the occlusion / degradation of visual cues. lvf occlusion was an acute change to visual condition, whereas the multifocal lens wearers had been wearing these types of spectacles for many years and were fully adapted to the change. It is interesting to note, that the acute adaptations of lvf occlusion (chapter 4) were reflected by an increasingly cautious stepping strategy. However, the chronic adaptations of degrading lvf (chapter 6) were reflected by an opposite effect of decreased level of caution, evidenced through participants ‘dropping’ onto the lower level. A likely explanation for these opposing effects is, when lvf was occluded (chapter 4) participants were not used to their visual field being occluded and adopted a cautious stepping strategy to compensate for the increased threat to maintaining stability during the descent. However, participants in chapter 6 had adapted (familiarised) to wearing multifocals and the associated degradation of the lvf over time, and as such did not perceive any threat to maintaining stability during the descent when switching between lvf degraded (i.e. multifocals) and non-degraded (i.e. single-vision distance spectacles) conditions, and therefore did not perceive the need to adopt a cautious stepping strategy.

Findings from chapter 4 highlight that step descent landing control is predominantly biased towards using feedforward visual mechanisms. However, in situations when task demand is increased, the role of online vision has been found to be increased

(Reynolds and Day, 2005; Marigold and Patla, 2008). It was therefore hypothesised, in chapter 5, that when descending a step carrying added mass the requirement to attenuate the increased downward momentum during the descent would increase the demands of the task, and this would affect the role of feedforward versus online vision. However, chapter 5 failed to observe differences in step descent landing control when vision was occluded from either immediately prior to MI or mid-swing onwards during step descents completed carrying added / no added mass. The lack of significant differences in how vision is used to regulate step descent landing control when carrying added mass was attributed to the adaptations that occurred due to carrying added mass. When step descents were completed carrying added mass, participants attenuated the increased momentum generated during the descent to ensure there was no increase in landing force on the lower level compared with descents completed carrying no added mass. Since the increased momentum was attenuated during the descent, this meant that there was no increased demand placed on the visual system to regulate step descent landing control. Subsequently, this meant that online visual mechanisms did not become more important when step descents were completed carrying added mass. These findings (chapter 5) suggest that the role of feedforward versus online vision used in regulating step descent landing control remains invariant when descents are completed with added mass. The wider implications of these findings is that the increased risk of falling associated with obesity and added mass (Owusu *et al.,* 1998; Qu and Nusbaum, 2008) are not associated with a change in visuomotor control, and thus must be attributed to other factors such as insufficient lower limb muscle strength (DeVita and Hortobágyi, 2003).

Since landings (chapter 5) also remained invariant when vision was occluded from the period immediately prior to MI compared to vision available throughout, findings extend upon those from chapter 4; suggesting that feedforward visual cues controlling step descent are typically acquired in the few seconds prior to MI and not during MI as previously suggested (chapter 4).

Findings from chapters 4 and 6 demonstrated that visual cues from lvf provide exproprioceptive information of the foot / lower leg relative to the floor, which subsequently enables landing to be regulated with increased certainty. Furthermore, this visual information is acquired in the few seconds prior to MI (see chapter 5). However, to determine when lvf cues are acquired prior to step descent, a different protocol was used in chapter 8. Participants were required to descend a step during on-going gait. Findings indicate that lvf cues are typically acquired in the penultimate step prior to descent and are used to ‘update’ exproprioceptive information of the foot / lower leg relative to the floor, and as a result allow individuals to plan to control landing with increased level of certainty. This was evidenced by landing control being significantly adapted when lvf was occluded from the penultimate step prior to step descent but not significantly adapted when vision was occluded from the final step. When lvf was occluded from the penultimate step, kneedrop decreased and occurred earlier during the descent and lateral CM velocity at the instant of landing and during initial landing was reduced and peak θ_{knee} and θ_{ank} during initial landing was increased. With no significant differences in step descent landing control when lvf was occluded in the final step prior to descent, these findings support previous work highlighting how the nervous system is able to utilise feedforward visual cues in advance of the final step to integrate an efference copy of the motor command to precisely control future stepping actions (see Marigold, 2008).

These findings (chapter 8) corroborate findings from Geruschat *et al.* (2003, 2006) that during step descent i.e. when stepping down from a kerb onto the road, visual information about the kerb / step edge and immediate lower ground area are not required to be sampled during the descent to ensure landings are completed successfully. Indeed visual cues pertaining to the step edge and immediate lower floor area are sampled prior to step descent approximately 2 steps in advance (see chapter 8); which is in agreement with previous research highlighting that gaze tends to be directed one or two walking steps ahead during adaptive locomotion (Patla and Vickers, 1997).

Findings from chapter 8 are consistent with findings of lvf occlusion prior to step descent from a stationary standing starting position (chapter 4). Furthermore, findings are similar to those of obstacle crossing under conditions of lvf occlusion, indicating that when the lower limbs and obstacle are occluded from 2 steps prior to crossing the obstacle, the ability to ‘fine tune’ lower-limb trajectory is significantly affected (Rietdyk and Rhea, 2006; Rhea and Rietdyk, 2007; Graci *et al.,* 2010). Therefore, the potential implication from this thesis is that there is a general pattern for visuomotor control of adaptive gait. Indeed, it is possible that visual cues from uvf are acquired to plan the general stepping strategy, whilst lvf cues (acquired in a feedforward manner) ensure that the movement is controlled with greater accuracy. In combination with previous findings, it is likely that the central part of the uvf is involved in planning the general stepping strategy (Graci *et al.,* 2010), as the centre of the retina (fovea) provides the highest detail of sampled visual information (Banks *et al.,* 1991).

Chapter 7 determined that irrespective of whether there was a high or low probability of visual occlusion during step descent, participants always planned to use online vision (if available) during the descent to regulate landing control. These findings are inconsistent with previous upper limb movement studies whereby participants plan for the ‘worse-case’ scenario, i.e. that online vision will be unavailable, and instead adopt a feedforward mode of control when uncertain if vision will be available throughout the movement (Zelaznik *et al.,* 1983; Elliott *et al.,* 1999). However, the aforementioned upper limb studies focused on rapid manual aiming movements which are predominantly regulated through online visual mechanisms (Elliott *et al.,* 1999), whereas chapter 4 and 5 determined that step descents are predominantly regulated through feedforward visual mechanisms. It was therefore possible that in chapter 7, participants planned to use online vision during step descent irrespective of the probability of visual occlusion, as the consequence of visual occlusion during the descent presented little threat to stability and thus the ability to control the movement. However, during precision movement tasks, as the role of online vision is increased, the consequence of planning to use online vision during the movement and it being unavailable will have significant implications on the ability to control the movement. The implication of this finding is that step descent paradigms which use a high probability of occluding online vision during the descent (i.e. chapters 4 and 5) has little / no effect on how online vision is used to regulate landing control in subsequent full vision trials. Findings from chapter 7 provide additional support that online vision is only used in the latter portion of step descent to subtly ‘fine tune’ landing, as per chapters 4 and 5.

Chapter 7 also determined that prior awareness and experience pertaining to a potential visual occlusion during step descent had little / no effect on step descent landing control. Whilst these results were inconsistent with previous research investigating the effects of prior awareness and / or experience of a slip / trip (i.e. Pijnappels *et al.,* 2001, 2006; Heiden *et al.,* 2006; Siegmund *et al.,* 2006), differences between chapter 7 and the aforementioned research were attributed to task demand. Indeed findings from chapter 7 indicated that task demand likely influences the effect of prior awareness and / or experience, and locomotor tasks with a reduced threat to stability may be less influenced by awareness and / or experience.

9.2 Limitations

In chapters 4, 5 and 7 participants were required to wear PLATO goggles throughout the experiment. When step descents were completed with full field vision available throughout, this was considered to represent participants' habitual stepping response. However, wearing the PLATO goggles occluded the outer most part of the peripheral visual field. The extent of peripheral visual field occlusion has been reported in the methods section (3.3.2). Since findings from this thesis highlight that visual cues acquired from lvf enable step descent landings to be controlled with increased level of certainty, it is possible that step descents completed wearing the PLATO goggles, which occluded a small area of the lvf, may not have truly reflected participant's habitual stepping strategy. This limitation does not question the validity of the findings from this thesis, rather, in the extreme case it is possible that the role of the lvf has a slightly greater importance in regulating step descent landing control than suggested. Future work should consider quantifying the effects of step descent landing control when wearing the PLATO goggles through analysing step descents completed with / without wearing the goggles.

It is also relevant to note, that the size and shape of each participant's face and nose altered the back vertex distance (distance between the lenses of the goggles and the participant's eye) thus slightly affecting the amount of peripheral visual field occlusion across each participant.

In chapters 4 and 8 the lvf was occluded by positioning, respectively the upper edge of black card / LCD sheet in line with the middle of participant’s pupil. To increase the reliability of accurately occluding only the lvf, participants were instructed to look directly ahead, fixating on a stationary object at eye level whilst the card / LCD sheet was attached. However, it is recognised that some element of human error could have occurred in positioning the lvf occlusion in line with the middle of participant’s pupil.

9.3 Future Work

The rationale behind this thesis which tasked participants with single step descent rather than descending multiple steps has been previously highlighted in the general discussion. Results from this thesis may provide some indication of the role of the lvf in regulating stair descent landing control since the highest risk of falling during stair descent occurs at the first or last few steps (Templer, 1992). However, further work is required to confirm whether the role of lvf in regulating step descent landing control is similar to stair descent landing.

The differences between chronic (chapter 4) and acute (chapter 6) adaptations in step descent landing control when lvf was manipulated, could suggest that the effects of temporary / acute lvf manipulation are very different to the effects of long term / chronic lvf changes. This may limit the validity of acutely manipulating lvf or indeed any part of the visual field and inferring conclusions to populations with habitual visual field loss i.e. patients with glaucoma. Thus future work is required to compare the adaptations of acute visual field loss and habitual visual field loss. This could be achieved through analysing step descents completed with patients with glaucoma and participants with full field vision who have the same amount of visual field occluded.

It is also possible that the differences between chapters 4 and 6 are not attributed to chronic and acute effects, rather participant age. In chapter 4 participants were young adults (age 24.4 ± 9.4 years) whereas in chapter 6 participants (who were habitual multifocal lens wearers) were older adults (age 71.9 ± 4.2 years). Previous research has highlighted differences in the utility of visuomotor control of step descent across participant age. For example Chapman and Hollands (2006) demonstrated that older adults (age 71.1 ± 3.1 years) rely more on vision to plan (feedforward control) and guide (online control) foot placement during a precision stepping task than young adults (age 23.5 ± 1.6 years). Thus future work is required to investigate the role of visuomotor control of step descent as the visual system ages.

Findings from the multifocal work presented in chapter 6 provided evidence based knowledge regarding how multifocal spectacles increase the risk of falling during step descent. Whilst these results provide important insight into the potential problems of multifocals during step descent, only a relatively small sample of healthy high level functioning older adults were used. Consequently these studies need replicating on older frailer adults with larger sample numbers.

Visuomotor control of step descent appears predominantly biased towards feedforward visual processing (chapters 4 and 5). It was therefore likely that in chapter 7, irrespective of the probability of visual occlusion during the descent participants planned to use online vision during the descent, as the consequence of visual occlusion during the descent presented little threat to stability and thus the ability to regulate landing control. Future work is required to investigate whether the probability of visual occlusion affects how participants plan to use online vision to control the movement when the role of visuomotor control is predominantly biased towards online visual processing i.e. during a precision stepping task.

Chapter 8 determined that lvf cues acquired in the penultimate step prior to descending a step during on-going gait are used in a feedforward manner to precisely determine lower floor height which subsequently allows landing to be regulated with increased level of certainty. Future research is required to determine when during the penultimate step such visual cues are acquired. Indeed this could be achieved through occluding lvf from periods relative to toe-off during the penultimate step i.e. from mid swing.

Chapters 4, 6 and 8 determined that visual cues acquired from uvf are able to plan the general stepping strategy when lvf is manipulated. This is attributed to expproprioceptive information regarding head position relative to the environment. However, the availability of lvf information enables step descent landings to be controlled with increased level of certainty, which is attributed to expproprioceptive information of the foot / lower leg relative to the floor. Future work is required to determine which information source, expproprioceptive information regarding the head in space or lower limb is more important in regulating step descent landing control. Through shifting the visual surround or the lower level that the participant intends to land on in the vertical direction (±), under condition of full field vision as the participant descends a step, this will determine which information source is more important when regulating step descent landing control.

References

Aglioti, S. DeSouza, J. F. X. and Goodale, M. A. (1995). Size-contrast illusions deceive the eye but not the hand, *Current Biology,* **5,** 679-685.

Allied Dunbar National Fitness Survey. (1992), Main Findings. The Sports Council and Health Education Authority, London.

Anand, V. Buckley, J. G. Scally, A. and Elliott, D. B. (2003). Postural Stability in the Elderly during Sensory Perturbations and Dual Tasking: The Influence of Refractive Blur, *Investigative Ophthalmology and Visual Science*, **44(7),** 2885-2891.

Arampatzis, A. Bruggemann, G. and Metzler, V. (1999). The effect of speed on leg stiffness and joint kinetics in human running, *Journal of Biomechanics,* **32,** 1349–1353.

Arampatzis, A. Bruggemann, G. and Klapsing, G. M. (2001). Leg stiffness and mechanical energetic processes during jumping on a sprung surface, *Medicine and Science in Sports and Exercise,* **33,** 923–931.

Arend, L. E. (1993). Mesopic lightness, brightness, and brightness contrast, *Perception and Psychophysics*, **54**, 469–476.

Arumi, P. Chauhan, K. and Charman, W. N. (1997). Accommodation and acuity under night-driving illumination levels, *Ophthalmic and Physiological Optics,* **17**, 291–299.

Bailey, L. (2006). Visual acuity. In Benjamin, W. J. ed. Borish's clinical refraction. 2nd ed. Missouri: Butterworth Heinemann.

Banks, M. S. Sekuler, A. B. and Anderson, S. J. (1991). Peripheral spatial vision: Limits imposed by optics, photoreceptors, and receptor pooling. *Journal of the Optical Society of America: A*, **8,** 1775–1787.

Barlow, H. B. Derrington, A. M. Harris, L. R. and Lennie, P. (1977). The effects of remote retinal stimulation on the responses of cat retinal ganglion cells, *Journal of Physiology*, **269**, 177–194.

Bear, M. F. Conners, B. W. and Paradiso, M. A. (2007). Neuroscience: exploring the brain*.* Baltimore, M. A: Lippincott, Williams and Wilkins.

Berg, W. P. Alessio, H. M. Mills, E. M. and Tong, C. (1997). Circumstances and consequences of falls in independent community-dwelling older adults, *Age and Ageing,* **26,** 261–268.

Breyer, A. Jiang, X. Rütsche, A. and Mojon, D. S. (2006). A New 3D Monitor–Based Random-Dot Stereotest for Children, *Investigative Ophthalmology and Visual Science,* **47,** 4842-4846.

Brown, B. Brabyn, L. Welch, L. Haegerstrom–Portnoy, G. and Colenbrander, A. (1986). Contribution of vision variables to mobility in age related maculopathy patients, *American Journal of Optometry and Physiological Optics,* **63,** 733–739.

Buckley, J. G. Heasley, K. J. Twigg, P. and Elliott, D. B. (2005). The effects of blurred vision on the mechanics of landing during stepping down by the elderly, *Gait and Posture,* **21,** 65-71.

Buckley, J. G. MacLellan, M. J. Tucker, M. W. Scally, A. J. and Bennett, S. J. (2008). Visual guidance of landing behaviour when stepping to a new level, *Experimental Brain Research,* **184,** 223–232.

Butler, R. J. Crowell III, H. P. and Davis, I. M. (2003). Lower extremity stiffness: implications for performance and injury, *Clinical Biomechanics,* **18,** 511–517.

Cagenello, R. Arditi, A. and Halpern, D. L. (1993). Binocular enhancement of visual acuity, *Journal of the Optical Society of America A, Optics, image science and vision,* **10,** 1841–1848.

Campbell, F. W. and Green, D. G. (1965). Monocular versus binocular visual acuity, *Nature,* **208,** 191-192.

Campbell, A. J. Reinken, J. Allan, B. C. and Martinez, G. S. (1981). Falls in old age: a study of frequency and related clinical factors, *Age and Ageing,* **10,** 264-270.

Carkeet, A. (2001). Modelling logMAR visual acuity scores: effects of termination rules and alternative forced-choice options, *Optometry and Vision Science,* **78,** 529-538.

Carpenter, R. H. S. (1992). Turning vision into action, *Current Biology,* **2,** 228-290.

Cham, R. and Redfern, M. S. (2001). Lower extremity corrective reactions to slip events, *Journal of Biomechanics,* **34 (11),** 1439–1445.

Chapman, G. J. and Hollands, M. A. (2006). Age-related differences in stepping performance during step cycle-related removal of vision, *Experimental Brain Research,* **174,** 613-621.

Charman, W. N. (1996). Night myopia and driving, *Ophthalmic and Physiological Optics,* **16,** 474–485.

Clarke, G. M. and Cooke, D. (1998). A basic course in statistics. 4th ed. London: Edward Arnold.

Coleman, A. L. Cummings, S. R. Yu, F. Kodjebacheva, G. Ensrud, K. E. Gutierrez, P. Stone, K. L. Cauley, J. A. Pedula, K. L. Hochberg, M. C. and Mangione, C. M. (2007). Binocular visual-field loss increases the risk of future falls in older white women, *Journal of the American Geriatrics Society,* **55(3),** 357-364.

Color and Light (2010) Exploring the science of light [online] Available from: http://www.optics4kids.com/teachersparents/articles/images/colorandlight06.gif [accessed 6 April 2010].

Connolly, M. and van Essen, D. (1984). The representation of the visual field in parvocellular and magnocellular layers of the lateral geniculate nucleus in the macaque monkey, *Journal of Comparative Neurology,* **226(4),** 544–564.

Consumer Safety Unit, DTI. (1997). Home accident surveillance system 1995: 19th Annual Report. Department of Trade and Industry.

Cook, P. B. Lukasiewicz, P. D. and McReynolds, J. S. (1998). Action potentials are required for the lateral transmission of glycinergic transient inhibition in the amphibian retina, *Journal of Neuroscience,* **18,** 2301–2308.

Cowie, D. Braddick, O. and Atkinson, J. (2008). Visual control of action in step descent, *Experimental Brain Research,* **186,** 343–348.

Craik, R. L. Cozzens, B. A. and Freedman, W. (1982). The role of sensory conflict on stair descent performance in humans, *Experimental Brain Research,* **45,** 399–409.

Crowe, A. Schiereck, P. de Boer, R. and Keessen, W. (1993). Characterisation of gait of young adult females by means of body centre of mass oscillations derived from ground reaction forces, *Gait and Posture,* **1,** 61-68.

Curcio, C. A. Sloan, K. R. Jr. Packer, O. Hendrickson, A. E. and Kalina, R. E. (1987). Distribution of cones in human and monkey retina: Individual variability and radial asymmetry, *Science*, **236(4801),** 579–582.

Curcio, C. A. and Allen, K. A. (1990). Topography of ganglion cells in human retina. *Journal of Comparative Neurology,* **300(1),** 5–25.

Dafoe, J. M. Armstrong, I. T. and Munoz, D. P. (2007). The influence of stimulus direction and eccentricity on pro- and anti-saccades in humans, *Experimental Brain Research*, **179(4),** 563–570.

Dancey, C. P. and Reidy, J. (2004). Statistics without maths for psychology; using spss for windows. 3rd ed. Essex: Pearson Education ltd.

Danckert, J. and Goodale, M. A. (2001). Superior performance for visually guided pointing in the lower visual field, *Experimental Brain Research,* **137(3–4),** 303–308.

Darker, T. and Jordan, T. R. (2004). Perception of words and non-words in the upper and lower visual fields, *Brain and Language,* **89,** 593–600.

Dassonville, P. and Bala, J. K. (2004). Perception, action, and Roelofs effect: A mere illusion of dissociation. *Plos Biology,* **2,** 1936–1945.

Daum, K. and McCormack, G. (2006). Fusion and binocularity. In Benjamin, W. J. ed. Borish's Clinical Refraction 2nd ed. Missouri: Butterworth Heinemann.

Davies, J. C. Kemp, G. J. Stevens, G. Frostick, S. P. and Manning, D. P. (2001). Bifocal/varifocal spectacles, lighting and missed-step accidents, *Safety Science,* **38,** 211–226.

Dempster, W. T. (1955). Space requirements of the seated operator. WADC Technical Report, Wright-Patterson Air Force Base, OH.

Department for transport (2010) [online] Available from www.dft.gov.uk [accessed 4 April 2010].

DeVita, P. and Hortobágyi, T. (2003). Obesity is not associated with increased knee joint torque and power during level walking, *Journal of Biomechanics*, **36,** 1355–1362.

Eames, M. H. A. Cosgrove, A. and Baker, R. (1999). Comparing methods of estimating the total body centre of mass in three-dimensions in normal and pathological gaits, *Human Movement Science,* **18,** 637-646.

Elliott, D. B. Bullimore, M. A. and Bailey, I. L. (1991). Improving the reliability of the Pelli–Robson contrast sensitivity test, *Clinical Vision Science,* **6,** 471-475.

Elliott, D. B. and Bullimore, M. A. (1993). Assessing the reliability, discriminative ability, and validity of disability glare tests, *Investigative Ophthalmology and Visual Science,* **34**,108–119.

Elliott, D. B. Patla, A. E. Flanagan, J. G. Spaulding, S. Rietdyk, S. Strong, G. and Brown, S. (1995). The Waterloo Vision and Mobility Study: postural control strategies in subjects with ARM, Ophthalmic and Physiological Optics, **15(6),** 553-559.

Elliott, D. Binsted, G. and Heath, M. (1999). The control of goal-directed limb movements: Correcting errors in the trajectory, *Human Movement Science,* **18,** 121-136.

Elliott, D. B. Patla, A. E. Furniss, M. and Adkin, A. (2000) Improvements in Clinical and Functional Vision and Quality of Life after Second Eye Cataract Surgery, *Optometry and Vision Science,* **77(1),** 13-24.

Elliott, D. B. (2006). Contrast sensitivity and glare testing. In Benjamin, W. J. ed. Borish's Clinical Refraction*.* 2nd ed. Missouri: Butterworth Heinemann.

Elliott, D. B. and Flanagan, J. (2007). Assessment of Visual Function. In. Elliott D. B. ed. Clinical procedures in primary eye care*. 3rd* ed. London: Butterworth Heinemann.

Elliott, D. B. Vale, A. Whitaker, D. and Buckley, J. G. (2009). Does My Step Look Big In This? A Visual Illusion Leads To Safer Stepping Behaviour, *PLoS One*, **4(2)**:e4577.

Elliott, D. B. and Chapman, G. J. (2010). Adaptive gait changes due to spectacle magnification and dioptric blur in older people. *Investigative Ophthalmology and Visual Science*, **51(2),** 718-22.

Enroth-Cugell, C. and Jakiela, H. G. (1980). Suppression of cat retinal ganglion cell responses by moving patterns, *Journal of Physiology,* **302,** 49–72.

Eysenck, M. W. and Keane, M. T. (2005). Cognitive psycholo______________________. East Sussex, England: Taylor and Francis group.

Farley, C. T. Blickhan, R. Saito, J. and Taylor, C. R. (1991). Hopping frequency in humans: a test of how springs set stride frequency in bouncing gaits, *Journal of Applied Physiology,* **71,** 2127–2132.

Felleman, D. J. and van Essen, D. C. (1991). Distributed hierarchical processing in the primate cerebral cortex, *Cerebral Cortex,* **1,** 1-47.

Ferris, F. L. 3rd and Bailey, L. (1996). Standardizing the measurement of visual acuity for clinical research studies: Guidelines from the Eye Care Technology Forum, *Ophthalmology,* **103,** 181-182.

Fischer, B. Krüger, J. and Droll, W. (1975). Quantitative aspects of the shift-effect in cat retinal ganglion cells, *Brain Research*, **83,** 391–403.

Fortenbaugh, F. C. Hicks, J. C. Hao, L. and Turano, K. A. (2007). Losing sight of the bigger picture: Peripheral field loss compresses representations of space, *Vision Research,* **47,** 2506–2520.

Fowler, J. Jarvis, P. and Chevannes, M. (2002). Practical Statistics for Nursing and Health Care. Sussex, England: Wiley.

Frank, H. and Althoen, S. C. (1994). Statistics: concepts and application. Cambridge: Cambridge University Press.

Franz, V. H. Gegenfurtner, K. R. Bulthoff, H. H. and Fahle, M. (2000). Grasping visual illusions: No evidence for a dissociation between perception and action, *Psychological Science,* **11,** 20–25.

Freedman, W. and Kent, L. (1987). Selection of movement patterns during functional tasks in humans, *Journal of Motor Behaviour,* **19,** 214–226.

Freitas, S. M. S. F. Prado, J. M. and Duarte, M. (2005). The use of a safety harness does not affect body sway during quiet standing, *Clinical Biomechanics,* **20,** 336–339.

Fukuchi, K. R. Arakaki, C. Veras Orselli, M. I. and Duarte, M. (2010). Evaluation of alternative technical markers for the pelvic coordinate system, *Journal of Biomechanics,* **43,** 592–594.

Geruschat, D. Turano, K. A. and Stahl, J. W. (1998). Traditional measures of mobility performance and retinitis pigmentosa, *Optometry and Vision Science,* **75,** 525–537.

Geruschat, D. Hassan, S. E. and Turano, K. A. (2003). Gaze Behavior while Crossing Complex Intersections, *Optometry and Vision Science,* **80(7),** 515-528.

Geruschat, D. Hassan, S. E. Turano, K. A. Quigley, H. A. and Congdon, N. (2006). Gaze Behavior of the Visually Impaired During Street Crossing, *American Academy of Optometry,* **83(8),** 550-558.

Glover, S. and Dixon, P. (2004). A step and a hop on the Muller-Lyer: illusion effects on lower-limb movements, *Experimental Brain Research,* **154,** 504–512.

Goodale, M. A. Milner, A. D. Jakobson, L. S. and Carey, D. P. (1991), *Nature,* **349,** 154-156.

Goodale, M. A. and Milner, A. D. (1992). Separate visual pathways for perception and action, *Trends in Neuroscience,* **15,** 20-25.

Goodale, M. A. and Haffenden, A. (1998). When vision is not sight: dissociation between perception and action in human vision. In M. Harris. And Michael Jenkin eds. Vision and Action. U.S.A: Cambridge University press, pp.270-295.

Goodale, M. A. and Humphrey, G. K. (1998). The object of action and perception, *Cognition,* **67,** 181-207.

Goodale, M. A. Westwood, D. A. and Milner, A. D. (2004). Two distinct modes of control for object-directed action, *Progress in Brain Research,* **144,** 131-144.

Graci, V. Elliott, D. B. and Buckley, J. G. (2009). Peripheral visual cues affect minimum-foot-clearance during overground locomotion, *Gait and Posture,* **30,** 370–374.

Graci, V. Elliott, D. B. and Buckley, J. G. (2010). Utility of Peripheral Visual Cues in Planning and Controlling Adaptive Gait, *Optometry and Vision Science,* **87(1),** 21–27.

Graham, N. Rico, M. Offen, S. and Scott, W. (1999). Texture segregation shows only a very small lower-hemifield advantage, *Vision Research,* **39(6),** 1171–1175.

Greenspan, S. L. Myers, E. R. and Kiel, D. P. (1998). Fall direction, bone mineral density, and function: risk factors for hip fracture in frail nursing home elderly, *American journal of medicine,* **104,** 539-545.

Greenspan, S. L. Myers, E. R. Kiel, D. P. Parker, R. A. Hayes, W. C. and Resnick, N. M. (2003). Fall direction, bone mineral density, and function: risk factors for hip fracture in frail nursing home elderly, *American Journal of Medicine,* **104,** 539–545.

Griffiths, D. Stirling, W. G. and Weldon, K. L. (1998). Understanding data, principles and practice of statistics. New York: Wiley.

Hamel, C. P. (2007). Cone rod dystrophies, *Orphanet Journal of Rare Diseases,* **1,** 2-7.

Hamill, J. and Knutzen, K. M. (1995). Biomechanical basis of human movement. 2nd ed. Baltimore: Lippincott Williams and Wilkins.

Hanavan, E. P. (1964). A mathematical model of the human body. Wright Patterson Air Force Base, OH: Aerospace Medical Research Laboratories.

Hansen, T. Pracejus, L. and Gegenfurtner, K. R. (2009). Color perception in the intermediate periphery of the visual field, *Journal of Vision,* **9(4),** 1-12.

Haran, M. J. Lord, S. R. Cameron, I. D. Ivers, R. Q. Simpson, J. M. Lee, B. B. Porwal, M. Kwan, M. M. and Severino, C. (2009). Preventing falls in older multifocal glasses wearers by providing single-lens distance glasses: the protocol for the VISIBLE randomised controlled trial, *BMC Geriatrics,* **26,** 9-10.

Harris, J. M. and Bonas, W. (2002). Optic flow and scene structure do not always contribute to the control of human walking, Vision Research, **42(13),** 1619-1626.

Hatze, H. (1975). A new method for the simultaneous measurement of the moment on inertia, the damping coefficient and the location of the centre of mass of a body segment in situ, *European Journal of Applied Physiology,* **34,** 217-226.

Hayhoe, M. Gillam, B. Chajka, K. and Vecellio, E. (2009). The role of binocular vision in walking, *Visual Neuroscience,* **26,** 73-80.

Haymes, S. A. Guest, D. Heyes, A. D. and Johnston, A. W. (1996). Mobility of people with retinitis pigmentosa as a function of vision and psychological variables, *Optometry and Vision Science,* **73,** 621–637.

Haymes, S. A. Johnston, A. W. and Heyes, A. D. (2002). Relationship between vision impairment and ability to perform activities of daily living, *Ophthalmic and Physiological Optics,* **22(2),** 79-91.

Hazel, C. A. and Elliott, D. B. (2002). The dependency of logMAR visual acuity measurements on chart design and scoring rule, *Optometry and Vision Science,* **79,** 788–792.

He, S. Cavanagh, P. and Intriligator, J. (1997). Attentional resolution, *Trends in cognitive sciences,* **1,** 115-121.

Heasley, K. Buckley, J. G. Scally, A. Twigg, P. and Elliott, D. B. (2004). Stepping up to a new level: effects of blurring vision in the elderly, *Investigative Ophthalmology and Vision Science,* **45,** 2122–2128.

Heiden, T. L. Sanderson, D. J. Inglis, J. T. and Siegmund, G. P. (2006). Adaptations to normal human gait on potentially slippery surfaces: The effects of awareness and prior slip experience, *Gait & Posture,* **24,** 237–246.

Helbostad, J. L. Vereijken, B. Hesseberg, K. and Sletvold, O. (2009). Altered vision destabilizes gait in older persons, *Gait and Posture,* **30,** 233-238.

Hennig, E. M. and Lafortune, M. A. (1991). Relationships between ground reaction force and tibial bone acceleration parameters, *International Journal of Sports Biomechanics,* **7,** 303–309.

Hensinger, R. N. (1986). Standards in paediatric orthopaedics*.* New York: Raven.

Heywood, S. and Churcher, J. (1980). Structure of the visual array and saccadic latency: Implications for oculomotor control, *Quarterly Journal of Experimental Psychology,* **32,** 335–341.

Hinton, P. R. McMurray, B. I. and Cozens. (2004). SPSS explained. Hove: Routledge.

Hof, A. L. Gazendam, M. G. and Sinke, W. E. (2005). The condition for dynamic stability, *Journal of Biomechanics,* **38,** 1–8.

Hollands, M. A. Marple-Horvat, D. E. Henkes, S. and Rowan, A. K. (1995). Human Eye Movements During Visually Guided Stepping, *Journal of Motor Behaviour,* **27(2),** 155-163.

Hollands, M. A, and Marple-Horvat, D. E. (1996). Visually guided stepping under conditions of step cycle-related denial of visual information, *Experimental Brain Research,* **109,** 343–356.

Hollands, M. A, and Marple-Horvat, D. E. (2001). Coordination of eye and leg movements during visually guided stepping, *Journal of Motor Behaviour,* **33(2),** 205-216.

Home, R. (1978). Binocular summation: a study of contrast sensitivity, visual acuity and recognition, *Vision Research,* **18**, 579–585.

Horowitz, A. (1994). Vision impairment and functional disability among nursing home residents, *Gerontologist,* **34,** 316–323.

Hortobágyi T, and DeVita P. (1999). Altered movement strategy increases lower extremity stiffness during stepping down in the aged, *The Journals of Gerontology series A, Biological sciences and Medical Sciences,* **54,** B63–70.

Hortobágyi, T. and DeVita, P. (2000). Muscle pre- and coactivity during downward stepping are associated with leg stiffness in aging, *Journal of Electromyography and Kinesiology,* **10,**117–126.

Howard, I. P. and Rogers, B. J. (1995). Binocular vision and stereopsis, Oxford Psychology Series no. 29. Oxford, England: Oxford University Press.

Inclusive mobility (2010a) Department for transport [online] Available from http://www.dft.gov.uk/transportforyou/access/peti/inclusivemobility?page=8 [accessed 4 April 2010].

Inclusive mobility (2010b) Department for transport [online] Available from http://www.dft.gov.uk/transportforyou/access/peti/inclusivemobility?page=6 [accessed 4 April 2010].

Ivers, R. Q. Cumming, R. G. Mitchell, P. and Attebo, K. (1998). Visual impairment and falls in older adults: the Blue Mountains Eye Study. *Journal of the American Geriatrics Society,* **46,** 58–64.

Jakobson, L. S. Archibald, Y. M. Carey, D. P. and Goodale, M. A. (1991). A kinematic analysis of reaching and grasping movements in a patient recovering from optic ataxia, *Neuropsychologia,* **29,** 803-809.

Johnson, L. Buckley, J. G. Scally, A. J. and Elliott, D. B. (2007). Multifocal spectacles increase variability in toe clearance and risk of tripping in the elderly, *Investigative Ophthalmology and Vision Science,* **48,** 1466-1471.

Johnson L. (2008). Biomechanics of step negotiation in the elderly: problems associated with using multifocal lenses. Thesis, University of Bradford.

Johnson, L. Buckley, J. G. Harley, C. and Elliott, D. B. (2008). Use of single-vision eyeglasses improves stepping precision and safety when elderly habitual multifocal wearers negotiate a raised surface, *Journal of the American Geriatrics Society,* **56,**178-180.

Johnson, L. J. Elliott, D. B. and Buckley, J. G. (2009). Effects of gaze strategy on standing postural stability in older multifocal wearers, *Clinical and Experimental Optometry,* **92,** 19-26.

Jones, S. F. Twigg, P. C. Scally, A. and Buckley, J. G. (2006). The mechanics of landing when stepping down in unilateral lower-limb amputees, *Clinical Biomechanics,* **21(2),** 184–193.

Kadaba, M. P. Ramakrishnan, H. K. and Wootten, M. E. (1990). Measurement of lower extremity kinematics during level walking, *Journal of Orthopaedic Research,* ***8*****(3),** 383-392.

Khan, M. A. and Lawrence, G. P. (2005). Differences in visuomotor control between the upper and lower visual fields, *Experimental Brain Research,* **164(3),** 395–398.

Kinnear, P. R. and Gray, C. D. (2000). SPSS for windows made simple. Hove: L. Erlbaum.

Klein, B. E. K. Moss, S. E. Klein, R. Lee, K. E. and Cruikshanks, K. J. (2003). Associations of visual function with physical outcomes and limitations 5 years later in an older population. The Beaver Dame Eye Study, *Ophthalmology,* **110,** 644–650.

Krigolson, O. and Heath, M. (2006). A lower visual field advantage for endpoint stability but no advantage for online movement precision, *Experimental Brain Research,* **170(1),** 127–135.

Kuyk, T. Elliott, J. L. and Fuhr, P. S. (1998). Visual correlates of obstacle avoidance in adults with low vision, *Optometry and Vision Science,* **75,** 174–82.

Lark, S. D. Buckley, J. G. Bennett, S. Jones, D. and Sargeant, A. J. (2003). Joint torques and dynamic joint stiffness in elderly and young men during stepping down, *Clinical Biomechanics* **18,** 848–855.

Latash, M. L. and Zatsiorsky, V. M. (1993). Joint stiffness: Myth or reality? *Human Movement Science,* **12,** 653–692.

Leat, S. J. and Woodhouse, J. M. (1993). Reading performance with low vision aids: relationship with contrast sensitivity, *Ophthalmic and Physiological Optics,* **13(1),** 9-16.

Lee, D. N. Lishman, J. R. and Thomson, J. A. (1982). Regulation of gait in long jumping, *Journal of Experimental Psychology: Human Perception and Performance*, **8(3),** 448-459.

Levine, M. W. and McAnany, J. J. (2005). The relative capabilities of the upper and lower visual hemifields, *Vision Research,* **45(21),** 2820–2830.

Li, W. X. and Matin, L. (2005). Two wrongs make a right: linear increase of accuracy of visually-guided manual pointing, reaching, and height-matching with increase in hand-to-body distance, *Vision Research,* **45,** 533–550.

Liebermann, D. G. and Goodman, D. (2006). Pre-landing muscle timing and post-landing effect of falling with continuous vision and in blindfold conditions, *Journal of Electromyogrpahy and Kinesiology,* **17,** 212-227.

Loftus, A. Servos, P. Goodale, M. A. Mendarozqueta, N. and Mon-Williams, M. (2004). When two eyes are better than one in prehension: monocular viewing and end-point variance, *Experimental Brain Research,* **158(3),** 317-327.

Lord, S. R. Clark, R. D. and Webster, I. W. (1991a). Visual acuity and contrast sensitivity in relation to falls in an elderly population, *Age and Ageing,* **20,** 175-181.

Lord, S. R. Clarke, R. D. and Webster, I. W. (1991b). Postural stability and associated physiological factors in a population of aged persons, *Journal of Gerontology,* **46,** M57-66.

Lord, S. R. Ward, J. A. Williams, P. and Anstey, K. J. (1993). An epidemiologic-study of falls in older community-dwelling dwelling women-the Randwick falls and fractures study, *Australian Journal of Public Health,* **17,** 240–245.

Lord, S. R. and Dayhew, J. (2001). Visual risk factors for falls in older people, *Journal of the American geriatric society,* **49,** 508-515.

Lord, S. R. Dayhew, J. and Howland, A. (2002). Multifocal glasses impair edge contrast sensitivity and depth perception and increase the risk of falls in older people, *Journal of the American Geriatrics Society,* **50,**1760–1766.

Lord S.R. (2006). Visual risk factors for falls in older people, *Age and Ageing,* **35(Suppl 2),** ii42–5.

Lovie–Kitchin, J. E. (1988). Validity and reliability of visual acuity measurements, *Ophthalmic and Physiological Optics,* **8,** 363–370.

Lovie–Kitchin, J. Mainstone, J. Robinson, J. and Brown, B. (1990). What areas of the visual field are important for mobility in low vision patients? *Clinical Vision Science,* **5,** 249–264.

Lyon, I. N. and Day, B. L. (1997). Control of frontal plane body motion in human stepping, *Experimental Brain Research,* **115,** 345-356.

Lythgo, N. Begg, R. and Best, R. (2007). Stepping responses made by elderly and young female adults to approach and accommodate known surface height changes, *Gait Posture,* **26,** 83–89.

Macuga, K. L. Loomis, J. M. Beall, A. C. and Kelly, J. W. (2006). Perception of heading without retinal optic flow, *Perception and Psychophysics,* **68(5),** 872-878.

Maehara, G. Okubo, M. and Michimata, C. (2004). Effects of background color on detecting spot stimuli in the upper and lower visual fields, *Brain and Cognition,* **55,** 558–563.

Marigold, D. S. and Patla, A. E. (2002). Strategies for dynamic stability during locomotion on a slippery surface: effects of prior experience and knowledge, *Journal of Neurophysiology,* **88,** 339–353.

Marigold, D. S. and Patla, A. E. (2007). Gaze fixation patterns for negotiating complex ground terrain, *Neuroscience,* **144,** 302–313.

Marigold, D. S. Weerdesteyn, V. Patla, A. E. and Duysens, J. (2007). Keep looking ahead? Re-direction of visual fixation does not always occur during an unpredictable obstacle avoidance task, *Experimental Brain Research,***176,** 32-42.

Marigold, D. S. (2008). Role of peripheral visual cues in online visual guidance of locomotion, *Exercise Sport Science Review,* **36(3),** 145-151.

Marigold, D. S. and Patla, A. E. (2008). Visual information from the lower visual field is important for walking across multi-surface terrain, *Experimental Brain Research,* **188(1),** 23-31.

Marron, J. A. and Bailey, I. L. (1982). Visual factors and orientation-mobility performance. *American Journal of Optometry and Physiological Optics,* **59,** 413–426.

Mather, G. (2006). <u>Foundations of Perception.</u> East Sussex: Psychology Press Ltd.

McClure, E. (1988). Visual Fields. In Edwards, K. and Llewellyn, R. eds. <u>Optometry.</u> London, England: Butterworths.

McFayden, B. J. and Winter, D. A. (1988). An integrated biomechanical analysis of normal stair ascent and descent, *Journal of Biomechanics*, **21,** 733–744.

McMahon, T. A. Valiant, G. and Frederick, E. C. (1987). Groucho running, *Journal of Applied Physiology*, **62,** 2326–2337.

McMahon, T. A. and Cheng, G. C. (1990). The mechanics of running: how does stiffness couple with speed? *Journal of Biomechanics,* **23(Suppl 1),** 65–78.

Melzer, I. Benjuya, N. and Kaplanski, J. (2004). Postural stability in the elderly: a comparison between fallers and non-fallers, *Age Ageing*, **33,** 602–607.

Menant, J. C. St George, R. J. Sandery, B. Fitzpatrick, R. C. and Lord, S. R. (2009). Older people contact more obstacles when wearing multifocal glasses and performing a secondary visual task, *Journal of the American Geriatrics Society,* **57,** 1833-1838.

Miall, R. C. and Wolpert, D. M. (1996). Forward models for physiological motor control, *Neural Networks,* **9,** 1265-1279.

Mille, M. L. Johnson, M. E. Martinez, K. M. Rogers, M. W. (2005). Age-dependent differences in lateral balance recovery through protective stepping, *Clinical Biomechanics,* **20,** 607–616.

Milner, A. D. and Goodale, M. A. (1995). The visual brain in action*.* Oxford, England: Oxford University Press.

Milner, A. D. Dijkerman, H.C. and Rogers, B. J. (1998). Motion parallax can provide information for visually-guided action in visual form agnosia, *Society for Neuroscience abstracts,* **24,** 2097.

Mohagheghi, A. A. Moraes, R. and Patla, A. E. (2004). The effects on distant and on-line visual information on the control of approach phase and set over obstacle during locomotion, *Experimental Brain Research,* **155,** 459-468.

Morin, J-B. Samozino, P. and Peyrot, N. (2009). Running pattern changes depending on the level of subjects' awareness of the measurements performed: A ''sampling effect'' in human locomotion experiments? *Gait & Posture*, **30,** 507–510.

Mullen, K. T. (1991). Colour vision as a post-receptoral specialization of the central visual field, *Vision Research,* **31,** 119–130.

National Safety Council. (1999). Injury Facts. The National Safety Council, Illinois.

Nigg, B. M. and Liu, W. (1999). The effect of muscle stiffness and damping on simulated impact force peaks during running, *Journal of Biomechanics,* **32(8),** 849-856.

Nigg, B. M. (2007). Modelling. In Nigg. B. M. and Herzog. W. eds. Biomechanics of the Musculo-skeletal System. United States: Wiley.

Noda, H. and Adey, W. R. (1974). Retinal ganglion cells of the cat transfer information on saccadic eye movement and quick target motion, *Brain Research,* **70,** 340–345.

Owusu, W. Willet, W. Ascherio, A. Spiegelman, D. Rimm, E. Feskanich, D. and Colditz, G. (1998). "Body anthropometry and the risk of hip and wrist fractures in men: Results from a prospective study," *Obesity Research*, **6,** 12–19.

Oyster, J. (1999). The Human Eye, Structure and Function. U.S.A: Sinauer Associates, Inc.

Palmer, S. E. (1999). Vision science; photons to phenomenology. Cambridge, England: The MIT Press.

Passaglia, C. L. Enroth-Cugell, C. and Troy, J. B. (2001). Effects of remote stimulation on the mean firing rate of cat retinal ganglion cells, *Journal of Neuroscience,* **21,** 5794–5803.

Patla, A. E. Prentice, S. D. Robinson, C. and Neufeld, J. (1991). Visual control of locomotion: strategies for changing direction and for going over obstacles, *Journal of Experimental Psychology Human Perception and Performance*, **17(3),** 603-634.

Patla, A. E. and Rietdyk, S. (1993). Visual control of limb trajectory over obstacles during locomotion: effect of obstacle height and width, *Gait and Posture,* **1,** 45-60.

Patla, A. E. Adkin, A. Martin, C. Holden, R. and Prentice, S. (1996). Characteristics of voluntary visual sampling of the environment for safe locomotion over different terrains, *Experimental Brain Research,* **112,** 513-522.

Patla, A. E. and Goodale, M. A. (1996). Obstacle avoidance during locomotion is unaffected in a patient with visual form agnosia, *Neuroreport,* **8,** 165-168.

Patla, A. E. (1997). Understanding the roles of vision in the control of human locomotion, *Gait and Posture,* **5,** 54-69.

Patla, A. E. and Vickers, J. N. (1997). Where and when do we look as we approach and step over an obstacle in the travel path? *Neuroreport,* **8,** 3661-3665.

Patla, A. E. (1998). How is human gait controlled by vision, *Ecological Psychology,* **10,** 287-302.

Patla, A. E. Niechwiej, E. Racco, V. and Goodale, M. A. (2002). Understanding the contribution of binoccular vision to the control of adaptive locomotion, *Journal of Experimental Brain Research,* **142,** 551-561.

Patla, A. E. and Vickers, J. N. (2003). How far ahead do we look when required to step on specific locations in the travel path during locomotion? *Experimental Brain Research,* **148,** 133-138.

Patla, A. E. and Greig, M. (2006). Any way you look at it, successful obstacle negotiation needs visually guided on-line foot placement regulation during the approach phase, *Neuroscience Letters,* **397,** 110-114.

Payne, W. H. (1967). Visual reaction times on a circle about the fovea, *Science,* **155(761),** 481–482.

Pelli, D. G. Robson, J. G. and Wilkins, A. J. (1998). The design of a letter chart for measuring contrast sensitivity, *Clinical Vision Science*, **2,** 187–199.

Pijnappels, M. Bobbert, M. F. van Dieën, J. H. (2001). Changes in walking pattern caused by the possibility of a tripping reaction, *Gait and Posture*, **14(1),** 11-18.

Pijnappels, M. Bobbert, M. F. van Dieën, J. H. (2006). EMG modulation in anticipation of a possible trip during walking in young and older adults, *Journal of Electromyography and Kinesiology,* **16,** 137–143.

Plainis, S. and Murray, I. J. (2002), Reaction times as an index of visual conspicuity when driving at night, *Ophthalmic and Physiological Optics,* **22(5),** 409-415.

Plainis, S. Murray, I. J. and Charman, W. N. (2005). The role of retinal adaptation in night driving, *Ophthalmic and Physiological Optics,* **82(8),** 682-688.

PLATO Visual Occlusion Spectacles (2010) Translucent technologies Inc. [online] Available from http://www.translucent.ca/plato.html [accessed 4 April 2010].

Podsiadlo, D. and Richardson, S. (1991). The timed “Up & Go”: a test of basic functional mobility for frail elderly persons, *Journal of the American Geriatrics Society,* **39,** 142–148.

Powell-Smith, V. and Billington, M. J. (1986). The building regulations explained and illustrated. London: Collins Professional books.

Qu, X. and Nussbaum, M. (2009). Effects of external loads on balance control during upright stance: Experimental results and model-based predictions, *Gait & Posture*, **29(1),** 23-30.

Quevedo, A. A. F. Patla, A. E. and Cliquet, A. Jr. (1997). Motor control for overcoming obstacles: Changes in obstacle height during the swing phase, *Medical and Biological Engineering and Computing,* **3(Suppl. 1),** 175.

Reeves, B. C. Wood, J. M. and Hill, A. R. (1993). Reliability of high- and low-contrast letter charts, *Ophthalmic and Physiological Optics*, **13(1),**17-26.

Reynolds, R. F. and Day, B. L. (2005). Visual guidance of the human foot during a step, *Journal of Physiology,* **569,** 677-684.

Rhea, C. K. and Rietdyk, S. (2007). Visual exterocpetive information provided during obstacle crossing did not modify the lower limb trajectory, *Neuroscience Letters*, **418(1),** 60-65.

Rietdyk, S. and Rhea, C. K. (2006). Control of adaptive locomotion: effect of visual obstruction and visual cues in the environment, *Experimental Brain Research,* **169(2),** 272–278.

Robertson, D. G. E. Caldwell, G. E. Hamill, J. Kamen, G. and Whittlesey, S. N. (2004). Research Methods in Biomechanics. United States: Human Kinetics.

Rodieck, R. W. (1998). The First Steps in Seeing. Sunderland, Massachusetts: Sinauer Associates, Inc.

Roys, M. S. (2001). Serious stair injuries can be prevented by improved stair design, *Applied Ergonomics,* **32,** 135-139.

Rubin, G. S. Bandeen-Roche, K. Huang, G. H. Muñoz, B. Schein, O. D. Fried, L. P. and West, S. K. (2001). The association of multiple visual impairments with subjective physical disability: SEE Project, *Investigative Ophthalmology and Vision Science,* **42**, 64-72.

Rubin, G. S. Ng, E. S. W. Bandeen-Roche, K. Key, P. M. Freeman, E. E. and West, S. K. (2007). A prospective, population-based study of the role of visual impairment in motor vehicle crashes among older drivers: The SEE study, *Investigative Ophthalmology and Vision Science,* **48(4),** 1483-1491**.**

Rushton, S. K. Harris, J. M. Lloyd, M. R. and Wann, J. P. (1998). Guidance of locomotion on foot uses perceived target location rather than optic flow, *Current Biology,* **8,** 1191-1194.

Ryan, T. (2007). Modern engineering statistics. Chichester: Wiley.

Saini, M. Kerrigan, D. C. Thirunarayan, M. A. and Duff-Raffaele, M. (1998). The vertical displacement of the centre of mass during walking: a comparison of four measurement methods, *Journal of Biomechanical Engineering,* **120,** 133-139.

Santello. M. McDonagh, M. J. N. and Challis, J. H. (2001). Visual and non-visual control of landing movements in humans, *Journal of physiology,* **537,** 313-327.

Schubert, M. Bohner, C. Berger, W. Sprundel, M. V. and Duysens, J. E. J. (2003). The role of vision in maintaining heading direction: effects of changing gaze and optic flow on human gait, *Experimental Brain Research,* **150,** 163-173.

Sherwood, M. B. Garcia-Siekavizza, A. Meltzer, M. I. Hebert, A. Burns, A. F. and McGorray, S. (1988). Glaucoma's impact on quality of life and its relation to clinical indicators, *Ophthalmology,* **105,** 561-566.

Shimba, T. (1984). An estimation of centre of gravity from force plate data, *Journal of Biomechanics,* **17,** 53-60.

Shumway-Cook, A. Brauer, S. and Woollacott, M. (2000). Predicting the probability for falls in community-dwelling older adults using the Timed Up & Go Test, *Physical Therapy,* **80,** 896–903.

Siegmund, G. P. Heiden, T. L. Sanderson, D. J. Inglis, J. T. and Brault, J. R. (2006). The effect of subject awareness and prior slip experience on tribometer-based predictions of slip probability, *Gait & Posture,* **24,**110–119.

Simoneau, G. G. Cavanagh, P. R. Ulbrecht, J. S, Leibowitz, H. W. and Tyrrell, R. A. (1991). The influence of visual factors on fall-related kinematic variables during stair descent by older women, *Journal of Gerontology,* **46,** M188–195.

Spanjaard, M. Reeves, N. D. van Dieën, J. H. Baltzopoulos, V. and Maganaris, C. N. (2008). Lower-limb biomechanics during stair descent: influence of step-height and body mass, *The Journal of Experimental Biology,* **211,** 1368-1375.

Stacoff, A. Dieze, C. Luder, G. Stussi, E. and Kramers-De Quervain, I. A. (2005). Ground reaction forces on stairs: effects of stair inclination and age, *Gait and Posture,* **21,** 24-38.

Startzell, J. K. Owens, A. D. Mulfinger, L. M. and Cavanagh, P. R. (2000). Stair negotiation in older people: a review. *Journal of the American Geriatrics Society,* **48,** 567–580.

Templer, J. (1992). The staircase. Studies of Hazards, falls and safer design. Cambridge: The MIT press.

The human eye (2010) Optics [online] Available from: http://images.google.co.uk/imgres?imgurl=http://www.internal.schools.net.au/edu/lesson_ideas/optics/images/rods_cones.gif&imgrefurl=http://www.internal.schools.net.au/edu/lesson_ideas/optics/optics_wksht2_p1.html&usg=__k9j1I9a5aKX9boXH6vx05aziUYU=&h=309&w=250&sz=6&hl=en&start=12&um=1&itbs=1&tbnid=-MAUyGF5pswqvM:&tbnh=117&tbnw=95&prev=/images%3Fq%3Dphotoreceptor%2Brods%2Band%2Bcones%26um%3D1%26hl%3Den%26tbs%3Disch:1 [accessed 6 April 2010].

The Public Service Vehicles Regulations (2002) Department for transport [online] Available from http://www.dft.gov.uk/transportforyou/access/buses/pubs/conduct/sstopublicservicevehicle5990.pdf [accessed 11 February 2010].

Thomson, J. A. (1983). Is continuous visual monitoring necessary in visually guided locomotion? *Journal of Experimental Psychology,* **9,** 427-443.

Thylefors, B. Négrel, A. D. Pararajasegaram, R. and Dadzie, K. Y. (1995). Global data on blindness, *Bulletin of the World Health Organisation,* **73,**115–121.

Thylefors, B. A. (1999). Mission for vision, *Lancet,* **354(Suppl),** SIV44.

Tinetti, M. E. Speechley, M. and Ginter, S. F (1988). Risk factors for falls among elderly persons living in the community, *New England Journal of Medicine,* **319,** 1701–1707.

Tresilian, J. R. (1999). Visually timed action: time-out for 'tau'? *Trends in Cognitive Science,* **3(8),** 301–310.

Turano, K. Rubin, G. S. Herdman, S. J. Chee, E. and Fried, L. P. (1994). Visual Stabilization of Posture in the Elderly: Fallers vs. Nonfallers, *Optometry and Vision Science,* **71(12),** 761-769.

Turano, K. A. Rubin, G. S. and Quigley, H. A. (1999). Mobility Performance in Glaucoma, *Investigative Ophthalmology and Vision Science,* **40,** 2803-2809.

Turano, K. A. Broman, A. T. Bandeen-Roche, K. Munoz, B. Rubin, G. S. and West, S. K. (2004). Association of Visual Field Loss and Mobility Performance in Older Adults: Sailsbury Eye Evaluation Study, *Optometry and Vision Science,* **81(5),** 298-307.

Turano, K. A. Yu, D. and Hao, L. (2005). Optic-flow and egocentric-direction strategies in walking: Central vs peripheral visual field, *Vision Research,* **45 (25-26),** 3117-3132.

UK DVLA visual standards (2010) Royal College of Ophthalmologists Guidelines [online] Available from: http://www.mrcophth.com/focus1/Visual%20Standards%20for%20Driving.htm [accessed 4 April 2010].

Ungerleider, L. G. and Mishkin, M. (1982). Two cortical visual streams. In Ingle, D. J. and Goodale, M. A. eds. Analysis of visual behavior. United States: The MIT press, pp.549-589.

Vale, A. Scally, A. Buckley, J. G. and Elliott, D. B. (2008a). The effects of monocular refractive blur on gait parameters when negotiating a raised surface, Ophthalmic and Physiological Optics, **28,** 135–142.

Vale, A. Buckley, J. G. and Elliott, D. B. (2008b). Gait Alterations Negotiating A Raised Surface Induced by Monocular Blur, *Optometry and Vision Science,* **85(12),** 1128-1134.

van der Linden, M. Marigold, D. S. Gabreels, F. J. M. and Duysens, J. (2007). Muscle reflexes and synergies triggered by an unexpected support surface height during walking, *Journal of Neurophysiology,* **97,** 3639–3650.

van Dieën, J. H. Spanjaard, M. Konemann, R. Bron, L. and Pijnappels, M. (2008). Mechanics of toe and heel landing in stepping down in ongoing gait, *Journal of Biomechanics,* **41,** 2417–2421.

Vicon Motion Systems Preparation Manual (2010) Vicon Motion Systems [online] Available from http://www.udel.edu/PT/Research/MAL/preparation_v1_2.pdf [accessed 7 April 2010].

Vision (2010) The Brain from top to bottom [online] Available from: http://thebrain.mcgill.ca/flash/a/a_02/a_02_cr/a_02_cr_vis/a_02_cr_vis.html [accessed 6 April 2010].

Von Noorden, G. K. and Campos, E. C. (2002). Binocular vision and ocular motility: theory and management of strabismus. St. Louis: Mosby.

Walsh, G. (2009). Vertical diplopia on downgaze with bifocals, *Optometry and Vision Science,* **86(9),** 1112-1116.

Wandell, B. A. (1995). Foundations of vision. U.S.A: Sinauer Associates, Inc.

Warren, W. H. Young, D. S. and Lee, D. N. (1986). Visual control of step length during running over irregular terrain, *Journal of Experimental Psychology, Human Perception and Performance,* **12(3),** 259-266.

Warren, W. H. (1998). Visually controlled locomotion: 40 years later, *Ecological. Psychology,* **10,** 177–219.

Warren, W. H. Kay, B. A. Zosh, W. D. Duncan, A. P. and Sahuc, S. (2001). Optic flow is used to control human walking, *Nature Neuroscience,* **4,** 213-216.

Warren, W .H. (2009). How do animals get about by vision? Visually controlled locomotion and orientation after 50 years, *The British Psychological Society,* **100,** 277-281.

Whittle, M. W. (1997). Three-dimensional motion of the centre of gravity of the body during walking, *Human Movement Science,* **16,** 347-355.

Wolfe, J. M. Kluender, K. R. Levi, D. M. Bartoshuk, L. M. Herz, R. S. Klatzky. R. L. and Nad Lederman, S. J. (2006). Sensation and perception. Sunderland, Massachusetts: Sinauer Associates, Inc.

Wood, J. M. and Troutbeck, R. (1995). Elderly drivers and simulated visual impairment, *Optometry and Vision Science,* **72,** 115–124.

Zachazewski, J. E. Reily, P. O. and Krebs, D. E. (1993). Biomechanical analysis of body mass transfer during stair ascent and descent of healthy subjects, *Journal of rehabilitation research,* **30,** 412-422.

Zatsiorsky, V. M. and Seluyanov, V. N. (1983). The mass and inertia charachteristics of the main segments of the human body. In Matsui, H. and Kobayashi, K. eds. Biomechanics VIII-B*.* Champaign, IL: Human Kinetics.

Zelaznik, H. N. Hawkins, B. and Kisselburgh, L. (1983). Rapid visual feedback processing in single-aiming movements, *Journal of Motor Behavior*, **15,** 217-236.

Appendices

Appendix 1: Participant information sheet used in chapter 4

Study title: Visuomotor control of step descent: evidence of specialised role of the lower visual field

You are being invited to take part in a research study. Before you decide, it is important for you to understand why the research is being done and what it will involve. Please take time to read the following information carefully and discuss it with friends and relatives. Ask us if there is anything that is not clear or if you would like more information. Take time to decide whether or not you wish to take part.

Purpose of the Study

Visuomotor control of step descent: evidence of specialised role of the lower visual field

Why have I been chosen?

You have been chosen because you are a young healthy individual that would be suitable to undertake the study.

Do I have to take part?

It is up to you to decide whether or not to take part. If you do decide to take part you will be given this information sheet to keep and be asked to sign a consent form. If you decide to take part, you are still free to withdraw from the study at any time and without giving a reason.

What will happen to me if I take part?

You will be asked to visit the Biomechanics Laboratory at the Department of Optometry, University of Bradford (this is within the university campus, off Richmond Road).

The visit to the Biomechanics Laboratory will be a onetime event lasting for ~2 hours. During your visit you will be asked to undertake the following:

- 3 simple eye tests
- Various stepping tasks that are encountered everyday under different visual conditions

The eye tests will consist of reading various letter charts and assessing depth perception through images presented in three dimensions (3-D).

The stepping tasks will require you to step down from different step heights with vision being removed at different times within the movement. In addition, some trials within the study will require you to step down without being able to see your lower body. During these tests you will have a number of small spherical markers placed on your clothing and cameras will track your body movements.

What do I have to do?

All you will be asked to do is turn up to the laboratory bringing shorts and flat, soft-soled shoes.

Is there any risk of harm to myself?

There is a hypothetical risk of you losing your balance when performing the physical tests, but we have never had anybody do so after many of these studies. In addition, a research assistant will stand next to you when you perform the tests and will help you (if required) to regain your balance. The eye tests are regularly used by opticians in general practice, and the physical tests are similar to ‘everyday’ tasks.

What should I do if I would like to help with the study?

Contact, by telephone or email, any of the researchers listed at the end of this form. The results of this study will be used for research purposes. If the research is published, you will remain anonymous.

Further information: if you would like more information about the study and what is being asked of you please contact
Matthew Timmis at Bradford University (tel. 01274 235926) or email at
M.A.Timmis@brad.ac.uk.

Thank you for reading this information sheet.

Research Team:
Dr John Buckley, Research Associate, Dept of Optometry, University of Bradford, Richmond Road, Bradford. BD7 1DP: tel 01274 234641, email j.buckley@bradford.ac.uk.

Appendix 2: Fitness and Body Image Health Questionnaire

Please circle the appropriate responses

Are you taking any medications or drugs that will prevent you from participating in this study?

Yes or No

Do you now, or have you had in the past 5 years:	
1. History of heart problems, chest pain or stroke.	Yes or No
2. Increased blood pressure.	Yes or No
3. Any chronic illness or condition.	Yes or No
4. Difficulty with physical exercise.	Yes or No
5. Advice from physician not to exercise.	Yes or No
6. Recent surgery (last 12 months).	Yes or No
7. Pregnancy (now or within last 3 months).	Yes or No
8. History of breathing or lung problems.	Yes or No
9. Muscle, joint, or back disorder, or any previous injury still affecting you.	Yes or No
10. Diabetes or thyroid condition.	Yes or No
11. Obesity (more than 20% over ideal body weight).	Yes or No
12. Increased blood cholesterol.	Yes or No
13. History of heart problems in immediate family.	Yes or No
14. Has your doctor ever said your blood pressure was too high?	Yes or No
15. Shortness of breath with or without exercise	Yes or No
16. Has your physician ever said you have heart trouble?	Yes or No
17. Do you often feel faint or have spells of severe dizziness?	Yes or No
18. Are you unaccustomed to vigorous exercise?	Yes or No
19. Has your doctor ever told you that you have a bone or joint problem that has been or could be made worse by exercise?	Yes or No
20. On average, do you participate in a form of physical	Yes or No

activity for 30 minutes five times per week?	
20. What types of exercise do you participate in? (please tick) Walking _____ Jogging _____ Swimming _____ Cycling _____ Gardening _____ Stationary Biking _____ Tennis _____ Other Aerobic ________________________	
Date ___/___/___ Signature __________________________	

Appendix 3: Consent form

Bradford University

Visuomotor control of step descent: evidence of specialised role of the lower visual field

Please print in block capitals

I, (**Subject's** full name)*..agree to take part in the above named project, the details of which have been fully explained to me and described in writing.

Signed (Subject) ....   Date ....

I, (**Investigator's** full name)* ....certify that the details of this project / procedure have been fully explained and described in writing to the subject named above and have been understood by him / her.

Signed (Investigator)..............................   Date ....

Appendix 4: Participant Details

Participant name…………………………………

Age……………………… Gender…………………………………

Height…………………… Mass…………………………………

Dominant Foot…………………………………

2nd toe and floor

Right…………………………………

Left…………………………………

2nd toe and shoe tip

Right…………………………………

Left…………………………………

Ankle Width
Right…………………………………

Left…………………………………

Knee width
Right…………………………………

Left…………………………………

Leg length
Right…………………………………

Left…………………………………

Inter ASIS distance…………………………………

Binocular Visual Acuity…………………………………

Binocular contrast sensitivity…………………………………

Stereopsis TNO tests…………………………………

Appendix 5: Macro used to analyse data in chapter 4

	Start Here
	Increment Line_Nb
Variables Analysed in this	=SELECT("Loop_1")
Macro:SEE COLUMN I	=FORMULA(Loop_1+1)
	=SELECT("Loop_2")
	=FORMULA(loop_2+1)
Subject_Name	=INDEX(C1:C3,loop_2)
File_Name_vel	=INDEX(E1,Loop_1)
	=NEW(5)
	=DIRECTORY("k:\Phd\macro\"&Subject_Name)
	=SAVE.AS("results\"&File_Name_Vel)
File_Name_text	=INDEX(D1:D1,Loop_1)
	=DIRECTORY("k:\Phd\macro\"&Subject_Name)
	=OPEN.TEXT("text\"&File_Name_text)
	=ACTIVATE()
	=SELECT("R1")
	=FORMULA.FIND("Time",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C1")
P_ForcePlate1:Z	=PASTE()
P_ForcePlate1:Y	
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column LHEE	=FORMULA.FIND("LHEE:Y",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)

	=SELECT("R1C3")
	=PASTE()
LHEEY	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column RHEE	=FORMULA.FIND("RHEE:Y",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C6")
	=PASTE()
RHEEY	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column ForcePlate1:z	=FORMULA.FIND("F_ForcePlate1:Z",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C9")
	=PASTE()
FZ1	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column ForcePlate2:z	=FORMULA.FIND("F_ForcePlate2:Z",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C12")
	=PASTE()
FZ2	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column Lheel:Z	=FORMULA.FIND("LHEE:Z",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")

	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C18")
	=PASTE()
LHEE:Z	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column RHeel:Z	=FORMULA.FIND("RHEE:Z",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C24")
	=PASTE()
RHEE:Z	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column RKnee:X	=FORMULA.FIND("RKNEEAngles:X",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C29")
	=PASTE()
RKNEEangles:X	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column LKnee:X	=FORMULA.FIND("LKneeAngles:X",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C33")
	=PASTE()
LKNEEangles:X	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column LAnk:X	=FORMULA.FIND("LAnkleAngles:X",1,2,1,1,FALSE)

	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C40")
	=PASTE()
LANK:X	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column RAnk:X	=FORMULA.FIND("RAnkleAngles:X",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C46")
	=PASTE()
RANK:X	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column CofM:Z	=FORMULA.FIND("CentreofMass:Z",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C52")
	=PASTE()
CofM:Z	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column LTOE:Y	=FORMULA.FIND("LTOE:Y",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C58")
	=PASTE()
LTOE:Y	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)

	=SELECT("R1")
finds column RTOE:Y	=FORMULA.FIND("RTOE:Y",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C64")
	=PASTE()
RTOE:Y	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column CofM:Y	=FORMULA.FIND("CentreofMass:Y",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C70")
	=PASTE()
CofM:Y	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column LheadAngle Z	=FORMULA.FIND("LHeadAngles:X",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C94")
	=PASTE()
LheadAngle:x	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column RheadAngle Z	=FORMULA.FIND("RHeadAngles:X",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C99")
	=PASTE()

RheadAngle:X	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column RToeZ	=FORMULA.FIND("RToe:Z",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C110")
	=PASTE()
	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column LToeZ	=FORMULA.FIND("LToe:Z",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C115")
	=PASTE()
LToe:Z	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column Cof P x	=FORMULA.FIND("P_ForcePlate2:X",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C135")
	=PASTE()
Cof P x	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column Cof P y	=FORMULA.FIND("P_ForcePlate2:Y",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)

	=SELECT("R1C140")
	=PASTE()
Cof P y	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column CofM X	=FORMULA.FIND("CentreofMass:X",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C125")
	=PASTE()
Cof M X	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column RANK Z	=FORMULA.FIND("RANK:Z",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C160")
	=PASTE()
RANK Z	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column LANK Z	=FORMULA.FIND("LANK:Z",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C165")
	=PASTE()
LANK Z	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column RKNE Z	=FORMULA.FIND("RKNE:Z",1,2,1,1,FALSE)

	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C170")
	=PASTE()
RKNE Z	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column LKNE Z	=FORMULA.FIND("LKNE:Z",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C175")
	=PASTE()
LKNE Z	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column RKNE Y	=FORMULA.FIND("RKNE:Y",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C200")
	=PASTE()
RKNE Y	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column LKNE Z	=FORMULA.FIND("LKNE:Y",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C201")
	=PASTE()
LKNE Y	=COLUMN(ACTIVE.CELL())

	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column LANK Y	=FORMULA.FIND("LANK:Y",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C202")
	=PASTE()
LANK Y	=COLUMN(ACTIVE.CELL())
	=ACTIVATE(File_Name_text)
	=SELECT("R1")
finds column RANK Y	=FORMULA.FIND("RANK:Y",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=COPY()
	=ACTIVATE(File_Name_Vel)
	=SELECT("R1C203")
	=PASTE()
RANK Y	=COLUMN(ACTIVE.CELL())
	=SELECT("r1C1")
	=SELECT.END(4)
End_Block	=ROW(ACTIVE.CELL())
	=SELECT("R1")
finds start and end on fz1	=FORMULA.FIND("F_ForcePlate1:Z",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")
	=FORMULA.FIND(".",1,2,1,1,FALSE)
start_FP1	=ROW(ACTIVE.CELL())
	=SELECT("R"&start_FP1&"c9:r"&End_Block&"c9")
	=SELECT.END(4)
End_FP1	=ROW(ACTIVE.CELL())
	=SELECT("R1")
finds start and end of fz2	=FORMULA.FIND("F_ForcePlate2:Z",1,2,1,1,FALSE)
	=COLUMN(ACTIVE.CELL())
	=SELECT("c")

	=FORMULA.FIND(".",1,2,1,1,FALSE)
start_FP2	=ROW(ACTIVE.CELL())
locates end of toe off trail leg	=FORMULA.FIND("0",2,1,1,1,FALSE)
End_FP2	=ROW(ACTIVE.CELL())
Calculation instant of	
heel-off of left foot	=SELECT("R"&start_FP1-150&"c19")
selects data between fz1 and 120 cells prior to fz1	=FORMULA("=average(r"&start_FP1-200&"c[-1]:r"&start_FP1-300&"c[-1])")
works out heel off by using the fz1 platform	=FILL.AUTO("RC:R"&End_FP1&"C")
	=SELECT("R"&start_FP1-150&"c20")
difference between average and actual marker	=FORMULA("=(rc[-2]-rc[-1])")
	=FILL.AUTO("RC:R"&End_Block&"C")
	=SELECT("R"&start_FP1-150&"c21")
determines if heel rises 1mm above average	=FORMULA("=IF(RC[-1]>3,2,1)")
	=SELECT("R"&start_FP1-150&"c21")
	=FILL.AUTO("RC:R"&End_Block&"C")
	=SELECT("R"&start_FP1-150&"c22")
	=FORMULA("=sum(R[4]C[-1]:rc[-1])")
	=SELECT("R"&start_FP1-150&"c22")
	=FILL.AUTO("RC:R"&End_Block&"C")
	=FORMULA.FIND("10",2,1,1,1,FALSE)
leftlead_heeloff	=ROW(ACTIVE.CELL())
Calculation instant of heel-off of right foot	=SELECT("R"&start_FP1-150&"c25")
selects data between fz1 and 120 cells prior to fz1	=FORMULA("=average(r"&start_FP1-200&"c[-1]:r"&start_FP1-300&"c[-1])")
this works out heel off by using the fz1 platform	=FILL.AUTO("RC:R"&End_Block&"C")
	=SELECT("R"&start_FP1-150&"c26")
difference between average and actual marker	=FORMULA("=(rc[-2]-rc[-1])")
	=FILL.AUTO("RC:R"&End_Block&"C")
	=SELECT("R"&start_FP1-150&"c27")
determines if heel rises 1mm above average	=FORMULA("=IF(RC[-1]>3,2,1)")
	=SELECT("R"&start_FP1-150&"c27")
	=FILL.AUTO("RC:R"&End_Block&"C")
	=SELECT("R"&start_FP1-150&"c28")
	=FORMULA("=sum(R[4]C[-1]:rc[-1])")
	=SELECT("R"&start_FP1-150&"c28")
	=FILL.AUTO("RC:R"&End_Block&"C")
	=FORMULA.FIND("10",2,1,1,1,FALSE)

rightlead_heeloff	=ROW(ACTIVE.CELL())
Calculation instant of toe off left foot	=SELECT("R"&leftlead_heeloff-100&"c59")
THIS NEEDS TO HAVE PLUS 3 CELLS DOWN ONIT	=FORMULA("=average(r"&start_FP1-200&"c[-1]:r"&start_FP1-300&"c[-1])")
TO REFLECT THE FZ SCORE	=FILL.AUTO("RC:R"&start_FP1+150&"C")
	=SELECT("R"&leftlead_heeloff-100&"c60")
	=FORMULA("=(rc[-2]-rc[-1])")
	=FILL.AUTO("RC:R"&start_FP1+150&"C")
	=SELECT("R"&leftlead_heeloff-100&"c61")
	=FORMULA("=IF(RC[-1]<-2,2,1)")
	=FILL.AUTO("RC:R"&start_FP1+150&"C")
	=SELECT("R"&leftlead_heeloff-100&"c62")
	=FORMULA("=sum(R[4]C[-1]:rc[-1])")
	=FILL.AUTO("RC:R"&start_FP1+150&"C")
	=FORMULA.FIND("10",2,1,1,1,FALSE)
left foot toe off	=ROW(ACTIVE.CELL())
right toe off	=SELECT("R"&End_Block&"c1")
	=SELECT("R[2]c[8]")
	=FORMULA("r toe")
	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("R[1]c")
	=FORMULA("=CELL(""row"",R"&End_FP2&"C)")
Left toe off	=SELECT("R"&End_Block&"c1")
	=SELECT("R[2]c[7]")
	=FORMULA("L Toe off")
	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("R[1]c")
	=FORMULA("=CELL(""row"",R"&left_foot_toe_off&"C)")
initiation of movement	=SELECT("r"&start_FP1-250&"c136")
Average CP X	=FORMULA("=average(r"&leftlead_heeloff-200&"c[-1]:r"&leftlead_heeloff-100&"c[-1])")
	=FILL.AUTO("RC:R"&start_FP1&"C")
	=SELECT("r"&start_FP1-250&"c137")
absolute value	=FORMULA("=(rc[-2]-rc[-1])")
	=FILL.AUTO("RC:R"&start_FP1&"C")
	=SELECT("r"&start_FP1-250&"c141")

Average CP Y	=FORMULA("=average(r"&leftlead_heeloff-200&"c[-1]:r"&leftlead_heeloff-100&"c[-1])")
	=FILL.AUTO("RC:R"&start_FP1&"C")
	=SELECT("r"&start_FP1-250&"c142")
absolute value	=FORMULA("=(rc[-2]-rc[-1])")
	=FILL.AUTO("RC:R"&start_FP1&"C")
Average CM Y	=SELECT("r"&start_FP1-250&"c71")
	=FORMULA("=average(r"&leftlead_heeloff-200&"c[-1]:r"&leftlead_heeloff-100&"c[-1])")
	=FILL.AUTO("RC:R"&start_FP1&"C")
	=SELECT("r"&start_FP1-250&"c72")
absolute value	=FORMULA("=(rc[-2]-rc[-1])")
	=FILL.AUTO("RC:R"&start_FP1&"C")
Average CMX	=SELECT("r"&start_FP1-250&"c126")
	=FORMULA("=average(r"&leftlead_heeloff-200&"c[-1]:r"&leftlead_heeloff-100&"c[-1])")
	=FILL.AUTO("RC:R"&start_FP1&"C")
	=SELECT("r"&start_FP1-250&"c127")
absolute value	=FORMULA("=(rc[-2]-rc[-1])")
	=FILL.AUTO("RC:R"&start_FP1&"C")
calculates CP CM X Y divergence	=SELECT("r"&start_FP1-250&"c138")
x divergence	=FORMULA("=(rc[-1]-rc[-11])")
	=FILL.AUTO("RC:R"&start_FP1&"C")
	=SELECT("r"&start_FP1-251&"c138")
	=FORMULA("x diverg")
	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("r"&start_FP1-250&"c143")
Y divergence	=FORMULA("=(rc[-1]-rc[-71])")
	=FILL.AUTO("RC:R"&start_FP1&"C")
	=SELECT("r"&start_FP1-251&"c143")
	=FORMULA("y diverg")
	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("r"&start_FP1-251&"c138:R"&left_foot_toe_off&"c138")
	=COPY()
	=SELECT("Rc[6]")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("R[1]c[1]")

X Y divergence	=FORMULA("=sqrt(rc[-2] ^2+rc[-1] ^2)")
	=FILL.AUTO("RC:R"&start_FP1&"C")
	=SELECT("r"&start_FP1-251&"c145")
	=FORMULA("x & y diverg")
	=FONT.PROPERTIES(,,,,,,,,,3)
locates initiation of movement	=SELECT("r"&start_FP1-250&"c146")
	=FORMULA("=IF(RC[-1]>20,2,1)")
	=FILL.AUTO("RC:R"&start_FP1&"C")
	=SELECT("r"&start_FP1-250&"c147")
	=FORMULA("=sum(R[4]C[-1]:rc[-1])")
	=FILL.AUTO("RC:R"&start_FP1&"C")
	=FORMULA.FIND("10",2,1,1,1,FALSE)
l initiation of movement	=ROW(ACTIVE.CELL())
init move cell no.	=SELECT("R"&End_Block&"c1")
	=SELECT("R[13]c[7]")
	=FORMULA("=CELL(""row"",R"&l_initiation_of_movement&"C)")
	=SELECT("R[-1]c")
	=FORMULA("l.init move")
	=FONT.PROPERTIES(,,,,,,,,,3)
create chart init of movement	=SELECT("r"&start_FP1-250&"c143:R"&start_FP1&"c145")
	=CREATE.OBJECT(5,"R"&End_FP1+18&"C15",33.75,10.5,"R"&End_FP1+38&"c26",30.75,1.5,1,TRUE)
	=CHART.WIZARD(TRUE,"R"&start_FP1-250&"c143:R"&start_FP1&"c145",4,2,2,,,1,,,,,0,0)
Head flexion of LheadAngle X	=SELECT("r"&leftlead_heeloff-20&"c95")
LheadAngle:X	=FORMULA("=average(r"&l_initiation_of_movement-50&"c[-1]:r"&l_initiation_of_movement-100&"c[-1])")
	=FILL.AUTO("RC:R"&start_FP1&"C")
	=SELECT("r"&leftlead_heeloff-20&"c96")
	=FORMULA("=(rc[-1]-rc[-2])")
	=FILL.AUTO("RC:R"&start_FP1&"C")
	=SELECT("r"&leftlead_heeloff-20&"c97")
looks for head flexion greater than 8mm	=FORMULA("=IF(RC[-1]>10,2,1)")
	=FILL.AUTO("RC:R"&start_FP1&"C")
	=SELECT("r"&start_FP1+1&"c97")
	=FORMULA("=average(r"&leftlead_heeloff-20&"c:r"&start_FP1&"c)")

l_head_flex	=ROW(ACTIVE.CELL())
left Step execution time	=SELECT("r"&leftlead_heeloff&"C1")
	=COPY()
	=SELECT("r2c84")
selects time point A column	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("R"&start_FP1&"c1")
	=COPY()
	=SELECT("r3c84")
selects time point A column	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("R[1]c")
	=FORMULA("=(r[-1]c-r[-2]c)")
left leg Step execution time	=ROW(ACTIVE.CELL())
	=SELECT("Rc[-1]")
	=FORMULA("step ex time")
	=FONT.PROPERTIES(,,,,,,,,,3)
left leg Stepping distance	=SELECT("R"&start_FP1-120&"c57")
	=FORMULA("=average(r"&l_initiation_of_movement-20&"c[1]:r"&l_initiation_of_movement-120&"c[1])")
one.1	=ROW(ACTIVE.CELL())
	=SELECT("R"&start_FP1-119&"c57")
	=FORMULA("=average(r"&start_FP1&"c[1]:r"&start_FP1+20&"c[1])")
two.1	=ROW(ACTIVE.CELL())
	=SELECT("r"&one.1&"c57")
	=COPY()
	=SELECT("R"&End_Block&"c1")
	=SELECT("R[2]c[2]")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("r"&two.1&"c57")
	=COPY()
	=SELECT("R"&End_Block&"c1")
	=SELECT("R[3]c[2]")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("R[1]c")
	=FORMULA("=(r[-2]c-r[-1]c)")
left leg Stepping distance	=ROW(ACTIVE.CELL())
	=SELECT("Rc[1]")
	=FORMULA("step distance")

	=FONT.PROPERTIES(,,,,,,,,,3)
left leg knee ang fz1 contact	=SELECT("r"&l_initiation_of_movement-200&"c34")
	=FORMULA("=average(r"&l_initiation_of_movement-20&"c[-1]:r"&l_initiation_of_movement-120&"c[-1])")
L_average_knee_ang	=ROW(ACTIVE.CELL())
	=SELECT("Rc[1]")
	=FORMULA("ave knee ang")
	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("r"&start_FP1-1&"c33")
	=COPY()
	=SELECT("r"&l_initiation_of_movement-199&"c34")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("R[1]c")
	=FORMULA("=(r[-1]c-r[-2]c)")
leftleg_kneeang_fz1_contact	=ROW(ACTIVE.CELL())
	=SELECT("Rc[1]")
	=FORMULA("kneang fz cont")
	=FONT.PROPERTIES(,,,,,,,,,3)
Max knee flex after fz contact	=SELECT("R"&start_FP1&"c34")
	=FORMULA("=max(r"&start_FP1&"C33:r"&End_FP1&"c33)")
	=FILL.AUTO("RC:R"&End_FP1&"C")
	=SELECT("R"&start_FP1&"c35")
	=FORMULA("=(rc[-2]-rc[-1])")
	=FILL.AUTO("RC:R"&End_FP1&"C")
	=FORMULA.FIND("0",2,1,1,1,FALSE)
	=SELECT("rc[-2]")
left highest_knee_ang_aft_fzcont	=ROW(ACTIVE.CELL())
	=COPY()
	=SELECT("r"&leftleg_kneeang_fz1_contact+9&"C34")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("r"&L_average_knee_ang&"c34")
	=COPY()
	=SELECT("r[10]c")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("R[2]c")
	=FORMULA("=(r[-1]c-r[-2]c)")

Left_max_knee_flex_aft_fz_cont	=ROW(ACTIVE.CELL())
	=SELECT("Rc[1]")
	=FORMULA("m.k.f aft fz cont")
	=FONT.PROPERTIES(,,,,,,,,,3)
peak vertical contact force left leg	
	=SELECT("R"&start_FP1&"c8")
	=FORMULA("=max(r"&start_FP1&"C9:r"&left_highest_knee_ang_aft_fzcont&"c9)")
	=FILL.AUTO("R"&start_FP1&"C8:r"&left_highest_knee_ang_aft_fzcont&"c8")
	=SELECT("R"&start_FP1&"c7")
	=FORMULA("=(rc[2]-rc[1])")
	=FILL.AUTO("RC:R"&left_highest_knee_ang_aft_fzcont&"C")
	=FORMULA.FIND("0",2,1,1,1,FALSE)
	=SELECT("rc[2]")
left_peak_vertical_contact_force	=ROW(ACTIVE.CELL())
	=SELECT("Rc[1]")
	=FORMULA("p.v.c.f")
	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("Rc[-1]")
	=COPY()
	=SELECT("R"&End_Block&"c1")
	=SELECT("R[8]c[6]")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("R[-1]c")
	=FORMULA("l.p.v.c.f")
	=FONT.PROPERTIES(,,,,,,,,,3)
left leg body weight support	=SELECT("r"&left_highest_knee_ang_aft_fzcont&"c12")
left_leg_body_weight_supp	=ROW(ACTIVE.CELL())
	=SELECT("Rc[1]")
	=FORMULA("weigh supp")
	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("Rc[-1]")
	=COPY()
	=SELECT("R"&End_Block&"c1")
	=SELECT("R[8]c[10]")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)

	=SELECT("R[-1]c")
	=FORMULA("l.body weight sup")
	=FONT.PROPERTIES(,,,,,,,,,3)
left leg weight transfer time	=SELECT("r"&start_FP1&"C1")
	=COPY()
	=SELECT("R2c88")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("r"&End_FP2&"c1")
	=COPY()
	=SELECT("R3c88")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("R[1]c")
	=FORMULA("=(r[-1]c-r[-2]c)")
Left_leg_weight_trans_time	=ROW(ACTIVE.CELL())
	=SELECT("Rc[-1]")
	=FORMULA("weight tran time")
	=FONT.PROPERTIES(,,,,,,,,,3)
left leg ankle ang fz1 contact	=SELECT("r"&leftlead_heeloff-120&"c41")
	=FORMULA("=average(r"&l_initiation_of_movement-20&"c[-1]:r"&l_initiation_of_movement-120&"c[-1])")
L_average_ankle_ang	=ROW(ACTIVE.CELL())
	=SELECT("Rc[1]")
	=FORMULA("aver ank ang")
	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("r"&start_FP1-1&"c40")
	=COPY()
	=SELECT("r"&leftlead_heeloff-119&"C41")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("R[1]c")
	=FORMULA("=(r[-2]c-r[-1]c)")
leftleg_ankleang_fz1_contact	=ROW(ACTIVE.CELL())
	=SELECT("Rc[1]")
	=FORMULA("ankang fz cont")
	=FONT.PROPERTIES(,,,,,,,,,3)

max ankle ang after FZ cont	=SELECT("r"&start_FP1&"c41")
	=FORMULA("=max(r"&start_FP1&"C40:r"&End_FP1&"c40)")
	=COPY()
	=SELECT("r"&leftleg_ankleang_fz1_contact+8&"C41")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("r"&L_average_ankle_ang&"C41")
	=COPY()
	=SELECT("r[9]c")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("R[2]c")
	=FORMULA("=(r[-1]c-r[-2]c)")
left_max_ankflex_after_fzcont	=ROW(ACTIVE.CELL())
	=SELECT("Rc[1]")
	=FORMULA("max aftfz cont")
	=FONT.PROPERTIES(,,,,,,,,,3)
vertical velocity CM Z	=SELECT("r1c52")
	=SELECT("c")
	=COPY()
	=SELECT("r1c178")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("r"&rightlead_heeloff-120&"c179")
	=FORMULA("=average(r"&l_initiation_of_movement-20&"c[-1]:r"&l_initiation_of_movement-120&"c[-1])")
	=FILL.AUTO("RC:R"&End_FP1&"C")
	=SELECT("r"&rightlead_heeloff-120&"c180")
	=FORMULA("=(rc[-2]-rc[-1])")
	=FILL.AUTO("RC:R"&End_FP1&"C")
	=SELECT("r"&rightlead_heeloff-118&"C181")
	=FORMULA("=(r[2]c[-1]-r[-2]c[-1])")
	=FILL.AUTO("RC:R"&End_FP1-5&"C")
	=SELECT("r"&rightlead_heeloff-118&"c182")
	=FORMULA("=(rc[-1]/0.04)")
	=FILL.AUTO("RC:R"&End_FP1-5&"C")
	=SELECT("r"&start_FP1-1&"c182")
L_a/p downward CM z	=ROW(ACTIVE.CELL())
	=SELECT("Rc[1]")
	=FORMULA("cm vel")

	=FONT.PROPERTIES(,,,,,,,,,3)
a/p velocity CM Y	=SELECT("r1c70")
	=SELECT("c")
	=COPY()
	=SELECT("r1c184")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("r"&rightlead_heeloff-120&"c185")
	=FORMULA("=average(r"&l_initiation_of_movement-20&"c[-1]:r"&l_initiation_of_movement-120&"c[-1])")
	=FILL.AUTO("RC:R"&End_FP1&"C")
	=SELECT("r"&rightlead_heeloff-120&"c186")
	=FORMULA("=(rc[-2]-rc[-1])")
	=FILL.AUTO("RC:R"&End_FP1&"C")
	=SELECT("r"&rightlead_heeloff-118&"C187")
	=FORMULA("=(r[2]c[-1]-r[-2]c[-1])")
	=FILL.AUTO("RC:R"&End_FP1-5&"C")
	=SELECT("r"&rightlead_heeloff-118&"c188")
	=FORMULA("=(rc[-1]/0.04)")
	=FILL.AUTO("RC:R"&End_FP1-5&"C")
	=SELECT("r"&start_FP1-1&"c188")
L_a_p_velocity CM Y	=ROW(ACTIVE.CELL())
	=SELECT("Rc[1]")
	=FORMULA("cm vel")
	=FONT.PROPERTIES(,,,,,,,,,3)
lower body extremity stiffness	=SELECT("r"&start_FP1&"c52")
locate fz1 contact on cmz	=COPY()
	=SELECT("r3c105")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("r"&start_FP1&"c53")
	=FORMULA("=min(r"&start_FP1&"c52:R"&left_highest_knee_ang_aft_fzcont&"c52)")
	=COPY()
	=SELECT("r4c105")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("r[1]c")
	=FORMULA("=(r[-2]c-r[-1]c)")
	=SELECT("Rc[1]")

	=FORMULA("cmz fz cont")
	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("Rc[1]")
	=FORMULA("- cmz end of cont")
	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("r5c105")
	=COPY()
	=SELECT("r15c105")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("r[1]c")
	=FORMULA("=max(r"&start_FP1&"c52:R"&Left_max_knee_flex&"c52)")
	=SELECT("r[1]c")
	=FORMULA("=(r[-1]c/r[-2]c)")
L_lower_body_extrem_stiffness	=ROW(ACTIVE.CELL())
	=SELECT("Rc[1]")
	=FORMULA("l.b.e.stiff")
	=FONT.PROPERTIES(,,,,,,,,,3)
velocity of Left ankle at fz contact	=SELECT("r1c40")
	=SELECT("c")
	=COPY()
	=SELECT("r1c148")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("r"&rightlead_heeloff-120&"c149")
	=FORMULA("=average(r"&l_initiation_of_movement-20&"c[-1]:r"&l_initiation_of_movement-120&"c[-1])")
	=FILL.AUTO("RC:R"&End_FP1&"C")
	=SELECT("r"&rightlead_heeloff-120&"c150")
	=FORMULA("=(rc[-2]-rc[-1])")
	=FILL.AUTO("RC:R"&End_FP1&"C")
	=SELECT("r"&rightlead_heeloff-118&"C151")
	=FORMULA("=(r[2]c[-1]-r[-2]c[-1])")
	=FILL.AUTO("RC:R"&End_FP1-5&"C")
	=SELECT("r"&rightlead_heeloff-118&"c152")
	=FORMULA("=(rc[-1]/0.04)")
	=FILL.AUTO("RC:R"&End_FP1-5&"C")
	=SELECT("r"&End_FP1+1&"c152")

	=FORMULA("=min(r"&start_FP1&"c152:R"&highest_knee_ang_aft_fzcont&"c152)")
vel Lank	=ROW(ACTIVE.CELL())
velocity of Left Knee	=SELECT("r1c33")
	=SELECT("c")
	=COPY()
	=SELECT("r1c153")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("r"&leftlead_heeloff-120&"c154")
	=FORMULA("=average(r"&l_initiation_of_movement-20&"c[-1]:r"&l_initiation_of_movement-120&"c[-1])")
	=FILL.AUTO("RC:R"&End_FP1&"C")
	=SELECT("r"&rightlead_heeloff-120&"c155")
	=FORMULA("=(rc[-2]-rc[-1])")
	=FILL.AUTO("RC:R"&End_FP1&"C")
	=SELECT("r"&rightlead_heeloff-118&"C156")
	=FORMULA("=(r[2]c[-1]-r[-2]c[-1])")
	=FILL.AUTO("RC:R"&End_FP1-5&"C")
	=SELECT("r"&rightlead_heeloff-118&"c157")
	=FORMULA("=(rc[-1]/0.04)")
	=FILL.AUTO("RC:R"&End_FP1-5&"C")
	=SELECT("r"&End_FP1+1&"c157")
	=FORMULA("=max(r"&start_FP1&"c157:R"&highest_knee_ang_aft_fzcont&"c157)")
vel Lknee	=ROW(ACTIVE.CELL())
CM positioning at instant of landing	=SELECT("r"&start_FP1&"c58")
	=COPY()
	=SELECT("r1c68")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("Rc[1]")
	=FORMULA("L toe y")
	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("r"&start_FP1&"c70")
	=COPY()
	=SELECT("r2c68")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("Rc[1]")
	=FORMULA("cm y")

	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("r"&start_FP1&"c64")
	=COPY()
	=SELECT("r3c68")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("Rc[1]")
	=FORMULA("r toe y")
	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("r5c68")
	=FORMULA("=(r[-2]c-r[-4]c)")
	=SELECT("Rc[1]")
	=FORMULA("base of sup")
	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("r6c68")
	=FORMULA("=(r[-3]c-r[-4]c)")
	=SELECT("r7c68")
	=FORMULA("=((r[-1]c/r[-2]c)*100)")
l_CM position instant landing	=ROW(ACTIVE.CELL())
	=SELECT("Rc[1]")
	=FORMULA("% step dist")
	=FONT.PROPERTIES(,,,,,,,,,3)
Knee Drop	=SELECT("r"&l_initiation_of_movement-120&"c209")
	=FORMULA("=ATAN2(rc[-8]-rc[-7],rc[-34]-rc[-44])")
	=FILL.AUTO("RC:R"&start_FP1&"C")
	=SELECT("r"&l_initiation_of_movement-120&"c210")
	=FORMULA("=(rc[-1])*180/pi()")
	=FILL.AUTO("RC:R"&start_FP1&"C")
	=SELECT("r"&l_initiation_of_movement-120&"c211")
	=FORMULA("=average(r"&l_initiation_of_movement-120&"c[-1]:r"&l_initiation_of_movement-20&"c[-1])")
	=COPY()
	=SELECT("r"&l_initiation_of_movement-119&"c211")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=FILL.AUTO("RC:R"&start_FP1&"C")
	=SELECT("r"&l_initiation_of_movement-120&"c212")
	=FORMULA("=(rc[-2]-rc[-1])")

	=FILL.AUTO("RC:R"&start_FP1&"C")
	=SELECT("r"&l_initiation_of_movement-118&"c213")
	=FORMULA("=(r[2]c[-1]-r[-2]c[-1])")
	=FILL.AUTO("RC:R"&start_FP1-5&"C")
	=SELECT("r"&l_initiation_of_movement-118&"c214")
	=FORMULA("=(rc[-1]/0.04)")
	=FILL.AUTO("RC:R"&start_FP1-5&"C")
	=SELECT("r"&l_initiation_of_movement&"c176")
	=FORMULA("=max(r"&l_initiation_of_movement&"c175:R"&start_FP1&"c175)")
	=FILL.AUTO("R"&l_initiation_of_movement&"C176:r"&start_FP1&"c176")
	=SELECT("r"&l_initiation_of_movement&"c177")
	=FORMULA("=(rc[-2]-rc[-1])")
	=FILL.AUTO("RC:R"&start_FP1&"C")
	=FORMULA.FIND("0",2,1,1,1,FALSE)
	=SELECT("rc[-2]")
L highest point of knee	=ROW(ACTIVE.CELL())
	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("r"&L_highest_point_of_knee+15&"c215")
	=FORMULA("=abs(rc[-1])")
	=FILL.AUTO("RC:R"&start_FP1-5&"C")
	=SELECT("r"&L_highest_point_of_knee+15&"c216")
	=FORMULA("=min(r"&L_highest_point_of_knee+15&"c215:R"&start_FP1&"c215)")
	=FILL.AUTO("RC:R"&start_FP1-5&"C")
	=SELECT("r"&L_highest_point_of_knee+15&"c217")
	=FORMULA("=(rc[-2]-rc[-1])")
	=FILL.AUTO("RC:R"&start_FP1-5&"C")
	=FORMULA.FIND("0",2,1,1,1,FALSE)
	=SELECT("rc[-3]")
L peak swing	=ROW(ACTIVE.CELL())
	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("R"&End_Block&"c1")
	=SELECT("R[2]c[15]")
	=FORMULA("L h knee")
	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("r"&L_highest_point_of_knee&"c175")

	=CUT()
	=SELECT("R"&End_Block&"c1")
	=SELECT("R[3]c[15]")
	=PASTE()
	=COPY()
	=SELECT("r"&L_highest_point_of_knee&"c175")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("R"&End_Block&"c1")
	=SELECT("R[2]c[16]")
	=FORMULA("L peak swing")
	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("r"&L_peak_swing&"c175")
	=CUT()
	=SELECT("R"&End_Block&"c1")
	=SELECT("R[3]c[16]")
	=PASTE()
	=COPY()
	=SELECT("r"&L_peak_swing&"c175")
	=PASTE.SPECIAL(3,1,FALSE,FALSE)
	=SELECT("R[2]c[17]")
	=FORMULA("l.kneedrop")
	=FONT.PROPERTIES(,,,,,,,,,3)
	=SELECT("R[1]c")
	=FORMULA("=(rc[-2]-rc[-1])/10")
l knee drop	=ROW(ACTIVE.CELL())
	=FONT.PROPERTIES(,,,,,,,,,3)
	=ACTIVATE(File_Name_text)
	=SELECT("r2c4")
	=COPY()
	=CLOSE()
	=ACTIVATE(File_Name_Vel)
	=SAVE()
	=CLOSE()

	=IF(Loop_1=90,GOTO(endloop2),GOTO(endloop1))
	=SELECT("c104")
	=CLEAR()
	=ACTIVATE(File_Name_text)
	=SELECT("r2c4")
	=COPY()
	=CLOSE()
	=ACTIVATE(File_Name_Vel)
	=SAVE()
	=CLOSE()
	=IF(Loop_1=90,GOTO(endloop2),GOTO(endloop1))
endloop2	endloop2
	=SELECT("Loop_1")
	=FORMULA(Loop_1-90)
	=IF(loop_2=9,GOTO(end),GOTO(nextsubject))
next subject	nextsubject
	=SELECT("loop_2")
	=FORMULA(loop_2+1)
	=GOTO(endloop1)
endloop1	endloop1
	=SELECT("Loop_2")
	=FORMULA(loop_2-1)
	=GOTO(start)
end	end
	=RETURN()

Appendix 6: Chapter 7, output from t-test comparing no awareness or experience trials (pre-test 1) and awareness + no experience trials (main test)

no awareness vs. awareness + no experience

Variable	P value	Explanation
%Kneedrop	0.200	
ank ang fz cont	0.880	
Vel CM fz cont a/p	0.710	
Bdywt sup fz cont	0.360	
Bdywt sup end land	0.110	
CM position % step	0.430	
CM Vel initial land m/l	0.760	
CM Vel initial land a/p	0.590	
DS Time	0.910	
Knee ang fz cont	0.140	
Kneedrop	0.070	
CM vel fz cont M/L	0.730	
PVCF	0.140	
Ss Time	0.170	
Step Distance	0.520	
Time to PVCF	0.890	
Ankle ang initial land	0.560	
Knee ang initial land	0.240	
Vel ankle initial land	0.270	
Vel knee initial land	0.140	
Vertical CM vel fz cont	0.200	
Weight Tran Time	*0.051*	

Appendix 7: Chapter 7, output from t-test comparing no awareness or experience trials (pre-test 1) and awareness + experience trials (main test full vision trials)

no awareness vs. awareness + experience full vision

Variable	**P value**	**Explanation**
%Kneedrop	0.260	
ank ang fz cont	0.790	
Vel CM fz cont a/p	0.950	
Bdywt sup fz cont	0.100	
Bdywt sup end land	*0.070*	
CM position % step	0.980	
CM Vel initial land m/l	0.170	
CM Vel initial land a/p	0.770	
DS Time	0.550	
Knee ang fz cont	*0.070*	
Kneedrop	*0.070*	
CM vel fz cont M/L	0.370	
PVCF	0.150	
Ss Time	**0.010**	reduced SS time in awa+exp condition
Step Distance	0.540	
Time to PVCF	0.230	
Ankle ang initial land	0.870	
Knee ang initial land	0.280	
Vel ankle initial land	0.210	
Vel knee initial land	0.940	
Vertical CM vel fz cont	0.150	
Weight Tran Time	*0.070*	

Appendix 8: Chapter 7, output from x 3 vision condition (full vision (FV) trial immediately before visual occlusion, visual occlusion trial, FV trial immediately after visual occlusion) x 2 test (early, late) ANOVA

Visual occlusion vs. FV trial immediately before and immediately after; early and late test comparison

Variable	Vision	post hoc	Test	post hoc	Interaction
%Kneedrop	0.754		0.282		0.71
ank ang fz cont	0.873		0.212		0.28
Vel CM fz cont a/p	**0.007**	2vs1&3- 2 reduced	*0.065*		0.67
Bdywt sup fz cont	0.49		0.18		0.14
Bdywt sup end land	0.198		**0.013**	reduced late	0.31
CM position % step	**0.035**	2vs1&3- 2 reduced	*0.059*		0.8
CM Vel initial land m/l	**0.010**	2vs1&3- 2 reduced	0.882		0.44
CM Vel initial land a/p	**0.006**	2vs1&3- 2 reduced	**0.007**	Increased anterior late	0.99
DS Time	0.842		0.141		0.07
Knee ang fz cont	*0.071*		0.145		0.19
Kneedrop	0.483		0.282		0.77
CM vel fz cont M/L	**0.021**	2vs1&3- 2 reduced	0.362		0.2
Net impulse entire	0.410		0.241		0.56
PVCF	0.764		0.128		0.56
SS Time	0.643		**0.047**	reduced late	0.36
Step Distance	0.481		0.708		0.31
Time to PVCF	0.275		0.919		0.38
Ankle ang initial land	**0.006**	2vs1&3- 2 reduced	0.147		0.7
Knee ang initial land	**0.003**	2vs1&3- 2 reduced	*0.072*		0.77
Vel ankle initial land	0.333		*0.082*		0.88
Vel knee initial land	0.171		**0.050**	increase late	0.59
Vertical CM vel fz cont	0.948		0.452		0.33
Weight Tran Time	0.320		**0.020**	reduced late	0.26

Key

Vision
1-FV trial immediately before visual occlusion
2-Visual occlusion
3-FV trial immediately after visual occlusion

Test
Early-FV trial immediately before and after 1st visual occlusion and 1st visual occlusion trial
Late- FV trial immediately before and after last visual occlusion and last visual occlusion trial

Appendix 9: Chapter 7, full vision trials with a high probability (67 %) of visual occlusion compared to trials in which there was zero probability of visual occlusion

High vs. Zero		
Variable	**P value**	**Explanation**
Ankle angle	0.709	
Bdywt support fz cont	0.780	
Bdywt support end of landing	0.553	
CM % step dist	0.376	
Double support	0.764	
Knee angle fz cont	0.103	
kneedrop	0.224	
Lower body extremity stiffness	0.627	
Ankle angle initial landing	0.920	
Knee angle initial landing	0.439	
pvcf	0.568	
SS time	**0.027**	quicker in zero
step distance	0.257	
step time	0.011	
time to pvcf	0.854	
% kneedrop	**0.014**	later in zero
CM a/p velocity fz cont	0.264	
CM vertical velocity fz cont	0.633	
Angular velocity ankle initial land	0.593	
Angular velocity Knee initial land	0.853	
WT time	0.432	